THE UNIVERSITY OF MANITOBA

THE EFFECTIVENESS OF LOOPS IN CONTINUOUS ARCH ORTHODONTIC THERAPY

by

MARJA KAISA INKERI HIRVONEN

A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES

IN PARTIAL FULFILMENT OF THE REQUIREMENTS FOR THE DEGREE

OF MASTER OF SCIENCE

DEPARTMENT OF PREVENTIVE DENTAL SCIENCE

WINNIPEG, MANITOBA

NOVEMBER, 1983

THE EFFECTIVENESS OF LOOPS IN CONTINUOUS ARCH ORTHODONTIC THERAPY

by

Marja Kaisa Inkeri Hirvonen

A thesis submitted to the Faculty of Graduate Studies of the University of Manitoba in partial fulfillment of the requirements of the degree of

MASTER OF SCIENCE

Abstract

In continuous archwire therapy initial alignment of malaligned teeth is usually carried out with the use of round low-modulus archwires. Previous experimental evidence shows, however, that these archwires (1) frequently generate unacceptably high forces, (2) offer no control over bucco-lingual tipping of teeth, and (3) cause significant movement of adjacent teeth which should serve as anchorage.

The present in vitro study was performed to investigate the possibilities of improving force and moment control on both the malaligned and the anchorage teeth by using multilooped archwires fabricated from rectangular wires. The three dimensional force and moment system generated by such archwires at the center of resistance of a selected tooth was measured at a given activation level. The following conclusions were drawn from the results of this study:

(1) Straight rectangular archwires when used in initial alignment provide poor control over bucco-lingual tipping of teeth.

(2) Loops in a rectangular archwire can provide good control over bucco-lingual tipping of teeth. The amount of improvement depends on the size and geometry of the loops.

(3) With looped archwires fabricated from stainless steel it is possible to create lower forces than with straight low-modulus wires.

(4) The rigidity of the archwire in the anchorage area relative to the loops in the active area greatly decreases the

expected amount of movement of the anchorage teeth.

The superior mechanical performance of looped archwires warrants efforts to overcome the difficulties of precise fabrication and placement that such wires present in practice.

Acknowledgements

First, I would like to thank my research supervisor Dr. K.R. McLachlan for his help and support.

I would also like to express my thanks to my committee members Drs. A.T. Storey and L. Gibson.

For typing this manuscript, I wish to thank Mrs. Joanne Gilbert.

My thanks to Mr. John Clark for his technical assistance.

My thanks and best wishes to my good classmates Greg Vigneux and Chung Yue.

TABLE OF CONTENTS

List of Figures

Figures Page

Figures		Page

Figures | Page

List of Tables

Tables Page

Introduction

In modern orthodontics continuous archwire mechanotherapy is commonly used. Various techniques have been developed to allow initial engagement of the wire into the brackets. These techniques range from incorporating loops and bends into the wire to accomodate severe malalignments to straight wire techniques employing triple control brackets. In triple control appliances the desired amount of torque angulation and offset are built into brackets and no bends are needed in the archwire. Preformed archwires are tied into edgewise bracket slots starting with small round flexible wires and working up to stiff, full-size rectangular wires. It is assumed that such a progression of wire sizes ensures acceptable force levels to be maintained throughout the treatment.

There is little experimental evidence which quantifies the force systems applied on teeth with an edgewise appliance. Every tooth tied into the archwire bears a certain amount of force and the force systems created are statically indeterminant. A machine has been developed which allows a three dimensional assessment of forces and moments on a tooth and Sullivan (1982) has used this machine to study the forces and moments generated in straight wire therapy. Sullivan's work suggests that the force levels with light round wires are unacceptably high. In addition, round wires give no direct control over bucco-lingual tipping during the bracket alignment phase. By inference rectangular wires which fit tightly into the bracket slots would have to be used to give direct control over bucco-lingual tipping during the finishing phase of treatment. Little is known, however, about the force delivery of continuous or looped rectangular archwires. Using the afore mentioned machine the purpose of the present investigation was to build on information gained by Sullivan and in particular:

(1) to study the increase of control gained by the use of rectangular archwires in rectangular bracket slots,

(2) to study the amount of flexibility which can be gained in both the bucco-lingual and occluso-gingival directions by the use of loops in the archwire, and

(3) to study force systems acting on teeth when continuous round archwires are tied into occluso-gingivally malaligned teeth.

Review of the Literature

Many studies have been performed to find optimal force levels for different types of tooth movement. An optimal force is one which would accomplish the fastest tooth movement consistent with acceptable levels of trauma to periodontal tissues. These studies show lack of agreement on optimum and acceptable force levels but in general they indicate that forces much above a few hundred grams when applied on a single tooth are excessive (Reitan, 1957).

Schwartz (1932) suggested that forces which exceed capillary blood pressure in the periodontal ligament (25 grams per square centimeter) should not be used. Ackerman et al. (1969) studied clinically a modification of edgewise technique using light continuous and interrupted forces. They felt that (1) there is no single optimal force but a broad range of forces that will produce the same rate of tooth movement under similar circumstances, and also that (2) duration of force application is more important than the magnitude of force.

Sullivan (1982) stated that when a straight archwire is tied into malaligned brackets the force systems acting on teeth are very complicated and statically indeterminant. The use of round archwires in edgewise brackets provides no control over bucco-lingual tipping of teeth. Undesirable forces and moments are acting not only on the malaligned teeth but also on the adjacent teeth which serve as anchorage. The forces and moments acting on adjacent teeth are comparable in magnitude with those acting on malaligned teeth. Thus the control over the final tooth position is lost.

The study reported here was aimed at investigating the possibilities of improving the force and moment delivery in the initial alignment of malaligned teeth by the use of loops in a rectangular archwire. Hence, the literature survey is subdivided into the following sections:

Continuous Archwires

Segmented Archwires

Looped Archwires

Rectangular Archwires

Commentary.

Continuous Archwires

Since the archwires used in orthodontics are intended to be activated only within their elastic limits they should return to their original form on total deactivation. A continuous arch can approach a malaligned tooth only by being distorted. When it is tied into the bracket energy is stored in the wire and will be delivered during deactivation (Steiner, 1953). The magnitude of the force applied for a given activation (the load-deflection rate of the appliance) and the distance over which the wire can be deflected (the range) depend on the properties of the wire material, the geometry of the wire and on the constraints to which it is subjected (Thurow, 1982).

According to Burstone and Koenig (1974) "the dogma of the ideal arch states, that if a wire is bent to the shape in which one would like the brackets to be found at the end of treatment, the tooth will move into a position on the ideal arch and thus produce the desired occlusion." They felt that this concept is more valid when a very rigid archwire is used: "it acts as a mold and the teeth are slowly displaced towards the desired position." Conversely, increased flexibility of the archwire causes undesirable movements of the teeth adjacent to the malaligned teeth. This becomes more noticeable with continuous archwires which have (1) a smaller cross-section, (2) a great deal of play between the archwire and the bracket, or (3) added flexibility produced by loops and other configurations of the wire. Thus any design factor which tends to reduce the rigidity of the archwire in the anchorage area will increasingly allow the reciprocal forces to displace the anchorage teeth.

Steiner (1953) felt that an ideal continuous archwire should be flexible enough to store energy well and to move malaligned teeth but it should be rigid enough to stabilize the anchorage areas and to resist environmental forces from the lips, the cheeks and the tongue and especially from the occlusion. In most cases, however, no archwire can meet these requirements and a compromise has to be made either in the flexibility or in the anchorage requirements.

Stoner (1960) analyzed theoretically the forces used in orthodontic practice. According to him the more the wire is deflected and the shorter the distance between the brackets on the adjacent teeth, especially on severely malpositioned teeth, the greater will be the force. This is supported by the work of Sullivan (1982). His results show that when initial alignment of malaligned teeth is done with a continuous archwire the force levels are significantly higher than is generally thought even when the deflections are fairly small and when so called flexible archwires (Nitinol, TMA or multistranded wires) are used. For example, he found that when the interbracket distances were 3 millimeters on each side and a 0.40 millimeter (0.016 inch) round Nitinol wire was activated through 1.5 millimeters it provided a force of over 600 grams.

Waters et al. (1981) felt that the force applied with a continuous archwire is directly proportional to the amount of deflection over a very limited range only. This is caused by the fact that most wires start to permanently deform at quite small deflections. Bracket binding caused by the friction between the archwire and the brackets and by possible bends in the archwire further complicates the unloading pattern.

