THE FORCE DECAY BEHAVIOR OF POLYURETHANE LIGATURE MODULES

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INTRODUCTION

In fixed appliance therapy, orthodontic forces are stored in the form of archwires and delivered to the teeth by means of brackets. The interface between archwire and bracket is mediated by ligatures. Small diameter wire has been the traditional means of ligation. For over fifteen years polyurethane modules have been marketed as timesaving substitutes for wire ligatures.

Plastic ligature modules represent more than a means of simplification of office procedure. These elastomeric units provide a potential supply of tooth moving force all their own. Early in treatment it is not unusual to find that full bracket engagement is difficult or impossible to achieve due to severe malposition or rotation of teeth. When stretched from bracket wings to archwire the polyurethane ligature unit stores potential energy which tends to draw the bracket (and tooth) toward the archwire.

The elastic character of the ligature module poses problems as well as advantages in clinical application. Clinicians often voice the need for use of inelastic wire ligatures applied to teeth adjacent to extraction sites during space closure. Their experiences indicate that the rotational components of force generated during space closure frequently exceed the capacity of the elastometric module to resist said rotational forces. Thus the ligature unit "gives" allowing unseating of the archwire within a portion of the bracket and loss of rotational control.

Polyurethane ligature modules are marketed by virtually every orthodontic supply company. Information available for each product is typically limited to the technique used in its manufacture (injection molded versus cut from an extruded sheet). Some companies provide more than one "grade" of module whose designation of "regular" versus "heavy" is a subjective appraisal based of relative ease or difficulty experienced in application and removal of ligature unit to bracket. The clinician frequently makes a similar appraisal when choosing which product to use in his or her practice. A module which requires greater effort in application and removal is deemed by the operator to be the superior product because it is presumed to provide a more positive seating force between archwire and bracket. Many clinicians will further resort to crossing the module over itself between bracket wing pairs to form a "figure eight". Difficulty in performing this task is interpreted as evidence that the maximum seating force has been delivered to the archwire.

The purpose of this study is: 1) to quantify the magnitude of force applied by polyurethane ligature modules to the archwire, 2) to determine whether or not relative clinical stiffness is correlated to measureable force applied to the archwire and, 3) to determine if applying the module in a "figure eight" pattern has any effect on the ultimate force applied to the archwire. All of the above will be observed over time under controlled in vitro conditions.

REVIEW OF THE LITERATURE

To the best of our knowledge nothing has been published in the dental literature to characterize forces associated with polyurethane ligature modules. All of the other forms of plastic orthodontic products have been the subject of investigation to include elastomeric thread, chains and the "K-type" space closure modules. Review of information available on these orthodontic supplies suggest that certain general aspects of force decay behavior are characteristic of the polyurethane material itself. There is, however, some considerable variation in individual product response over time seen when comparing elastomerics of different form and those of similar configuration produced by different manufacturers. Such variation between products greatly limits our ability to predict the pattern of force decay versus time behavior for a form of elastomeric module not previously studied.

The first two articles documenting the study of forces produced by elastomeric modules were published in 1970 by Andreasen and Bishara[1,2]. The first of these evaluated Unitek AlastiK chain material over a three week test period. A pilot portion of the study measured force decay over time under six in vitro test conditions to include room temperature air, water and saliva and 37 degrees C air, water and saliva. Body temperature saliva (37 degrees C) was the standard to which the other five sets of data were compared statistically since 37 degrees C saliva was the test medium most like the oral environment. Only body temperature water was found to yield results of no significant difference to body temperature saliva for the entire test period. Based on this finding 37 degree C water was selected as the test medium for the remainder of the study.

The balance of their investigation made comparison of forces generated over time by standard and heavy AlastiK chain segments of equal length stretched over five fixed stretch distances. The range of stretch distance simulated typical maxillary first molar to first molar dimensions as determined by the authors clinically by measurement of sixty patients. The standard chain generated less mean force than the heavy chain up to eight hours after which the standard graded material consistently exerted the greater mean force at each measurement interval up to the three week termination point. This would imply when comparing products that a higher initial force is not predictive of a greater sustained force. Pooled data for the standard and heavy chain test results yielded the following mean retained force versus time figures expressed as percent remaining initial force: 44.3% at one hour, 25.8% at twenty four hours and 17.6% at three weeks. Based on these data the authors concluded that force decay is rapid during the first day after which the force applied is fairly constant. They suggested that the operator apply an initial force on the order of four times the desired long term force in order to compensate for the seveny five percent force loss occuring the first day.

In their second article Bishara and Andreasen[2] evaluated Unitek's K-1 and K-3 standard and K-2 heavy AlastiK space closure modules under test conditions identical to those used in the first study. The range of stretch distance was selected based on a clinically determined range of separation between second molar bracket hooks and hooks soldered to archwires mesial to canine brackets. Force levels at the end of three weeks expressed as a percent of initial force were lowest for the modules graded by the supplier as heavy but variation for each size and grade of module was so great that no significant difference could be demonstrated statistically. Mean remaining force figures representing pooled data for all sizes and grades of modules and all stretch distances were as follows: 54.7% at one hour, 45.3% at twenty four hours, 39.5% at one week and 25.1% at three weeks. Standard deviations figures were not provided in the article but the authors described the variation as being far in excess of that seen in latex elastics. This marked variation was said to reduce the operator's ability to predict long term force levels as compared to latex elastics. However, patient cooperation in the form of daily elastic application was effectively eliminated as a factor in treatment when elastomerics were used. This latter observation was cited as justification for continued use of plastic modules in spite of their inherent force variability. The authors suggested improved manufacturer quality control as a means of reducing said variability. As a general, qualitative observation, the authors noted that modules tended to exert greater force as they were allowed to dry.

The effect of tooth movement on polyurethane chain force decay behavior was first evaluated by Hershey and Reynolds in 1975[3]. They compared five different products representing three suppliers under conditions of fixed stretch distance and simulated space closure at rates of .25mm and .50mm per week. All products were stored in 37 degrees C water (as per Andreasen and Bishara) during the six week test period. While the Unitek grey and clear AlastiK chains consistently showed higher mean force values over time the differences between products amounted to a few percent of initial force levels at each measurement interval. This minimal variation between product means led the authors to the conclusion that all five were "clinically equivalent" though the Unitek clear and TP Laboratory's chain products showed three times the intraproduct variation. As for the effects of simulated space closure, the .25mm per week samples retained 33% of initial force at four weeks and the .50mm per week rate of closure yielded 25% remaining initial force at four weeks (these figures are based on pooled data for all five products). The fixed stretch distance control samples retained a mean 40% of initial force for the same period. At six weeks the corresponding figures were 42%, 28% and 18% for space closure rates of zero, .25mm and .50mm per week respectively. The authors did not specify force levels in absolute terms but did report that forces produced by all five products under both rates of simulated space closure should be adequate clinically to produce tooth movement as per Storey and Smith[4], Reitan[5], Begg[6], Hixon et al[7] and Boester and Johnston[8].

Varner[9,10] conducted a fixed stretch distance study to determine the force decay behavior of Unitek's KX-2 space closure modules. Based on clinical measurement of forty treatment patients a range of 26mm to 40mm was judged to cover virtually all clinical applications. Jigs were constructed which corresponded to eight fixed stretch distances covering the stated range by two millimeter increments. Ten KX units were stretched on each of the jigs and stored in 37 degrees C water for four weeks and removed only for periodic force measurements. Regression curves for the mean force versus logrithmic time transforms were constructed for each fixed stretch distance beginning with two hour force measurements. The slopes of these straight line regression curves were determined by analysis of covariance not to be significantly different from one another at a 99% confidence level. This would indicate that for the range studied KX-2 modules exhibit the same rate of force decay from two hours to four weeks regardless of initial elongation. Further statistical evaluation of the data using two-way analysis of variance and Scheffe' contrast numbers demonstrated a change in force decay rate occuring sometime during the second week. The author refers to this rate change as a "plateau" beyond which the force level was relatively constant up to four weeks. The mean decrease in force for the pooled data was 26% from two hours to four weeks which Varner judge to be "relatively small". Mean force levels at four weeks (ranging from 0.8 pounds to 1.3 pounds depending on stretch distance) were considered adequate for continued space closure beyond the test period up to six or perhaps even eight weeks.

