

Force Generation and Decay
of
Elastometric Ligatures of Varying Sizes and Configurations

by
Peter C. Wagner

OHSU School of Dentistry
Department of Orthodontics

June 1995

WU4
W134
1995

Force generation and decay of elastomeric ligatures of varying sizes and configurations. Wagner, Peter C. Certificate Thesis, OHSU Department of Orthodontics, Portland, Oregon.

The effect that elastomer size and configuration have on force generation and force decay was evaluated. Ormco Power O's in sizes 090, 110, and 120 were placed in one of three configurations around orthodontic brackets engaging a rigid archwire segment. Conventional, figure-eight, and double (one elastomer per wing) were the three configurations studied. Using an Instron Universal Testing Machine the archwires were distracted from the bracket slot base and the force was measured. Measurements were made at intervals over an eight week period. Samples were stored in distilled water at 37° C. A statistical analysis was completed which included ANOVA's and Newman-Keuls tests. Over time Ormco Power O's experienced force decay with the majority occurring in the first day. Percent force decay was least for those elastomers stretched the least, and mean force was greater with increased initial stretch of an elastomer. Figure-eighted elastomers generated more force than conventionally applied elastomers, and doubled elastomers generated forces between those produced by conventional and figure-eighted elastomers. Force generation decreased in the elastomer size order 110 > 090 > 120 for both figure-eight and conventional samples over the course of this study. It was concluded that elastomer material mass, elastomer dimension, elastomer stretch with application, and friction may play a role in force generation and its measurement. The figure-eight elastomeric configuration may enhance light archwire engagement and resist forces which cause displacement of the archwire from the bracket slot.

Dental

CONTENTS

Acknowledgements	ii
Introduction	1
Literature Review	3
Materials and Methods	18
Results	20
Discussion	21
Clinical Implications	25
Conclusions	27
Bibliography	28
Appendix	
A. Figures 1-10	31
B. Figures 11-16	36
C. Tables I - IV	42
D. Statistics	44

ACKNOWLEDGEMENTS

I would like to thank the efforts of the following individuals who contributed to the completion of this project:

Dr. Larry Doyle	Dr. Jack Ferracane
Jerry Adey	John Condon
Dr. Dave Phillips	Dr. Dave May
Cathey Schroeder	

INTRODUCTION

Chain elastomers have been the focus of numerous studies, both published and unpublished.¹⁻²³ The effects that environment, additives, prestretching, and manufacture have upon force delivery and force decay have been evaluated. The primary use of chain elastomers is for individual tooth translation and space closure in conjunction with orthodontic treatment; therefore, most researchers related the measurements they obtained to proposed forces necessary for retraction of a canine tooth. Occasionally the opposing effects of friction between the archwire and the bracket/ligature unit were also taken into consideration and discussed. With the vast body of knowledge that is available with regard to the properties of chain elastomers, it is surprising to find how little information is available with regard to individual elastomeric ligature modules.

The single elastomeric ligature module is probably the simplest, most utilized item in modern orthodontic practice. It is simply a small, donut-shaped, polyurethane "rubber band" that engages the archwire to the bracket by locking behind overhanging projections, or "wings." It is taken for granted that once applied, the elastomeric ligature unit holds the archwire intimately against the base of the bracket slot. But does it? Unless the archwire is passive in the slot, some force is necessary to produce the full engagement necessary to arrive at the desired tooth movement. Without full engagement, no or only partial tooth movement may occur. The result is inefficient progression of treatment. Full bracket engagement has become much easier with the advent of resilient shape-memory alloys.

In practice today, clinicians take advantage of the elasticity of the ligature modules by distorting them in different fashions, hoping to improve their ability to achieve full engagement of the archwire in the bracket slot and result in the desired tooth movement. Figure-eighted elastomeric ligature modules are applied to the bracket and archwire in hopes of increasing the force by which the archwire is held to the bracket slot. This configuration is slightly more difficult to place and remove

than a conventionally placed elastomeric ligature, and the elastomer must be stretched, giving the clinician a sense of enhanced force. But is more force actually generated, and if the force is enhanced does it retain such a relationship throughout the treatment period?

Another approach utilized to increase the force produced by elastomeric ligatures is to use those of a smaller lumen size since, again, they "feel" like they hold the archwire more tightly to the bracket than larger elastomeric ligature modules since they must be stretched farther during placement. To accomplish just the opposite, that is, reduce the amount of force applied to the archwire and improve patient comfort, some have utilized a single elastomeric unit on each wing of a twin bracket. The thought is that the elastomers do not feel as tight during placement so the elastomer must be exerting less force, and the archwire must have more freedom from the bracket slot.

Currently, there is no published research relating elastomeric ligature tension to force generation against the archwire. The purposes of this study are: 1) to determine if elastomeric ligature modules placed in a figure-eight configuration deliver more force to the archwire than conventionally applied elastomeric ligatures, 2) to determine if single conventionally applied elastomeric ligatures produce more force than two elastomers of the same size applied to the wing of a bracket, and 3) to evaluate the effect that the size of the elastomer has on its ability to generate force in each of the three configurations under investigation. If differences in force generation are established, the ability of an elastomer to maintain this advantage over time will be evaluated.

LITERATURE REVIEW

Andreasen and Bishara¹ were among the first to evaluate the behavior of the new Alastik material and describe its force ranges under clinical use. When compared to latex elastics, Alastik chains were felt to be less effective clinically in retracting canines than in closing generalized or anterior spaces, and they deformed and discolored over time. To evaluate these observations, the authors stretched two sizes of elastics and two strengths of Alastik chain at 10 mm intervals from 65 to 105 mm distances. From thirty extraction and non-extraction cases 65 mm was determined to be the minimum arch length or distance the elastomers might be stretched in an extraction case and 105 mm was the maximum arch length. Forces were measured at the length stretched at intervals over three weeks. Samples were stored in one of six combinations of conditions that included water, saliva, dry air, room temperature, and 37° C.

Samples were found to test best in water at 37° C, although statistically no difference existed with those tested in saliva at 37° C. Wetness was found to decrease the measured forces relative to dry conditions. Most of the force decay which occurred happened within 24 hours with Alastiks losing 74% of their force and the two elastic sizes about 42%. Most force decay occurred in the first day. After 24 hours the Alastiks had only 8% further loss of force while the elastics experienced 5% loss of force over the next three weeks. The greatest percent of force decay occurred within one hour (56% for Alastiks and 27% for elastics). The Alastik chains delivered greater force than the elastics throughout the experiment, even after the 74% force decay after 24 hours. At three weeks force values were still 90-154 grams for Alastiks (down from initial forces of 263-567 grams) while elastics delivered only 48-74 grams (down from initial forces of 87-198 grams). Heavy Alastik chains displayed higher initial force values, but they also experienced greater force decay so that by the end of the experiment the lighter chains produced more force. The greater the initial extension of both Alastiks and elastics, the larger the initial force. After three weeks elastics continued to exhibit this relationship while the trend was less defined for

the Alastiks. The Alastik material was found to permanently deform by 50% of its length compared to only 23% for elastics. Deformation increased with time and distance stretched. Both materials were found to stain. Three times higher variability was observed in the Alastik materials than in the elastics. Andreasen and Bishara felt the Alastiks offered several advantages, including their ligature-like placement over the brackets and archwire and the requirement for no patient cooperation as with elastics. They also suggested using Alastiks with four times the initial desired force so that after the first day of use the force would be at the required level.

In the same year Bishara and Andreasen² similarly evaluated and compared Alastiks of three sizes and elastics of many lengths and forces. The materials were stretched in 6 mm increments from 22 to 40 mm in 37° C water to simulate the distances clinically encountered when utilizing elastomers in class II and III applications. As before, force measurements were made at intervals over a three week period. As before, most of the force decay occurred in the first day, and both materials experienced changes in color. After 24 hours the Alastiks had experienced force decay of 55% while elastic forces decayed about 17%. By the end of the study, Alastik forces had dropped 67% compared with 25% for elastics. Again, the Alastik material experienced greater permanent deformation which increased with the distance stretched. Forces generally increased as the distances that the elastomers were stretched increased, but this relationship was not linear. The authors suggested that the initial discomfort expressed clinically when using Alastiks disappeared due to the large initial force decay. They cautioned that should force drop excessively then tooth movement may not occur, so clinicians should take into account the potential force decay in determining initial force levels applied to the teeth. Also, elastics might not be changed daily since their force delivery was relatively constant.

Hershey³ studied force decay of elastomeric modules under conditions simulating tooth movement rather than at a fixed elongation. Influence of initial force magnitude on the force decay curves was also evaluated as was variability within modules and manufacturers (TP,

Ormco, and Unitek). On a metal frame capable of being approximated by a threaded mechanism, elastomeric chain of several manufacturers was initially stretched varying distances in two millimeter increments between 12 and 34 mm. The samples were stored in 37° C water during the six week course of the experiment, and the frame was closed .25 mm or .50 mm per week to simulate tooth movement. Force levels were measured at intervals with a Carpo gauge. From the measurements a statistical analysis was completed.

Initial force levels were approximately 300 and 500 grams. All modules lost about 50% of their initial force in 24 hours, and they lost about 60% of their initial force at four and six weeks at a zero rate of space closure. One third of the initial force was retained in modules closed .25 mm per week while those closed .5 mm per week retained only one quarter of their initial force. Percent force loss appeared independent of initial stretch. As expected there was noted variation between modules of the same type, but products were similar between manufacturers. Though force decay of some samples approached 75%, the remaining force was felt sufficient to elicit the tooth movement reaction.

Kovatch, et al,⁴ studied Unitek's K2 regular Alastik modules with respect to load-extension and load decay relative to the rate and degree of extension. An Instron Universal Testing Machine was utilized to test the modules under a variety of conditions. The modules were stretched to breaking at room temperature at crosshead speeds of .2, 2, and 20 inches per minute. To simulate clinical oral use the units were also stored one week at 37° C in saliva and stretched one inch at the same crosshead speed, maintained several weeks, and the force decay monitored. Eight specimens were tested under each set of variables. They found that rapidly stretched modules initially displayed higher forces, but after 1.5 inches of extension there was a crossover so that they failed at lower forces than those that were more slowly stretched. For the one inch extensions the initial forces for the rapidly stretched samples was higher but the force decay was initially much more rapid so that the slowly stretched modules generated more force after the first minute. The authors suggested that elastomers be stretched slowly into place to take

advantage of the slightly enhanced force generation, even though it may not prove clinically significant.

Varner⁵ attempted to quantify the force generation and force decay of KX2R elastomeric units over the course of a typical interappointment time period. Randomly chosen KX units were stretched on brass jigs between 26 and 40 mm in 2 mm increments with 10 samples per elongation. Forces were measured with an Instron at intervals over a four week span, and the data were statistically analyzed. Samples were stored at 37° C in water. It was found that although up to 38% variation was measured at the initial measurements, this variance decreased markedly with time. After 24 hours forces generated ranged from .861-1.437 pounds, while at four weeks the range was .820-1.165 pounds. At all elongations the rate of force decay was not significantly different. A 26% decrease in force production occurred between two hours and four weeks with most occurring before one week. It was concluded that with the relatively constant force levels of approximately one pound at four weeks that clinically significant tooth movement could result, and perhaps the KX2R units did not require replacement for perhaps another four weeks.

Brantley, et al,⁶ investigated the effects of prestretching elastomeric modules. Alastik C chain andOrmco Power Chain II were both tested. The chains were extended to about 100% elongation on a metal framework, providing initial forces of 300-500 grams. Control chains were not prestretched, while those prestretched were either prestretched for 24 hours or three weeks. Rest periods of zero to one week followed prestretching prior to force measurements over a three week period. The units were stored at 37° C in water or in air at room temperature. The results showed that although Unitek chains had higher initial forces, the remaining force at three weeks was greatest for the Ormco chain. Most force decay occurred within the first hour. Prestretching in water for three weeks and 24 hours resulted in only a 10 and 35 gram loss of force, respectively, during the three weeks of testing. Prestretching in air was less effective. Longer rest periods between prestretching and testing resulted in loss of prestretching effects by 30-40%. Although prestretched specimens exhibited less force decay during testing, their

initial forces following prestretching were significantly less than units which had not been prestretched. After three weeks the remaining force in the control and the prestretched chains were essentially the same, so that prestretching resulted in the production of a more constant force over the period, not an elevated force.