Waters et al. (1975) presented theoretical calculations of the force levels applied on a malaligned tooth with plain and looped archwire spans. They made five assumptions to simplify the calculations: (1) no force was dissipated through interdental contacts, (2) there was a single point contact between the bracket and the archwire and no force was lost through friction, (3) the distances between the brackets on the adjacent teeth were equal, (4) the wire was deformed elastically, and (5) the span was loaded centrally. Although these assumptions cause the model to differ substantially from the real clinical situations some conclusions could be drawn. These authors felt that the force levels created by plain round archwire spans are not significantly different for horizontal and vertical deflections. In symmetrical situations tying additional teeth into the archwire on either side of the central span will increase the stiffness of it. The load-deflection characteristics of the central span are not significantly affected by extra constraints after the first two teeth on either side.

Creekmore (1976) analyzed the stiffnesses of round and rectangular archwire spans using single brackets and extra wide Siamese brackets. He found that "light forces and amounts of tooth movements per adjustment depend more on interbracket width than archwire size." He calculated that a 50 percent increase in interbracket width decreases stiffness 3.37 times for bending and 1.5 times for torsion. He felt , however, that the disadvantage of using narrow brackets is the loss of rotation and tipping control. According to Burstone (1962) and Sullivan (1982) broad brackets with narrow slots offer more precise control of tooth position, but, if they are used force levels with continuous arches may become excessive even when so called flexible wires are used. Waters et al. (1975) concluded from their calculations of the stiffness of plain continuous archwire spans relative to the length of the span that the stiffness of the archwire span greatly depends on its length and therefore the width of the brackets becomes very significant. If the wire is fully engaged into brackets 3 millimeters wide the stiffness of a span 13 millimeters long increases 6.4 times compared to the point contacts they assumed in their theoretical analyses.

Waters et al. (1981) stated that unlike the experimental situation, clinically "some protection for the patient against excessive force levels may be given by immediate small movements of the teeth within the limit of the periodontal ligament, by dissipation of the force through the interdental contacts, by plastic deformation of the archwire, by binding between the archwire and brackets and even by bending of the alveolar process". He suggests, however, that the simple use of continuous archwires is not suitable for the initial alignment of very irregular teeth, especially if wide brackets are used. Serious consideration should be given to the use of looped archwires or other techniques in such cases.

<u>Segmented Archwires</u>

Burstone (1966) developed his segmented arch technique "in order that continuous forces could be utilized without the loss of control". According to him the use of segmented arches makes it possible to control the tooth movement with known moment to force ratios and to aim at optimal biologic response by delivery of relatively constant forces at an optimal magnitude.

"Unlike the typical continuous archwire, the segmented arch usually consists of multiple wire cross-sections so that varying wire sizes may be found in different portions of the arch" (Burstone, 1962). It is possible to use more flexible wire materials and smaller cross-sections in the areas where more flexibility is required to accomplish tooth movements. Conversely, where more rigidity is needed, for better control as in stabilizing arches in the anchorage areas, heavier steel wires can be used (Burstone, 1962). He suggested that when anchorage areas are stabilized with heavier wires and possibly with palatal and lingual arches, the tooth movement can be accomplished by using larger segments as anchorage. In this way better control is gained. In contrast only the teeth adjacent to the malaligned tooth serve as anchorage units when a conventional continuous arch is used (Burstone, 1962 and Sullivan, 1982).

When a force is applied on a malaligned tooth an opposite reaction force is developed in the anchorage area. Sufficient resistance has to be employed there to prevent the anchor teeth from moving. Begg (1956) originated the concept of "differential force" which he describes as a force great enough to move certain teeth but not great enough to move others. According to Jarabak and Fizzel (1972) differential forces are equal and opposite forces which are developed by a single active element in an appliance.

Differential forces are produced by distribution of one force of a given magnitude over a certain root area and its reciprocal force over a significantly larger root area (Weinstein, 1967). According to Burstone (1962) this can be accomplished by the use of segmented arch technique. If the rigidity of the archwire is maintained in the anchorage area forces are distributed over a larger root area and the force levels on individual teeth would be below the threshold level and therefore not capable of causing tooth movement.

Hixon et al. (1969 and 1970) felt that there is no data to support the theory of optimal force and so there is no basis for the differential force theory. They agree that "higher forces per unit root area increases the rate of biological response." They feel that "higher forces produce more rapid tooth movement but there are large differences between patients."

When continuous archwires are used in multibanded appliances the distance between brackets on adjacent teeth is limited. This results in high load-deflection rates (Sullivan, 1982). On the other hand, the segmented arch philosophy allows much greater flexibility in choosing the distance between points of force application since archwires are not necessarily tied into adjacent teeth (Burstone, 1966).

Looped Archwires

The practice of the segmented arch technique can be approached by incorporating loops into a continuous archwire to reduce the stiffness in the areas where lower load-deflection rates are required. If these looped arches are fabricated in rigid wire it is possible to control anchorage areas quite precisely. Thus, using looped arches instead of plain continuous arches makes it easier to meet the two requirements Steiner (1953) placed on orthodontic archwires which are: (1) enough flexibility in active areas and (2) enough rigidity in achorage areas.

Burstone (1961) tested the load-deflection rates of various orthodontic springs with an instrument incorporating a force gauge and a flexor tester. He felt that the release of relatively constant forces depends primarily on the structural design of the appliance and secondly on the mechanical properties of the wire. Jarabak and Fizzel (1972) stated that by incorporating loops it is possible to alter the load-deflection characteristics much more effectively than by varying the wire material or cross-section. By changing the configuration of the loop it is possible to reduce the stiffness in the plane in which more flexibility is required and maintain the rigidity in the other planes. For example, when vertical loops are used the flexibility is primarily increased in the horizontal plane.

According to Stoner (1960) any loop will reduce force (load-deflection rate) and increase the range (maximum elastic deflection). The force reduction effected by contouring most loops is directly related to the increase in the amount of wire between the brackets. The force of any loop may be further reduced by coiling the wire one or more times at the apex. He stated that "the load-deflection characteristics are dependent on two types of forces built up in it:

(1) The spring of the legs themselves which act as independent levers and may be activated in any direction at right angles to its deflection but it may not be deflected easily through its length, and

(2) The activity developed at the curvature of the apex of the loop. A loop may act as a rigid lever taking deflection at the apex, or it may act as an independent spring depending upon the resiliency of the wire in the loop to give it activity. Usually, the action of any loop is dependent on the

combination of the two" (Stoner, 1960).

The results of Meador's study (1982) indicated that selection of loop design is a very important factor in determining force systems which will be created at the center of resistance of the involved teeth. He used the same instrumentation as was used by Lack (1980). He found that for vertical activations of alignment loops low stiffness may be most efficiently attained by addition of wire in the horizontal plane. Conversely, vertical addition of wire is most effective in altering the stiffness for horizontal activations. Helical additions to a loop are of little value in force control and may result in an increase in spurious moment magnitudes.

Waters et al. (1975) performed a theorectical analysis of the effectiveness of vertical loops in a continuous archwire. They determined the reduction in stiffness in the horizontal and vertical deflections given by such a loop over that provided by a straight wire. They studied the behavior of two vertical loops symmetrically placed on either side of a malaligned tooth in a segment involving three teeth. The length of the span was 13 millimeters, which is the average span length in the lower incisor area. To simplify the calculations they made the same assumptions as when they studied a plain archwire span. They found that when the loops are deflected into bucco-lingual direction the vertical components of the loops bend much more than the horizontal components. The base of the loops undergo torsion. The flexibility of the span for horizontal deflections increased by lengthening the vertical components of the loops and by widening the base. When the length of the vertical components was increased from 6 millimeters to 7 millimeters the stiffness was reduced by 30 percent. An increase in the base width from 2 millimeters to 3 millimeters reduces the span stiffness by 20 percent under horizontal loading.

They stated that when the tooth in the middle is deflected verticaly (into occluso- gingival direction) bending occurs in the horizontal components. The width of the loop base is most important and the length of the vertical components makes only a minor contribution. They concluded that "a multilooped arch with vertical loops is always much stiffer in the vertical plane than in the horizontal plane." Vertical loops are effective in altering the stiffness of a wire to horizontal deflections with vertical deflections, however, a looped span of 0.40 millimeter (0.016 inch) stainless steel wire is always stiffer than a plain span of multiple strand wire.

Waters et al. (1981) characterized archwires for initial alignment of irregular teeth. In addition to theoretical analyses he used a rig and weights to test the load-deflection rates of various archwire spans. According to his results looped arches offer better control and more favourable force levels for initial alignment of very irregular teeth than continuous flexible arches. This becomes more apparent if wide brackets with narrow slots are used to offer more precise control over tooth position. He thought, however, that looped arches cause more patient discomfort and are more time consuming to fabricate, "and unless the mechanics of these complex archwires is fully understood, unwanted tooth movement may be produced both when the loop is activated and when the force distribution (the moment to force ratio) changes with tooth movement."