In 1976 Wong[11] evaluated two brands of polyurethane chain (Ormco and Unitek) comparing the force decay characteristics of long and short segments of each brand. As was expected the short segments exerted a greater mean force than the long segments when both were stretched over the same distance (17mm interbracket distance; length of chain segments unspecified). The rates of force decay were comparable for long segments of each product with ultimate three week force decay levels at 67% and 69% loss of initial force. Data obtained in the short segment phase of the study showed marked dissimilarity between each brand's force decay behavior. Comparable length short segments of Ormco Power Chain and Unitek AlastiK C_2 double loop chain yielded initial force levels of 342 grams and 641 grams respectively. At twenty four hours the Ormco product had lost 50% of its initial force while the Unitek chain experienced 73% force decay. The mean three week force levels for the two products were the same (128 grams) which represented 37% retained initial mean force for the Ormco chain and only 21% of the mean initial force for the Unitek product. The author speculated that the 641 gram initial force exerted by the Unitek chain was much more likely to cause patient discomfort than the mean 342 gram figure registered by comparable lengths of Ormco's product. Since both products generated the same mean force at three weeks Wong suggested that the precipitous early force loss might be minimized by "prestretching" modules prior to their clinical application.

Kovatch et al[12] conducted a study to determine the effect that rate of stretch during application has on the force versus log time curve for AlastiK K-2 modules. Use of the Instron Universal Testing Maching allowed elongation of the space closure units at precisely controlled rates of 0.2, 2.0 and 20.0 inches per minute. Eight modules were stretched an additional 1.00 inch beyond a resting length of 0.63 inch at each of the three rates, held at this fixed stretch distance for two weeks in a 37 degree C saliva bath and periodically removed for force measurements on the Instron machine. The slowest rate of stretch yielded the lowest mean initial force (0.8 pounds) while a hundred fold elongation rate increase produced the highest mean initial force (1.2 pounds). The rapidly stretched modules also experienced the most rapid initial force decay such that the force versus log time plots for the rapid and slow rates of elongation crossed each other at two minutes. Thereafter the modules applied at the slowest rate maintained the highest mean force versus time. While the slowest rate of application did result in a sustained higher force over time it should be noted that a ten fold difference in elongation rate gave rise to a difference in mean force at two weeks of less than one ounce. Thus the authors recommend stretching plastic modules slowly to place even though the inevitable force decay is likely delayed a few hours at best.

Ash and Nikolai[13] were the first investigators to compare force decay behavior of elastomeric modules tested in vitro versus in vivo. Unitek's AlastiK CK chain and K-I modules were tested at a fixed stretch distance of 28mm in 37 degrees C air and water and in the mouths of eleven adolescent orthodontic patients. Fifteen of each module were tested in each of the three media. The air and oral environment mean force versus time data were significantly different from the first hour and throughout the remainder of the three week test period. Comparison of force decay data for the modules stored in the 37 degrees C water bath and those maintained in the oral cavity failed to demonstrate a significant difference up to one week for the CK chain and up to two weeks for the K-1 units. At three weeks both module types produced significantly lower mean force when maintained in the oral environment as compared to those stored in 37 degrees C water. The authors concluded that data obtained by in vitro testing in body temperature water can be used as a predictor of oral force behavior up to one week post-insertion. Beyond that point such data is at least suspect as a valid indicator of in vivo performance. However, this conclusion was qualified by the authors' own observation that the difference in mean force level observed in the two media at three week amounted to just over one and one half grams, a difference which was "difficult to judge to be clinically significant in view of the initial load (magnitudes) and the very substantial degree of relaxation overall".[13]

Two articles were published in 1979 which tested the efficacy of prestretching as a means of reducing precipitous inital force decay as suggested by Wong[11]. Young and Sandrick[14] utilized CK and C_2 AlastiK injection molded chain and the Instron Universal Testing Machine in their study. Four link segments of each were selected at random and stretched to various prestretch distances which were arbitrarily selected by the authors. The prestretching procedure was described as being performed "quickly by hand" after which "the chain was returned immediately to its original position (length)", all under dry conditions. Control chain segments were not prestretched prior to force decay testing on the Instron machine. Modules were stored in body temperature water during the brief twenty four hour force decay testing period. Statistical comparison of CK chain force decay data for the two prestretch sample groups (14mm and 23mm prestretch distances) and controls showed significantly decreased rates of force reduction for the prestretch groups at a 99% confidence level. The two prestretch sample groups were significantly different from one another at a 95% confidence level with the 23mm prestretch samples exhibiting the lower mean force decay rate. The C_2 chain samples, however, failed to show significantly different force reduction rates between any of three prestretch sample groups and controls. The authors could not explain the dissimilarity of response to prestretching exhibited by two chain materials produced by the same manufacturer using the same molding process. In spite of the unpredictable benefit demonstrated the authors recommended that elastomeric modules be prestretched either by the operator or the manufacturer. Though not fully explained, it appeared that this recommendation was based on the finding that the prestretching procedure was of some benefit in one product and of no apparent detriment in the other.

Brantley et al[15] conducted a study of the effects of prestretching in which the impact of prestretch storage medium and duration of prestretch were tested. Segments of two brands of spool chain (Unitek's AlastiK C Chain and Ormco's Power Chain) were prestretched to twice their passive length and maintained at that extention for either twenty four hours or three weeks. Two prestretch storage media were used: 37 degrees C water and 24 degrees C air. At the end of the prescribed prestretch period samples were removed from their respective prestretch environments and either applied immediately to fixed stretch distance jigs for force decay testing or allowed to relax in room temperature air for one or seven days prior to force decay evaluation. Data were grouped according to prestretch storage medium and duration and according to time elapsed between termination of prestretch and commencement of force decay testing. All experimental groups and controls (not prestretched) were stored in 37 degrees C water during the force decay evaluation period. Comparison of control group data to data obtained for each experimental group showed significant reduction in rate of decay when prestretched modules were stored in body temperature water and applied immediately upon removal from the water bath. Such prestretching techniques produced modules which provided "nearly constant forces" corresponding to a mean reduction in force over three week of as little as seven grams (as in the case of Ormco Power Chain samples prestretched in 37 degrees C water for three weeks and applied immediately to test jigs for force decay testing). Prestretching in room temperature air for up to three weeks was judged ineffective as a means of producing modules capable of delivering a near constant force. For each brand of chain the mean three week force levels were not significantly effected by method of prestretch such that the various means differed from one another by no more than five percent. Thus it was shown that prestretching elastomerics under stringently controlled conditions can effectively reduce the rate of force reduction without diminished long term force generating capability.

Two separate studies conducted by Doyle[16] and Porter[17] evaluated Unitek's C_1 grey AlastiK chain under in vitro conditions using a test apparatus which simulated typical chain and fixed orthodontic appliance relationships. In the former study[16] Plexiglass models were constructed which incorporated bands and brackets for canine, second bicuspid, first and second molar teeth, a rectangular archwire and a seven millimeter first bicuspid extraction space. Porter's study[17] utilized the same model, reducing the extraction site to five millimeters of residual space. He further adapted the apparatus to simulate an approximate one millimeter space closure during the four week test period. In both studies chain was applied over siamese bracket wings in an attempt to closely approximate clinical patterns of usage. Doyle's fixed stretch distance evaluation showed a 36% reduction in initial force at two hours, 39% at twenty four hours and 44% at four weeks. These figures represent a rate of mean force reduction somewhat less severe as compared to previous chain and K-type module studies conducted at fixed stretch distance on non-prestretched elastomerics.[1,2,3,9,10,11] The mean force at four weeks was approximatley 1.7 pounds which was judged by the author to be fully capable of elicting tooth movement. Based on this latter finding Doyle did not recommend replacement of chain clinically prior to four weeks.