Doyle⁷ studied both the force production and the rate of force decay of C1 Alastik chain modules. Utilizing a Plexiglas model simulating an extraction scenario with adjacent molar and second premolar bands, a seven mm extraction site, and a canine band, forces were measured with a hand held gauge at intervals over a four week period. Four unit chains were placed on both the buccal and lingual of the test models on all but one trial set. Trials were run with and without an archwire in place, and the models were stored at 37° C in water. Standard error of the measure was determined to be 3.8%. From an initial mean force of nearly three pounds, the mean force decayed 35% in two hours and 49% after four weeks. Forces measured from models without an archwire were slightly lower, apparently due to decreased friction. Samples cooled to room temperature after two hours in 37° C water yielded force measurements exceeding their prior measurements by .14 pound and .32 pound after ten and thirty minutes, respectively. For a single chain the rate of force decay was found to be 40% at two hours and 46% after one week. It was concluded that temperature influenced the rate of force decay for C1 Alastik chain. Also the minimal 5-15% force decay occurring between 24 hours and four weeks experienced by this investigator and others did not necessitate replacement of Alastik chains prior to three or four weeks, since they still produced sufficient force for tooth movement to occur.

Degenova, et al,⁸ investigated force degradation of three commercially available elastomeric chains -Ormco Power Chain II, RMO Energy Chain, and TP Elast-O-Chain. In addition, the effects that thermocycling, initial load, and simulated tooth movement (decreasing extension) had on the properties of the chains were evaluated. Short and long chains of each product were extended slowly to 20 mm on either a sliding jig to simulate tooth movement with space closure or a fixed jig. Samples were stored in Oralube synthetic saliva. Half of the samples

were maintained at 37° C and the other half cycled between 15° C and 45° C. For samples undergoing decreased extension, three decreases of .5 mm were used at 24 hours, 7 days, and 18 days. Force measurements were made utilizing an electronic force gauge at intervals over three weeks. A statistical analysis was completed. The authors found that the thermocycled samples retained a significantly higher percentage of force at 21 days than the samples stored at 37° C (42.5-60.8% vs. 39.1-56%). Since thermocycling favored elevated force readings, the remainder of the tests used the data compiled from thermocycled samples. RMO E chain experienced less force decay than TP Elast-O-Chain, and Ormco power chain experienced the most force decay. Short chains generated greater initial forces, and they still produced significantly more force at three weeks than the longer chains. Short chains also retained a higher percentage of initial force. Simulated tooth movement significantly reduced the force measurements obtained from the extended chains at 14 and 21 days.

The new synthetic Energy chain manufactured by RMO was compared to original plastic chain produced by American Orthodontics by Killiany and Deplessis⁹ in order to distinguish any real advantage of the "improved" material. Equal lengths of each chain was stretched to twice its original length and stored in 37° C saliva. Measurements were made with an Instron Universal Testing Machine at intervals out to eight weeks. The results showed that the new energy chain, though having less force initially, experienced less force decay. At all measurement periods beyond the initial measurement it exhibited much greater force than the plastic chain. At eight weeks the plastic chain retained only 27% of its initial force while Energy chain retained 55% of its original force. The authors felt that this new generation of elastomeric chain offered promise for clinical use.

Rock, et al,¹⁰ set out to investigate the force/extension characteristics of thirteen commercially available elastomeric chains. Chains were cut into two, three, and four loop sections and tested in tension on an Instron Universal Testing Machine. The chains were extended at 20 mm per minute, and the force was recorded on the chart,

yielding a force-extension curve. The data showed that most specimens developed 4-5 Newtons (N) of force at 100% extension. The relationship between force and extension was also not linear, that is, the initial part of the curve was linear until a transition occurred at which point the curve flattened before increasing again. Although the curve again steepened, it never reached the height of the initial linear part of the curve if it had been extrapolated. The authors concluded that an extension of 50-70% would be adequate to produce the desired tooth moving force of about 3 N.

Rock, et al,¹¹ then looked at the effect that four weeks in the oral environment had upon the properties of eight of the previously tested chains. Chains of three and four loops were used to close first premolar extraction spaces in association with edgewise mechanics with medium twin brackets and a flat .016" Australian wire arch. The chains had been stretched 55-140% of their initial length to produce forces ranging from 2.9-5.9 N. At four weeks the elastomer stiffness was reduced 34-67% (degradation), and the forces exerted reduced 20-59% (tooth movement and degradation). Tooth movement ranged from 0-5.1 mm. It was concluded that all materials retained sufficient integrity to produce forces capable of physiologic tooth movement.

Wolfgang and Droschle¹² evaluated the force decay of orthodontic elastics relative to time and extension. Four sizes of elastics in groups of ten were stretched at different distances in 2 mm increments from 13-35 mm and stored in 37° C saliva. Force measurements using a spring scale were taken immediately and at 30, 60, 180, and 480 minutes. They showed that there was a significant loss of force over the first half hour especially, with no significant loss of force between 180 and 480 minutes. The authors suggested measuring the intraoral extension required of an elastic and referencing a chart they produced to arrive at the estimated force generation after the initial large force decay.

Nordberg¹³ evaluated individual elastomeric ligating modules in order to quantify the force magnitude that would be applied to the archwire by the elastomer and to determine if the clinical impression of

"stiffness" correlated with the actual level of force. As an aside, a few samples were tied in a figure-eight configuration, and their force magnitudes measured. Modules from five different manufacturers were stored in 37° C water, and forces were measured with an Instron Universal Testing Machine at intervals over a six week period. The standard error of measure was determined to be 2.5%. He found that conventionally applied modules lost 34% of their initial force in 24 hours compared to a loss of 43% for the figure-eight samples. At four weeks 41 and 56% force decay was experienced, respectively. Figure-eight samples had significantly elevated force values; however, it was not felt that the difference would clinically enhance tooth movement. The minimum force generated by the figure-eight sample and three of the five conventionally applied samples exceeded 360 grams, which would allow engagement of an .016 nickel titanium archwire displaced .020" from the bracket slot. ACo. and GAC modules were found superior to those of RMO and Ormco with Unitek being intermediate. Initial module stiffness did not correlate with the force magnitudes measured at increasing intervals as the forces generated by those modules deemed stiff fell below those which were less stiff at the time of placement.

Citing inconsistent force levels, water absorption, discoloration, and bulk as problems plaguing existing elastomeric chains, Brown¹⁴ compared Unitek Alastik chain with the new generation RMO Energy chain and GAC Chainette. Utilizing an Instron Universal Testing Machine she measured the force generated by each chain at four different elongations at intervals over four weeks. Samples were stored in a 37° C water bath. A statistical analysis was completed, and standard error of measure was estimated at 1.1%. She found that Unitek chain imparted less force initially and it decayed 46-55% over 24 hours compared to approximately 30% for the other chains. Overall, Unitek Alastik chain experienced the greatest percent force decay, and percent force decay decreased with increased elongation. GAC Chainette experienced force decay nearly independent of percent elongation, and force decay for RMO Energy chain increased with increased elongation. From the data collected regression line equations were produced to allow prediction of the final force values from initial measurements and vice versa. It was noted that force decay

must be taken into account when activating elastomeric chain due to the large initial rate of decay; however, increased initial force levels can be associated with pain as well as PDL ischemia which might result in undermining resorption rather than frontal resorption.

Huget, et al,¹⁵ evaluated the elasticity and degradation of a synthetic elastomer.Ormco Grey Power Chain II segments were stored in distilled water at 37° C for periods of up to seventy days. A dry control was utilized. An Instron Universal Testing Machine was used to determine loading and unloading curves, used in determining instantaneous elastic recovery (IER), permanent set (PS), and delayed recovery (DR). Also, a gas chromatograph mass spectrometer was used to determine the presence of organic compounds in the storage water. A two way ANOVA revealed that the extensions, distance and water storage duration affected load requirement, IER, PS, and DR. Interactions of the extension distance and storage duration affected load requirement, IER, and PS, but not DR. Organic materials were detected on the fourteenth day and thereafter, suggesting that exposure to water (absorption) first weakened the intermolecular forces and subsequent chemical degradation caused release of organic molecules from the elastomer.

Lu, et al,¹⁶ evaluated the relaxation of elastomeric chain over time, comparing force decay of three stretched lengths and grey versus clear chains. RMO clear chains and American Orthodontic clear and grey chains were tested. RMO E chains of six units in length were stretched either 30, 35, or 40 mm over a metal framework for testing of force decay. To compare clear with grey chain, both American chains of eight units in length were similarly stretched to 40 mm. Weekly decreases in extension of .5 mm simulated tooth movement. All force measurements were made with a dial type dynamometer at intervals over a six week period. Samples were stored in water at 37° C and pH 7. Statistical analysis of the data was accomplished. The results showed that initial force and the force remaining were significantly greater for greater extension of the chains (40>35>30), but the percentage of remaining force had just the opposite relationship with samples stretched the least exhibiting a larger percentage of remaining force. The most force decay occurred within the

first hour. Transparent chains displayed significantly larger remaining forces, and their percentages of remaining forces at all time periods were also greater. For all samples, the most force decay occurred within the first hour, and relaxation continued at a slower rate throughout the experiment. Remaining force and percentage of remaining force for RMO clear E chain was significantly more than that of either of the American Orthodontics chains. Based on previous studies on tooth movement, the authors concluded that RMO E chain stretched 40 mm would be effective for retracting a canine for three weeks before the chain would need to be replaced.

More recently, Baty, et al,¹⁷ have investigated the force delivery capabilities and dimensional stability of the more recently developed colored elastomeric chains. For the force delivery study groups of five chain specimens of four differently colored four loop chains from three manufacturers were suspended from hooks on a universal testing machine and the distraction required to adequately generate either 150 or 300 grams measured. Testing was conducted at intervals over three weeks. Samples were stored either in air, distilled water, or artificial saliva at 37° C, and their length, width, lumen diameter, and thickness measured weekly.

Due to the large number of observations, small differences were shown to be statistically significant when subjected to three and four factor ANOVAs and Bonferroni simultaneous t-tests. Arbitrarily, a ten percent difference was determined to be a clinically significant difference. From the data collected it was found that an extension of 52-57% would still yield forces capable of tooth movement even with force decay taken into account. All chains tested appeared capable of providing the necessary force for tooth movement to occur. The amount of distraction did vary between manufacturers, but the colored chains tended to behave as the grey chain of the same source with the exception of theOrmco purple and green chains which required significantly more distraction to yield the required force levels. Although not noted by the authors, there was a tendency in all chains after one week or less for the amount of distraction required to produce 150 grams of force to decrease,

possibly due to stiffening of the chain over time or elastic recovery after the initial, closely spaced tests. All chains experienced significant dimensional changes of less than six percent; however, this was not felt to be clinically significant.

Wong¹⁸ investigated several elastomeric materials (latex and polyurethane) to determine the changes in force and physical properties that may occur in the mouth. He tested the fracture strength, force changes for a given stretch, changes in the modulus of elasticity, and force decay. The force decay and force change experiments consisted of elastomer samples having an initial stretch of 17 mm or an initial force of 300 or 450 grams, respectively. These were stored in a water bath at 98° F, and they were monitored at intervals out to three weeks. Among other findings, the synthetic elastomers experienced up to 73% force decay the first day. One material (Alastik C2 double links) with an initial force of 641 grams exhibited a force of 171 grams at 24 hours, identical to another material, Ormco power chain, which only had an initial force of 342 grams. The greatest amount of force decay occurred within the first three hours for all materials with forces thereafter remaining relatively constant. The modulus of elasticity was shown to decrease substantially when materials were subjected to the water bath conditions. The authors also noted permanent deformation in shape in all materials. Suggestions for clinical use included taking into consideration the force decay of elastomeric materials and prestretching.