Rectangular Archwires

Johns (1953) studied the forces applied by different orthodontic appliances on a palatally displaced maxillary lateral incisor using an electric strain gauge. He felt that only with the use of rectangular archwires in an edgewise appliance can forces be applied on any tooth in any direction. His results indicate, however, that "when stainless steel wires are used in an edgewise appliance force levels in all three orders of bends are higher than with any other appliance". Therefore he felt that continuous rectangular stainless steel archwires are not suitable for alignment of severely malaligned teeth.

Moyers and Bauer (1950) studied the histological changes in the periodontal ligament of teeth that had been moved orthodontically. His results indicated that the amount of obstruction of blood supply and thus the rate of bone remodeling in periodontal ligament differs depending on the type of tooth movement attempted. Therefore for example tipping and extrusion are more easily carried out than bodily movement and intrusion. Halderson et al. (1953) felt that it is most difficult to predict the exact direction and amount of tooth movement achieved by a rectangular archwire. This was based on the data of Moyers and Bauer (1950).

In order to make the load-deflection characteristics of rectangular stainless steel arches more favourable Carey (1956) developed laminated edgewise and ribbon arches. The rationale behind his "double edgewise" and "double ribbon" is that when the archwire was split into two each half would distort equally under the same load. An archwire divided vertically would deflect horizontally eight times as much as an undivided arch. According to him it would provide more control for bucco-lingual tipping than a round archwire. His laminated arches were not experimentally tested.

Burstone (1981) stated that a rectangular archwire provides better control during the initial alignment of irregular teeth than a round archwire. He studied the elastic properties of various orthodontic archwire materials. Based on his studies he developed his concept of "variable modulous orthdontics." According to this concept the wire material should be changed instead of changing from smaller to larger cross-sections using the same wire material during the progressive steps of orthodontic treatment. This would allow the use of rectangular archwires during the entire treatment.

But as already mentioned Sullivan's (1982) results indicate that when continuous arches are used for initial alignment of severely malaligned teeth the force levels created even with round flexible wires are excessive. Changing from round to rectangular wire cross-section makes the force levels even higher. So, inspite, of their potential to provide better control of bucco-lingual tipping of the teeth commercially available rectangular wires are seldom suitable for initial alignment.

The ability of a continuous arch to return to its original configuration after it has been engaged into the bracket on a malaligned tooth depends on its ability to slide through the brackets on the adjacent teeth (Jarabak and Fizzel, 1972 and Sullivan, 1982). Therefore the use of larger rectangular wires which fit tightly into the bracket slots is undesirable during the initial alignment with straight archwires.

Commentary

Edgewise appliances with continuous straight archwires are the most commonly used technique in modern orthodontics. However, there is little experimental evidence to describe the force systems applied on teeth when this technique is used.

Studies done on optimal force levels for tooth movement show lack of agreement but it can be concluded that forces should not exceed a few hundred grams on a single tooth. When continuous straight archwires are used for initial alignment of severely malaligned teeth force levels are often excessive even when so called low modulus archwires are used. In many cases the elastic limits of wires are exceeded and they are permanently distorted.

When a tooth is tied into a continuous archwire it is seldom undergoing one single type of movement. Round archwires provide no control over bucco-lingual tipping of teeth but even when rectangular archwires are used it is more difficult to predict the exact amount and direction of tooth movement. Therefore the tooth movement achieved can be quite different from that which is intended especially during initial alignment when the distances over which the forces are active are fairly large. The teeth adjacent to malaligned teeth automatically serve as anchorage and large forces and moments are acting on them as well. Thus control over the final tooth position is lost.

By incorporating loops into rigid archwires forces can be reduced and range increased in the areas where more flexibility is needed to accomplish tooth movement while rigidity can be maintained in the anchorage areas. Therefore initial alignment with multilooped archwires would allow the use of tightly fitting rectangular archwires in edgewise brackets to gain better control over tooth movement.

The purpose of the present study was to find out if the traditional multilooped archwires fabricated from rectangular wires provide better control and lower force levels than round low modulus archwires in the initial alignment of malaligned teeth.

Materials and Methods

Introduction

The data for this investigation was obtained using instrumentation similar to that used by Sullivan (1982) except for the activation system which was modified to allow the testing of rectangular as well as round wires. It consisted of three components:

a measuring system,

a data acquisition system, and

a minicomputer.

The instrumentation is shown in Figure 1. Because the measuring system does not allow the model teeth to tip, the torsional stiffness of the rectangular wires and loops were tested in separate experiments.

Measuring System

The measuring system makes use of six transducers arranged in a geometrical configuration which permits the computation of three forces and moments by a "linear combination of the six transducer responses" (Paquien, 1978).

The measuring instrument consists of four parts: the frame, an internal, suspended ring, a central triangular block and an electromagnetic vibrator (Figure 2). The relationship between the axes of the measuring system and those of the measured tooth were the same as described by Sullivan (1982). It is shown in Figure

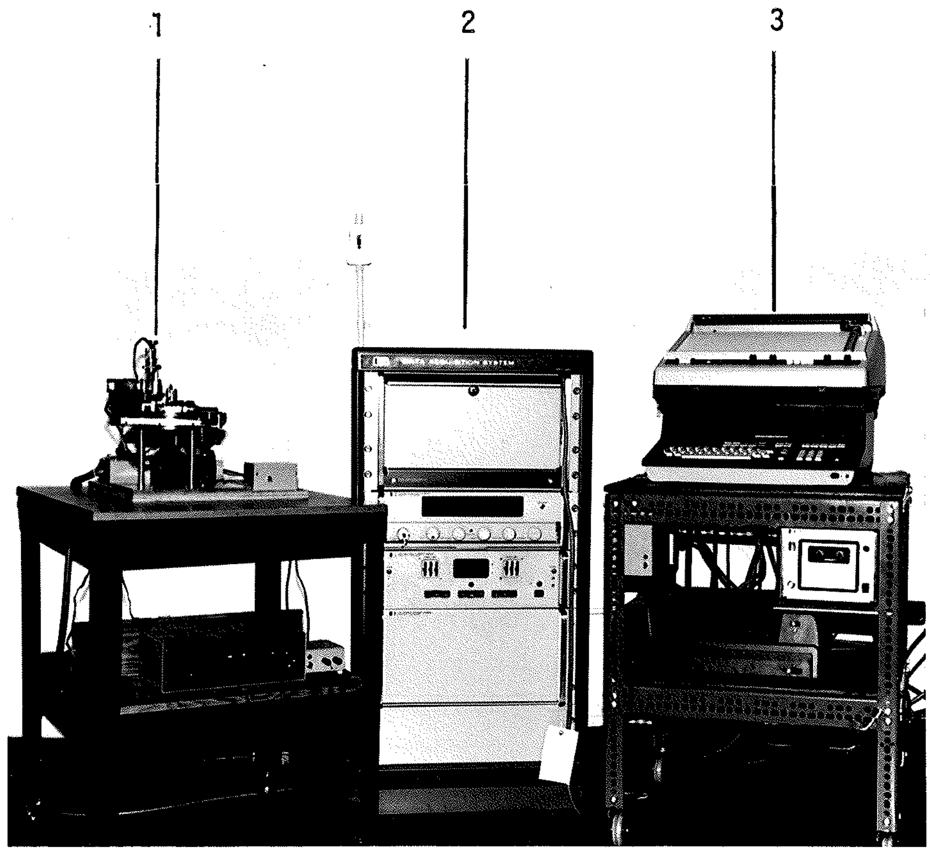


Figure 1. General view of the instrumentation.