Porter's simulated space closure study[17] utilized the same injection molded AlastiK C chain used in the former study. For purposes of comparison he also tested chain from the same supplier of the same material and dimensional specifications manufactured by an alternate method; that is, cut from an extruded sheet of polyurethane material. Initial and four week force levels for the two products were dissimilar at a 95% confidence level as demonstrated statistically by use of two-way analysis of variance. The injection molded material produced a greater initial mean force (1057 grams) than the "punched" chain (986 grams) but a lesser mean force at four weeks (214 grams and 264 grams respectively). These absolute measures correspond to relative rates of force loss of 80% and 73% respectively. These rates of force reduction were somewhat higher than those reported by Hershey and Reynolds in their simulated space closure evaluation of multiple brands of elastomeric chain.[3] Porter concluded that while there was demonstrable difference between AlastiK C_1 chain manufactured by these two methods on a statistical basis, the final force levels indicate capabilities for production of tooth movement such that the two should perform comparably in a clinical setting.

The force decay behavior of elastomeric thread was evaluated in two articles, one by Howard and Nikolai[18] and the other by Persson, Killaridis and Lennartson[19]. In the former study in vitro and in vivo comparison of a polyurethane product (TP Lab's Zing String) and a nylon covered latex thread was carried out over a twelve week test period. Statistical comparison of force decay data showed no significant difference between the two through six weeks product means in vivo. Six weeks was considered by the authors to be the maximum duration of usefulness for a single application of either product. The lack of difference in force decay behavior between the latex and polymeric threads should be considered in light of earlier studies which did demonstrate statistically significant differences between elastomerics (chain and K-type modules) and their latex counterparts[1,2]. These early studies consistently showed greater force reduction over time for elastomerics as compared to latex elastics. The rate of force decay reported by Howard and Nikolai for Zing String also contrasts sharply with rates reported in earlier studies. In vivo test samples retain 74% of their initial force at three weeks while in vitro levels reported ranged from 70% to 81% depending on initial force magnitude. These figures represent mean rates of force reduciton substantially lower than those reported previously for chain and K-type modules.[1,2,3,9,10,11]

Perrson, Killaridis and Lennartson[19] compared four polyurethane thread products (including Zing String by TP Labs) to two pure latex and two nylon covered latex products. In vitro testing was carried out over a three week period with samples stored in 37 degrees C saline solution. Rates of force decay for the four elastomeric threads varied substantially from one another exhibiting three week force reduction rates ranging from 20% (for Rocky Mountain's "Super Thread") to 65% ("Power Tube" by Ormco) loss of initial force. TP Lab's Zing String showed over 50% force reduction at three weeks as compared to 26% reduction in mean initial force reported by Howard and Nikolai for the same product.[18] The force curves for the four latex products were interspersed between those for the elastomerics with the latter at both extremes. Analysis of these data led the authors to much the same conclusion as was reached by Howard and Nikolai; that is, latex and polyurethane thread force decay behaviors are not significantly different.

The effects of temperature manipulation on in vitro test results were evaluated by De Genova et al[20]. In the pilot portion of a study comparing three different chain products the authors subjected samples to force decay evaluation under two storage conditions: constant temperature (37 degrees C) artificial saliva and the same 37 degree C storage medium subjected to twice daily "thermal cycling". The 15 degree C to 45 degree C temperature cycling range was selected based on a previous publication's report that these define the extremes in oral temperature resulting from ingestion of hot coffee and ice water[21]. The cycling procedure consisted of two half hour sessions per day of temperature modulation between the stated extremes with a thirty second dwell time. The author's interest in the effects of temperature on elastomeric force behavior stemmed from Ash and Nikolai's finding that comparative 37 degrees C water and oral environment testing yielded force decay data which were dissimilar beyond the first week[13]. Data from the present study demonstrated that thermal cycling resulted in retention of a significantly higher percent of initial force as compared to constant temperature sample storage. Based on this statistically significant difference in force decay behavior the authors concluded that thermal cycling better simulated oral conditions and was more suitable for in vitro testing than constant temperature storage. More careful review of Ash and Nikolai's results, however, show that storage of samples in the oral cavity resulted in retention of a lesser percent of initial force over time as compared to constant temperature water storage. If thermal cycling as prescribed by De Genova et al had truly created an environment more comparable to oral conditions one would have expected a more rapid rate of force decay as compared to fixed temperature storage.

Brooks and Hershey[22] reported that an unspecificied procedure simulating ingestion of hot fluids resulted in a force decay rate higher than that exhibited by control samples stored under constant temperature conditons. Although this pattern mirrors the results of Ash and Nikolai[13] these findings weren't compared to in vivo data for verification that this thermal cycling procedure better simulated conditions in the oral environment. The authors seemed more interested in the potential for counteracting the effects of heating by prestretching the elastomerics. Lacking evidence to the contrary, it would seem that constant body temperature water storage of samples as prescribed by Andreasen and Bishara[1] is as a valid a technique for in vitro testing of plastic modules as any prescribed to date.

A "second generation" polyurethane material has recently appeared on the market. Specific information related to chemical differences between polymers and additive modifiers is considered proprietary information[23]. Information regarding force decay behavior is, on the other hand, available since these materials have been the subject of two recently published articles. The first of these, by De Genova et al[20], was already mentioned with regard to thermal cycling and the second was conducted by Killiany and Deplessis[24]. The former[20] compared two "traditional" polymer products, Ormco Power Chain II and TP Lab's Elastochain, to the second generation Rocky Mountain Energy Chain. Under in vitro conditions already discussed, the authors found that the Rocky Mountain product retained higher force levels throughout the three week test period for all degrees of initial stretch at a statistically significant level ($p < 0.0l$). The latter study[24] showed similar results when comparing Rocky Mountain Energy Chain with American Orthodontic's Short Plastic Chain. While not subjected to t-test analysis the Energy Chain notably retained a mean 67.4% of it's initial force at three weeks as compared to 32.8% for the American Orthodontics product. The mean initial force level for the Energy Chain (329.8 grams) was lower than the mean initial force produced by comparable lengths of the traditional material (374.9 grams). After only one day, the Energy Chain generated the greater mean force (255.0 grams versus 169.2 grams), a pattern which continued throughout the eight week period. In fact, the Energy Chain retained a greater mean force at eight weeks (182.8 grams) than was exhibited by the traditional product at the end of one day (169.2 grams). The authors characterized the second generation polymer as a significantly improved alternative to traditional competitors not proven clinically to date.

Rock et al[25] conducted an in vivo evaluation of eight chain products in an attempt to correlate pre- and post-treatment force production to measured space closure. The eight products were selected based on a pilot study[26] as representative of the full range of initial force generating capabilities. Three and four link segments of each were randomly selected and utilized clinically in space closure. Initial stretch distance was determined clinically for each case and the module was tested to said distance on the Instron Universal Testing Machine. After twenty eight days the chains were removed and stored in tap water until force levels could be measured again on the Instron machine. Attempts were made to correlated magnitude of tooth movement with any of the following variables: initial force level, percent initial extention and percent reduction in force. No meaningful correlations were discovered which the authors attributed to "the several (unspecified) variables involved".[25]

Another in vivo investigation, conducted by Sonis et al[27], set out to test the relative effectiveness in space closure of a nylon covered latex thread, a traditional polyurethane chain (Unitek's AlastiK) and Rocky Mountain's Energy Chain. Twenty five patients undergoing symmetrical extraction orthodontic treatment were selected for the study. Latex thread was applied to each of forty quadrants with one of the two chain products on the contralateral side. Initial force levels ranged from 350 to 400 grams as determined by strain gauge measurement at the time of application. For both elastomeric chain materials space closure rates were not significantly different from rates seen in contralateral quadrants treated with elastic thread as determined by t-test. Three way analysis of variance similarly showed no treatment difference in spite of predicted differences in force decay rates. These findings are in agreement with those of Boester and Johnston[8] but did not corroborate those of Hixon et al[7].

