A METHOD FOR EVALUATING MULTIPLE STRAND ORTHODONTIC WIRES

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INTRODUCTION

Controlled movement of teeth is a fundamental procedure for the clinical dentist dealing with the correction of malocclusion. Because of the poor understanding of the biologic process of tooth movement, no data are available which allows us to pre-select forces for given needed tooth movement changes. It seems ironic that more easily obtained pertinent data are lacking as well. Little has been reported concerning the physical properties or clinical behavior of multiple strand wires even though they are widely used in the early stages of multiband orthodontic treatment for "leveling and aligning." Only one study known to us addresses itself to the physical properties of multiple strand wire (Stephens and co-workers[1]), and that report by British investigators compares two grades of Orthoflex* twisted wire against Bundled Arch** (now out of production), the Johnson Twin-Wire Arch, and one of the

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currently used American-made twisted wires, Twistflex.*** Stephens[1] describes the wires tested in terms of their "elastic energy characteristics" which may be a meaningful measurement to an engineer, but not much use to a clinical dentist.

Since the introduction of Twistflex, other twisted wires have been introduced with claims of superior properties and superior handling characteristics (Multi-Strand,**** Swiss Wire II,***** Wildcat Wire,****** and Multiflex Wire.*******).

The purpose of this study is to devise a model for the evaluation of multiple strand wires in an experiment that closely resembles clinical conditions and attempt to gain an insight into the actual forces produced as well as some idea of their elastic limit in clinical terms. The method can then be used to compare the various wires commercially available with each other and with comparable-sized solid wires.

*** Unitek Corporation, Monrovia, California

**** Summit Orthodontic Services, Cuyahoga Falls, Ohio

***** Ormco, Glendora, California

****** GAC International, Farmington, Long Island

******* Ortho International Services, Wilmington, Delaware

REVIEW OF LITERATURE

Johnson[2] in 1934, suggested the use of two small wires instead of one heavier wire in the bracket engagement of an orthodontic appliance. He stated that a small wire was more resilient than a large one and, whereas one small wire would have insufficient power to move the teeth, by using two wires one could double the force exerted but retain the desired resiliency. That concept was the basis for the Johnson Twin-Wire appliance which had inherent disadvantages of its own, but the idea undoubtedly stimulated someone to use a multiple strand, highly resilient wire for leveling and aligning. It is during this early treatment procedure that a highly resilient archwire that will not be "set" easily has its usefulness because the objective is to achieve maximum tooth movement with a minimum of adjustment. Johnson also made mention of measuring with a spring-activated scale some of the forces exerted by wires. He found the forces to range between two and ten ounces with his "twin wire" system, whereas an .018 wire often exerted from two to four pounds.

The Bundled Arch was made commercially available after development by Brandhorst, Fogel and Magill. These arches consisted of either four or seven strands of small diameter wire lying parallel to each other. They were available in four diameters ranging from .016 to .021. Fogel and Magill[3,4] published two articles advocating the use of a multistrand light "bundled arch" as a separate second phase of treatment for leveling and aligning. Their bundled arch consisted of six strands of .008 wire which is a deviation from the commercially available Bundled Arch which carries their names. The manufacturer discontinued Bundled Arch in early 1972, due to low sales volume probably due to the greater convenience of the available twisted multiple strand wires.[5]

Mathews[6] mentioned Bundled Arches in an article in 1961, in which he was suggesting the use of lighter forces. He went on to make the point that, "the operator should not be deluded into thinking that such an archwire exerts less force and is therefore more gentle than, for example, a .016 wire." He did a crude experiment with a typodont setup by loading several different wires to a 1.0 millimeter deflection over an 18 millimeter span and measuring the force with a spring-activated scale. He found the measured force to be 100 grams with a .012 wire, 400 grams with a .016 wire, 475 grams with the Bundled Arch (four .010 wires) and about 1900 grams for a .021 x .025 arch. Again, this author makes a point of the large amount of force applied "between teeth" about which more will be said later.

Eirew[7] in 1969, described a technique that relied heavily on the Twistflex Arch. His description, however, was entirely clinical and he made no mention of the amount of force being produced by these wires. His two rules for selecting the proper size of archwire were 1) the greater the tooth irregularity and rotation to be corrected, the thinner the gauge of archwire used, and 2) the longer the unsupported lengths of arch between banded teeth, the thicker the archwire.

As mentioned before, the only article published to date dealing directly with the physical properties of multiple strand arches is that of Stephens and co-workers.[1] They measured the minimum radius from which there was complete elastic recovery and calculated the energy stored per unit length at this stress (after the method of Stephens and Waters[8]). The resulting elastic energy characteristic has little significance to the clinical dentist, for example, in helping one to determine how much force is being exerted in a given situation or guiding one in deciding which size wire to use. Many articles and textbooks mention and advocate the use of multiple strand wires, but none of them give any indication of the relative merits of any particular brand nor do they discuss the type and amount of force produced with these wires.

Multiple strand wires are generally used in a straight configuration, that is, without loops. The forces produced during "leveling and aligning" are "between teeth" forces. As shown in Fig. 1, one of the teeth is not "level" with the two adjacent teeth. Thus, placing a straight wire in the brackets will produce forces as shown "between teeth." Addition of other "unlevel" or "unaligned" teeth obviously complicates the situation, but the forces produced are still "between teeth."

METHOD AND MATERIALS

For this study, a simple hypothetical clinical situation was selected for testing the "between teeth" forces of the type generated by a multiple strand wire during "leveling and aligning." The situation simulated was an intact, "level and aligned" maxillary arch with a "high cuspid." This case was selected because of its simplicity in that it involved only one tooth out of alignment. A "standard" wood arch form was cut from one-half inch walnut. This form was derived from a commercially available archwire formation card* (Fig. 2). Brackets were welded on a one-half-inch flat strip of .005 inch stainless steel stock. The position of the brackets was determined by using average tooth widths (taken from the fiftieth percentile)[9] and calculating the midpoint of the labial or buccal surface of each tooth on both sides of the midline. Extra wide .022 twin edgewise brackets (Unitek #001-695) were used for the central incisors and first molars. Standard .022 twin edgewise brackets (Unitek #001-697) were used for the lateral incisors,

*Huntington Laboratories, Pasadena, California left cuspid, and first and second premolars. Single rectangular 5° angulation .022 molar tubes (Ormco #597, 598-228) were used for the second molars. No bracket was placed in the test area of the right cuspid. The brackets were carefully welded in position so that a straight piece of .0215 x .028 wire lay passively in all the brackets. The strip was then bent around the walnut arch form and secured with screws and nails. The strip was cut in the test area and the walnut block cut away to allow room for the testing pen (Fig. 3,4).

The testing ~~pen~~ pin (Fig. 5) was made from one-half-inch diameter aluminum stock turned down on a lathe and then drilled out to reduce its mass. A standard .022 twin edgewise bracket was glued to the end of the testing ~~pen~~ pin with epoxy glue. The total weight of the testing ~~pen~~ pin was 11.50 grams (.0253 pounds).

The test arch was positioned in a test jig and secured with a bolt threaded on both ends (Fig. 5). Two steel pins in the test jig were used to accurately position the test arch in the test jig (Fig. 6).

Wires to be tested were placed in the test arch and ligated with

A-1 Ālastik modules (Unitek Corp.). All wires were used straight; that is, no attempt was made to put an arch form in them. This was done for uniformity as well as the belief that most clinicians use them this way. The ends of the wires were left straight in all cases. The testing pen was placed with its bracket bearing on the wire and the apparatus placed in an Instron compression testing instrument (Fig. 7,8) which automatically records a force-deflection curve.

The Instron (Fig. 7) consists of a moveable crosshead for loading or unloading the sample (by deflecting the wire in this case), a load cell for detecting the load or force, and a pen to register the force on a moveable chart. The chart is synchronized to the crosshead; that is, as the crosshead moves down or up loading or unloading the wire, the chart moves down or up at a proportional speed. When the crosshead starts or stops, the chart starts or stops. The resulting chart is a force-deflection curve (Fig. 9).

In this study, the crosshead was set to "load or unload" the wire at a rate of .05 inches per minute. The chart speed was set at 10 inches per minute. Hence values for the vertical axis or deflection in the resulting curve can be calculated from the ratio .05/10. That is, one square on the graph represents .005 inch of deflection. The horizontal axis or force is read directly from the graph in pounds.

Wildcat wire was used for all of the tests done in this study. The wire was put in the test apparatus which was placed in the Instron. The crosshead was manually moved down until first contact with the test pen (as noted by a deflection of the registering pen). The automatic crosshead controls were then set to produce the desired amount of deflection and the machine started. When the given amount of deflection was reached, the crosshead stopped. A period of 10-15 seconds was allowed for the wire to "stabilize" and unloading was begun by raising the crosshead at the same rate until it was out of contact with the test pen.

At first a fresh pre-cut length of wire was placed in the test arch for each trial. However, the wire-changing procedure became very time consuming, so an alternative method was tested. One end of a 14 inch length of wire was placed in the test arch. After a trial, approximately one inch of the wire was "pulled through" the brackets and cut off. This procedure put a fresh section of wire in the test area for each test. For one wire size and one deflection (.0195 wire at 0.50 mm. deflection) ten trials were done with separate wires and ten trials were done with one long piece of wire "pulled through" as just described. The means for the measurements used in the study were t-tested and there was no significant difference at the one percent level between the two methods (see Appendix A).

FINDINGS

Evaluation of the Force-Deflection Curves

The original intention of this study was to measure the elastic force limit and determine the elastic deformation-force ~~ration~~ ratio for each wire size after the method used by Mahler and Goodwin[10] for evaluating solid round wires. With these figures, one could then determine the amount of force produced by a given amount of deflection or vice versa. The figures could also be used to compare various wires.