1. Measuring system
2. Data acquisition system
3. Minicomputer

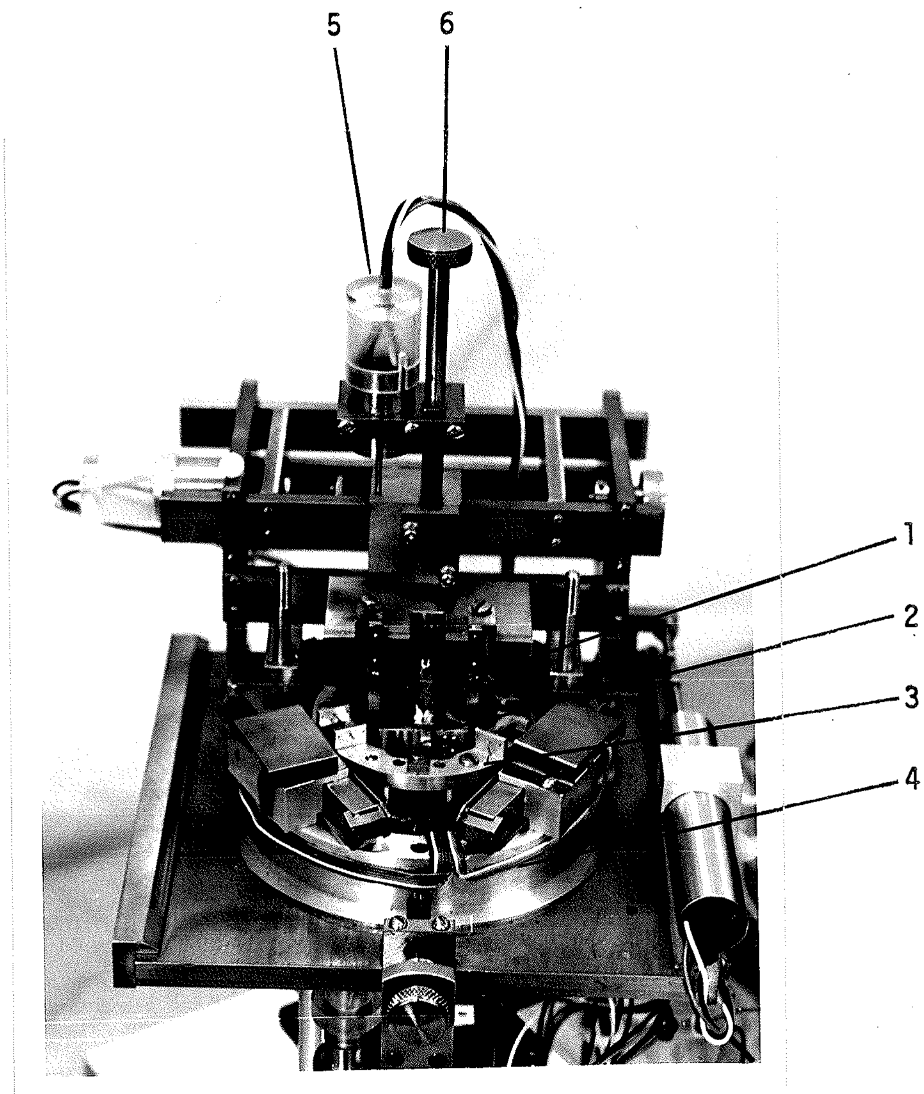


Figure 2. The measuring instrument.

1. Measured tooth
2. Triangular block
3. Suspended ring
4. Frame
5. Linear voltage displacment transducer
6. Adjustment screw

3. The measuring instrument determines the forces and moments exerted at the center or resistance of the measured tooth. To this end the center of measurement of the measuring system and the center of resistance of the measured tooth are coincident. The perpendicular distance between the center of resistance and the bracket on the measured tooth is 4.2 millimeters in the horizontal, buccal direction and 12.5 millimeters in the vertical, occlusal direction.

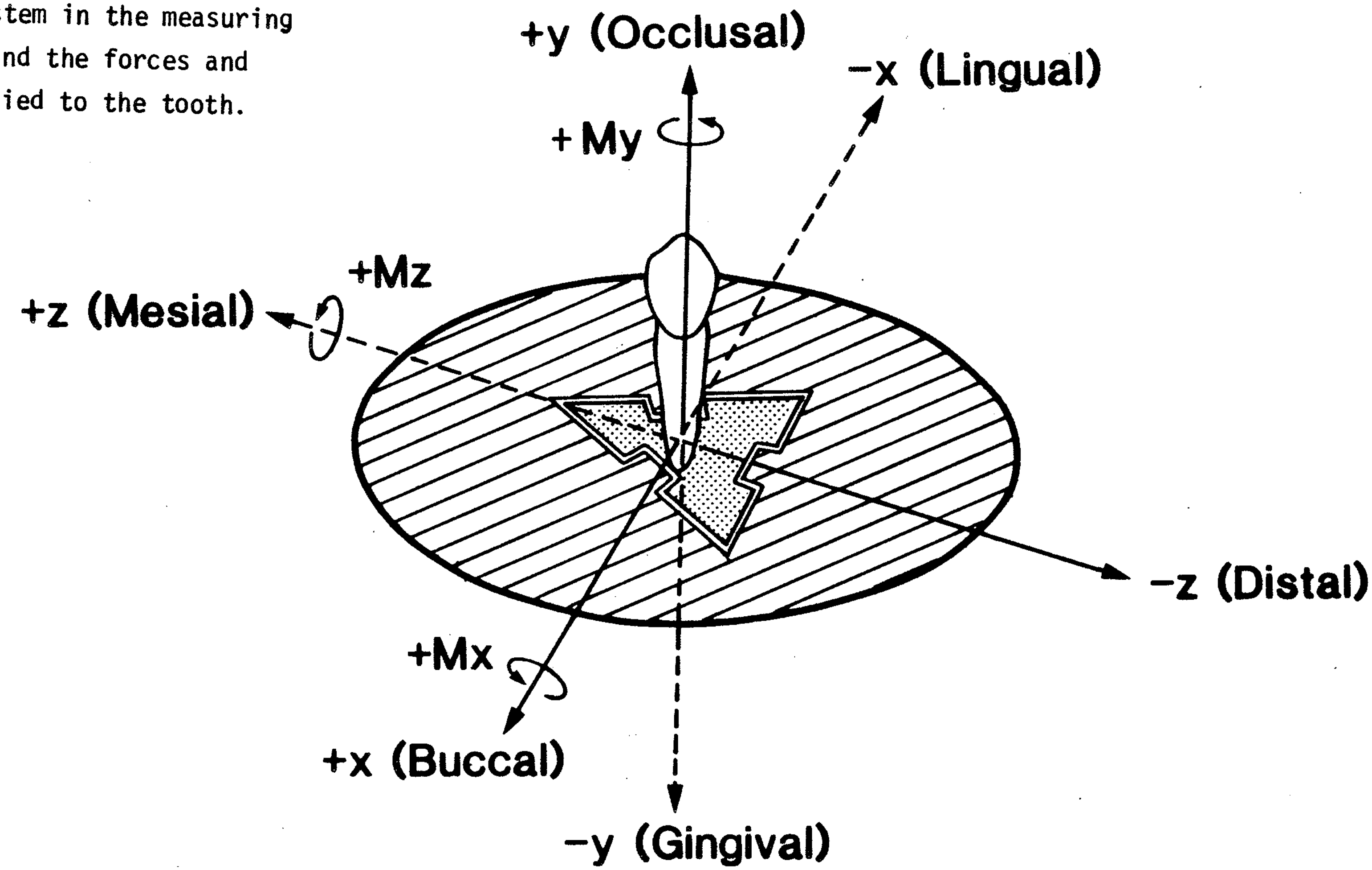
Two types of transducers are used, type A and type B. The type A transducers are attached to the central triangular block and are free to slide horizontally through their contacts located on the suspended ring. They measure horizontal forces and the pivoting moment. The type B transducers are attached to the frame and to the suspending ring. They measure vertical forces and tipping moments.

The instrument has a maximum force range of 1000 grams and a maximum moment range of 23,000 gram-millimeters. The accuracy of the measurements was confirmed, using dead weight calibration method, to be within plus or minus three percent of maximum force or moment levels.

Data Acquisition System

The data acquisition system is utilized to measure the electrical outputs of the transducers in the measuring instrument. It possesses a scanner which allows the sequential measurement of the transducer outputs to be made in about one second. These forces

Figure 3. The relationship between the axis system in the measuring instrument and the forces and moments applied to the tooth.



are relayed to the computer to allow the calculation of the forces and moments generated when the measuring instrumentation is loaded (Lack, 1980).

Minicomputer

The minicomputer is a Hewlett-Packard 9830A interfaced with an x-y plotter, a printer, a recording system, and the data acquisition system. It was used to collect, calculate, store and present the data.

Computer Programs

Modified versions of the two computer programs written by McLachlan (1979) were utilized in this study. They were similar to those used by Sullivan (1982). Both programs were written in BASIC, and consisted of a data acquisition program and a data analysis program.

The data acquisition program was used to collect data from the transducers and also to control the data collection. It also provided a graphic display of one variable (for example, the total force generated) on the x-y plotter. The program allowed for storage of the digital data on magnetic tape at the end of each experimental run.

The data analysis program was used to convert the raw data obtained from the transducers into three forces and three moments at the center of measurement of the measuring system. Chosen relationships could then be displayed on the x-y plotter or printed on a high speed printer. The program allows the horizontal axis to be proportional to the activation or to the force and the vertical axis to moments or forces.

Activation System

The activation system consisted of two interchangeable assemblies and two interchangeable sets of attachments. They were made of brass. A combination of three, four or five teeth could be tested.