METHODS AND MATERIALS

As stated in the introduction, significance is often attached to the operator's subjective appraisal of ligature module "stiffness". The relative difficulty encountered in application of module to bracket is frequently taken as a predictor of long term performance. This presumed relationship between initial force applied by a previously unstretched, dry module and longitudinal force production is the premise upon which this study's experimental design was based. In phase one of the study randomly selected samples of commercially available products were stretched to a predetermined distance approximating the distention required for application of ligature modules to the selected test bracket. Resultant force means and standard deviation figures were analysed statistically to determine which product means constituted significantly different "initial treatment effects". These data were in turn used to select five products for longitudinal force evaluation. This evaluation of force production over time will be referred to in subsequent discussion as phase two of the study.

PHASE ONE

Samples of twelve ligature module products were obtained from seven different suppliers (see table 1). A few were obtained by departmental purchase order while most were donated by the respective supply representative. Those companies donating samples of their product were apprised of the use intended for their elastomerics. Upon receipt all products were kept sealed in their plastic shipment wrappings until the commencement of force testing. Exposure to light was minimized during storage in an attempt to reduce surface oxydation of the polyurathane material.

An Instron Universal Testing Machine was used for all force testing in both phases one and two of the study. The Instron machine allows precise measurement of force at a selected linear displacement. This displacement is achieved by controlled vertical movement of the machine's lower table relative to the fixed upper member. Attachments for the upper and lower elements of the machine must be fabricated to fit the specific needs of the material being tested. In phase one of this study two hooks made of 0.045 inch diameter stainless steel wire were adapted for attachment to both upper and lower members of the Instron machine. A micrometer was used to determine the dimensions of the orthodontic bracket selected for use in the phase two test model. Based on these measurements it was determined that 0.16 inch displacement between test hooks would simulated the distention required for placement of an elastomeric module over the test bracket and archwire. A sample of twelve modules randomly selected from each of the twelve product samples were stretched 0.16 inch at a constant rate of stretch or "crosshead speed" of 2.0 inches per minute. The force means and standard deviations derived from this procedure are listed in descending mean order in table 1. Comparison of these data by one-way analysis of variance (ANOVA) allowed grouping of those force means not significantly different from one another. This analysis also identified those product means which constitute significantly different "initial treatment effects" at a ninety five percent confident level.

Based on the above data four products were selected for phase two testing. These four products represented the full range of mean initial force production, each mean characterized as a distinctive "initial treatment effect". Other criteria for selection included the requirement that the four products should come from four different supply companies and both injection molded and extruded products be included. It also seemed appropriate that products which have seen some use in the OHSU Graduate Orthodontics Department should be tested where possible (that is, without violating the other selection criteria).

PHASE TWO

Several criteria guided the process of designing a test model for longitudinal force testing. Reliable maintenance of a constant stretch dimension for each module was imperative. Given the small size of the ligature modules it seemed appropriate that the design allow scheduled force testing without the necessity of removal of module from test model. Planned storage of ligature modules in small vials of water imposed certain dimensional limitations on the test model. A model configuration that mimics the clinical relationship of ligature module to archwire and bracket seemed desirable. Finally, ease of fabrication was important in light of the large number of test models which would eventually be required.

The design chosen fulfilled all the above requirements (see figures 1-4). American Orthodontics No. 001-698 siamese brackets with 0.022 inch slots were welded to one half inch segments of 0.242 inch inside diameter stainless steel tubing. Short segments of 0.022 x 0.028 inch stainless steel wire were used as the models' "archwire". The relative rigidity of this archwire material and consequent minimal flex during force testing prompted its choice for application in the test model. Segments of 0.020 inch round stainless steel wire were maintained in the bracket slots under the rectangular "archwires" during storage, being removed just prior to and replaced immediately following scheduled force testing. These round wire segments provided a force tending to unseat the archwire from the bracket slot and simulated clinical use of elastomeric ligature modules as a means of achieving a first order tooth movement. The 0.020 inch diameter wire was chosen specifically because it produced a considerable unseating force between archwire and bracket while still allowing partial maintenance of the rectangular archwire within the confines of the bracket slot.

Test models were affixed to the lower member of the Instron machine by means of a stainless steel yoke and removeable horizontal pin assembly (see figure 2). The pin fits simultaneously through a horizontal track in the yoke and the lumen of the tubing thereby attaching the test model to the Instron's moveable lower member. A double hook assembly was fabricated which attaches to the Instron's upper member by means of a horizontal pin. The hook assembly was designed such that the two hooks fit snugly on either side of the test models orthodontic bracket while resting under the rectangular archwire segment (see figure 3). The interlocking loops in the center of the shaft of the hook assembly serves as a universal joint, preventing binding in three plains of space (see figure 4). When the lower table of the Instron is mechanically depressed, the dual hook system effectively "lifts" the archwire from the test model's bracket slot.

Precise control of the amount of lift is critical to each scheduled force measurement. After installation of the test model in the yoke, the hook assembly is engaged and the lower table is slowly depressed by manual control until the force recorder first registers measureable force. The point at which positive force is first registered corresponds to the moment at which the archwire is first lifted from the bracket slot. After locating this starting point the controls of the Instron machine are set to automatically lower the test model 0.020 inch at a rate of 0.5 inch per minute crosshead speed. Upon arrival at the prescribed 0.020 inch displacement, the machine automatically reverses directions and returns the test model to the unloaded position.

The above outlined measurement procedure was followed for each test model at each of nine scheduled periodic test intervals. Previous elastomeric studies [1,2,3,11] indicate that precipitious loss of force should be expected during the first few hours after placement of the polyurethane modules and that up to three fourths of initial force loss may occur during the first twenty four hours[1]. Thus it seemed appropriate to plan multiple force measurements during the first day followed by weekly measurements thereafter. Most studies have reported initial force figures; that is forces measured immediately at the point of achieving fixed stretch distance. The time required for assembly of each test model/archwire/module complex added to the time required for location of a starting point for each model made it impossible to measure force at the moment of "initial activation". Therefore the first force measurements were made ten minutes after module placement. Subsequent measurements were scheduled at two hours, six hours, twenty four hours and weekly up to four weeks. Six week measurements were also accomplished for all sample groups.

The first two sample sets of ten modules each served as a pilot group in that the force testing of these twenty modules served to identify certain technical and logistical problems which were subsequently worked out. This first run also permitted familiarization with the management of the Instron machine itself. Due to the presence of these variables many of the early measurements are suspect. As a result, data for these two sample sets was used for apparatus familiarization only and will not be reported in this paper.

The main group of test samples were started one week after the pilot group. Five samples of ten modules each representing four products comprised the main test group. Four of the sample sets were characterized by ligature modules applied to test models in the conventional "O" configuration. The fifth sample set, whose ligature modules were crossed over themselves to form a "figure 8", consisted of modules of the product whose mean force value in phase one was the lowest of all the modules tested.

A third group of samples were tested beginning two weeks after the main group. This supplemental group was started in response to a problem encountered with one brand of test module. This particular product showed a marked propensity to fracture such that thirty percent of the modules of this brand failed during the first two weeks of testing. As it happened, this was also the product selected for "figure 8" applicaiton. Of the latter sample, sixty percent had failed two weeks into the test period. Since this problem greatly limited the potential comparative value between crossed (or "figure 8") and uncrossed module data, a fifth product was selected for testing from the lower end of the phase one mean force hierarchy. It should be noted that this product also met all the previously mentioned selection criteria. Two sample sets were randomly selected from this fifth product: fourteen modules of this product were applied in "figure 8" fashion and twelve were applied conventionally. The sample sizes were increased somewhat as compared to sample sets used in the main test group (ten each). The increase was intended as a means of compensation for potential module fracture such that a reasonable sample size at four weeks would be assured. This precaution was taken in spite of the fact that module failure had not as yet occurred among any of the other three main test group product sets.