However, it became readily apparent in the preliminary stages that the model being used did not measure only the physical property of the wire. Friction of the wire with the brackets and the ĀlastiK modules as well as possible slippage of the strands of wire entered into the situation. Indeed, one of the patent holders of Wildcat wire had stated, "It is very difficult to measure physical properties...on multi-stranded wire, so I do not have figures."[11]

It then became necessary to determine what parameters were meaningful within the scope of the hypothetical experimental model. A look at some preliminary curves (for example, Fig. 9) showed a loading portion similar to that expected from other studies.[10,12] However, when greater amounts of deflection were produced, the loading curve became less uniform due to the influence of friction and slippage. There was also a "plateau" at the top of all the curves that signified an almost instant reduction of force upon cessation of loading and beginning of unloading. The unloading curve then proceeded down at a somewhat flatter slope displaced to the right of the loading curve. The greater the amount of deflection, the flatter the unloading curve meaning the faster the reduction of force as well as the greater the amount of "non-recovery" or failure of the wire to return to zero.

Since the hypothetical clinical situation assumed that a wire was displaced into a "high" cuspid bracket and we wanted to measure the force exerted on the tooth, the amount of force necessary to get the wire to the bracket was not important, rather the force <u>after</u> the wire was there and "stabilized" was important. This involved subtracting the "plateau" to get a meaningful value for the force exerted by the wire at a given deflection.

Another value that would have significance would be some measure of the force remaining after the tooth had moved some distance. This value would come from along the unloading curve, since in actual fact it is this force that produces the tooth movement; that is, the wire attempting to return to its unloaded state.

In preliminary trials, the question arose as to the necessity of "Mexicaning back" or bending over the ends of the wire against the second molar tubes. In the clinical case, the wire would already be loaded having been placed in the "high" cuspid bracket and unloading would tend to extend the wire out the ends of the molar tubes, hence Mexicaning back would have no purpose. Therefore, none of the wires were Mexicaned back.

One other measure that was thought to be useful was the amount that the wire failed to return to zero or its "non-recovery." This would give some idea of elastic limit in clinical terms because it is a measure of any possible "set" or plastic deformation of the wire plus the friction in the model.

Measurements

Due to the testing ~~pen~~ pin weight (.0253 pounds), all of the curves began at a force level equal to that weight. The curves were extrapolated to zero with a French curve (Point A, Fig. 9). Since the actual deflection (crosshead travel) did not start from zero, it was not possible to set the machine to deflect the wire the exact desired amount. Hence it was deflected slightly more than the specified amount. Starting from the derived zero then, the exact amount of deflection was measured (Point B, Fig. 9). For example, one square on the chart paper represents .005 inch or .0127 mm. of deflection. Hence 0.50 mm. of deflection equals 39.4 squares. (The example shown in Fig. 9 is for a 0.195 Wildcat wire deflected 0.50 mm.).

The "plateau" at the top was subtracted from this point and the resulting value (Point C, Fig. 9) was taken as the force exerted by the wire at the specified deflection (0.50 mm. in the example).

The value obtained from the unloading curve (force remaining after the wire had unloaded 0.125 mm. in the example) was determined by locating the force on the unloading curve (Point D, Fig. 9) corresponding to the derived deflection force (Point C, Fig. 9). The amount of "unloading" was subtracted (0.125 mm. = 19.7 squares) and the value read (Point E, Fig. 9).

The last measurement was the amount of non-recovery which was simply Point F minus Point A. All values were converted to grams and millimeters.

Testing

Four force-deflection curves were determined for each of the four sizes of Wildcat wire. Deflections of 0.50, 1.00, 1.50, and 2.00 millimeters were done for the two smaller sized wires, .015 and .0175. For the larger sizes, .0195 and .0215, deflections of 0.25, 0.50, 0.75, and 1.00 millimeters were run. Ten trials were done for each deflection.

The "force remaining after unloading a given amount" was determined by subtracting 0.50 mm. where the deflection was 0.75 mm. or more. For the cases where 0.25 and 0.50 mm. deflections were tested, unloading was measured at half that amount, or 0.125 and 0.25 mm. respectively. These figures were arbitrarily chosen on the assumption that 0.50 mm. of tooth movement would be representative of a good clinical tooth movement response in a three to four week appointment period. The practical question to be answered was: Did a "tooth movement" force still exist after a reasonable amount of movement had occurred? Perhaps more importantly: Is a new and possibly heavier wire indicated at the second appointment?

RESULTS

Mean, variance, standard deviation and standard error of the mean were computed from the raw data. Results are shown in Table I.

This data was further reduced to Tables II, III, and IV which show the 95% confidence interval ($t = 2.262$ for nine degrees of freedom) for each of the parameters studied.

Regression lines were calculated for deflection versus force (Table II) and deflection versus non-recovery (Table IV) for each wire size. A regression line was also calculated for wire size versus unloading (Table III) for the two situations for which data was available for all four wire sizes. The regression formulae and standard errors of the slope are shown in Table V and the lines plotted on Graphs I, II, III, and IV.

Interpretation of Results

Parameter of deflection versus force: As shown in Graph I, force is directly related to deflection and wire size. The separation of the lines with their "standard slope error envelopes"* indicates that the

*slope $\pm$ one standard error of the slope lines are significantly different. These results are in keeping with the expected physical properties of wire of the same composition. It can be seen from Table II that the forces ranged from 109.4 ± 5.93 grams to 917.5 ± 34.79 grams for the deflections and wire sizes studied.

Parameter of unloading force: Unloading force would be expected to be directly related to initial deflection and wire size. Regression analysis could not be performed as it was for deflection versus force because the unloading amount was variable. However, a look at the mean force values in Table III shows the expected trend and the fact that none of the confidence intervals "overlap" may be taken as an indication that the differences are real. For two unloading conditions (0.50 mm. deflection unloaded 0.25 mm. and 1.00 mm. deflection unloaded 0.50 mm.), values were measured for all four wire sizes, so regression lines were calculated for wire size versus unloading force for those conditions. A plot of the mean values (Table III) showed the relationship to be logarithmic rather than linear, so a log transformation was done on the y values (force). Graph IV shows that unloading force is directly related to wire size at least for the two conditions tested.

Table III shows that the amount of force remaining after "unloading" ranged from 41.04 $\pm$ 2.53 grams to 280.1 $\pm$ 15.77 grams for the deflections and wire sizes studied.

Parameter of non-recovery: It would be expected that non-recovery be directly related to the amount of deflection and inversely related to wire size.

Plotting the mean values in Table IV of deflection versus non-recovery produced two groups, the .0195 and .0215 in a linear relationship and the .015 and .0175 in ~~a logarithmic~~ an exponential relationship. Hence, two separate pairs of regression lines were calculated.

A log transformation was done on the y values (non-recovery) of the

The resulting Graph III shows a direct relationship between deflection and the log of non-recovery but the lines are reversed up to 1.6 mm. deflection from what would be expected. That is, the .0175 wire demonstrates more non-recovery than the smaller .015 up to 1.6 mm. deflection. Slope error envelopes do not overlap indicating that the lines are different. This was checked by calculating a t statistic for the difference of the slopes from the formula:

$$t = \frac{S_1 - S_2}{\sqrt{\frac{E_{s_1}^2 + E_{s_1}^2}{2}}}$$

where S_1 = slope for .015

S_2 = slope for .0175

E_{s_1} = standard error of slope for .015

E_{s_2} = standard error of slope for .0175

The value obtained was t = 3.294 which is significant at the 95% level of confidence for 18 degrees of freedom.

The regression lines calculated for the larger sized wires are plotted in Graph II and show the same expected direct relationship between deflection and non-recovery. However, their error envelopes (noted by red brackets) overlap for the entire length of the lines indicating no significant difference between wire sizes. The ~~logarithmic~~ * regression lines for the smaller sized wires are also plotted on Graph II to show their relationship to the larger wires. The "band" of slope error is marked in blue and it can be seen that only the .0175 is different from the other three by a very small amount.

Although figures are available for all four wire sizes for several deflections (0.50 mm. and 1.00 mm.), it was not possible to generate a wire size versus non-recovery regression for those deflections because the grouping of the wire sizes for practical purposes reduces the number of sizes to two and at least three points are necessary to produce a valid regression line.

Table IV shows that non-recovery ranged from .017 $\pm$ .0027 mm. to .630 $\pm$ .1950 mm. for the deflections and wire sizes studied.

DISCUSSION

Sources of error in this study would include machine error and operator errors. The Instron[13] is capable of measuring to ± 0.25% of full load scale. Most of the measurements were done with a one pound full load scale which would be accurate within ± 1.6 grams. The two and five pound full load scales were also used which would be accurate to within ± 2.7 grams and ± 6.1 grams respectively. The range of values reported that were determined with the one pound full load scale was from 41.04 to 424.5 grams. The ± 1.6 gram error noted varies from ± 3.9 to 0.38% of the forces being measured. For the two pound scale the range was 509.0 to 819.9 grams or ± 0.53 - 0.32%. The five pound scale was used for only one measurement, 917.5 grams, the error for which would be ± 0.66%. These amounts of error are considered negligible.

Operator error could potentially be the greatest source of error. Manipulation of the apparatus and measuring errors could be considered. Measuring errors would include extrapolating the curves, estimating and reading values on the chart, and subtracting the "plateau" with bow dividers. Manipulation errors would include calibrating and running the machine and positioning of the wires and the test pen.

Since all of the ~~factors mentioned are randomly distributed, the assumption can be made that errors made would tend to cancel each other out.~~ Also, since all the work was performed by one operator, any consistent error introduced would be the same for all the trials.

Any model system patterned after a clinical situation must eventually be related back to the clinic. How closely the model fits the clinical situation helps determine how clinically applicable the result will be. This system was a static in vitro model of a dynamic in vivo situation. Every effort was taken, however, to duplicate the clinical situation except the possible use of a saliva substitute to act as a lubricant and the incorporation of some kind of elastic component to the medium carrying the brackets to simulate the cushioning effect of the periodontal ligament. Muscle forces would undoubtedly affect the clinical situation too, but the amount and direction of involvement is unknown and would be so variable as to be impossible to duplicate in a laboratory model.

A lubricant was considered for this study, but the idea discarded for the sake of convenience since this was a pilot study and more basic questions needed to be answered. All things considered, for the situation simulated, it is felt that the force values recorded can be considered "ball park" estimates of actual forces that would be applied in the mouth.