Further modifications of the model teeth described by Sullivan (1982) were made to allow for the investigation of rectangular archwires and looped archwires. The attachments consist of brass frames where two identical stainless steel blocks are mounted in such a way that a rectangular space is left between them to simulate the bracket slot. They are four millimeters wide thus simulating a bracket width of four millimeter. One set forms a rectangular slot into which a 0.41 x 0.56 millimeter (0.016 x 0.022 inch) wire fits tightly. The other set forms a 0.46 x 0.61 millimeter (0.018 x 0.025 inch) rectangular slot which allows 0.41 x 0.56 millimeter wire to slide through. The attachment made for testing round wires are described in Apppendix B.

Assembly I is shown in Figure 4. When it is used the rest of the teeth are moved with respect to the measured tooth. They are attached to a horizontal plate in such a way that when their fixing screws are tightened the alignment of their slots (brackets) is automatically achieved. When assembly I is used the measured

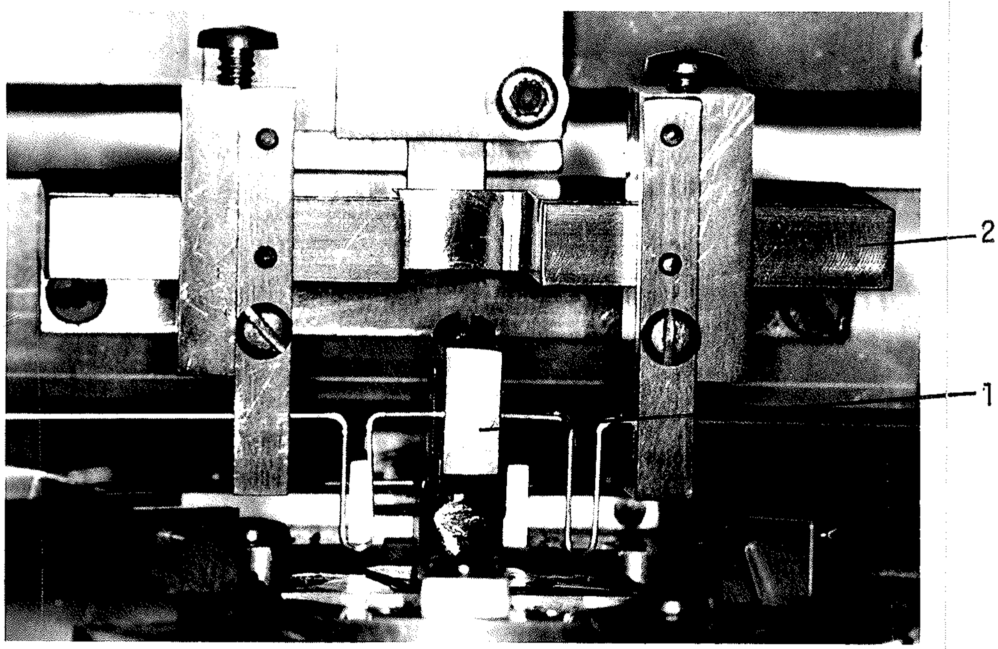


Figure 4. Activation assembly I.

1. Measured tooth
2. Horizontal activation plate

tooth acts as a malaligned tooth (active tooth).

Assembly II is shown in Figure 5. When it is used the tooth adjacent to the measured tooth is attached to a vertical extension of a horizontal plate. It can be moved independently from the other teeth which are attached to a fixed horizontal plate. When assembly II is used the measured tooth acts as a tooth adjacent to the malaligned tooth (reactive tooth).

The activation system can activate the appliance which is being tested in both the positive and the negative directions in three planes of space. In this investigation bucco-lingual and occluso-gingival activations only were used. Activation directions and distances are monitored by the appropriate linear voltage displacement transducer. Activation is accomplished manually through a screw mechanism.

In this investigation an activation distance of one, two or three millimeters was chosen. The amount of activation in each case depended on the load-deflection rate of the appliance being tested. The appliance was deflected as far as possible without overloading the measuring system or causing permanent deformation to the appliance.

Round Wires

A study was done on force systems acting vertically on malaligned teeth when a round low-modulus archwire was tied into brackets. It is discussed in Appendix B.

Rectangular Wires

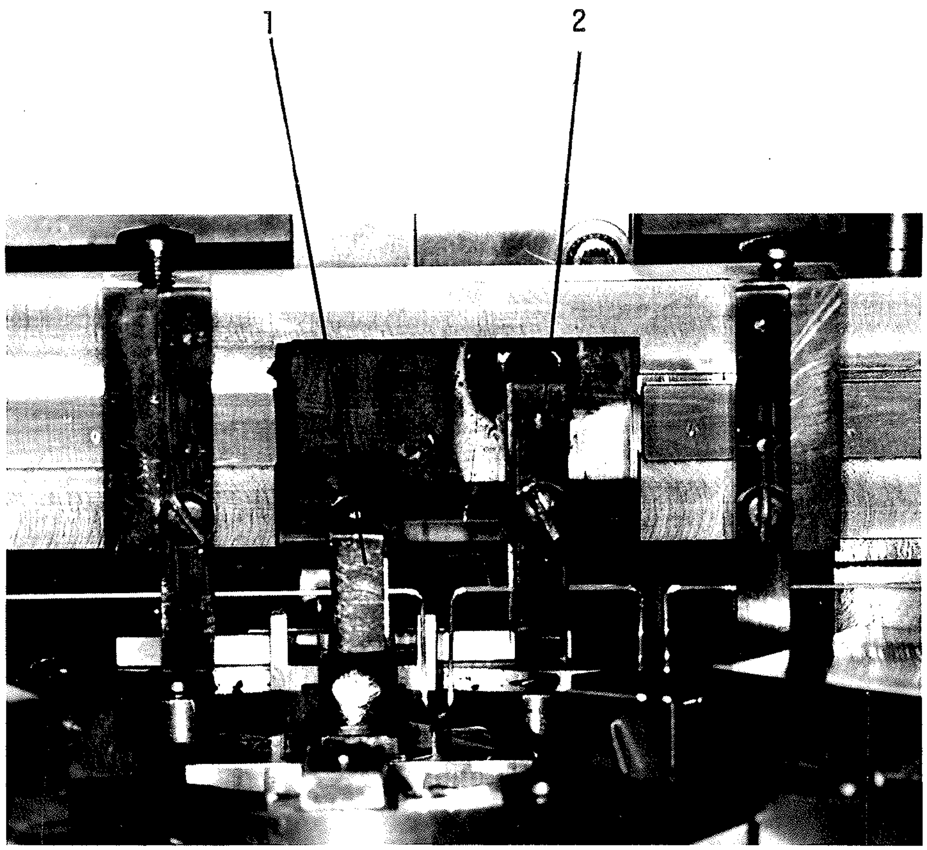


Figure 5. Activation assembly II.

1. Measured tooth
2. Malaligned tooth

(1)	Stainless Steel	0.41 x 0.56 mm.	(0.016 x 0.022 inch)
(2)	Nitinol	0.41 x 0.56 mm.	(0.016 x 0.022 inch)
(3)	TMA	0.41 x 0.56 mm.	(0.016 x 0.022 inch)

The rectangular wire size 0.41 x 0.56 millimeter (0.016 x 0.022 inch) was chosen because it was the smallest rectangular wire size available in all commonly available wire materials. It was felt to be desirable to use the smallest possible wire size in order not to overload the measuring instrument.

Fabrication of Loops

The loops tested in this investigation were fabricated by hand. Rectangular 0.41 x 0.56 millimeter stainless steel wire was used in the edgewise configuration. A template was first drawn on millimeter paper. Care was taken to keep all parts of the loop in one plane and not to build in any torque. Sharp bends were avoided. Only one example of each looped wire was used. The design and dimensions of the loops are shown in Figure 6.

Vertical loops tested in this study:

(1) Height 8 mm., base width 1.5 mm. (Will be referred to as vertical loops 8 x 1.5 mm.)

(2) Height 10 mm., base width 1.5 mm., 3 mm. and 4.5 mm. (Will be referred to as vertical loops 10 x 1.5 mm., 10 x 3 mm. and 10 x 4.5 mm.)