RESULTS

Data obtained during the six week test period including force means, standard deviations and sample sizes are listed for each of seven sample sets and nine scheduled test intervals in table 2. These data reflect a 48% loss of force from ten minutes to four weeks (based on data pooled from all seven sample groups) with a 28% to 60% range of decay rates demonstrated by individual sample sets. Overall force decay rates (again based on pooled data) at 24 hours and one week are 38% and 44% respectively. The figure eight applied samples showed somewhat higher rates of force loss at 24 hours, one week and four weeks: 43%, 52% and 56% respectively. For purposes of comparison conventionally applied sample groups showed 34%, 37% and 41% loss of force for the same respective measurement intervals. The force decay patterns for all seven samples are represented graphically in figure 5.

Early force measurements (those taken prior to 24 hours) were made in this study to characterize the entire force decay cycle. The early figures allow comparison of decay rates to those reported in other studies[9,10,16,17] whose data also include multiple first day measurements. Pre-twenty four hour measures are also of some interest clinically since elevated early force levels may subject the patient to a greater potential for discomfort. These early figures are not, however, of particular interest when considering force magnitude as it relates to tooth movement. Forces in effect during the latter part of the 30 day cycle of the tooth movement reaction are of greater interest since tooth movement is generally observed in that portion of the cycle. Thus, the statistical treatment of the experimental data will utilize those data obtained at 24 hours and at weekly intervals thereafter.

1. PRODUCT SELECTION AND FORCE PRODUCTION

Force means for the five conventionally applied sample sets were compared statistically using two way analysis of variance (see Appendix 1). This analysis demonstrated the overall effects of the two main variables (time and material or product) using Tukey contrast numbers[28] for comparison of composite product and timed force means. We will consider first the effects of time on force behavior. The force mean associated with the 24 hour measurement interval was shown to be similar on a statistical basis to the one week mean but differed significantly from all subsequent timed force means. The one week mean was not dissimilar from any of the subsequent force means through six weeks which indicates that forces from one to six weeks were, for the samples tested, relatively constant.

Analysis of the five product force means (Appendix 1) identified three statistically discrete levels of long term force production. The GAC and A Company products were indistinguishable on a statistical basis and produced the highest overall forces. Ormco and Rocky Mountain samples were also similar to one another with regard to their long term force behavior producing force levels at the other end of the spectrum. The UniteK product mean was intermediate and distinct from the other four products by the same contrast number analysis. Since these three statistically distinct "treatment effects" can be identified it would be reasonable to state that choice of product significantly influence force production.

2. PREDICTION OF LONG TERM FORCE BASED ON INITIAL FORCE

The preceeding analysis utilized force means for each of five products which had been generated by the ANOVA program. These means characterize the long term force behavior of each product sample as tested. If initial or early force were highly correlated with this long term behavior, it would be possible to select a "best product" from a group of products based on initial force levels. To test this notion, the ten minute means for each of the five products were paired with the corresponding ANOVA generated product means (see Appendix 1). The correlation coefficient generated by this procedure was $r=0.609$. Comparison of this figure to the appropriate table value ($r=0.878$)[33] indicated that early force figures were not predictive of long term force behavior for the samples tested in this study. It should be pointed out, however, that our ability to show significant correlation was somewhat limited by the small number of paired means. Future evaluation using a larger number of samples may be warrented.

3. MODULE CONFIGURATION AND FORCE PRODUCTION

A second two way analysis of variance (ANOVA) was performed to test the effect of pattern of module application (see Appendix 3). Due to a high module failure rate, Unitek samples were deleted from this analysis. Only conventionally applied and figure eight GAC sample force means were used. The two overall sample means generated in this analysis were significantly different from one another by Tukey contrast number analysis[28] as were individual cell means at each measurement interval. This finding indicates that a statistically significant benefit was realized through use of the higher force producing figure eight module configuration.

4. ERROR ANALYSIS

The four week force assessment of one product set's modules (Rocky Mountain) was repeated to determine the magitude of error associated with the force measurement procedure. The two measurement sessions were separated by approximately forty five mintues. In the intervening period other samples were tested and the Instrom machine was recalibrated and rebalanced. The standard error of measure ($\sqrt{\frac{\sum d^2}{2N}}$) was determined to be 0.0208 pounds (9.4 grams) or 2.5% of the overall mean force at four weeks.

DISCUSSION

Earlier elastomeric studies [1,2,3,11,12] have shown that force decay in polyurethane products is generally most precipitous during the first twenty four hours of activation after which force loss up to four weeks is progressive at a much reduced rate. Data from the present study exhibited a similar pattern of force decay. In the previous section this force reduction was expressed for data pooled from all sample sets in terms of percent of ten minute force levels. Similar relative rates of force reduction were reported for pooled data segregated on the basis of module application pattern: that is, figure eight versus conventionally applied sample sets. Let us now consider force decay over time in terms of absolute force measurements (table 2). At one day the mean force for all conventionally applied sample groups was 0.88 lbs. One and four week means were 0.83 and 0.79 lbs respectively. The sample exhibiting the lowest force production at six weeks registered a mean of 0.68 pounds. This last figure is pivotal in application of these data to clinical orthodontics as it represented a relative lower limit of ultimate force potential in the samples tested.

How appropriate are these forces from the standpoint of potential for producing tooth movement? Continuous force on the order of magnitude of 30 grams (0.066 lbs) has been reported as capable of producing tooth movement[5]. Closing loop mechanics utilizing Bull loops for space closure have been reported to product 500 grams (l.l lbs) of force with the typical one millimeter activation[5]. Clearly a wide range of force magnitudes have been shown capable of producing orthodontic tooth movement. Concepts of optimal force[4] and differential force[6] which pronounce certain force levels to be most appropriate from the standpoint of physiologic responses or maintenance of anchorage have been invalidated by recent reports[7,8]. Thus it would seem that the forces observed in the present study can be considered fully capable of eliciting tooth movement and would not be characterized as excessive.

One of the stated objectives of this study was to determine the relative benefit of the figure eight module application procedure. As previously discussed, side by side comparison of module performance data for the two GAC samples showed a statistically significant advantage conferred by use of the figure eight technique. A benefit would only be realized, of course, if force magnitude and rate of tooth movement were proportionatly related to one another. Hixon et al[7]proposed such a relationship between magnitudes of force and differential rates of tooth movement while Boester and Johnston[8] demonstrated that forces ranging from 140 to 310 gms (0.31-0.68 lbs) produced similar rates of tooth movement independent of specific force magnitude. Whether such a relationship exists or not it is doubtful that an overall mean force difference of 0.18 lbs (82 gms - see Appendix 3) would result in clinically apparent acceleration of tooth movement.[25,27] Based on the above considerations it could be said that, from the standpoint of enhanced tooth movement use of the figure eight technique has little potential for producing measureable clinical benefit.

Another potential advantage to be gained from higher forces through use of the figure eight procedure is maintenance of full bracket engagement. Miuri et al[29] reported that a .020 inch (0.5mm) deflection of .016 stainless steel archwire corresponds with approximately 500 grams or 1.1 pounds of resultant force. The five conventionally applied module samples produced mean forces at or below this level after only two hours of use while the figure eight samples fell below 500 grams sometime between one day and one week. Forces required for full bracket engagement would be substantially higher for rectangular stainless steel archwires[30]. In the same article Miuri et al reported that the various nickle titanium alloy archwires of .0l6 inch diameter exert 260-360 grams when deflected .020 inch (0.5mm) as in our experimental model. Three of the five conventionally applied samples and both of the figure eight samples exhibited mean force at all measurement intervals in excess of 360 grams (0.79 lbs). Thus it would seem appropriate to apply modules in a figure eight when using round superelastic archwires for the sake of sustained bracket engagement. This same procedure could not, however, be relied upon to maintain full bracket engagement for most stainless steel archwires not fully engaged passively at the time of module application.