For the everyday use of multiple strand wire in leveling and aligning, there are many teeth involved. Rotations, buccal-lingual, mesial-distal, and intrusive-extrusive movements may be necessary in varying combinations. Force systems "between teeth" become very complicated when all these factors are considered. Burstone and Koenig[14] in a recent article discussed the ramifications of the possible forces produced between just two teeth. When more teeth are considered, the possibilities become astronomical especially when it is realized that the force systems change continually as teeth move. The effect that multiple malposed teeth would have on the forces produced by these wires raises a question that might stimulate an improved model system.

The amount of force necessary or desireable to move teeth is questionable. Several studies have been made of the amount of force required for specific tooth movements. Storey and Smith[15] have reported that the optimum range of force for retraction of the lower canine tooth is between 150-200 grams. Burstone and Groves[16] retracted anterior teeth by simple tipping and stated that optimum rates of tooth movement were observed when 50-75 grams of force were applied. Reitan[17] has stated that the maximum force needed during any stage of a continuous bodily movement of the canine is approximately 250 grams. Reitan[18] has also stated that the force exerted to extrude teeth must not exceed 25-30 grams. Buck and co-workers[19,20] used 70-75 grams $\pm$ 10 percent to tip maxillary bicuspids in human histologic tooth movement studies. Hixon and co-workers[21] used a range from 64-1515 grams for cuspid retraction in a cephalometric study.

The range of forces mentioned by these various studies is very broad. Leveling and aligning with a multiple strand wire must certainly take a minimum of force per tooth since most of the movement is simple tipping or extrusion. It is interesting to note that all the initial forces measured in this study were greater than 100 grams (even for a .015 wire deflected only 0.50 mm.) and the upper extreme was a force of over two pounds generated by deflecting a .0215 wire only one millimeter.

For the usual clinical situation involving a number of malposed teeth, it would be logical to assume that these forces would be reduced. However, the experimental model simulated a condition that does occur clinically and it is possible that excessive forces are being applied to "high" cuspids. Other clinical conditions, such as correcting individual teeth during finishing with heavy arch wires (for example "stepping" a tooth up or down) are undoubtedly generating some extremely large "between teeth" forces. ~~The necessity for corrections of this type is generally improper band placement at start.~~ Whether or not large forces are detrimental is not known scientifically, but intuitively and clinically most operators would like to avoid them.

Unloading forces as defined and measured in the study ranged from 41-280 grams. For the "high" cuspid situation any of these forces would probably be sufficient to effectively extrude the tooth. Again, the usual case with several or many malposed teeth would produce different force values and would have to be examined with a modified testing model. The relative amount of force remaining after unloading for the conditions set up for this study averaged 30.7% with a range from 18.2 - 41.9% (column 12, Table I).

The parameter termed "non-recovery" produced rather equivocal results. The "grouping" of the smaller and larger sized wires can be attributed to the deflection conditions selected for the wire sizes. Frictional factors became more apparent as the amount of deflection was increased. No macroscopic plastic deformation was noted for any of the wires tested, although at deflections of 1.50 mm. and 2.00 mm. the wires were retained in a "deformed" position by friction after the testing ~~pen~~ pin was disengaged. Upon removal of the ĀlastiK ligatures, the wires would return to their straight configuration.

It is felt that if tests had been done on the smaller wires at the smaller deflections, the deflection versus non-recovery graph would have been linear up to about 1.0 mm. and then proceed on as a curvilinear function. Likewise, if the larger sized wires had been tested at the larger deflections, their curves would have become more curvilinear beyond some point, perhaps 1.0 mm., due to friction.

The apparent difference between .0175 and the other three sizes for the deflection range 0.6 - 1.0 mm. is not readily explained. It could have been due to batch variability or some difference in the manufacturing procedure such as looser twisting or being under less tension during twisting although one would expect variations in the other parameters studied as well if that were the answer.

An idea of the reliability of the model system can be gained by looking at the standard deviation as a percentage of the mean for the parameters studied. In a physical system such as this, the percentage would be expected to be about 10%.

For the deflection force the standard deviation averaged 6.0% of the mean values with a range from 2.9 - 11.0%. For the unloading force the average was 7.4% with a range from 3.0 - 13.7%. Non-recovery averaged 28.3% with a range from 11.9 - 50.0%.

Deflection force and unloading force appear to be reliable measures in the model system. Non-recovery of the other hand appears not to be reliable. This plus the equivocal results noted previously make this parameter suspect as a useful measurement for this model system. Further testing and evaluation are indicated, however, before dropping non-recovery as a useful parameter.

A comment should be made concerning the "plateau" noted at the top of all of the force-deflection curves (see Fig. 9). This "plateau" signified an almost instant reduction of force upon cessation of loading and beginning of unloading. The fact that a "plateau" was also present on the force-deflection curves generated for some solid round wires in a preliminary test would lead one to believe that it was due to something other than the multiple strand configuration of the wires being studied. It was most likely due to a relative motion of the wire in the brackets, a friction and slippage of the wire "stabilizing" itself when motion was stopped or started. It is possible that there was some slippage or reorganizing of the individual strands of the wires being tested, but that could not be determined in this study.

Questions that could possibly be answered using the proposed model system or a variation of the system would be batch uniformity of wires as received from the manufacturer; possible differences, advantages or disadvantages between "brands" of wire; force comparison with or simply force values for solid wires (both round and rectangular) for "between teeth" forces, and generation of desireable specifications for wires from the clinicians point of view.

Possible variations in the model system would include altering the "span between teeth" (for this study the span was 12.5 mm.), including other unlevel or malposed teeth, altering the model so the wire is straight in the test area (for this study the wire was on a slight curve as seen in Fig. 2), using a lubricant to simulate saliva, and incorporating an elastic component to the medium carrying the brackets to simulate the cushioning effect of the periodontal membrane.

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SUMMARY

A model system has been presented for evaluating multiple strand orthodontic wires. The model is designed to measure forces "between teeth" of the type generated by placing a straight wire into brackets that are no aligned. Three measurement parameters were suggested, namely "deflection force," "unloading force," and "wire non-recovery."

One brand of multiple stand wire, Wildcat Wire, was tested and values tabulated for the parameters noted. The results were presented in Tables and Graphs to show the relationships between deflection, force, wire size, and non-recovery.

CONCLUSIONS

1. Of the three measurement parameters suggested, non-recovery was not considered very reliable.

2. Values measured for deflection versus force were quite high, ranging from just over 100 grams to just over 900 grams.

3. The force remaining after unloading about 0.50 mm. averaged about 30% of the initial force.

4. The model system presented appears to be a useful method for testing "between teeth" forces.

BIBLIOGRAPHY

1. Stephens, C.D., Houston, W.J.B., Waters, N.E. Multiple strand arches. Dent. Practit. 22:147, 1971.

2. Johnson, J.E. A new orthodontic mechanism: The twin wire alignment appliance. Int. J. Ortho. and Dent. for Children 20:946, 1934.

3. Fogel, M.S., Magill, J.M. The combination technique. Am. J. Orthod. 49:801, 1963.

4. Fogel, M.S., Magill, J.M. A fundamental re-appraisal of popular techniques with a collective approach toward appliance therapy. Am. J. Orthod. 55:705, 1969.

5. Brunson, B. Personal communication. Rocky Mountain Dental Products, Denver, Colorado, 1974.

6. Mathews, J.R. Clinical management and supportive rationale in early orthodontic therapy. Angle Orthod. 31:35, 1961.

7. Eirew, H.L. The twistflex arch. The Orthodontist 1:14, 1969.

8. Stephens, C.D. Waters, N.E. The elastic properties of unformed orthodontic wires. Dent. Practit. 22:2, 1971.

9. Horowitz, S.L. and Hixon, E.H. The nature of orthodontic diagnosis. St. Louis, C.V. Mosby, 1966, p.156.

10. Mahler, D.B., Goodwin, L. An evaluation of small diameter orthodontic wires. Am. J. Orthod. 37:13, 1967.

11. McCauley, D.S. Personal communication. GAC International, Inc., Farmington, New York, 1973.

12. Bass, T.P., Stephens, C.D. Some experiments with orthodontic springs. Dent. Practit. 21:21, 1970.

13. Instron Catalog 1-1, Canton, Mass., Instron Engineering Corp. p. 2.

14. Burstone, C.J. and Koenig, H.A. Force systems from an ideal arch. Am. J. Orthod. 65:270, 1974.

15. Storey, E. and Smith, R. Force in orthodontics and its relation to tooth movement. Aust. D.J. 56:18, 1952.

16. Burstone, C.J. and Groves, M.H., Jr. Threshold and optimum force values for maxillary anterior tooth movement. Abstract. J.Dent. Res. 39:695, 1960.

17. Reitan, K. Some factors determining the evaluation of forces in orthodontics. Am. J. Orthod. 43:32, 1957.

18. Reitan, K. Clinical and histologic observations on tooth movement during and after orthodontic treatment. Am. J. Orthod. 53:721, 1967.

19. Buck, D.L., Church, D.H. A histologic study of human tooth movement. Am. J. Orthod. 62:507, 1972.

20. Buck, D.L., Griffith, D.A., and Mills, M.J. Histologic evidence for lipids during human tooth movement. Am. J. Orthod. 64:619, 1973.

21. Hixon, E.H., Atikian, H., Callow, G.E., McDonald, H.W., and Tacy, R.J. Optimum force, differential force, and anchorage. Am. J. Orthod. 55:437, 1969.