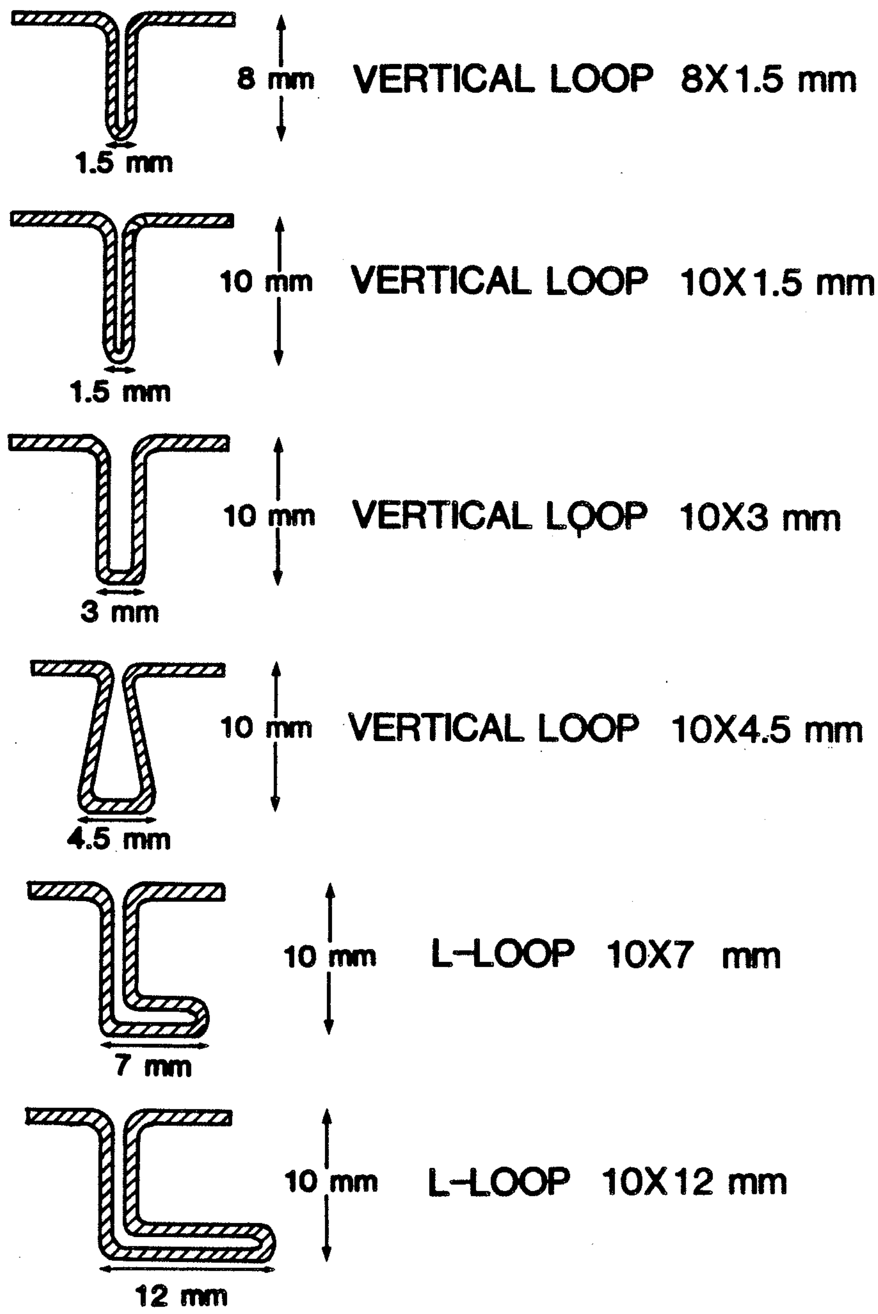


FIGURE 6 : TYPES AND DIMENSIONS OF LOOPS.

L-loops tested in this study:

(1) Height 10 mm., length of horizontal component 7 mm. (Will be referred to as L-loops 10 x 7 mm.)

(2) Height 10 mm., length of horizontal component 12 mm. (Will be referred to as L-loops 10 x 12 mm.)

Experimental Procedure

The experiments were started with the use of three teeth in assembly I. The effects of activation on the malaligned tooth could be determined. The larger 0.46 x 0.61 millimeter slots were used and the attachments adjusted to establish symmetrical 5 and 10 millimeter interbracket distances on either side of the measured tooth. The attachments were aligned and the wire samples were passively inserted into them in such a way that there was a loop in the middle of the archwire span on either side of the malaligned tooth. During the experimental run the horizontal plate with the two adjacent teeth attached was moved in the positive horizontal (buccal) or vertical (occlusal) direction over ten steps and then deactivated back to zero. After that it was activated in the negative (lingual or gingival) direction and deactivated again. Depending on the stiffness of the wire and the size of the loops being tested forces and moments were generally recorded over a range of one or two millimeter of activation.

To investigate the forces and moments exerted on the tooth adjacent to the malaligned tooth assembly II using four teeth was used. The active tooth attached to the vertical extension of the horizontal plate was moved in the positive and negative horizontal or vertical directions. Because the malaligned tooth was moved in assembly II the sign of the x-axis was reversed relative to that when assembly I was used, consequently, a switch was incorporated to signal the appropriate x-axis sign to the data collection system. The four tooth situation was tested using symmetrical 5 millimeter and 10 millimeter interbracket distances.

The measuring instrument set limitations to the dimensions of the loops which were examined and to the range of activation. The loops have to have enough space to move freely when they are activated without touching any parts of the measuring instrument. These limitations were as follows: (a) the widest vertical loops (10 x 4.5 millimeters) could not be tested with 5 millimeters interbracket distances, (b) the measurements on the reactive tooth could not be made with 10 x 3 millimeter vertical loops and 10 x 12 millimeter L-loops and (c) the bucco-lingual activation range was limited to 2 or 3 millimeters and the occluso-gingival activation range to 1 millimeter.

During the second part of the study straight lengths of rectangular wires were tested in the larger (0.46 x 0.61 millimeter) slots which allow the wire to slide through. Only assembly I was used. Experiments were started using interbracket distances of 20 millimeters. The distances were then reduced to 15 , 10 and 5 millimeters. These interbracket distances were chosen in order not to overload the measuring instrument. For the same reason Nitinol wires were used instead of using stainless steel wires.

Data Analysis

As mentioned, the analysis program allowed for the initial analysis of the data and also for the plotting of the data. The relationships chosen to represent the data when the horizontal activations were used were:

Px/Ax : the force to activation ratio (bucco-lingual)

Py/Ax : the force to activation ratio (occluso-gingival)

Pz/Ax : the force to activation ratio (mesio-distal)

Mx/Ax : the moment to activation ratio (around bucco-lingual axis)

My/Ax : the moment to activation ratio (around occluo-gingival axis)

Mz/Ax : the moment to activation ratio (around mesio-distal axis).

When the vertical activations were used the relationships were:

Px/Ay : the force to activation ratio (bucco-lingual)

Py/Ay : the force to activation ratio (occluso-gingival)

Pz/Ay : the force to activation ratio (mesio-distal)

Mx/Ay : the moment to activation ratio (around bucco-lingual axis)

My/Ay : the moment to activation ratio (around occluso-gingival axis)

Mz/Ay : the moment to activation ratio (around mesio-distal axis).

Torsional Stiffness

In the present form the activating arrangements of the instrument does not allow for rotational activation. Consequently the force and moment effect caused by the tipping of the teeth cannot be directly taken into account. Therefore it was felt to be necessary to measure the torsional stiffness of the rectangular archwires and loops which were tested in this study so that this effect could be added to the force and moment results obtained with the measuring instrument. Over the deflection range of interest in orthodontics it is assumed that the torsional stiffness is not affected by the amount of horizontal activation.

Tests were made with bracket spacings and dimensions identical to those used in the measuring instrument for testing the forces and moments. The measuring arrangement is shown diagrammatically in Figure 7. Simple, dead loads were applied to a hook on one side of a bilateral lever arm and measurements of deflection taken from two points on the other side of this lever arm. From these deflection measurements and the known separation between the two points angular deflection could be precisely calculated. To calculate the torsional stiffness the following formula (Timoshenko, S., 1940) was used:

$$\text{Torsional Stiffness} = \frac{T}{\theta} = \frac{G\,J}{D}$$

Where T is the twisting moment applied by the load W; θ is the angle of twist over which the cantelever XL moves; G is the shear