The effects of heat and prestretching on elastomeric modules of three manufacturers was evaluated by Brooks and Hershey.¹⁹ Modules were mounted on a device that simulated tooth movement, and force measurements were made at intervals out to four weeks. Those modules subjected to heat (simulating a hot beverage) experienced a 70% decrease in force at one hour and a 80% decrease in force at four weeks when compared to controls. They found that prestretching the modules partially offset the reduction of force, reducing it to only 50% at one hour and 69% at four weeks.

Young and Sandrik²⁰ studied the effects of prestretching on the relaxation of orthodontic elastic chains. Using Unitek CK grey and C2 grey Alastiks, they prestretched the chains 14 or 23 mm and 18, 36 or 48 mm, respectively, before subjecting them to loads of 90 grams at 37° C in distilled water. A group of C2 chains was subjected to 181 grams, and another group was tested in 100% humidity. By 24 hours the CK control had force decay averaging 56%, while the samples prestretched to 23 and 14 mm had force decay of 45% and 49%, respectively. All differences were statistically significant. The C2 control modules experienced average force decay of 44% at 24 hours compared to 47%, 43%, and 43% for prestretches of 48, 36, and 18 mm, respectively. Differences for the C2 chains were all insignificant. Force decay for modules stored at 100% humidity showed no significant differences, but those modules with initial loads of 181 grams experienced 51% force decay which was significant. Based on the essentially linear force decay that occurred after six hours, the authors predicted the remaining force at four weeks for all groups. They calculated that the prestretched CK units would exhibit 64-93% more force than the controls at four weeks, while for the C2 units there would be no significant difference. Differences between the two chains were attributed to differences in their shape and/or polymer composition. The authors suggested that prestretching be accomplished prior to placement by either the operator or manufacturer.

The in vitro testing of elastomeric modules was challenged by Ash and Nikolai,²¹ because mechanical and chemical damage to elastomers that occurred in vivo was not accurately portrayed by the in vitro tests. Alastik CK grey chain and K1 standard modules were stretched on jigs and stored at 37° C in both air and water, and they were ligated to fixed orthodontic appliances intraorally. A hand held tension gauge was used to measure forces at incremental periods out to three weeks. They found that force degradation was significantly more rapid in water and the oral environment than in air for essentially the entire testing period. The force decay for the oral sample became significantly greater than that of the 37° C water sample at one week for the CK Alastik and at two weeks for the K1 Alastik. The authors concluded that the oral environment had

an unfavorable effect on the elastomeric modules, but they were unsure of its clinical significance.

Ferriter, et al,²² evaluated the effect of pH on force degradation of elastomeric chains. Seven commercially available chains were extended to three different distances to allow both comparison of the elastic products at equal distances and equal initial force levels. Samples were stored in either pH 4.95 or pH 7.26 solutions at 37° C for four weeks. Force measurements were made with a Correx gauge at intervals over the four week period. A statistical analysis was completed. The basic solution produced significantly greater decay in the chain elastomers than the acidic solution at four weeks. Three of the four products extended different distances experienced significantly greater force decay for chains extended to a lesser degree in the basic solution.

VonFraunhofer, et al,²³ set out to study the effects of topical fluoride treatments and artificial saliva on the load relaxation and elastic properties of elastomeric chains of three manufacturers (Ormco Generation II Power Chain, Unitek Alastik C1 modules, and TP Orthodontics E Chain). Six test environments were used: air, distilled water, artificial saliva, Gelkam (.4% SnF₂), .31% APF, and .4% KCl. Using a universal testing machine the units were tested in tensile until fracture to arrive at displacements required to generate forces of 150 and 300 grams. Chains were tested at intervals out to four weeks and stored in the medium at 37 C. Force decay was also studied in both prestretched and unstretched chains. Prestretched chains were stretched slowly to twice their length before testing. Force decay was only measured over a 36 minute interval. A statistical analysis was completed. The authors found that permanent deformation occurred when forces exceeded 300 grams. Both artificial saliva and fluoride increased the distraction necessary to produce forces of 150 and 300 grams, but only the differences for distractions of 300 grams were felt to be large enough to be clinically significant. Prestretching the elastomeric chains was not determined to enhance their load relaxation behavior. Also, load relaxation was less for the distilled water and APF groups than for chains exposed to air.

Echols²⁴ studied the frictional forces required to linearly displace .016", .018", .020", and .019" x .025" stainless steel archwire held in an edgewise bracket with elastomeric ligatures. The elastomers had been used intraorally in active orthodontic treatment for two weeks, then they were removed and replaced on the test modules for measurement. The results showed that 50+/-11 grams was required to displace the .016" wire while 121+/-21 grams was required to displace the .019" x .025" archwire. He concluded that larger archwires required greater forces for linear movement, and he suggested that elastomers be avoided where sliding mechanics were being attempted.

Baker, et al,²⁵ studied the effects of a saliva environment on friction of different sizes of archwires tied with elastomeric ligatures. An Instron Universal Testing Machine measured the force required to initiate movement of either a .018", .020", or a .018" x .025" stainless steel orthodontic wire in a perpendicularly mounted .022" x .028" edgewise bracket tied with a size 110 elastomer. Environmental conditions were either dry (control), a saliva substitute and 37° C, or glycerine at 37° C. Measurements were made at three and seven days to allow for maximal relaxation of the ligatures. A statistical analysis was completed. No significant differences existed between the three and seven day measurements. The artificial saliva showed a significant 15-19% reduction in friction compared with the dry controls and glycerine samples. The authors suggested this knowledge might be useful in treating patients with reduced salivary flow.

A study on the friction present between archwire, bracket, and ligature tie was carried out by Sims, et al.²⁶ He compared frictional values between the A Company Activa bracket, the Strike Industries Speed bracket, conventional elastomeric ligation of a Minitwin, and ligation with a figure eight elastomeric tie on Minitwin brackets. Rectangular archwires of four sizes were degreased, placed, and ligated into the brackets. Then they were tested on a vertically mounted Instron Universal Testing Machine. Elastomeric modules were placed immediately prior to testing. Each bracket/archwire combination was tested six times. A statistical analysis was completed. The results showed that the Activa

brackets had the least friction. The Speed brackets exhibited approximately fifteen times more friction. Conventionally ligated Minitwin brackets showed 225-300% more friction than the Speed brackets (except for .016 x .022 archwire), and figure eight ligation increased friction 70-220% over conventional elastomeric ligation. Overall, friction generally increased as wire size increased. The authors did point out that these figures for elastomeric ligation were maximal as force decay over time in the oral environment may reduce force levels as much as 73% in the first day.

MATERIALS AND METHODS

Ormco grey Power O's of three different sizes were utilized for the study: 120, 110, and 090. (See fig. 1.) All units of each size were taken from the same envelope as supplied by the manufacturer to eliminate the product variation occurring between lots. For each sample of ten modules tested, modules from three sticks (50 units per stick) were carefully separated with a Bard Parker blade to avoid uneven tears and unnecessary distortion of the modules. The units were mixed, and ten were randomly selected for study. This was repeated ten times, yielding 100 sample units, 40 of size 120, 20 of size 110, and 40 of size 090. For each size of elastomer ten modules were placed in a conventional fashion, and ten were placed in a figure-eight fashion. This yielded six groups of ten samples each. Also, for the sizes 120, and 090, twenty modules were placed in a double fashion using two modules, one per bracket wing. This contributed two additional groups of ten samples. The samples were stored at 37° C in distilled water after the initial force measurements.

Modules were loaded into a Mathieu style ligature plier (Orthopli) which had a slotted tip to prevent damage from occurring to the elastomeric modules. Modules were placed over a one half inch long .022" x .028" stainless steel wire segment fully engaging an American Orthodontics No. 001-698 siamese .022 inch standard bracket which was welded to a numbered, half inch section of .242 inch inside diameter stainless steel tubing. The elastomers were slowly stretched just enough to insure placement on the bracket without damage to the ligatures. (See fig. 2.)

The testing jig consisted of a solid, stainless steel cylinder milled to accept the mounted bracket unit. This was secured to the descending beam of an Instron Universal Testing Machine. To the stationary beam of the Instron was secured a doubly jointed claw-like arm fashioned of .050" stainless steel. The claw engaged the .022" x .028" archwire segment immediately adjacent to the bracket without binding. (See figures 3-8.) The Instron was set to displace the wire approximately .023 inches at .2 inches per minute, assuring at least a .020 inch displacement occurred

without excessive extension. Paper speed was set at ten inches per minute with a full scale reading of five pounds. Upon initiation of a test the displacement would occur, then the crosshead automatically returned to zero, resulting in the recording of a sloped line on the graph paper which represented the force generated by the elastomer on the archwire on the vertical axis and the displacement of the archwire on the horizontal axis.

After securing a set of ten elastomers of a single size and configuration, the group was tested immediately, then placed in distilled water in a sealed glass container stored in an incubator at 37° C. (See fig. 9.) This was repeated for each group. Subsequent measurements were taken at one hour, one week, two weeks, three weeks, four weeks, six weeks, and eight weeks. Prior to testing, the Instron was balanced, calibrated, and zeroed in the same manner. After the initial measurement, water was drained from the glass container, and the units were blotted twice with paper towels to remove excessive amounts of water, but the units were not dried. After testing each group of a single size and configuration the units were immediately returned to their storage container at 37° C and exchanged for the container having the next group. Only one group was removed from the 37° C incubator at a time for testing. Testing time averaged approximately four minutes per group. Measurements proceeded in this manner for each of the eight groups.

After testing the entire lot, the graph paper was collected. On each plot, ten squares were counted on the horizontal axis from the initial detection of force generation, representing a .020" displacement. The force in pounds was read on the vertical axis and recorded. (See fig. 10.) Means, standard deviations, and two three-way ANOVA's were calculated from the raw data utilizing a computerized statistical program (Systat) to determine significance. A Newman-Keuls test was completed for each ANOVA to enhance the interpretation of significance.

RESULTS

The means and standard deviations of force values for each combination of elastomeric size and configuration are presented in Table I. A graphic representation of the data is displayed in figure 11. Based on a three-way ANOVA and a Neuman Keuls test, significant differences ($p < .01$) existed between all three of the elastomeric configurations. (See Appendix D.) Size 110 elastomers were not included in this ANOVA since this size was not tested in a double configuration. Throughout the study, the figure-eight configuration yielded higher forces than the double configuration which in turn yielded higher forces than the regular configuration, irrespective of elastomer size. Figures 12 - 16 graphically demonstrate that the forces generated by both figure-eight and doubled elastomers were significantly different than those generated by conventionally placed elastomeric ties.

Significant differences ($p < .01$) also existed between the three sizes of elastomers. (See Appendix D.) Size 110 elastomers outperformed size 090 elastomers, and both exceeded the forces produced by size 120 elastomers. The double configurations were not included in this ANOVA since they were not tested for size 110 elastomers. As can be seen in Figure 11, the data for the doubled samples (sizes 090 and 120) were essentially identical. As expected from the results of previous investigators, the most dramatic decay of force occurred by the first day for all samples. After the first day forces remained essentially constant as evidenced by Figure 11 and Table II.

Mean force decay as both a gross number and a percentage for each combination was grouped by configuration and presented in Table II. At both day one and week eight time periods figure-eighted elastomers experienced significantly more gross degradation of force than conventionally applied elastomers, and the conventional ligatures had greater gross force decay than the doubled samples. As a percentage, mean force decay was similar for both conventional and figure-eight samples, and both lost a larger percentage of force than did the doubled elastomers.

DISCUSSION

In order to closely approximate clinical use, brackets were chosen which were similar in size to appliances commonly marketed today. A full-size stainless steel wire attachment mechanism (the "archwire") was utilized to minimize flexure which might have resulted in inaccurate displacements and measurements of force. Since multiple authors have reported on both the possible positive and negative effects of prestretching and rate of stretch, elastomers were slowly extended just adequately to facilitate placement.^{5,18,19,20} The samples were purposely measured without completely drying them to more closely approximate oral conditions.^{1,21,25}

Several samples were measured on the Instron utilizing a hand zero procedure followed by the .020 inch displacement. This was found to be no more accurate than having a 10% error. At the expense of slightly overstretching the samples, the displacement was calibrated into the Instron slightly in excess, and then the exact .020 inch displacement measured on the graph paper chart. The error of this method was far superior, being essentially limited to that of the Instron mechanism itself once hand zeroing was eliminated. The small standard deviations reflect the small measurement error even with material variability included in the measurement.