The high rate of module failure exhibited by figure eight applied samples of one of the products tested in the present study raises the possibility of a detrimental effect of the figure eight procedure. The Unitek convetionally applied samples showed a 30% fracture rate while the Unitek figure eight sample sustained 70% module failure over the entire test period. None of the other sample groups exhibited any such module failure.

These observations raise the following question: does the figure eight procedure significantly increase fracture potential? The two module failure rates were compared by chi square analysis. The calculated chi square (3.21) was smaller than the appropriate table value (3.84) indicating that no significant difference in rates of failure had been demonstrated.[31,32]

SUMMARY AND CONCLUSIONS

Twelve elastomeric ligature module products were obtained for evaluation. Five of these were selected for longitudinal force evaluation based on initial force production demonstrated in a linear dry stretch test. Selection of the five products was based on statistical analysis of the dry stretch test data using one way analysis of variance (ANOVA). Seven sample sets of ten or more modules were applied to test units: five sets were applied in the conventional "O" configuration while the samples in the two remaining sets were crossed between bracket wings to form a "figure 8". All samples were stored in 37 degrees C water and tested periodically to determine the mean force production over time for each sample set. Force measurements were made at a relative wire-to-bracket displacement of 0.020 inch (0.5mm). This same bracket-to-archwire relationship was also maintained during constant temperature water storage. Data thus obtained was analysed using the following statistical procedures: two way analysis of variance (ANOVA) and linear regression/correlation coefficient analysis.

The following conclusions were made based on the above outlined data and subsequent analyses:

1. A 48% reduction in force was observed from ten mintues to four weeks, most of which (38%) had occured by the end of the first day (based on data pooled from all main and supplemental sample sets). The four week force decay figures were somewhat higher for the figure eight samples (56%) while the conventionally applied sample sets registered 41% force loss for the same period.

2. The overall force behavior of the figure eight samples was significantly superior to that of the conventionally applied samples on a statistical basis. However given the magnitude of mean differences it is doubtful that this superior force generating capacity will enhance tooth movement.

3. Appraisal of "stiffness" during module application is not likely to be an effective or reliable means of selecting a product capable of producing predictably superior force over time.

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TABLE 1

Phase I (Dry Stretch) Force Means

Product	Mean Force (lbs.)	Standard Deviation
AI: A Co. (injection molded)	2.38	0.172
RI: Rocky Mountain (inj.)	2.25	0.079
RE: Rocky Mountain (extruded)	2.13	0.084
AE: A Co. (extruded)	2.11	0.069
LI: Lancer Pacific (inj.)	1.97	0.060
OI: Ormco (inj.)	1.89	0.064
LE: Lancer Pacific (extr.)	1.79	0.098
GI: GAC (inj.)	1.70	0.077
GE: GAC (extr.)	1.63	0.023
OE: Ormco (extr.)	1.58	0.080
SI: OIS (inj.)	1.56	0.072
UI: Unitek (inj.)	1.46	0.059

Contrast Number (from one way Analysis of Variance): 0.113

Comparison of Means using Contrast Number Analysis*

	AI	RI	RE	AE	LI	OI	LE	GI	GEI	OE	SI	UI
AI	–	s	s	s	s	s	s	s	s	s	s	s
RI	s	–	s	s	s	s	s	s	s	s	s	s
RE	s	s	–	ns	s	s	s	s	s	s	s	s
AE	s	s	ns	–	s	s	s	s	s	s	s	s
LI	s	s	s	s	–	ns	s	s	s	s	s	s
OI	s	s	s	s	ns	–	ns	s	s	s	s	s
LE	s	s	s	s	s	ns	–	ns	s	s	s	s
GI	s	s	s	s	s	s	ns	–	ns	s	s	s
GE	s	s	s	s	s	s	s	ns	–	ns	ns	s
OE	s	s	s	s	s	s	s	s	ns	–	ns	s
SI	s	s	s	s	s	s	s	s	ns	ns	–	ns
UI	s	s	s	s	s	s	s	s	s	s	ns	–

* "ns" indicates no significant difference between means while "s" signifies a statistically significant difference at a probability level $p < 0.05$.

TABLE 2

CONVENTIONALLY APPLIED SAMPLES

	10 min.	2 hr.	6 hr.	1 day	1 week	2 weeks	3 weeks	4 weeks	6 weeks
Unitek (inj.)	1.09 n=10 s=.070	0.92 n=10 s=.036	0.88 n=10 s=.035	0.85 n=9 s=.017	0.86 n=7 s=.014	0.84 n=7 s=.024	0.79 n=7 s=.022	0.78 n=7 s=.017	0.78 n=7 s=.024
ACo. (inj.)	1.51 n=10 s=.136	1.14 n=10 s=.094	1.05 n=10 s=.077	0.94 n=10 s=.033	0.92 n=10 s=.028	0.90 n=10 s=.032	0.89 n=10 s=.024	0.87 n=10 s=.028	0.86 n=10 s=.024
Ormco (inj.)	1.21 n=10 s=.070	0.90 n=10 s=.020	0.87 n=10 s=.006	0.84 n=10 s=.028	0.73 n=10 s=.033	0.70 n=10 s=.032	0.68 n=10 s=.033	0.68 n=10 s=.087	0.68 n=10 s=.033
Rocky Mtn. (ext.)	1.33 n=10 s=.120	0.93 n=10 s=.069	0.86 n=10 s=.042	0.81 n=10 s=.067	0.76 n=10 s=.050	0.76 n=10 s=.068	0.72 n=10 s=.078	0.71 n=10 s=.081	0.68 n=10 s=.082
GAC (ext.)	1.46 n=12 s=.180	0.93 n=12 s=.030	0.98 n=12 s=.024	0.93 n=12 s=.022	0.88 n=12 s=.017	0.88 n=12 s=.014	0.88 n=12 s=.007	0.87 n=12 s=.014	0.87 n=12 s=.010
Timed Means from ANOVA (see Appendix 1)				0.86	0.83	0.82	0.80	0.79	0.78

FIGURE EIGHT APPLIED SAMPLES

	10 min.	2 hr.	6 hr.	1 day	1 week	2 weeks	3 weeks	4 weeks	6 weeks
Unitek (inj.)	1.92 n=10 s=.131	1.39 n=7 s=.089	1.28 n=7 s=.092	1.16 n=7 s=.063	1.03 n=5 s=.062	1.05 n=4 s=.039	1.01 n=4 s=.020	1.00 n=3 s=.010	1.00 n=3 s=.041
GAC (ext.)	2.45 n=14 s=.142	1.32 n=14 s=.129	1.48 n=14 s=.139	1.33 n=14 s=.132	1.07 n=14 s=.060	1.02 n=14 s=.041	1.01 n=14 s=.056	0.99 n=14 s=.046	1.00 n=14 s=.045

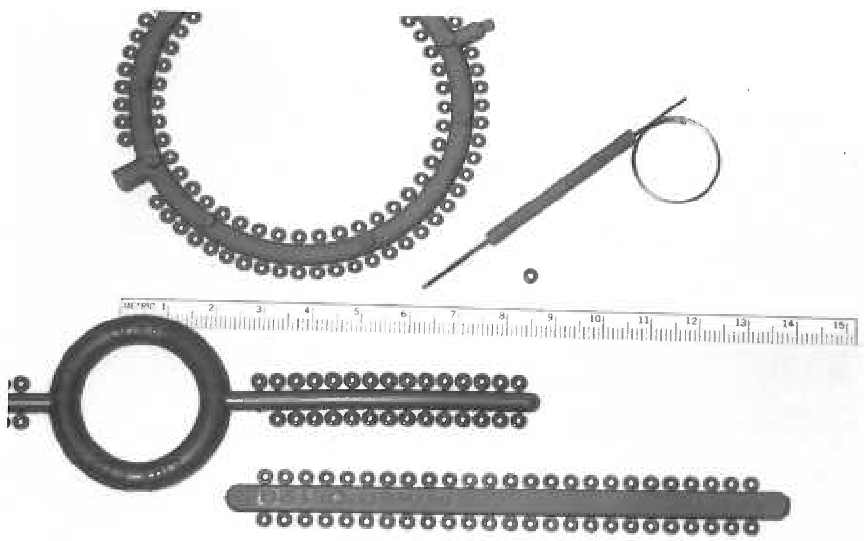


Fig. 1A Polyurethane ligature modules are supplied in various configurations by the different supply companies.