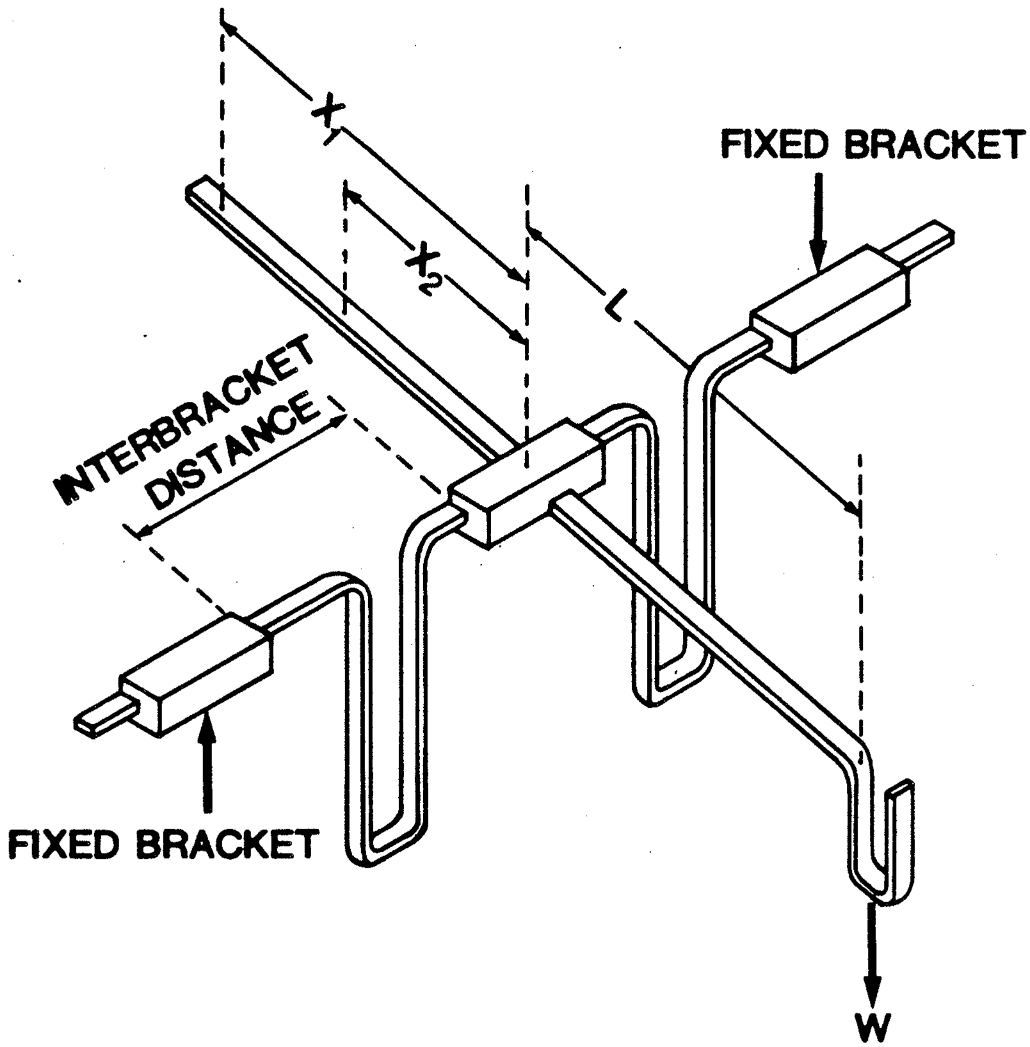


FIGURE 7 : TORSIONAL STIFFNESS MEASURING ARRANGEMENT

modulus of the wire material being tested; J is the polar second moment of area of the cross-section of the wire being tested; and D is the length of the wire.

The twisting moment is the product of the load W and the moment arm L:

$$T = WL$$

To simplify calculations θ was kept small through all the torsion tests so that we may write for their numerical values:

$$\text{Tan}\,\theta = \sin\theta = \theta$$

(Warnock, F.V. and Bendham, P.P., 1965). And hence, in this case,

$$\theta = \frac{\delta_1 - \delta_2}{x_1 - x_2}$$

where δ_1 and δ_2 are the deflections of the cantelever measured, in millimeters, at points x_1 and x_2 from the centerline of the brackets (Figure 7).

Thus the torsional stiffness can be calculated as:

$$\text{Torsional Stiffness} = \frac{T}{\theta} = \frac{WL}{\theta} = \frac{WL(X_1 - X_2)}{\delta_1 - \delta_2}$$

The length of the cantilever L was 25 millimeters, x_1 was 44 millimeters and x_2 was 13 millimeters. The load, W, was increased in five equal steps of 2.5 grams to a maximum of 12.5 grams. The maximum torsion in each case was, then,

$$T = WL = 12.5 \text{ g} \times 25 \text{ mm.} = 312.5 \text{ g.-mm.}$$

For the test conditions given above the angle of twist in each wire was

$$\theta = \frac{\delta_1 - \delta_2}{x_1 - x_2} = \frac{\delta_1 - \delta_2}{31}$$

Thus, in these tests the torsional stiffness of each wire was:

$$\frac{T}{\theta} = \frac{312.5 \times 31}{\delta_1 - \delta_2} \quad \text{(g-mm./rad)}$$

When the torsion generated by the wire or loops is added to the bucco-lingual tipping moment, Mz, obtained by the measuring instrument the total tipping effect around the bucco-lingual axis, Mz_T, can be calculated.

If the measuring instrument would allow the teeth to tip they would tip to an angle θ', which can be calculated as

$$\theta' = \frac{Ax}{CR} = \frac{2 \text{ mm.}}{12.5 \text{ mm.}} = 0.17 \text{ rad}$$

Where Ax is the maximum bucco-lingual activation or deflection for each experiment (2 millimeters) and CR is the distance between the center of resistance and the bracket of the measuring tooth (12.5 millimeters).

Torsion or countermoment generated by a wire or loop is

$$T' = \frac{T}{\theta} \times \theta'$$

where T/θ is the torsional stiffness of the wire or loop. Thus the total tipping effect around the bucco-lingual axis, Mz_T, can be calculated by adding the countermoment, T', to the tipping moment, Mz, obtained with the measuring instrument, or,

$$Mz_T = Mz + T'.$$

Results

The force and the moment results obtained with the measuring instrument for 10 x 1.5 millimeter vertical loops and 10 x 7 millimeter L-loops can be seen in the computer plots shown in figures 8 to 15. The rest of the plotted results for 10 millimeter interbracket distances are shown in the Appendix A. All the plots for 5 millimeter interbracket distances are not shown because, as the plots in Figures 8 to 15 show, they are similar to those obtained with 10 millimeter interbracket distances except that the forces and moments are higher. The plots in this thesis are photocopies of the computer plots obtained using the analysis program described in "Materials and Methods".

All results are plotted as a function of activation (bucco-lingual activation, Ax, or occluso-gingival activation, Ay). The horizontal axis of each plot represents activation distance and the maximum activation scale is indicated at the negative end of the horizontal axis. The vertical axis represents both forces and moments. The forces are marked off at 200 gram intervals reaching a maximum of 1000 grams. The moments are marked off at 4000 gram-millimeter intervals reaching a maximum of 20,000 gram-millimeters.

The plotted results in Figures 8 to 15 and Appendix A show that the forces and moments are, except for a small dead zone, linearly related to the activation. The results for the primary parameters, Px and Mz in bucco-lingual activation and Py and Mz in occluso-gingival activations, are symmetrical for positive and negative activations. For this reason only the positive activation is shown. To make comparison easier the results on the active and the reactive tooth are shown in the same plot, the active tooth on the right and the reactive tooth on the left. It should be noted that the two measurements are not taken at the same time but care was taken to ensure correct replacement of the loops in each case to make comparison valid.

The magnitudes of forces and moments are shown in Tables 1, 2, 3, and 4. The abreviations used in the tables are explained on page 50. The means of the values obtained with positive and negative activations are shown. There were differences between the values for positive and negative activations due to small misalignments of the model teeth in the beginning of experiments or to small errors in the fabrication of loops. They did not usually exceed 10 percent of the maximum value. The numbers in the brackets represent the force systems of the reactive tooth. The magnitudes quoted are those for two millimeters activation in the bucco-lingual direction and one millimeter activation in the occluso-gingival direction. However, it should be noted that in order to avoid overload of the measuring instrument 8 x 1.5 millimeter vertical loops were deflected only 0.5 millimeters in the occluso-gingival direction. The values for the forces are rounded to the nearest 10 grams and for the moments to the nearest 100 gram-millimeters to avoid the presentation of spurious accuracy. The stiffness of each wire was calculated by dividing the maximum force by the maximum activation assuming that the relationships between the forces and the activations were linear.

The torsional stiffness of the plain and the looped rectangular archwires was tested using the instrumentation described in the section on materials and methods. Results are shown in Tables 5 and 6. The tangent of the angle over which the cantilever x moved when it was loaded is shown in the first column (tan θ). The torsional stiffness of each wire is in the second column (T/θ). In table 5 the stiffness of the looped wires are presented as percentages of the stiffness of straight stainless steel wires and in Table 6 the stiffness of the Nitinol and beta titanium wires are presented as percentages of the stiffness of stainless steel wires.

Tables 7 and 8 present the results of torsional stiffness measurements combined with those obtained with the measuring instrument. The bucco-lingual tipping of the measured tooth was calculated using the method presented in "Materials and Methods". The countermoment created by the torsional stiffness of the wire which tends to resist bucco-lingual tipping is denoted by T'. Mz_T is the actual bucco-lingual tipping moment after the uprighting effect of the rectangular archwire is taken into account. Mz_T/Px shows the actual moment to force ratio in each case.