It must be pointed out that force measurements made during the present study were elevated somewhat due to their measurement .020 inches beyond the point of the bracket contact with the full size wire during storage. However, a previous study utilizing near identical experimental protocol as the present study showed that the forces measured at the same level at which the figure-eight elastomer was displaced during storage were still in excess of 450 grams at six weeks.¹³ It is possible that even these numbers may be excessive as others have shown that the oral environment (thermocycling, fluoride, pH, etc.) had a more detrimental effect on elastomers than an in vitro environment.^{8,19,21,22,23}

As others have shown, force decay occurred over time with most happening in the first day.^{1-3,5,7,8,11,13,14,16,18-20,22} After the first day, the amount of force decay was small, ranging from -2.4 - 5.8%. Figure-eighted samples experienced the largest force decay, but on a percentage basis it was essentially identical to that of the conventional samples. The slopes of the lines in Figure 11 show the rate of decay for the doubled samples to be less than that of either the conventional or the figure-eighted elastomers. As shown in Table IV, both conventional and figure-eighted samples were stretched approximately twice that of the doubled sample during initial placement. Several investigators have shown that increased stretching results in enhanced force production and greater force decay with either an increased or decreased percent force decay over elastomers stretched to a lesser degree.^{1,2,8,16,22} In general, force decay has been attributed to structural elastomeric relaxation, chemical breakdown, and absorption of water.

Force generation curves from which the elastomer displacement and resultant forces were measured closely resembled those reported by Rock, et. al.¹⁰ The initial part of the curve was linear until a transition occurred where the curve flattened. Shortly after the plot steepened but never reached the height of the initial linear portion had it been extrapolated.

Significant differences in mean force generation (figure-eight > doubled > conventional and 110 > 090 > 120) can be attributed to elastomer mass, elastomer dimension, friction, and the distance stretched. These differences for the three elastomers studied are depicted in Tables III and IV. Increased mass should result in greater force generation as more material resists stretching better than less material. The size 120 elastomers would be expected to resist displacement better than either size 110 or 090 elastomers based on mass alone. The doubled configuration would be expected to yield elevated force values since the elastomer mass was twice that of the other two configurations for the same size elastomer. Within their elastic limit, elastomers having smaller dimensions (lumen diameter) would be expected to generate more force than those of equal mass but with larger

dimensions since they would require increased stretch to engage the same size bracket. If diameter was the only variable, size .090 elastomers would be expected to generate more force than either of the other two sizes of elastomers.

For two equal size elastomers, the one stretched the greatest should generate more force as indicated by Hooke's Law, $F=kx$, where x is the elongation.²⁸ The values in Table IV were determined by measuring the length of .090 inch diameter stainless steel ligature wire necessary to encompass the bracket in the desired configuration when tightly adapted with con pliers and a plugging instrument. Here, it can be seen that the figure-eight configuration required that these elastomers be stretched more than the other two configurations, and regular elastomers were stretched more than the doubled elastomers.

Increased friction should also increase the force measured. Sharp bracket corners might contribute to the friction acting between the elastomer and the metal. During extension the figure-eight elastomers had to overcome eight bracket corners per elastomer while conventional and doubled samples only had to negotiate four bracket corners per elastomer. Also, both figure-eight and conventional elastomers had to slide against eight flat bracket surfaces whereas the doubled elastomers had only six flat surfaces to drag against per elastomer during extension. Elastomers have been shown to deform over time.^{1,2,17,18,23} The bracket corners might actually crease or form a "foothold" in the elastomer at the point of contact, causing elastomer irregularity and increased friction. Others have shown that figure-eight ligation increased friction 70-220% over conventional elastomeric ligation.²⁶ Conventional elastomeric ligation has been shown to require forces of over 100 grams to draw a .019" x .025" archwire through a stainless steel bracket slot.²⁴

Although the statistics for the main effects were quite clear, several interactions were significant. That between size and configuration in the ANOVA testing three different sizes in two configurations can be discounted since a plot of means showed that the lines are not converging rapidly; however, their slight divergence was

sufficient to imply mathematical significance. For both ANOVA's all interactions with time were significant as would be expected due to the force decay that occurred over the course of the study.

CLINICAL IMPLICATIONS

Mean force production even after eight weeks never dropped below one pound. Based on the data produced by others, this is far in excess of that required for tooth movement and for displacement of a .016 inch nickel-titanium archwire .020 inches. Values ranging from as little as 30 grams to over 500 grams have been reported to cause tooth movement.^{3,11,29,30,31,32} Miura reported that approximately 260-360 grams was required to deflect the nickel-titanium wire as described above.³³ About 500 grams (1.1 pounds) was required to deflect a .016 inch stainless steel wire similarly. The sufficient force measurements at eight weeks suggest that the elastomer may retain its ability to actively hold the archwire to the bracket slot even at eight weeks and possibly beyond. The monthly change of modules may be unnecessary for a patient with partial appliance (without need for adjustment) that might preferably be monitored on a more extended recall schedule. This patient may be seen every other month or perhaps every third or fourth month without fear that the elastomers have stopped functioning.

As previously noted, friction has been shown to be elevated for figure-eight ligation when compared to conventional elastomeric ligation.²⁶ It has also been shown that elastomers may produce less of a binding force than stainless steel ligatures due to their ability to deform under force, instead of binding.³⁴ It must be kept in mind that elastomeric forces are generally low, and the use of overpowering retracting forces may overcome the ability of the elastomer to retain the archwire in the bracket slot. It would seem that figure-eight ligation would be appropriate for holding a canine to the archwire during space closure, and it might even be more efficient than ligation with stainless steel as long as the elastomer is not overpowered, which could result in the tooth being rotated off the archwire.

Placement and removal of figure-eight elastomers may require slightly more time than use of conventionally placed elastomeric ties. Use of a placement instrument with a small hooked tip and recessed grasping surfaces void of sharp edges allows placement of elastomers

with minimal elastomer damage, extension, and slippage. Removal can be facilitated with the use of a sharp shepherd's hook explorer. Investment of a few seconds of time to place selected figure-eight ligatures may improve treatment efficiency by improving initial archwire engagement and later preventing unwanted rotations from occurring which might necessitate delay of a month or more to step down in archwire sizes to recapture the rotation.

Based on this study, the placement of two elastomers per bracket (one per wing) would not serve its proposed clinical purpose of decreasing archwire engagement or "making it easier on the patient." The use of smaller sizes of elastomers would seem an effective means of enhancing applied force to the archwire to improve engagement, but there may be offsetting disadvantages to their lessened mass such as breakage during placement and increased susceptibility to damage and fracture in the oral environment.

From an oral hygiene perspective, Forsberg has shown that teeth ligated with elastomers have significantly larger bacterial populations than teeth ligated with stainless steel.²⁷ Elastomeric ligation might be a poor choice for the patient with hygiene difficulties.

CONCLUSIONS

1. Like other elastomers, Ormco power O's experienced force decay over time with the majority of force decay occurring in the first day.

2. Percent force decay was least in those elastomers stretched the least, and mean force decay was greater with increased initial stretch of an elastomer.

3. Over the course of this study, figure-eighted elastomers generated more force than conventionally applied elastomers, and doubled elastomers generated forces between those produced by conventional and figure-eighted elastomers.

4. Force generation decreased in the elastomer size order 110 > 090 > 120 for both figure-eight and conventional samples over the course of this study.

5. Elastomer material mass, elastomer dimension, elastomer stretch with application, and friction may play a role in force generation and its measurement as made during this study.

6. The figure-eight elastomeric configuration may enhance light archwire engagement and resist forces which cause displacement of the archwire from the bracket slot.

BIBLIOGRAPHY

- 1 Andreassen GF and Bishara S. Comparison of Alastik chains involved with intra-arch molar to molar forces. AO 40:151-158, 1970
- 2 Bishara S and Andreassen GF. Comparison of time related forces between plastic Alastiks and latex elastics. AO 40:319-328, 1970
- 3 Hershey HG and Reynolds WG. The plastic module as an orthodontic tooth moving mechanism. AJODO 67:554-62, 1975
- 4 Kovatch JS, Lautenschlager EP, Apfel DA, and Keller JC. Load-extension-time behavior of orthodontic Alastiks. JDR 55:783-6, 1976
- 5 Varner RE and Buck DL. Force production and decay rates in Alastik modules. J Biomed Res 12:361-6, 1978
- 6 Brantley SA, Salander S, Merers CL, and Winders RV. Effects of prestretching on force degradation characteristics of plastic modules. AO 49:37-43, 1979
- 7 Doyle L. Force production and decay rate in C1 Alastik modules. Certificate Thesis, OHSU, 1979
- 8 DeGenova DC, McInnes-Ledoux P, Weinberg R, and Shave R. Force degradation of orthodontic elastomeric chains - a product comparison study. AJO 87:377-84, 1985
- 9 Killiany DM and Deplessis J. Relaxation of elastomeric chains. JCO 19:592-3, 1985
- 10 Rock WP, Wilson HJ, and Fisher S. A laboratory investigation of orthodontic elastomeric chains. BJO 12:202-7, 1985
- 11 Rock WP, Wilson HJ, and Fisher S. Force reduction of orthodontic elastomeric chain after one month in the mouth, BJO 13:147-50, 1986
- 12 Bertle W and Droschl H. Forces produced by orthodontic elastics as a function of time and distance extended. EJO 8:198-201, 1986
- 13 Nordberg RC. The force decay behavior of polyurethane ligature modules. Certificate Thesis, OHSU, 1987
- 14 Brown M. The force decay behavior of orthodontic elastomeric chains as a function of time and elongation. Certificate Thesis, OHSU, 1989

- 15 Huget EF, Patrick KS, and Nunez CJ. Observations on the elastic behavior of a synthetic orthodontic elastomer. JDR 69:496-501, 1990
- 16 Lu TC, Wang WN, Tamg TH, and Chen JW. Force decay of elastomeric chain - a serial study. Part II. AJODO 104:373-7, 1993
- 17 Baty DL, Valz JE, and von Fraunhofer JA. Force delivery properties of colored elastomeric modules. AJODO 106:40-6, 1994
- 18 Wong A. Orthodontic elastic materials. AO 46:196-205, 1976
- 19 Brooks DG and Hershey HG. Effect of heat and time on stretched plastic orthodontic modules. JDR 55B:363, 1976
- 20 Young J and Sandrik JL. The influence of preloading on stress relaxation of orthodontic elastic polymers. AO 49:104-8, 1979
- 21 Ash JL and Nikolai RJ. Relaxation of orthodontic elastomeric chains and modules in vitro and in vivo. JDR 57:685-90, 1978
- 22 Ferriter JP, Meters CE, and Lorton L. The effects of hydrogen ion concentration on the force degradation rate of orthodontic polyurethane chain elastics. AJODO 98:404-10, 1990
- 23 von Fraunhofer JA, Coffelt MT, and Orbell GM. The effect of artificial saliva and topical fluoride treatments on the degradation of the elastic properties of orthodontic chains. AO 62:265-74, 1992
- 24 Echols PM. Elastic ligatures, binding forces, and anchorage taxation. AJO 67:219-20, 1975
- 25 Baker KL, Nieberg LG, Weimer AD, and Hanna M. Frictional changes in force values caused by saliva substitution. AJODO 91:316-20, 1987
- 26 Sims AP, Waters NE, Birnie DJ, and Pethybridge RJ. A comparison of the force required to produce tooth movement in vitro using two self ligating brackets and a pre-adjusted bracket employing two types of ligation. EJO 15:377-85, 1993
- 27 Forsberg CM, et al. Ligature wires and elastomeric rings: two methods of ligation, and their association with microbial colonization of Strep mutans and lactobacilli. EJO 13:416-20, 1991
- 28 Sears FW and Zernansky MW. University Physics, 4th Ed. Addison-Wesley Pub Co, 1970, p157
- 29 Storey EE and Smith R. Force in orthodontics and its relation to tooth movement. Aust J Dent 56:11-8, 1952