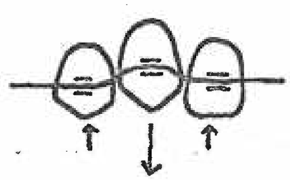


Fig. 1 Arrows indicate the "between teeth" type of forces produced by placing a straight wire in "unlevel" brackets

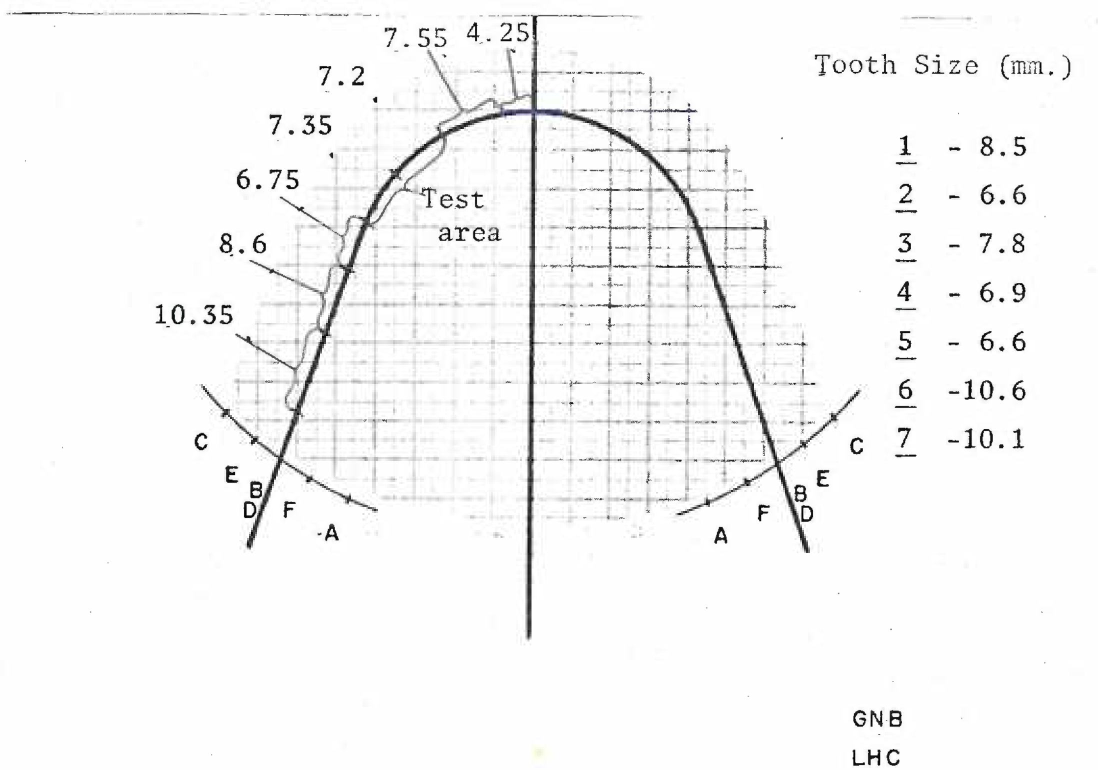


Fig. 2 Archwire formation card used to determine the "standard" arch. Tooth sizes shown are from the fiftieth percentile.

Fig. 3 Completed "test arch" with wire in place

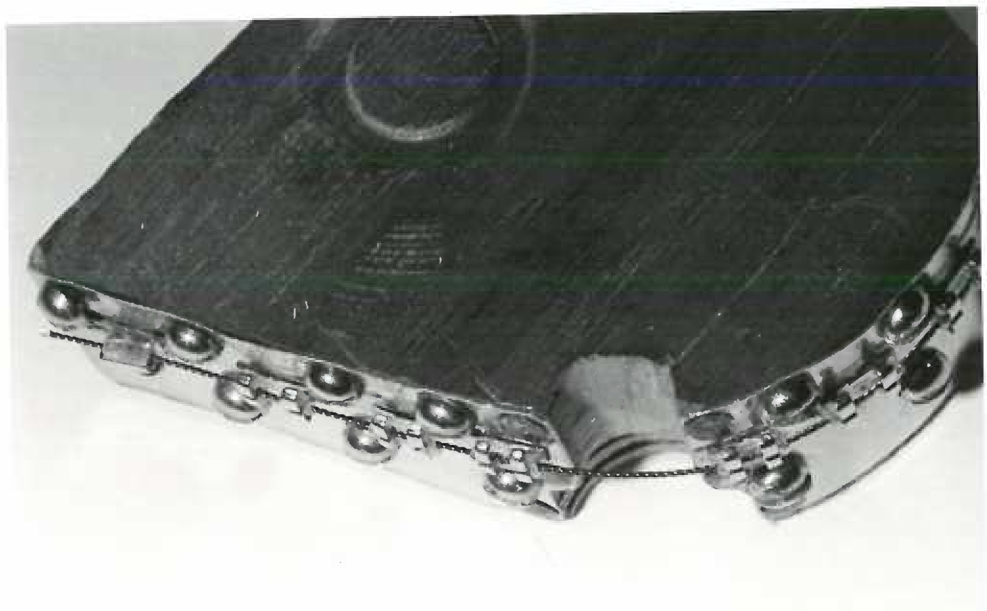
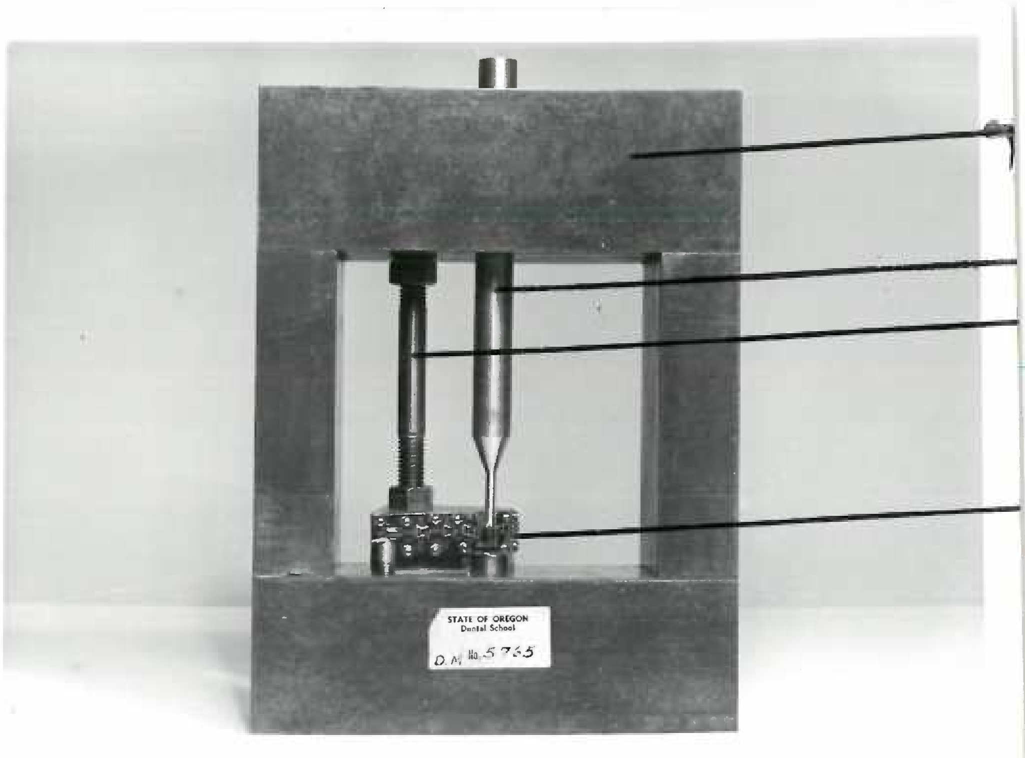