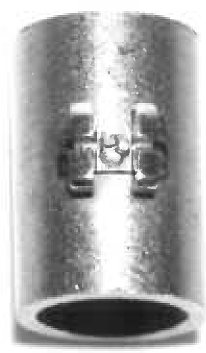


Fig. 1B The test model.

Fig. 2 Stainless steel yoke and horizontal
pin assembly with test model in place.

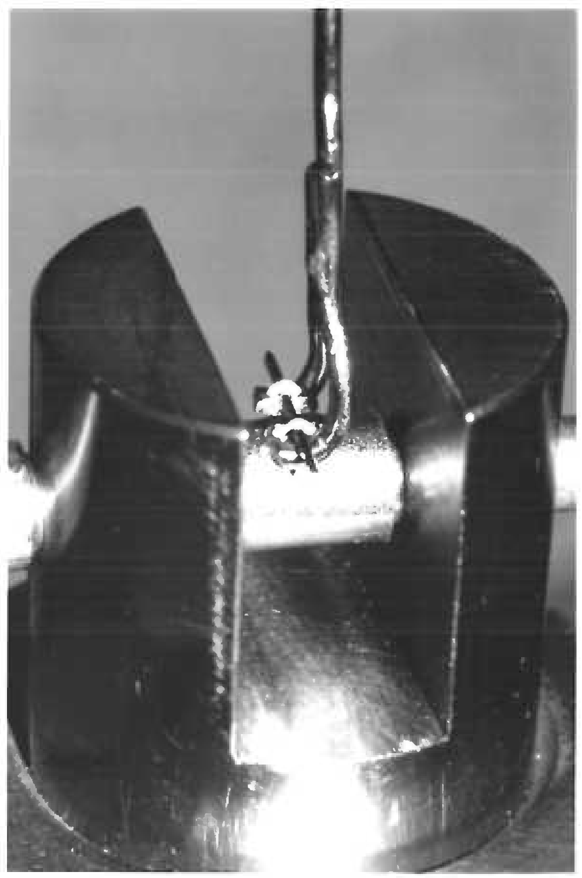


Fig. 3 Close-up view of test model with hook assembly attached.

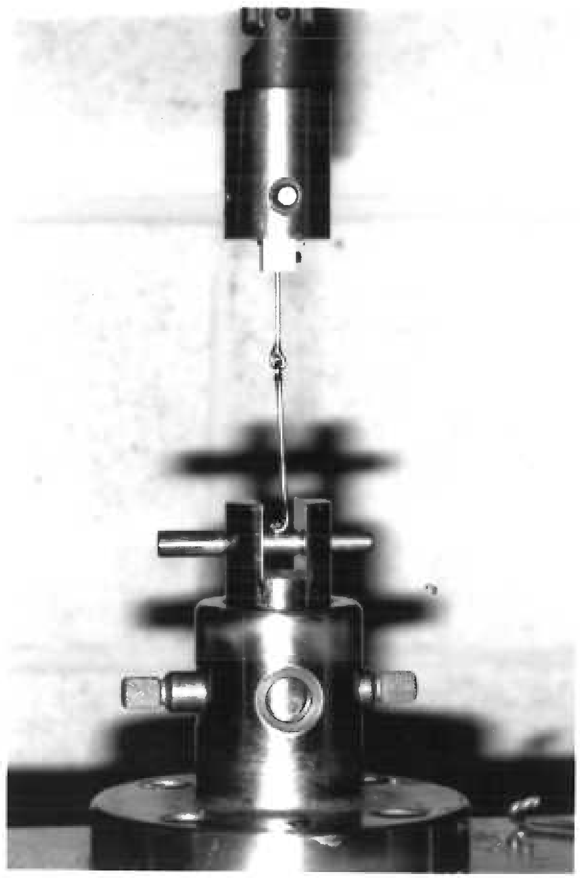
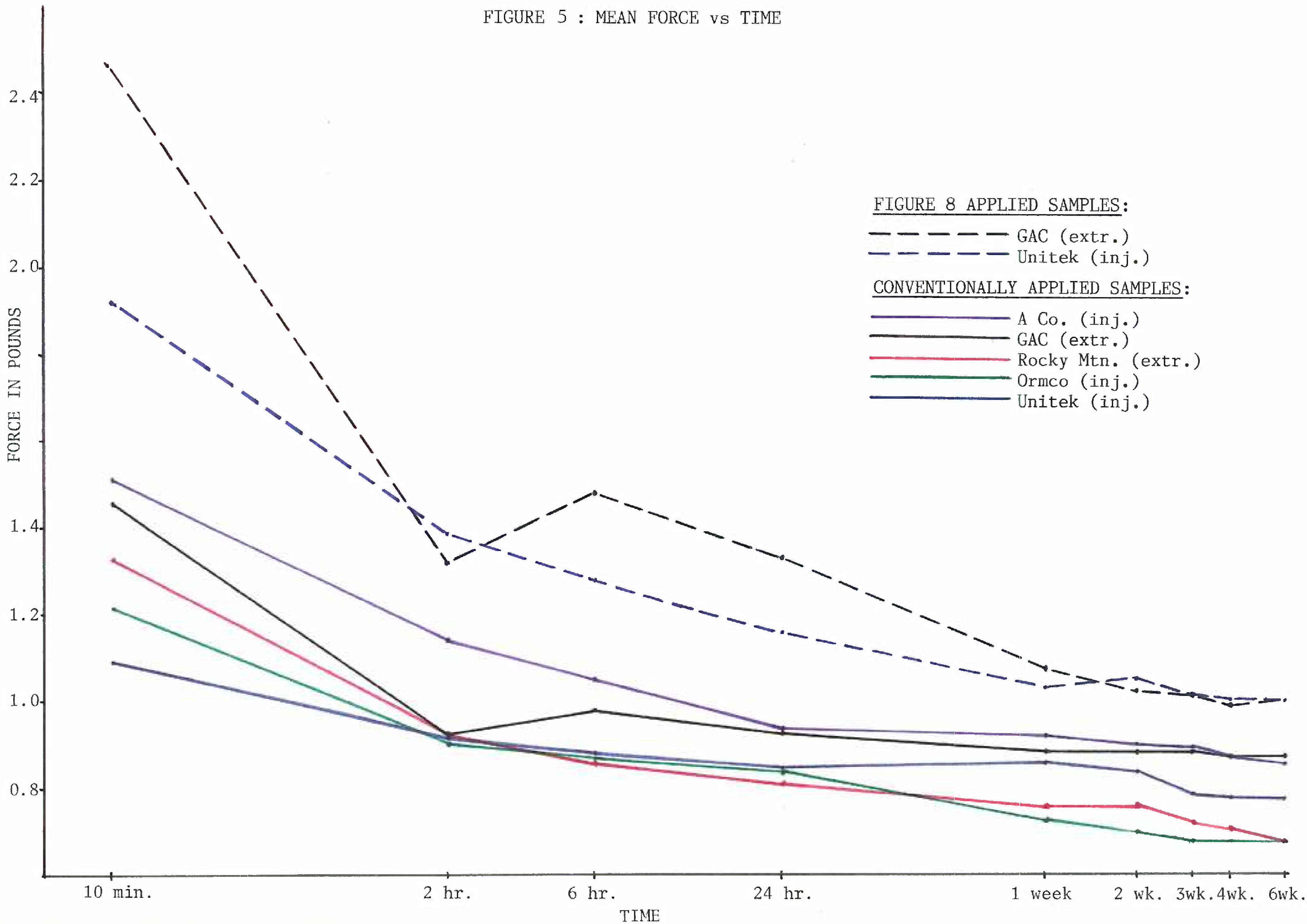


Fig. 4 All elements of force testing apparatus in place ready for phase two test proceedure.