- 30 Reitan K. Some factors determining the evaluation of forces in orthodontics. AJO 43:32-45, 1957
- 31 Hixon EH, Atikan H, Callow GE, McDonald HW, and Tracy RJ. Optimum force, differential force and anchorage. AJO 55:437-57, 1969
- 32 Boester CH and Johnston LE. A clinical investigation of the concepts of differential and optimal force in canine retraction. AO 44:113-9, 1974
- 33 Miura F, Mogi M, Ohura Y, and Hamanaka H. The superelastic property of Japanese NiTi alloy wire for use in orthodontics. AJODO 90:1-10, 1986
- 34 Thurow R. Letters to the Editor - Elastic ligatures, binding forces, and anchorage taxation. AJO 67:694, 1975

Appendix A

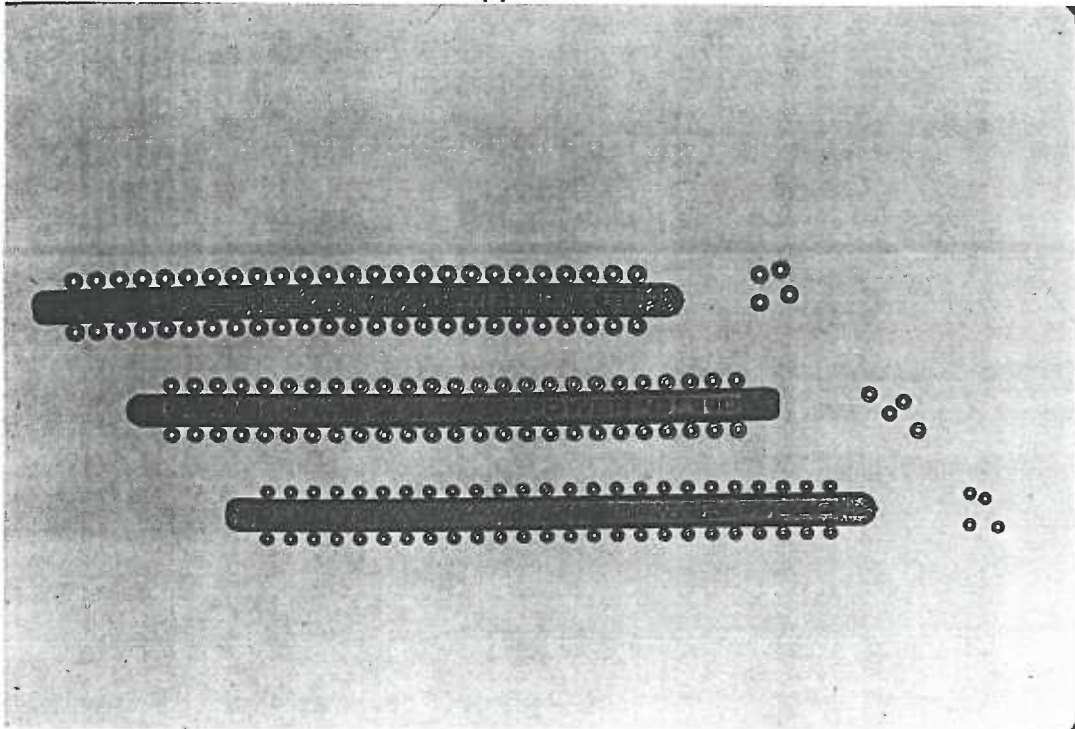


Figure 1. Top to bottom, Ormco Power O's in sizes 120, 110, and 090.

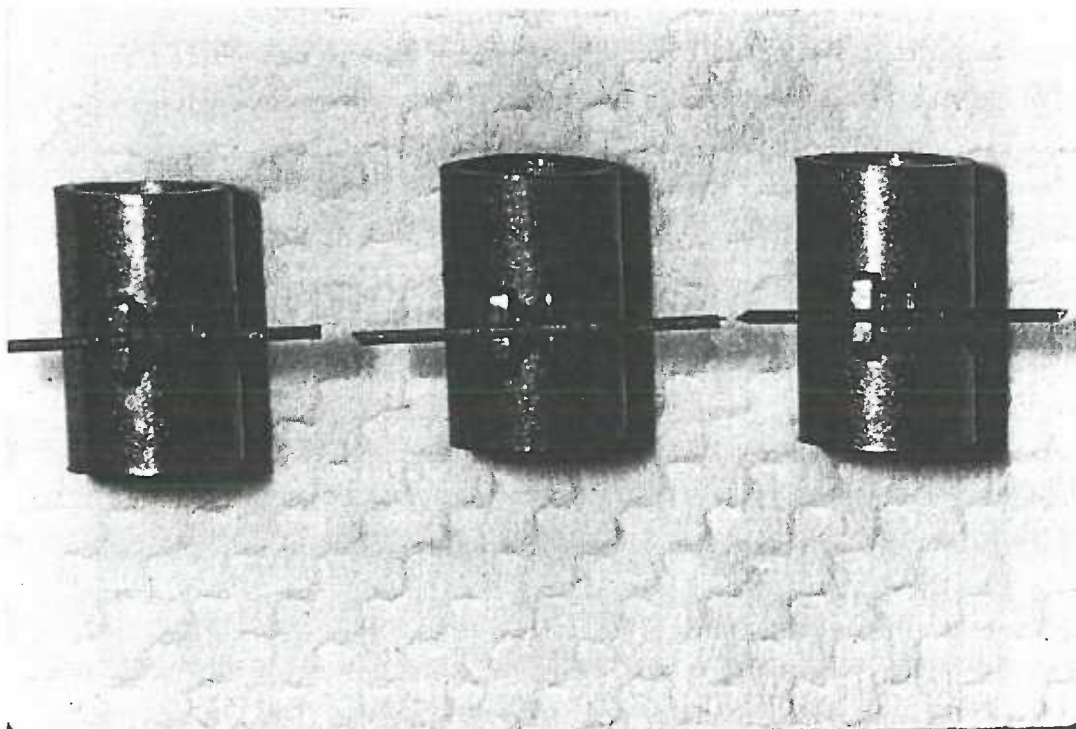
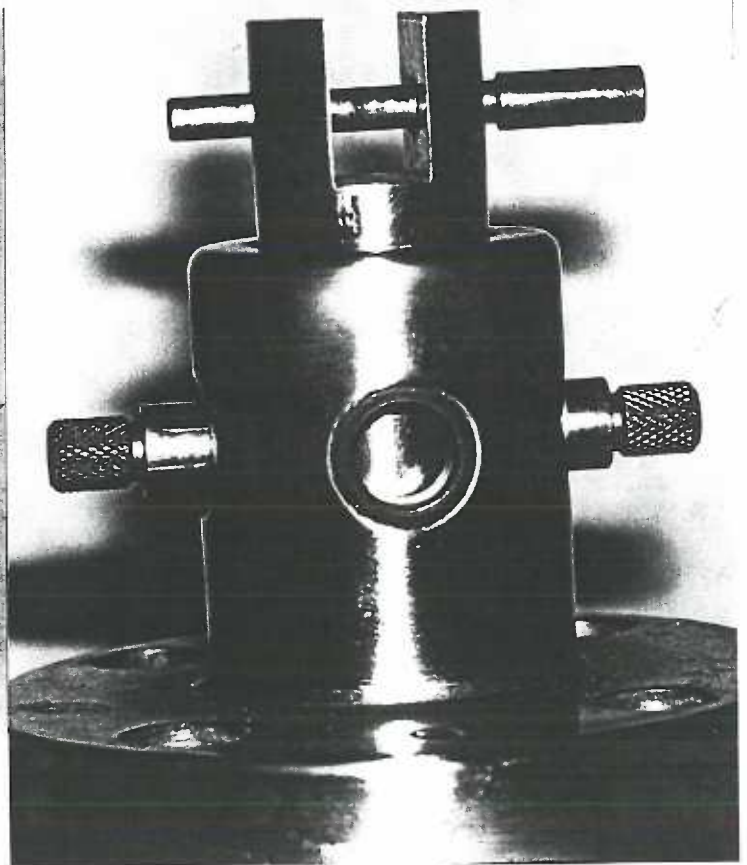
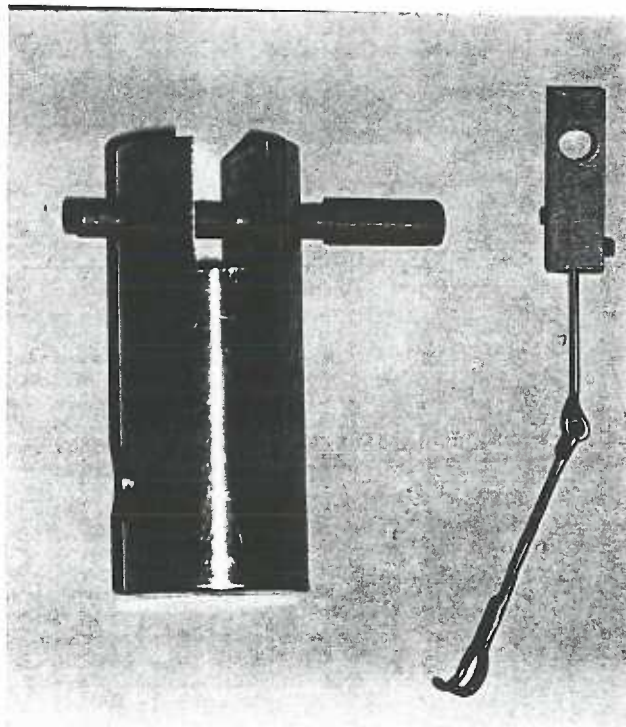


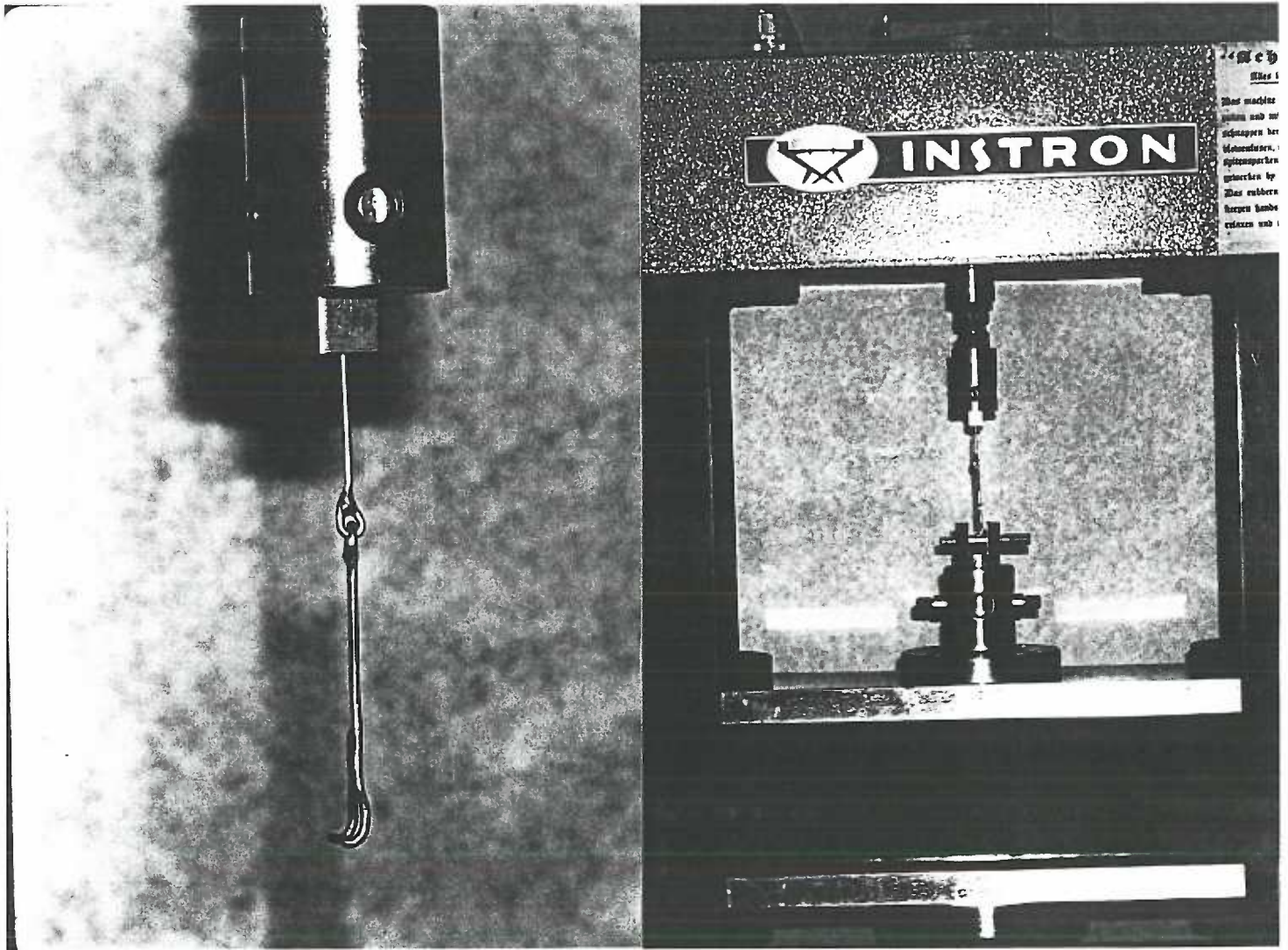
Figure 2. Power O's doubled, figure-eighted, and conventionally applied.