Fig. 4 Closer view of "test area"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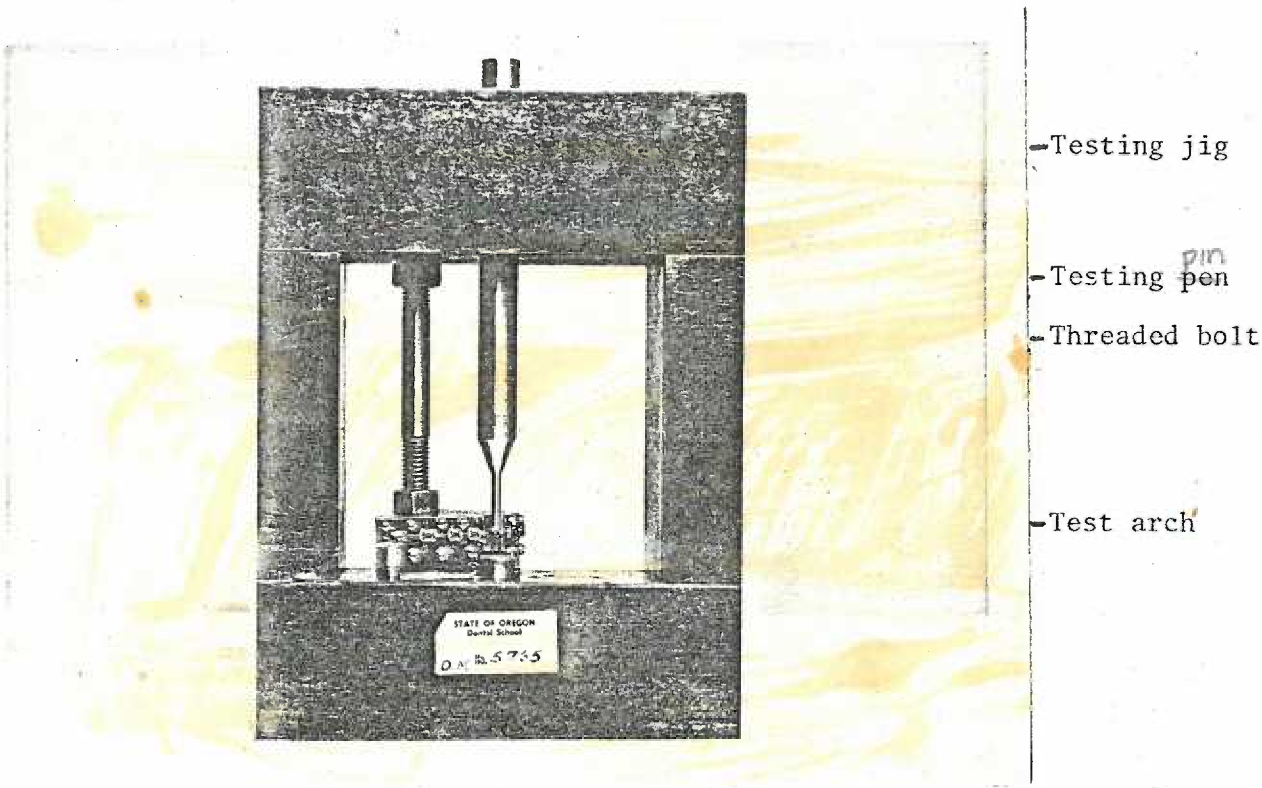
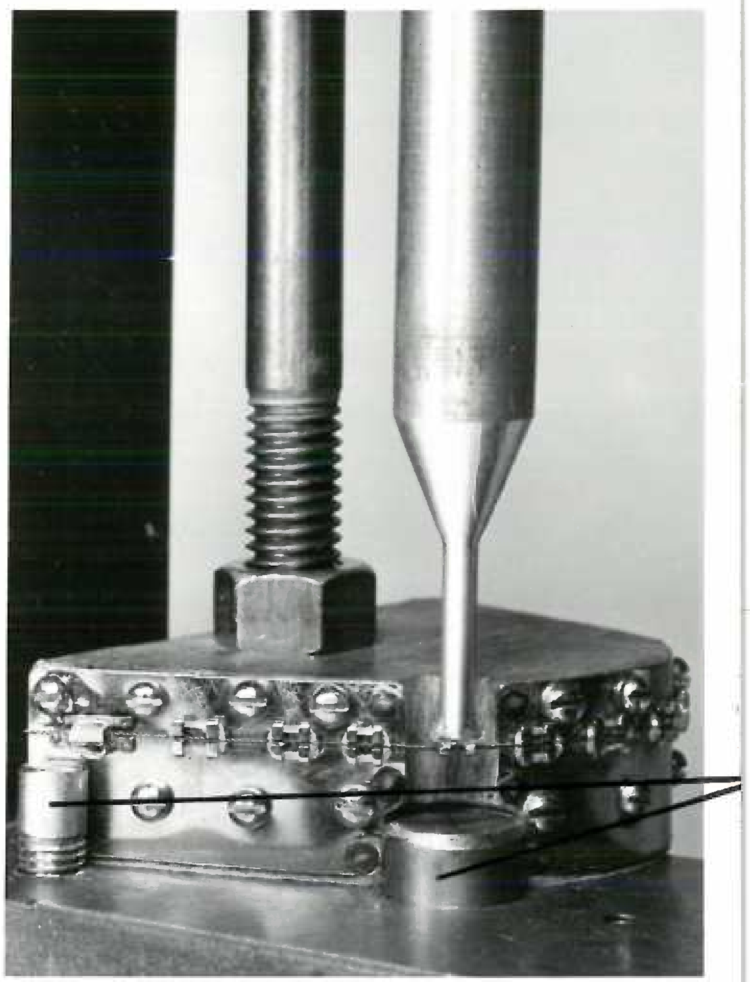
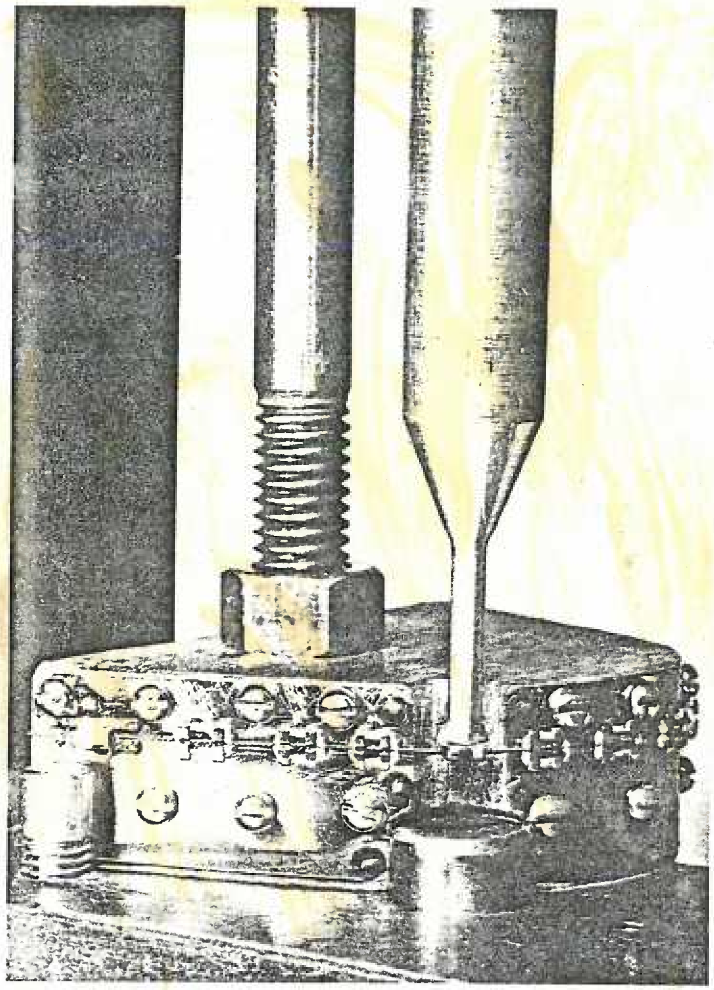


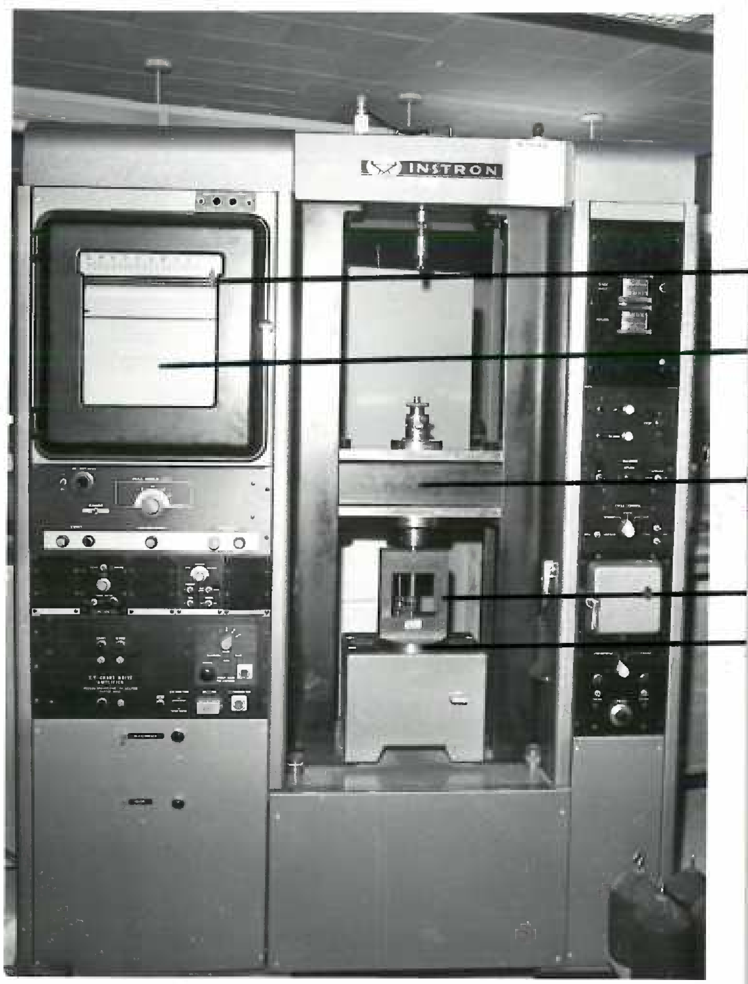
Fig. 5 Testing jig with test arch and testing pen in position