FIGURE 5 : MEAN FORCE vs TIME



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N FACTOR ANALYSIS OF VARIANCE

06/11/87

	A	B	C	n	MEAN	VAR
1	1	1	1	9	.85	.0003
2	2	1	1	7	.86	.0002
3	3	1	1	7	.84	.0006
4	4	1	1	7	.79	.0005
5	5	1	1	7	.78	.0003
6	6	1	1	7	.78	.0006
7	1	2	1	10	.94	.0011
8	2	2	1	10	.92	.0008
9	3	2	1	10	.9	.001
10	4	2	1	10	.89	.0006
11	5	2	1	10	.87	.0008
12	6	2	1	10	.86	.0006
13	1	3	1	10	.84	.0008
14	2	3	1	10	.73	.0011
15	3	3	1	10	.7	.001
16	4	3	1	10	.68	.0011
17	5	3	1	10	.68	.0076
18	6	3	1	10	.68	.0011
19	1	4	1	10	.81	.0045
20	2	4	1	10	.76	.0025
21	3	4	1	10	.76	.0046
22	4	4	1	10	.72	.0061
23	5	4	1	10	.71	.0065
24	6	4	1	10	.68	.0068
25	1	5	1	12	.93	.0005
26	2	5	1	12	.88	.0003
27	3	5	1	12	.88	.0002
28	4	5	1	12	.88	.00005
29	5	5	1	12	.87	.0002
30	6	5	1	12	.87	.0001

SOURCE	SS	DF	MS	F
A	.3342990	5	.0668598	38.013
B	1.651415	4	.4128538	234.731
AXB	.0951221	20	.0047561	2.704
ERROR	.46785	266	.0017588	

------- MEANS AS ENTERED -------

TIMES/MTL		TIMES		MTL	
24HRS/UI	.85	24HRS	.876666	UI	.818181
1WK/UI	.86	1WK	.830204	AI	.896666
2WK/UI	.84	2WK	.817142	OI	.718333
3WK/UI	.79	3WK	.795714	RE	.74
4WK/UI	.78	4WK	.785714	GE	.885
6WK/UI	.78	6WK	.777551		
24HRS/AI	.94				
1WK/AI	.92				

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N FACTOR ANALYSIS OF VARIANCE

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2WK/AI	.9
3WK/AI	.89
4WK/AI	.87
6WK/AI	.86
24HRS/OI	.84
1WK/OI	.73
2WK/OI	.7
3WK/OI	.68
4WK/OI	.68
6WK/OI	.68
24HRS/RE	.81
1WK/RE	.76
2WK/RE	.76
3WK/RE	.72
4WK/RE	.71
6WK/RE	.68
24HRS/GE	.93
1WK/GE	.88
2WK/GE	.88
3WK/GE	.88
4WK/GE	.87
6WK/GE	.87

-------- MEANS ORDERED --------

TIMES/MTL		TIMES		MTL	
3WK/OI	.68	6WK	.777551	OI	.718333
4WK/OI	.68	4WK	.785714	RE	.74
6WK/OI	.68	3WK	.795714	UI	.818181
6WK/RE	.68	2WK	.817142	GE	.885
2WK/OI	.7	1WK	.830204	AI	.896666
4WK/RE	.71	24HRS	.876666		
3WK/RE	.72				
1WK/OI	.73				
2WK/RE	.76				
1WK/RE	.76				
4WK/UI	.78				
6WK/UI	.78				
3WK/UI	.79				
24HRS/RE	.81				
2WK/UI	.84				
24HRS/OI	.84				
24HRS/UI	.85				
6WK/AI	.86				
1WK/UI	.86				
6WK/GE	.87				
4WK/AI	.87				
4WK/GE	.87				
1WK/GE	.88				
3WK/GE	.88				
2WK/GE	.88				
3WK/AI	.89				
2WK/AI	.9				
1WK/AI	.92				

DENTAL MATERIALS SCIENCE OHSU
N FACTOR ANALYSIS OF VARIANCE

06/11/87

24HRS/GE	.93
24HRS/AI	.94

CONTRASTS FOR MEANS OF CELLS:
Q(30 . 266) = 5.434 CONTRAST = 0.07355701

CONTRASTS FOR MEANS OF FACTOR A:
Q(6 . 266) = 4.096 CONTRAST = 0.05469347

CONTRASTS FOR MEANS OF FACTOR B:
Q(5 . 266) = 3.917 CONTRAST = 0.05296099

DENTAL MATERIALS SCIENCE OHSU
N FACTOR ANALYSIS OF VARIANCE

06/11/87

	A	B	C	n	MEAN	VAR
1	1	1	1	12	.93	.0005
2	2	1	1	12	.88	.0003
3	3	1	1	12	.88	.0002
4	4	1	1	12	.88	.00005
5	5	1	1	12	.87	.0002
6	6	1	1	12	.87	.0001
7	1	2	1	14	1.33	.0173
8	2	2	1	14	1.07	.0036
9	3	2	1	14	1.02	.0017
10	4	2	1	14	1.01	.0031
11	5	2	1	14	.99	.0021
12	6	2	1	14	1	.002

SOURCE	SS	DF	MS	F
A	.8422769	5	.1684553	60.304
B	1.326876	1	1.326876	475.003
AXB	.3783230	5	.0756646	27.086
ERROR	.40225	144	.0027934	

------- MEANS AS ENTERED -------

TIMES/CONFIG		TIMES		CONFIG	
24HRS/0	.93	24HRS	1.14538	0	.885
1WK/0	.88	1WK	.982307	8	1.07
2WK/0	.88	2WK	.955384		
3WK/0	.88	3WK	.95		
4WK/0	.87	4WK	.934615		
6WK/0	.87	6WK	.94		
24HRS/8	1.33				
1WK/8	1.07				
2WK/8	1.02				
3WK/8	1.01				
4WK/8	.99				
6WK/8	1				

-------- MEANS ORDERED --------

TIMES/CONFIG		TIMES		CONFIG	
6WK/0	.87	4WK	.934615	0	.885
4WK/0	.87	6WK	.94	8	1.07
3WK/0	.88	3WK	.95		
2WK/0	.88	2WK	.955384		
1WK/0	.88	1WK	.982307		
24HRS/0	.93	24HRS	1.14538		
4WK/8	.99				
6WK/8	1				
3WK/8	1.01				

DENTAL MATERIALS SCIENCE OHSU
N FACTOR ANALYSIS OF VARIANCE

06/11/87

2WK/8	1.02
1WK/8	1.07
24HRS/8	1.33

CONTRASTS FOR MEANS OF CELLS:
Q(12 , 144) = 4.714 CONTRAST = 0.06930640

CONTRASTS FOR MEANS OF FACTOR A:
Q(6 , 144) = 4.096 CONTRAST = 0.06004198

CONTRASTS FOR MEANS OF FACTOR B:
Q(2 , 144) = 2.8 CONTRAST = 0.04116630

APPENDIX 4 - FORCE vs TIME: FORCE MEANS AND STANDARD DEVIATIONS FOR CONVENTIONALLY AND FIGURE 8 APPLIED SAMPLES (GAC only).