Appendix A



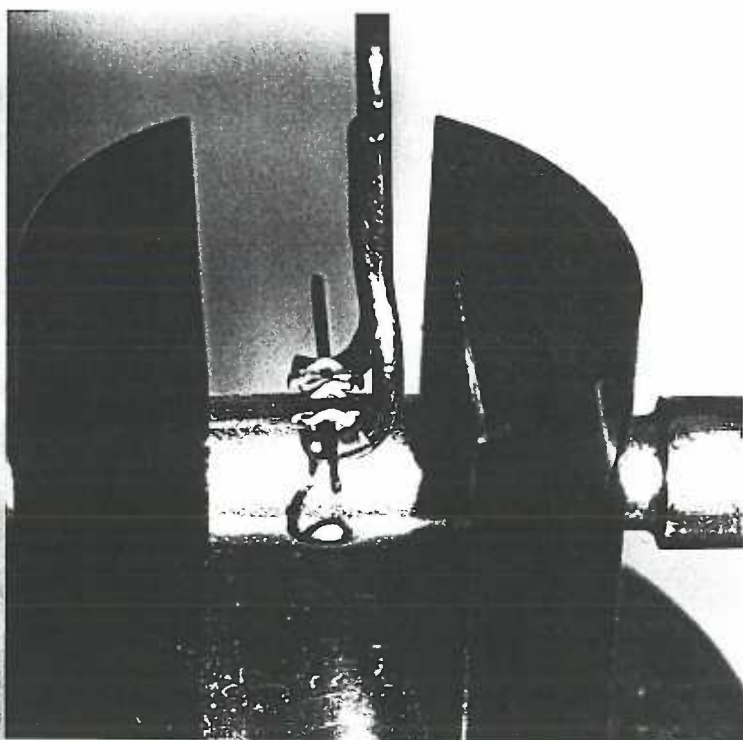
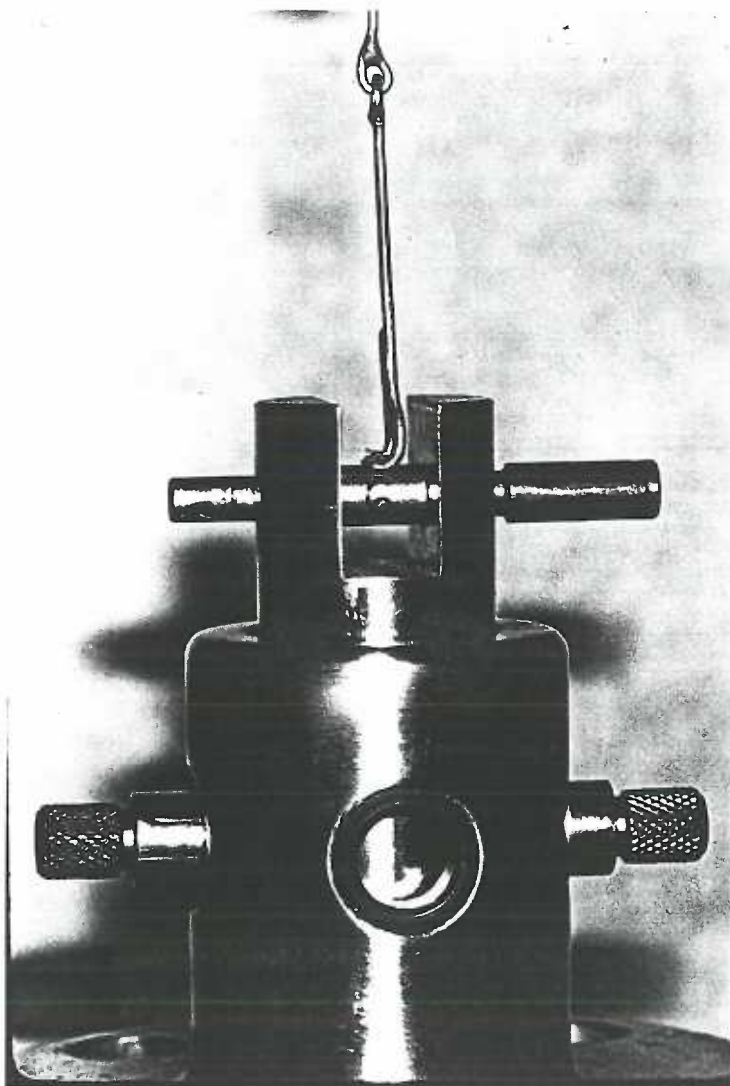
Figures 3-4. The mechanical testing apparatus and its attachment to the Instron Universal Testing Machine.

Appendix A



Figures 5-6. The mechanical testing apparatus and its attachment to the Instron Universal Testing Machine.

Appendix A



Figures 7-8. Engagement of the attachment arm to the archwire segment.

Appendix A



Figure 9. Storage of ten Power O's in place on separate bracket units.

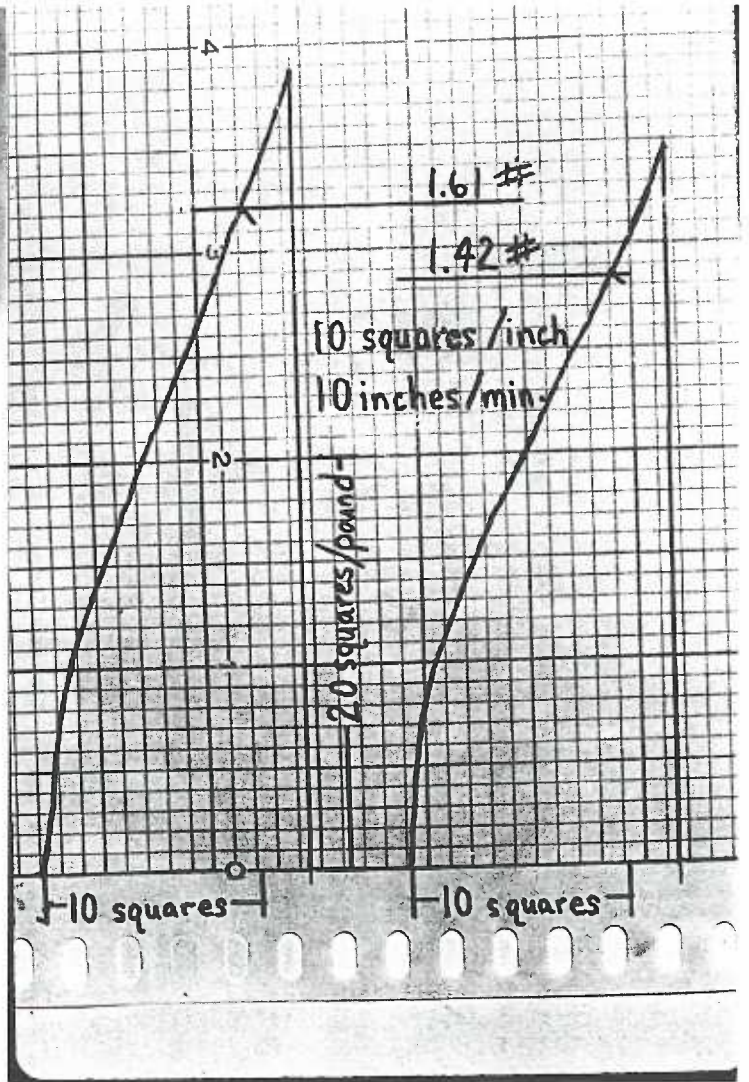


Figure 10. Curves depicting force (y-axis) versus time (x-axis) and the determination of force at .020 inches of displacement.

Appendix B

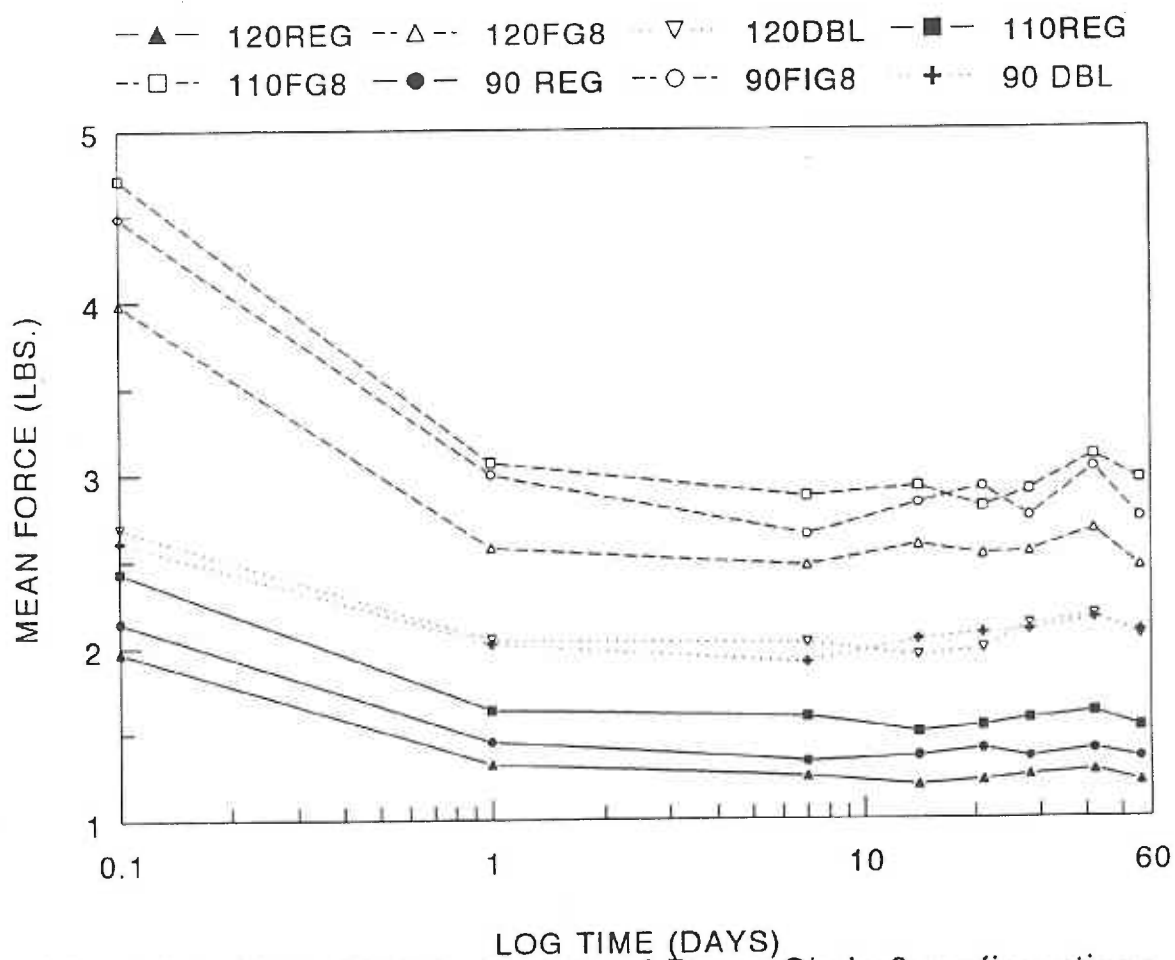


Figure 11 Mean force for 3 sizes of Power O's in 3 configurations.