Steel positioning pins

Fig. 6 Closer view of "test area" with testing pen in position on the wire


INSTRON

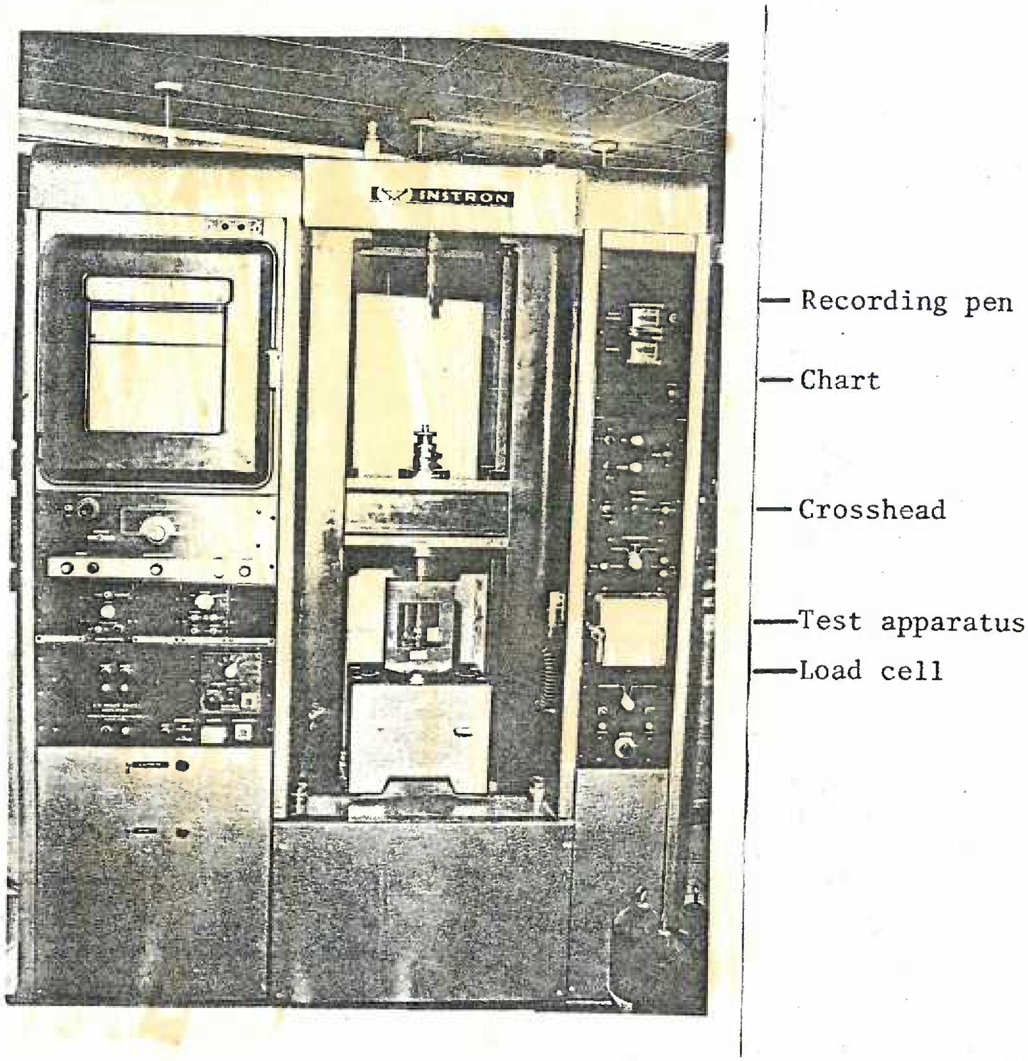


Fig. 7 Instron compression testing instrument with test apparatus in place

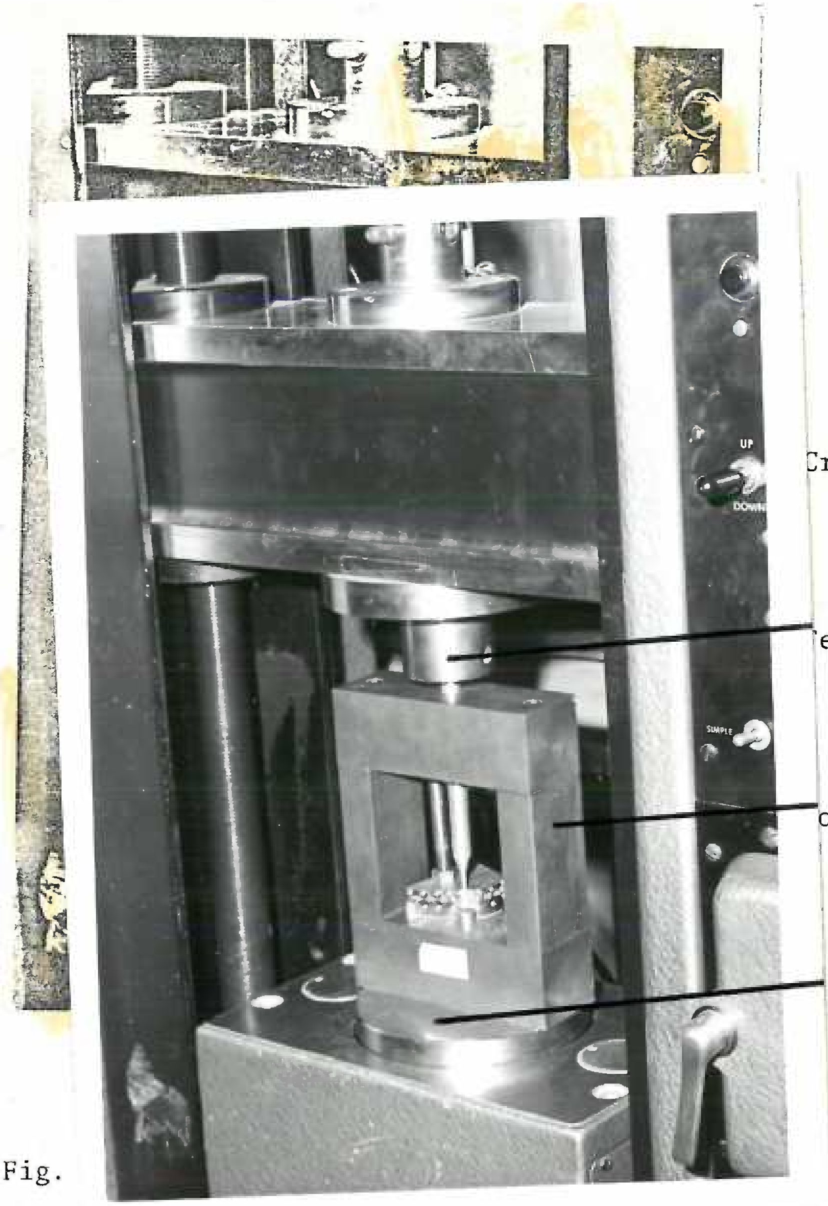


Fig. in the Instron

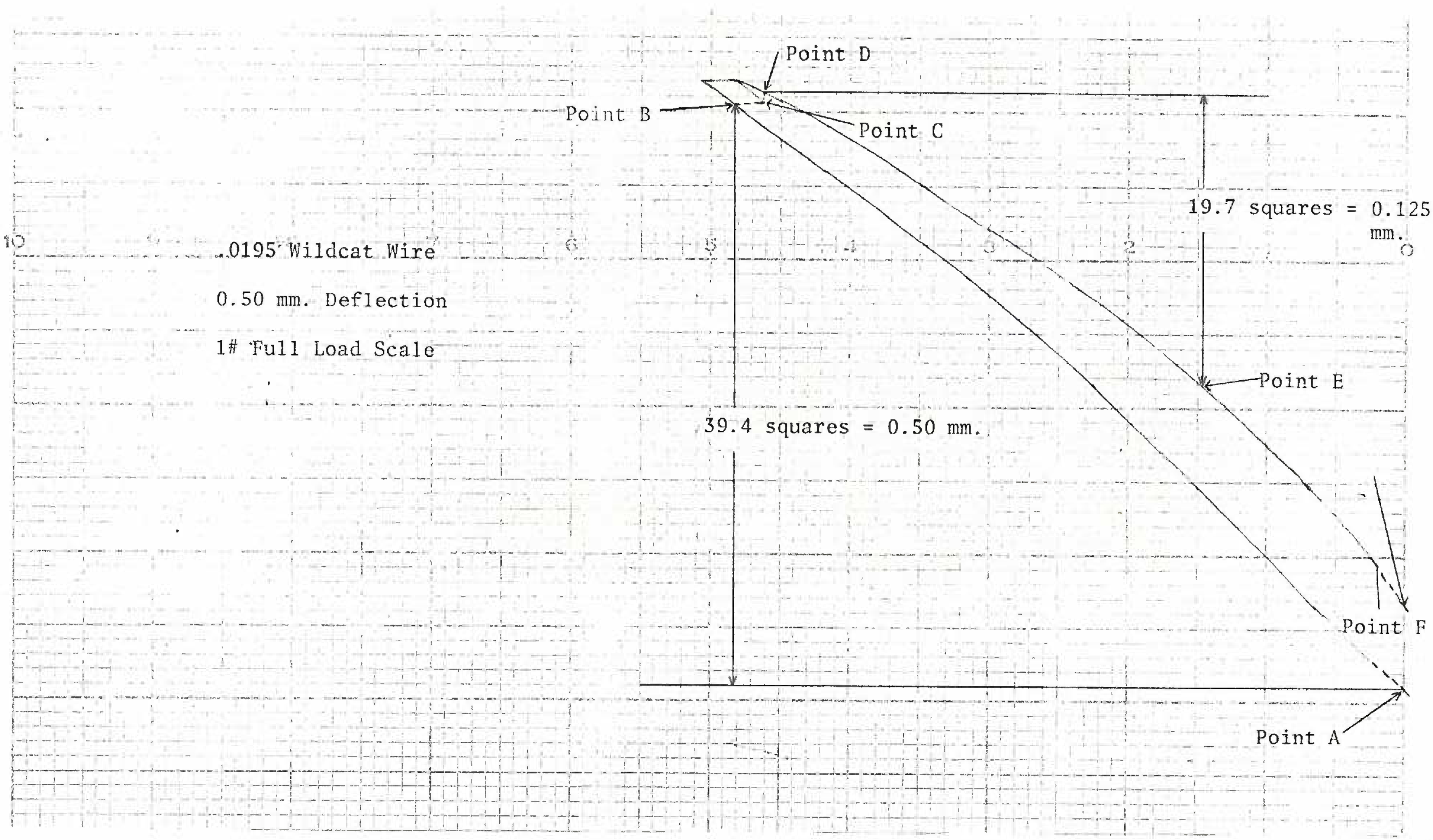


Fig. 9 A force-deflection curve re-drawn from an Instron recording on Instron chart paper.

Table I Data taken from force-deflection curves

WIRE SIZE	DEFLECTION (mm.)	FORCE AT GIVEN DEFLECTION (Gms.) Point C, Fig. 9				UNLOADING AMOUNT (mm.)	FORCE REMAINING AFTER UNLOADING GIVEN AMOUNT (Gms.) Point E, Fig. 9					NON-RECOVERY (mm.) Point F-Point A, Fig. 9			
		$\bar{x}$	S^2	S	SEM		$\bar{x}$	S^2	S	SEM	%	$\bar{x}$	S^2	S	SEM
.015	0.50	109.4	68.47	8.27	2.62	0.25	41.04	12.47	3.53	1.12	37.5	.024	.00014	.012	.0037
	1.00	254.5	498.5	22.32	7.06	0.50	56.84	39.98	6.32	2.00	22.3	.11	.0015	.038	.012
	1.50	327.3	464.2	21.55	6.81	0.50	93.26	163.5	12.78	4.04	28.5	.25	.0034	.059	.019
	2.00	509.0	3159.0	56.21	17.77	0.50	127.2	135.8	11.65	3.68	24.9	.63	.073	.27	.086
.0175	0.50	160.9	131.2	11.46	3.62	0.25	59.43	26.22	5.12	1.62	36.9	.038	.00014	.012	.0037
	1.00	333.1	355.1	18.84	5.96	0.50	89.08	41.38	6.43	2.03	26.7	.12	.00072	.027	.0085
	1.50	598.1	819.4	28.62	9.05	0.50	187.6	47.25	6.87	2.17	31.4	.21	.00061	.025	.0078
	2.00	819.9	3826.2	61.86	19.56	0.50	246.1	255.3	15.98	5.05	30.0	.62	.015	.12	.039
.0195	0.25	105.4	42.97	6.56	2.07	0.125	41.63	9.07	3.01	0.95	39.5	.017	.000014	.0038	.0012
	0.50	237.6	59.47	7.71	2.44	0.250	84.63	19.58	4.42	1.40	35.6	.032	.00015	.012	.0039
	0.75	424.5	430.9	20.76	6.56	0.50	77.41	22.29	4.72	1.49	18.2	.042	.00019	.014	.0043
	1.00	591.6	701.1	26.48	8.37	0.50	172.9	87.27	9.34	2.95	29.2	.068	.000099	.0100	.0032
.0215	0.25	157.4	98.40	9.92	3.14	0.125	65.97	22.52	4.75	1.50	41.9	.019	.000035	.0059	.0019
	0.50	360.3	287.50	16.96	5.36	0.250	134.6	93.65	9.68	3.06	37.4	.024	.000055	.0074	.0024
	0.75	607.9	320.3	17.90	5.66	0.50	124.4	13.77	3.71	1.17	20.5	.030	.000093	.0097	.0030
	1.00	917.5	2365.9	48.64	15.38	0.50	280.1	485.4	22.03	6.97	30.5	.061	.000080	.0089	.0028