Appendix B

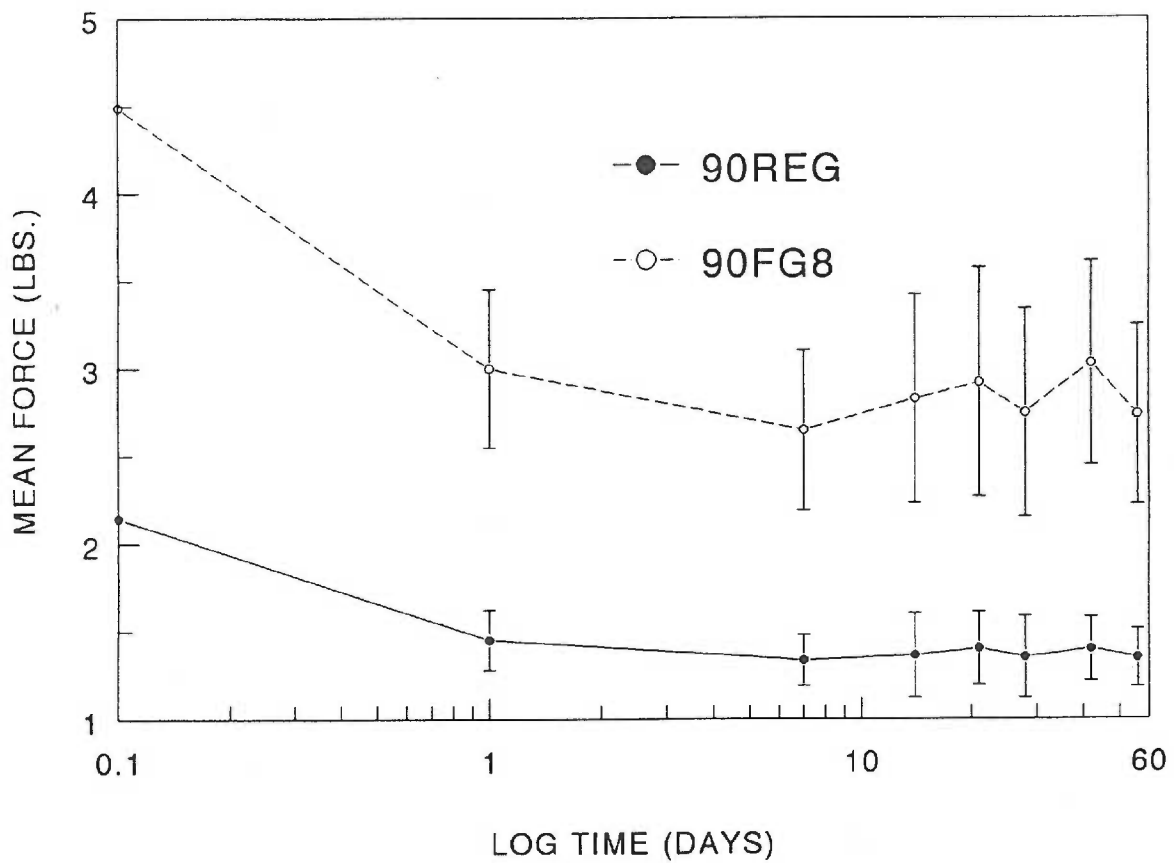


Figure 12 Mean force $\pm 2SD$ for size 90 figure 8 and regular Power O's.

Appendix B

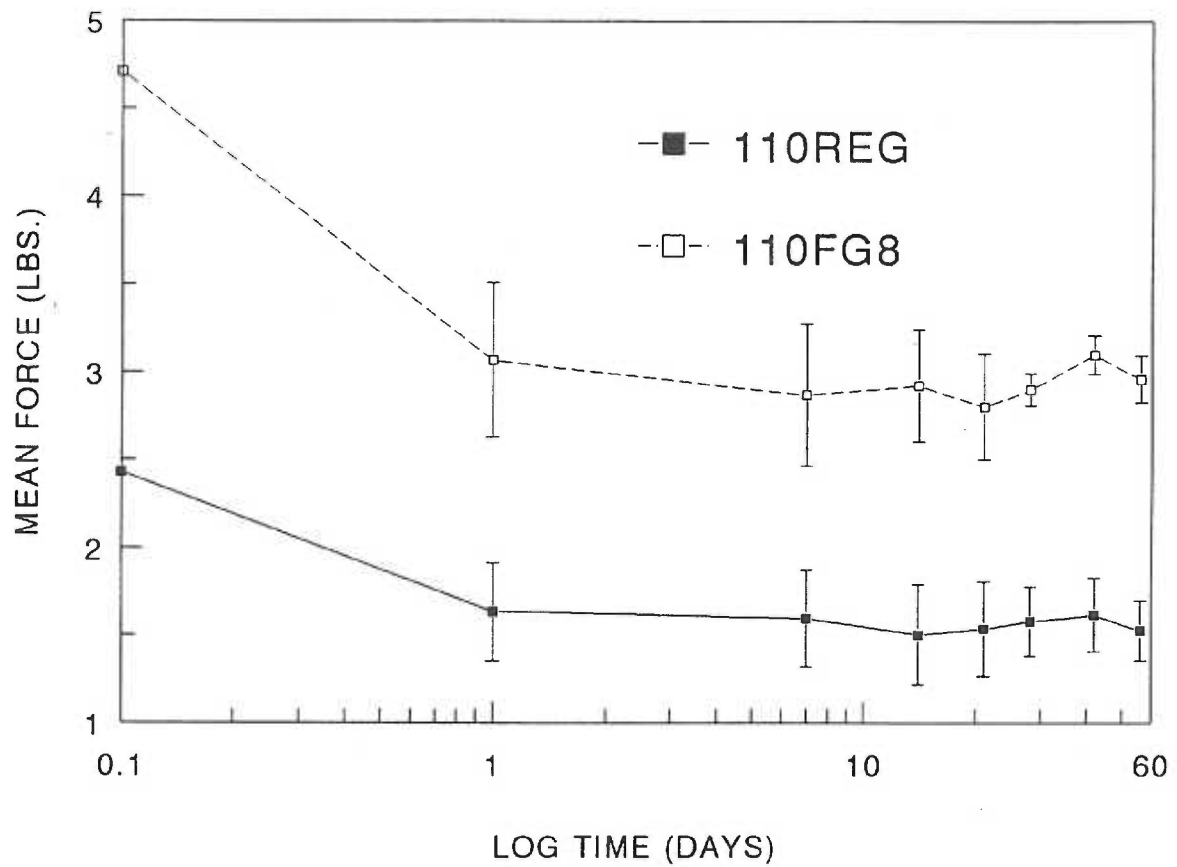


Figure 13 Mean force $\pm 2SD$ for size 110 figure 8 and regular Power O's.

Appendix B

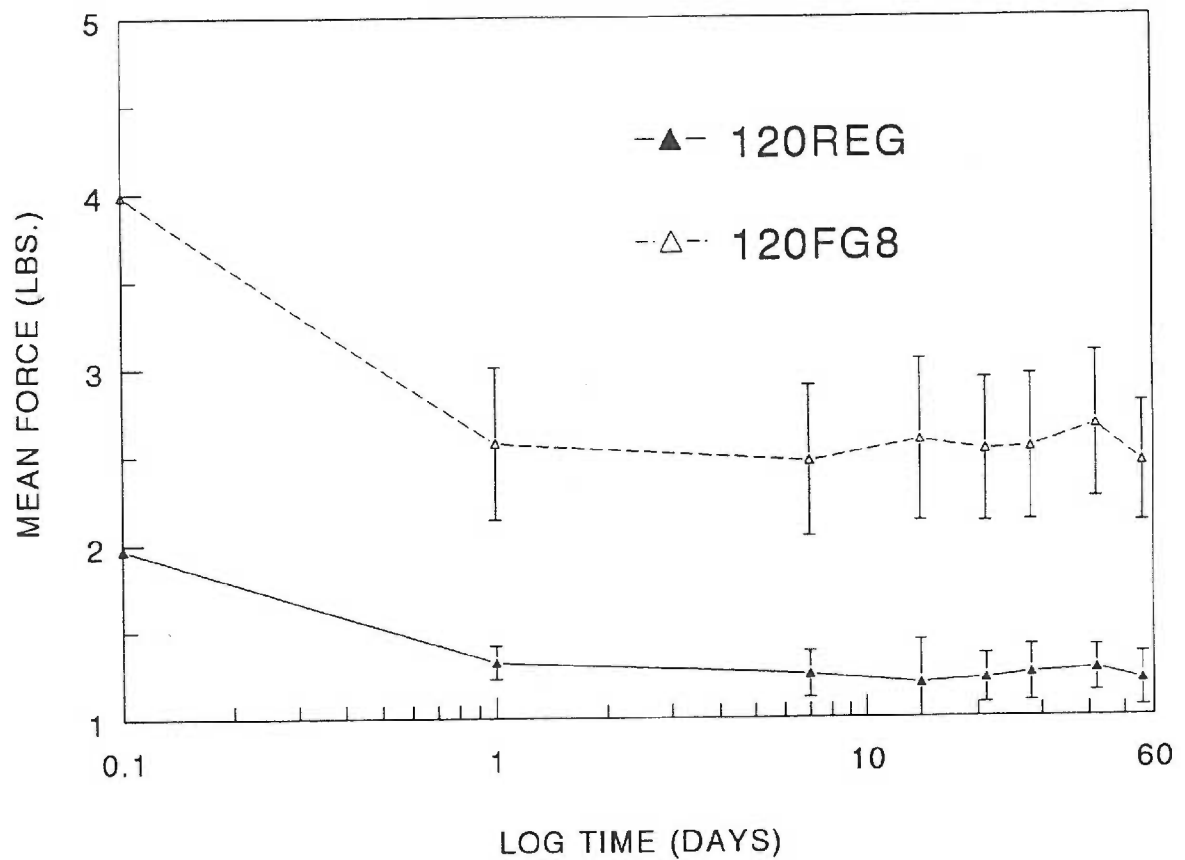


Figure 14 Mean force $\pm 2SD$ for size 120 figure 8 and regular Power O's.