$\bar{x}$ = mean = $\frac{\Sigma x}{n_x}$

S^2 = variance = $\frac{\Sigma (x_1-\bar{x})^2}{n_x-1}$

S = standard deviation = $\sqrt{S^2}$

SEM = standard error of mean = $\frac{S}{\sqrt{n}}$

Table II Deflection versus force (in grams ± 2.262 SEM) for the four wire sizes

Deflection (mm.)	WIRE SIZE .015	.0175	.0195	.0215
0.25			105.4 ± 4.68	157.4 ± 7.10
0.50	109.4 ± 5.93	160.9 ± 8.19	237.6 ± 5.52	260.3 ±12.12
0.75			424.5 ± 14.84	607.9 ±12.80
1.00	254.5 ± 15.97	333.1 ± 13.48	591.6 ± 18.93	917.5 ±34.79
1.50	327.3 ± 15.40	598.1 ± 20.47		
2.00	509.0 ± 40.20	819.9 ± 44.24		

Table III Unloading force (in grams ± 2.262 SEM) for the four wire sizes

Unloading Amount (mm.)	WIRE SIZE .015	.0175	.0195	.0215
0.25-0.125			41.63 ± 2.15	65.97 ± 3.39
0.50-0.250	41.04 ± 2.53	59.43 ± 3.66	84.63 ± 3.17	134.60 ± 6.92
0.75-0.500			77.41 ± 3.37	124.40 ± 2.65
1.00-0.500	56.84 ± 4.52	89.08 ± 4.59	172.90 ± .67	280.10 ± 15.77
1.50-0.500	93.26 ± 9.14	187.60 ± 4.91		
2.00-0.500	127.20 ± 8.32	246.10 ± 11.42		

Table IV Deflection versus non-recovery (in millimeters ± 2.262 SEM) for the four wire sizes

Deflection (mm.)	WIRE SIZE .015	.0175	.0195	.0215
0.25			.017 ± .0027	.019 ± .0043
0.50	.024 ± .0084	.038 ± .0084	.032 ± .0088	.024 ± .0054
0.75			.042 ± .0176	.030 ± .0068
1.00	.110 ± .0270	.120 ± .0192	.068 ± .0072	.061 ± .0063
1.50	.250 ± .0430	.210 ± .0176		
2.00	.630 ± .1950	.620 ± .0882		

Table V

Regression data calculated from Tables II, III, and IV

Deflection versus force (from Table II)

Wire Size	Regression Formula	Standard Error of the Slope
.015	y = 254.33000 x + - 17.88000	10.99883
.0175	y = 448.42400 x + - 82.555	11.38832
.0195	y = 658.14680 x + - 71.56400	11.94494
.0215	y = 1011.19600 x + - 121.21000	21.77176

Deflection versus non-recovery (from Table IV)

Wire Size	Regression Formula	Standard Error of the Slope
.015	y = .94743 x - 2.06876	.06170
.0175	y = .78449 x - 1.79815	.03296
.0195	y = .06576 x - .00130	.00622
.0215	y = .05336 x + .00040	.00592

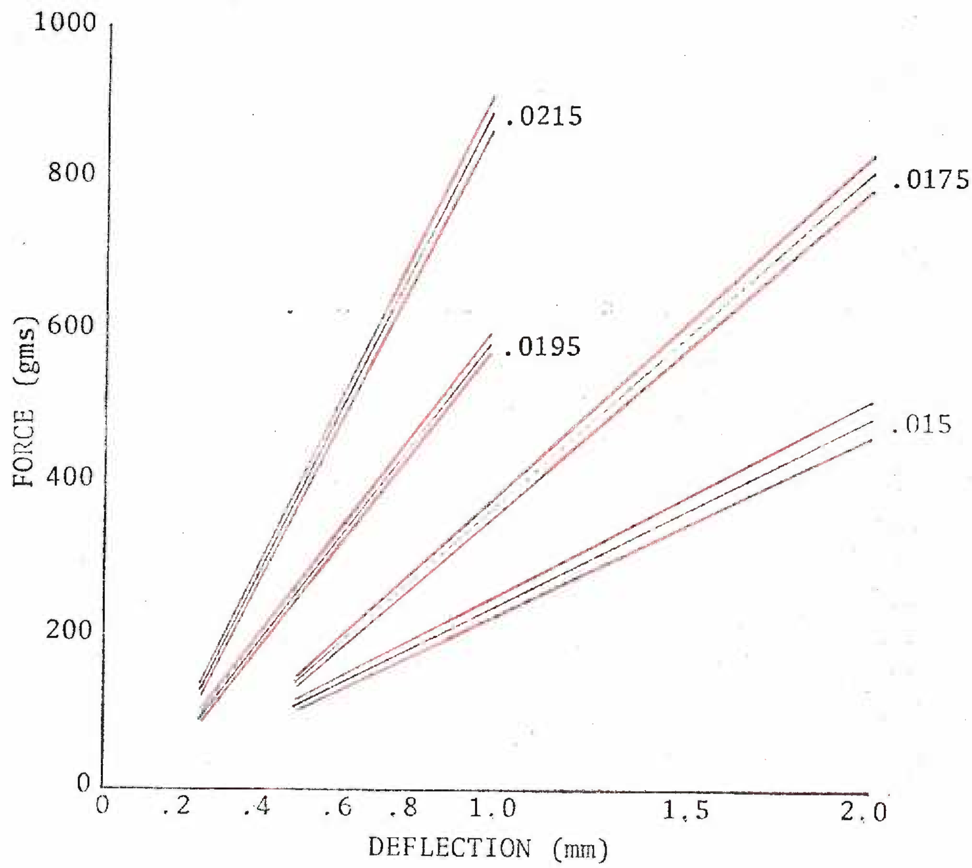
Log transformation done on y values for .015 and .0175 (see text)

Wire size versus unloading (from Table III)

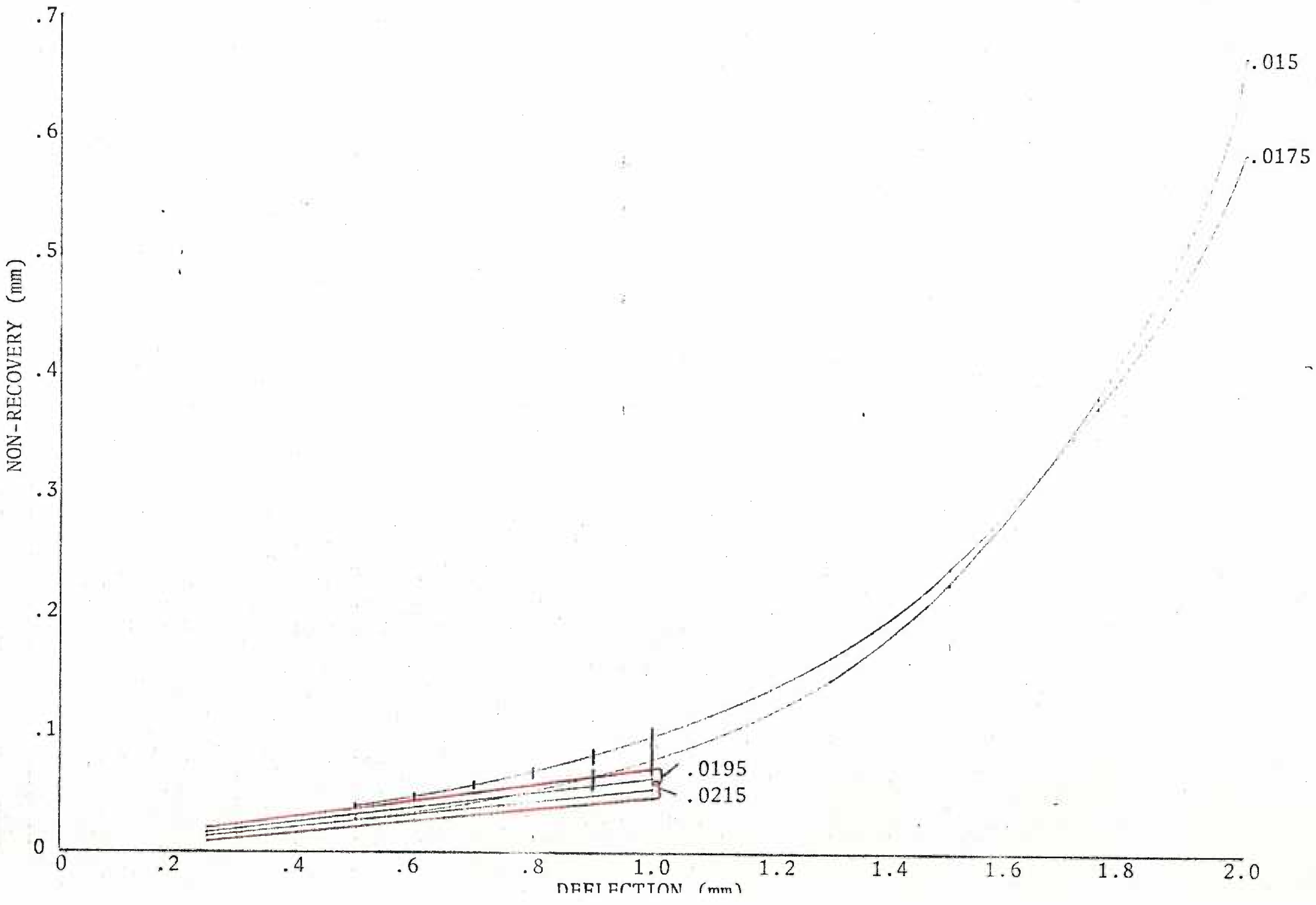
Unloading Amount	Regression Formula	Standard Error of the Slope
0.50-0.25 mm.	y = 78.78016 x + .41220	2.48473
1.00-0.50 mm.	y = 109.63169 x + .08154	3.11583