Appendix B

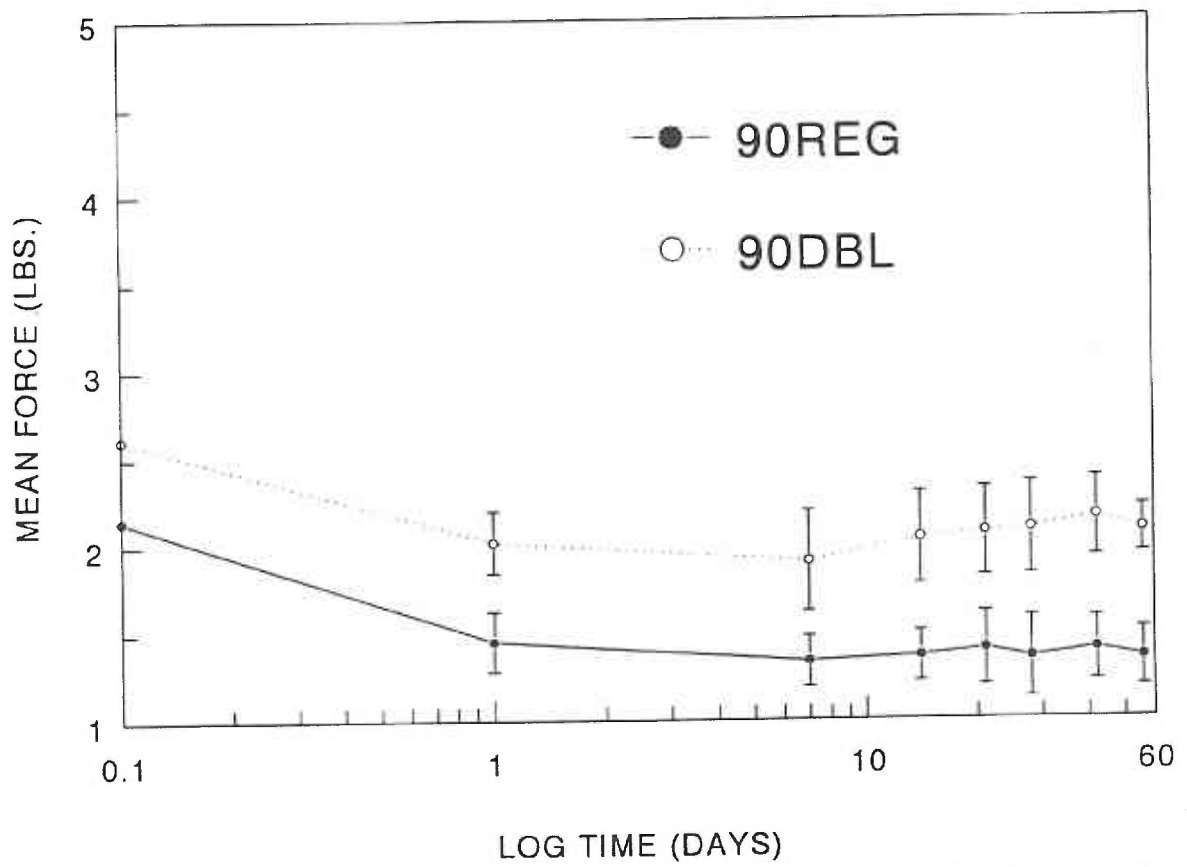


Figure 15 Mean force $\pm 2SD$ for size 90 double and regular Power O's.

Appendix B

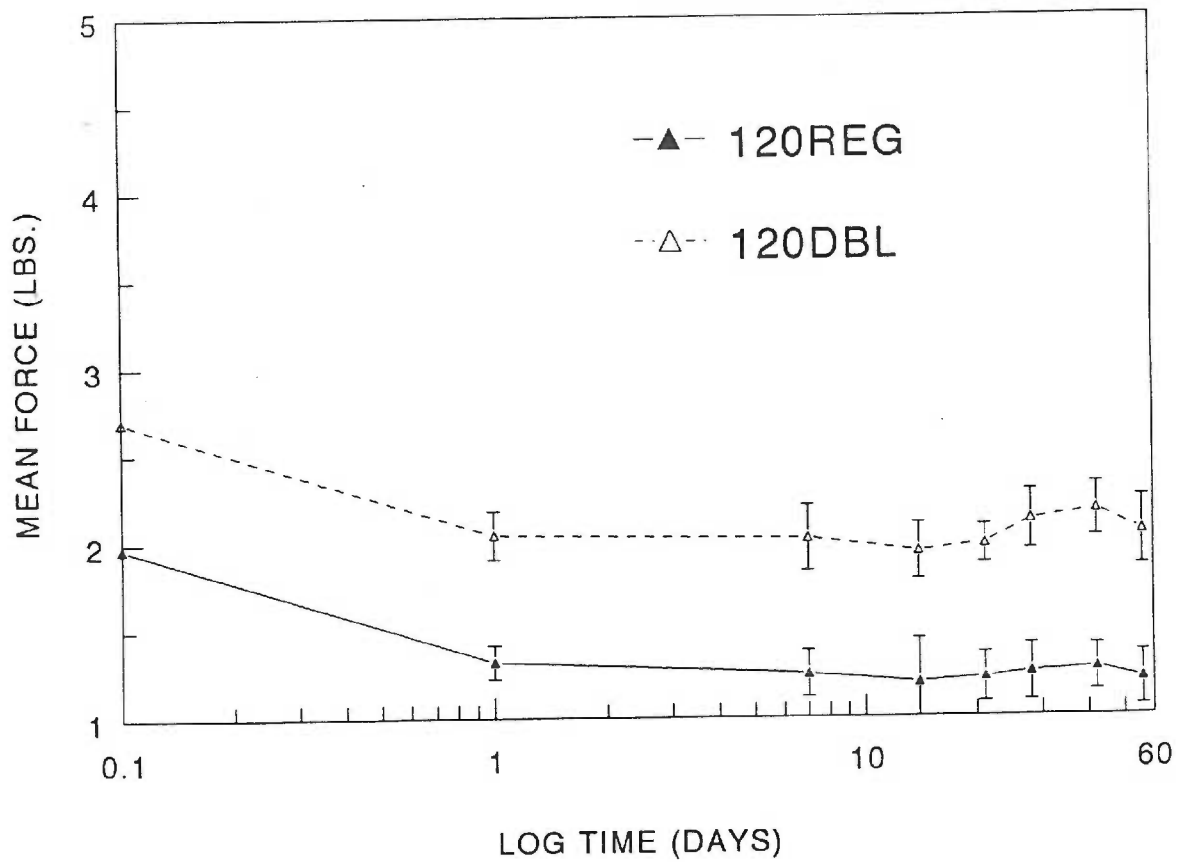


Figure 16 Mean force $\pm 2SD$ for size 120 double and regular Power O's.

Appendix C

TABLE I Mean force (lbs) \pm 1 SD at each measurement period for three sizes of Ormco Power O's in three configurations.

size	config.	initial	24 hrs.	1 wk	2 wk	3 wk	4 wk	6 wk	8 wk
120	regular	1.971 \pm .059	1.322 \pm .048	1.248 \pm .067	1.196 \pm .125	1.221 \pm .071	1.251 \pm .081	1.274 \pm .066	1.213 \pm .077
120	figure 8	3.987 \pm .285	2.575 \pm .217	2.471 \pm .215	2.588 \pm .231	2.532 \pm .205	2.544 \pm .208	2.670 \pm .208	2.458 \pm .171
120	double	2.690 \pm .137	2.045 \pm .069	2.023 \pm .093	1.946 \pm .080	1.984 \pm .054	2.121 \pm .084	2.175 \pm .076	2.053 \pm .097
110	regular	2.432 \pm .150	1.635 \pm .140	1.596 \pm .139	1.504 \pm .144	1.537 \pm .135	1.579 \pm .099	1.618 \pm .106	1.527 \pm .086
110	figure 8	4.712 \pm .261	3.070 \pm .220	2.872 \pm .202	2.926 \pm .160	2.805 \pm .151	2.902 \pm .213	3.101 \pm .234	2.961 \pm .260
090	regular	2.144 \pm .143	1.452 \pm .086	1.336 \pm .073	1.364 \pm .122	1.404 \pm .105	1.354 \pm .117	1.400 \pm .092	1.349 \pm .083
090	figure 8	4.490 \pm .377	3.001 \pm .226	2.650 \pm .228	2.829 \pm .297	2.921 \pm .326	2.748 \pm .297	3.030 \pm .291	2.737 \pm .257
090	double	2.610 \pm .140	2.021 \pm .089	1.910 \pm .143	2.041 \pm .131	1.074 \pm .127	2.090 \pm .131	2.156 \pm .113	2.082 \pm .067

TABLE II Percentage force decay and mean force decay (lbs) at day one and week 8 for three sizes and three configurations of Ormco Power O's. Percentage change in force decay is also shown.

size	config.	DAY 1		WEEK 8		
		% F decay	mean F decay	% F decay	% change	mean F decay
120	regular	32.9	0.649	38.5	5.6	0.758
110	regular	32.8	0.797	37.2	4.4	0.905
090	regular	32.3	0.692	37.1	4.8	0.795
120	figure 8	35.4	1.412	38.2	2.8	1.529
110	figure 8	34.8	1.642	37.2	2.4	1.751
090	figure 8	33.2	1.489	39.0	5.8	1.753
120	double	24.0	0.645	23.7	-0.3	0.637
090	double	22.6	0.589	20.2	-2.4	0.528

Appendix C

TABLE III Physical characteristics for three sizes ofOrmco Power O's.

size	mass/module (mg)	lumen dia (mm)
120	4.25	1.10
110	3.90	0.90
090	2.48	0.71

TABLE IV Ligature length required for a .090" steel
ligature to encompass the test bracket in
three configurations.

configuration	lig. length (mm)
regular	15.0
figure 8	18.3
double	9.2

Appendix D

DEPENDENT VARIABLE MEANS

TIME(1)	TIME(2)	TIME(3)	TIME(4)	TIME(5)
3.289	2.176	2.029	2.068	2.070
TIME(6)	TIME(7)	TIME(8)		
2.063	2.182	2.041		

UNIVARIATE AND MULTIVARIATE REPEATED MEASURES ANALYSIS

BETWEEN SUBJECTS

SOURCE	SS	DF	MS	F	P
CONFIG\$	264.820	1	264.820	1450.646	0.000
SIZE	12.361	2	6.181	33.856	0.000
CONFIG\$*SIZE	0.680	2	0.340	1.863	0.165
ERROR	9.858	54	0.183		

WITHIN SUBJECTS

SOURCE	SS	DF	MS	F	P	G-G	H-F
time(76.961	7	10.994	760.924	0.000	0.000	0.00
time(
*CONFIG\$	9.590	7	1.370	94.817	0.000	0.000	0.00
time(*SIZE	0.904	14	0.065	4.469	0.000	0.000	0.00
time(
*CONFIG\$							
*SIZE	0.323	14	0.023	1.597	0.077	0.111	0.09
ERROR	5.462	378	0.014				

Three-way ANOVA with repeated measures. Three size of Ormco Power O's (090, 110, and 120) and two configurations (conventional and figure-eight) were analyzed.

Appendix D

DEPENDENT VARIABLE MEANS

TIME(1)	TIME(2)	TIME(3)	TIME(4)	TIME(5)
2.982	2.069	1.940	1.994	2.023
TIME(6)	TIME(7)	TIME(8)		
2.018	2.118	1.982		

UNIVARIATE AND MULTIVARIATE REPEATED MEASURES ANALYSIS

BETWEEN SUBJECTS

SOURCE	SS	DF	MS	F	P
SIZE	2.753	1	2.753	18.438	0.000
CONFIG\$	176.052	2	88.026	589.589	0.000
SIZE*CONFIG\$	2.178	2	1.089	7.295	0.002
ERROR	8.062	54	0.149		

WITHIN SUBJECTS

SOURCE	SS	DF	MS	F	P	G-G	H-F
time()	49.772	7	7.110	616.786	0.000	0.000	0.00
time()*SIZE	0.323	7	0.046	4.006	0.000	0.002	0.00
time()*CONFIG\$	9.795	14	0.700	60.694	0.000	0.000	0.00
time()*SIZE*CONFIG\$	0.372	14	0.027	2.305	0.005	0.013	0.00
ERROR	4.358	378	0.012				

Three-way ANOVA with repeated measures. Two sizes of Ormco Power O's (090 and 120) in three configurations (conventional, figure-eight, and double) are analyzed.

Appendix D

	2.033	2.263	2.424
2.033		.230 **	.391 **
2.263			.161 **
2.424			
	q(r,54)	3.760	4.280
	q(sq.rt.(MS/n)	0.127	0.145

	1.406	2.126	2.889
1.406		.720 **	1.483 **
2.126			.763 **
2.889			
	q(r,54)	2.830	3.400
	q(sq.rt.(MS/n)	0.244	0.294

Newman-Keuls test for each ANOVA, respectively.