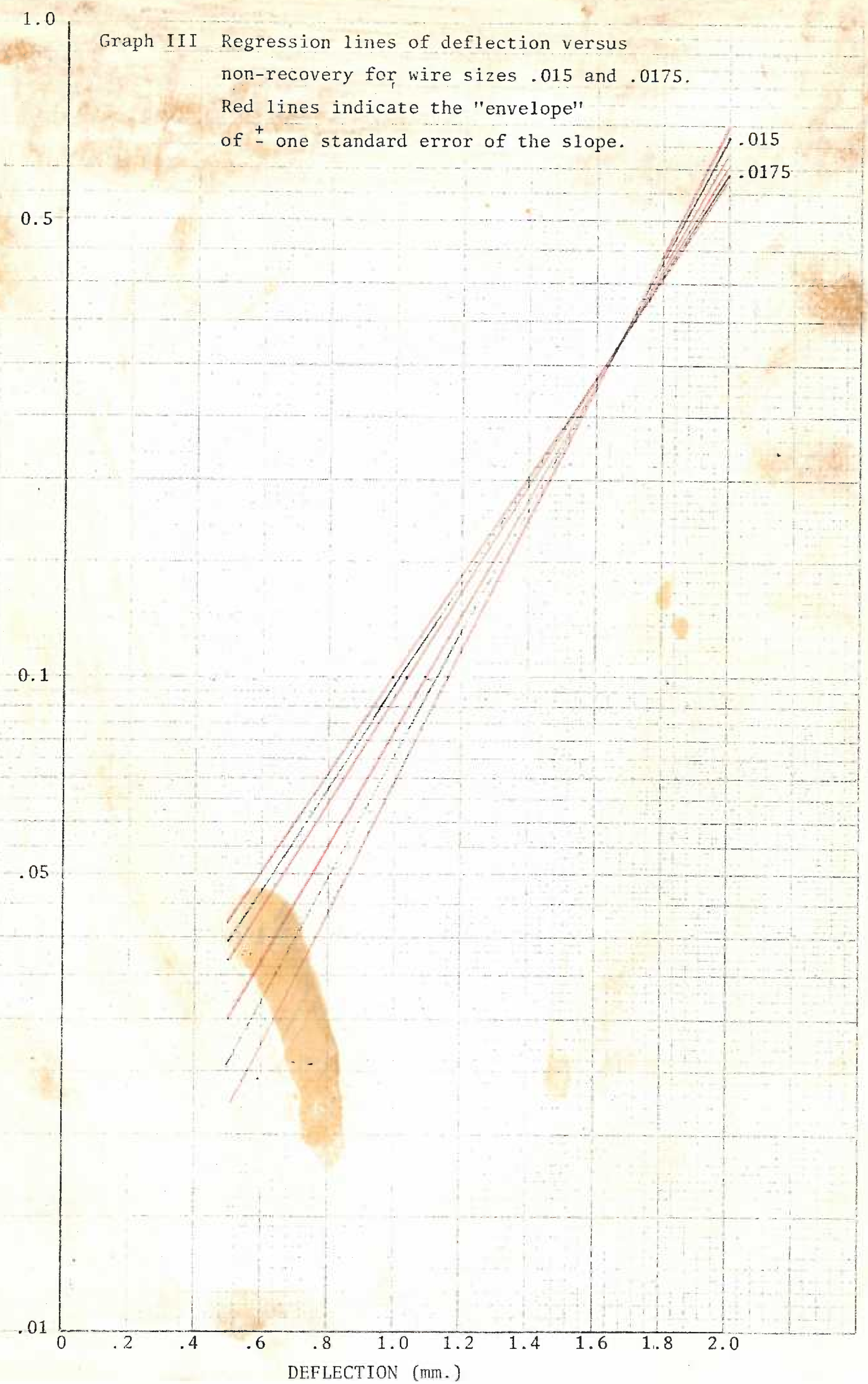
Log transformation done on y values (see text)

Graph I Regression lines of deflection versus force for four wire sizes. Red lines indicate the "envelope" of ± one standard error of the slope.

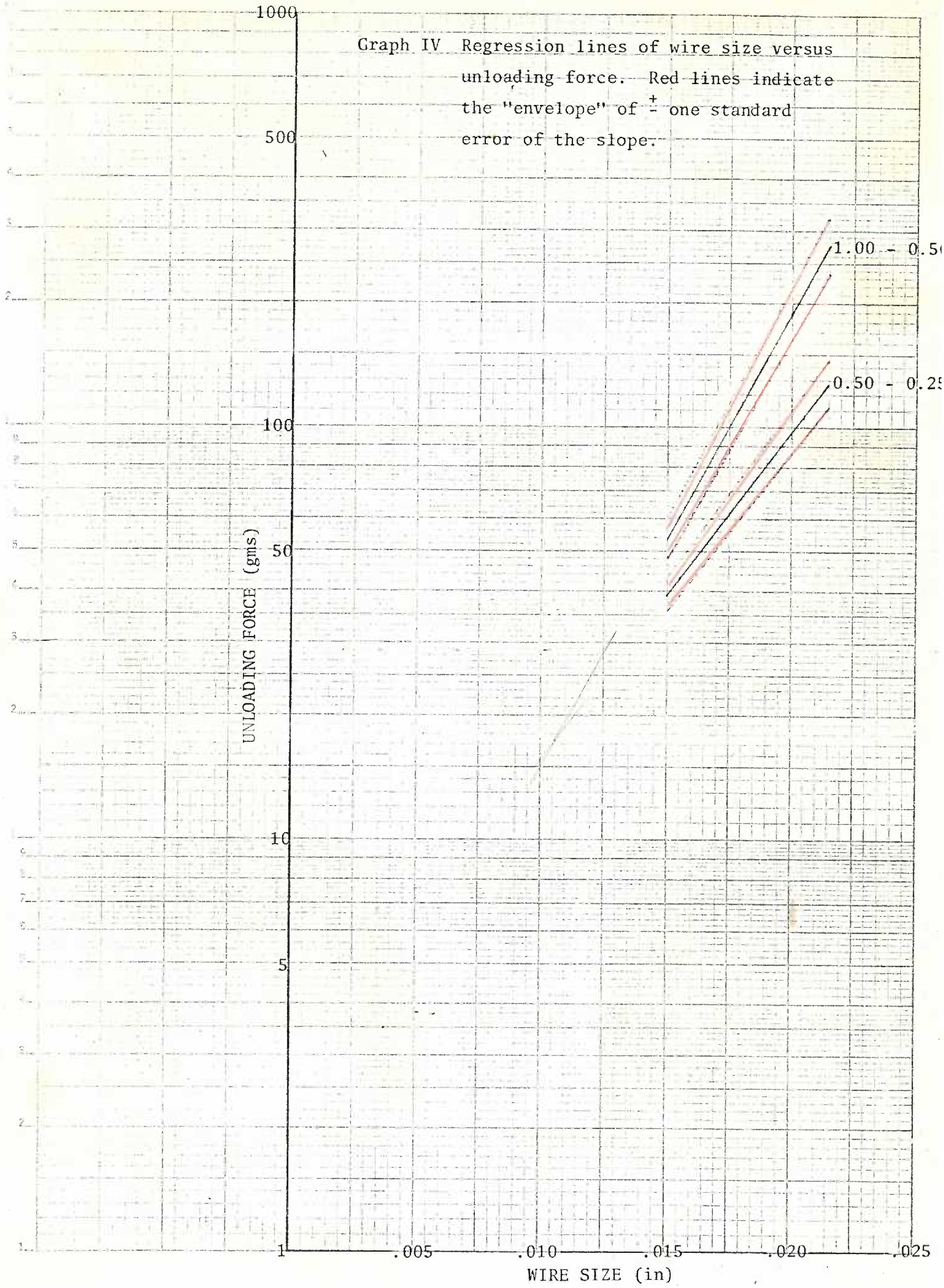


Graph II Regression lines of deflection versus non-recovery for four wire sizes.





Graph III Regression lines of deflection versus non-recovery for wire sizes .015 and .0175. Red lines indicate the "envelope" of ± one standard error of the slope.
.015
.0175
1.0
0.5
0.1
.05
.01
0
.2
.4
.6
.8
1.0
1.2
1.4
1.6
1.8
2.0
DEFLECTION (mm.)



Graph IV Regression lines of wire size versus unloading force. Red lines indicate the "envelope" of ± one standard error of the slope.

APPENDIX A

Test method A: separate .0195 Wildcat wires used for each of 10 trials

Test method B: one 14-inch piece of .019 Wildcat wire "pulled through" and used for 10 trials

		Method A	Method B	t value*
FORCE AT 0.50 mm.	$\bar{x}$	245.3	238.8	0.8943
DEFLECTION (Gms.)	S^2	331.71	207.24	
	S	18.2	14.4	
	SEM	5.76	4.55	
FORCE REMAINING	$\bar{x}$	84.04	83.67	0.1140
AFTER 0.25 mm.	S^2	43.26	58.74	
UNLOADING (Gms.)	S	6.58	7.66	
	SEM	2.08	2.42	
	$\bar{x}$	.059	.046	2.10
NON-RECOVERY (mm.)	S^2	.00017	.00021	
	S	.013	.014	
	SEM	.0042	.0046	

*18 degrees of freedom

$\bar{x}$ = mean

S^2 = variance

S = standard deviation

SEM = standard error of mean