Bucco-lingual tipping was only calculated for the horizontal activations. It was not considered necessary to calculate it for vertical activations because the tipping moments created by vertical forces are much smaller than those created by horizontal activations due to the position of the bracket relative to the center of resistance of the measured tooth. Theoretically, if an appliance provided no control over bucco-lingual tipping of the tooth the moment to force ratio for horizontal activation, Mz/Px, would be 12.5 and for vertical activations, Mz/Py, 4.2. Therefore, if the uprighting effect of a rectangular wire is taken into account the bucco-lingual tipping moments created by the vertical forces becomes insignificant.

As a continuation of Sullivan's work (1982) low-modulus archwires were tested on occluso-gingivally malaligned teeth. These results are presented and discussed in Appendix B.

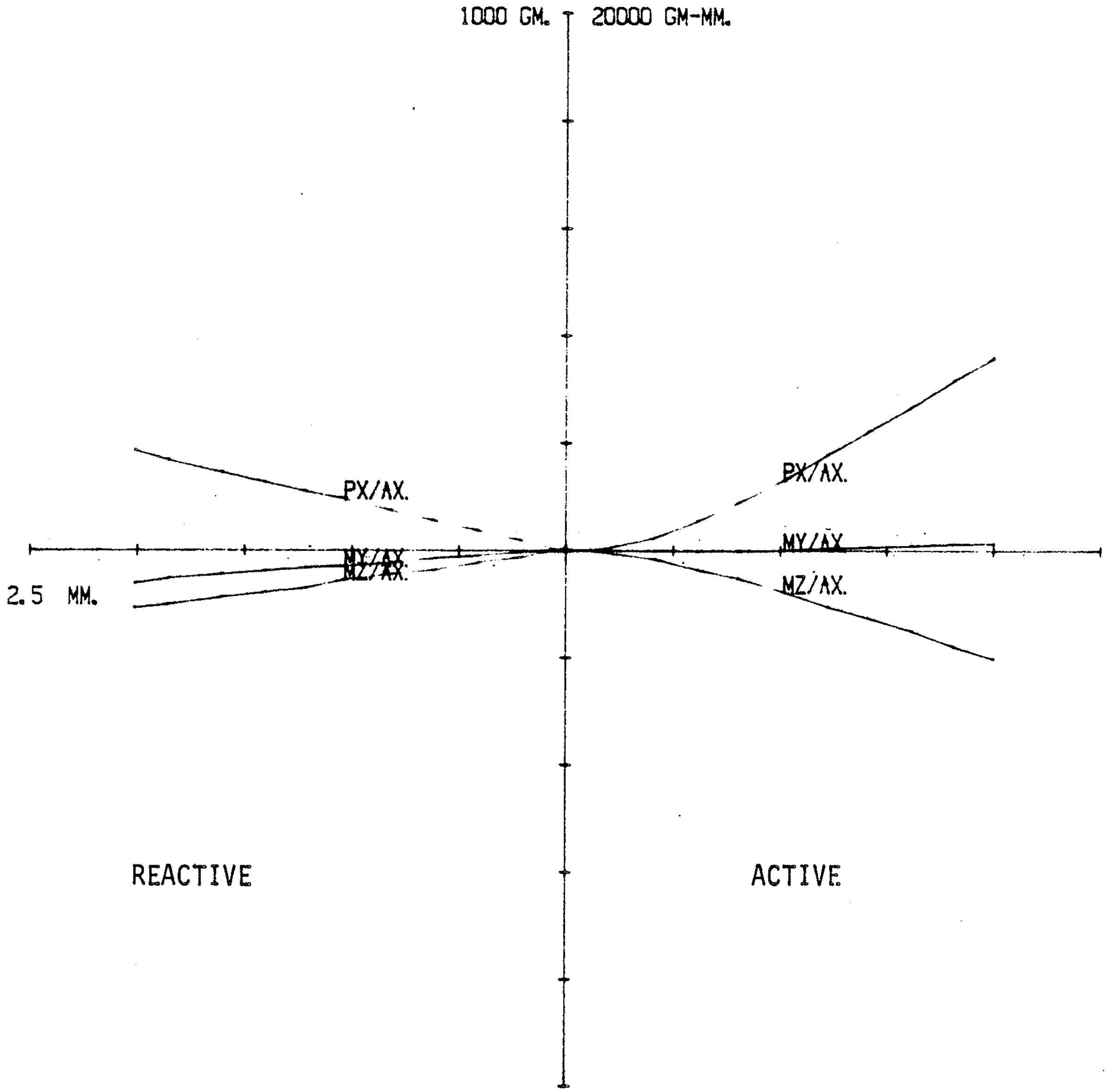


FIG. 8. VERTICAL LOOPS 10X1.5MM.
INTERBRACKET DISTANCE 10MM.
HORIZONTAL ACTIVATION.

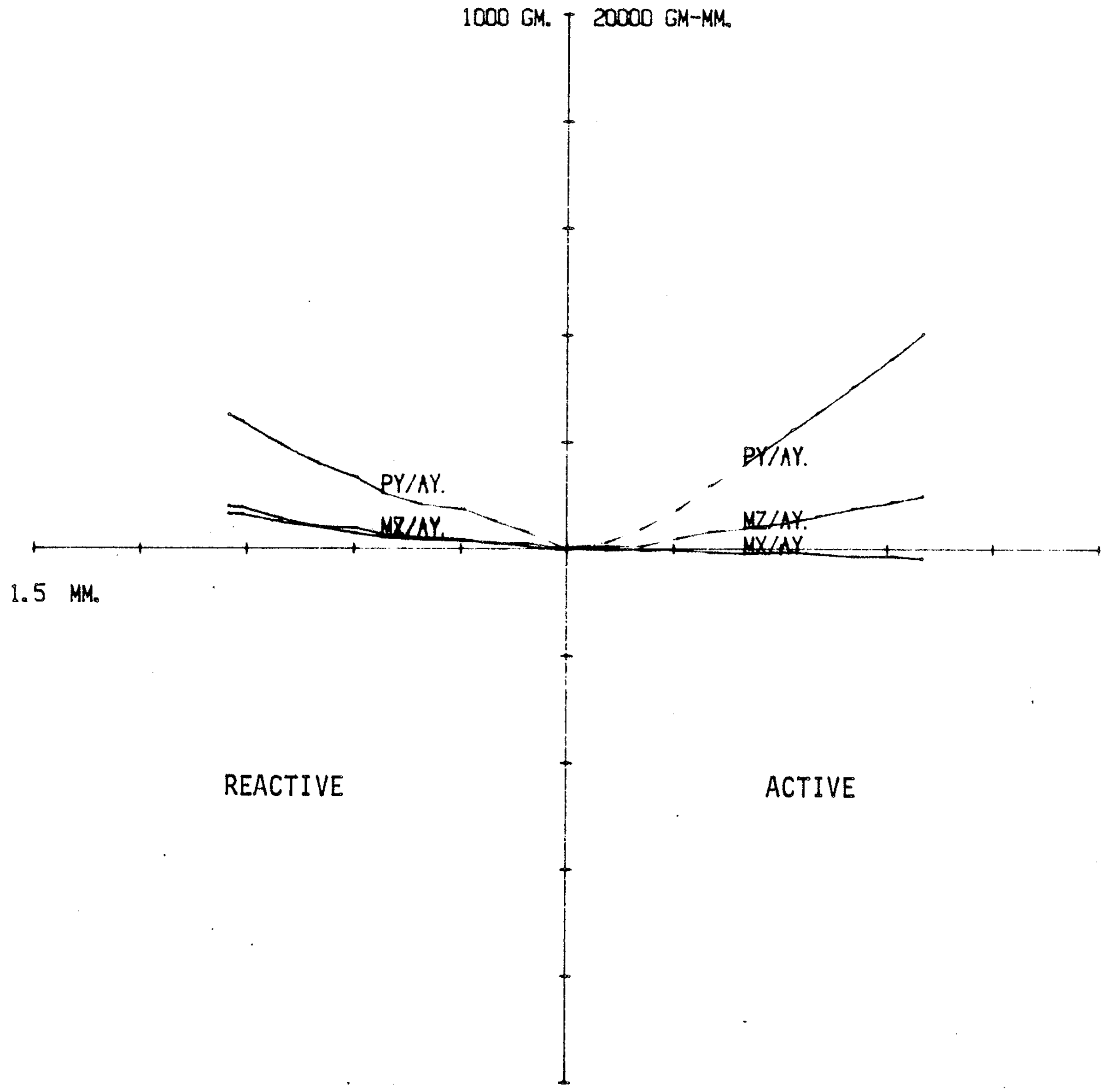


FIG. 9. VERTICAL LOOPS 10X1.5MM.
INTERBRACKET DISTANCE 10MM.
VERTICAL ACTIVATION.

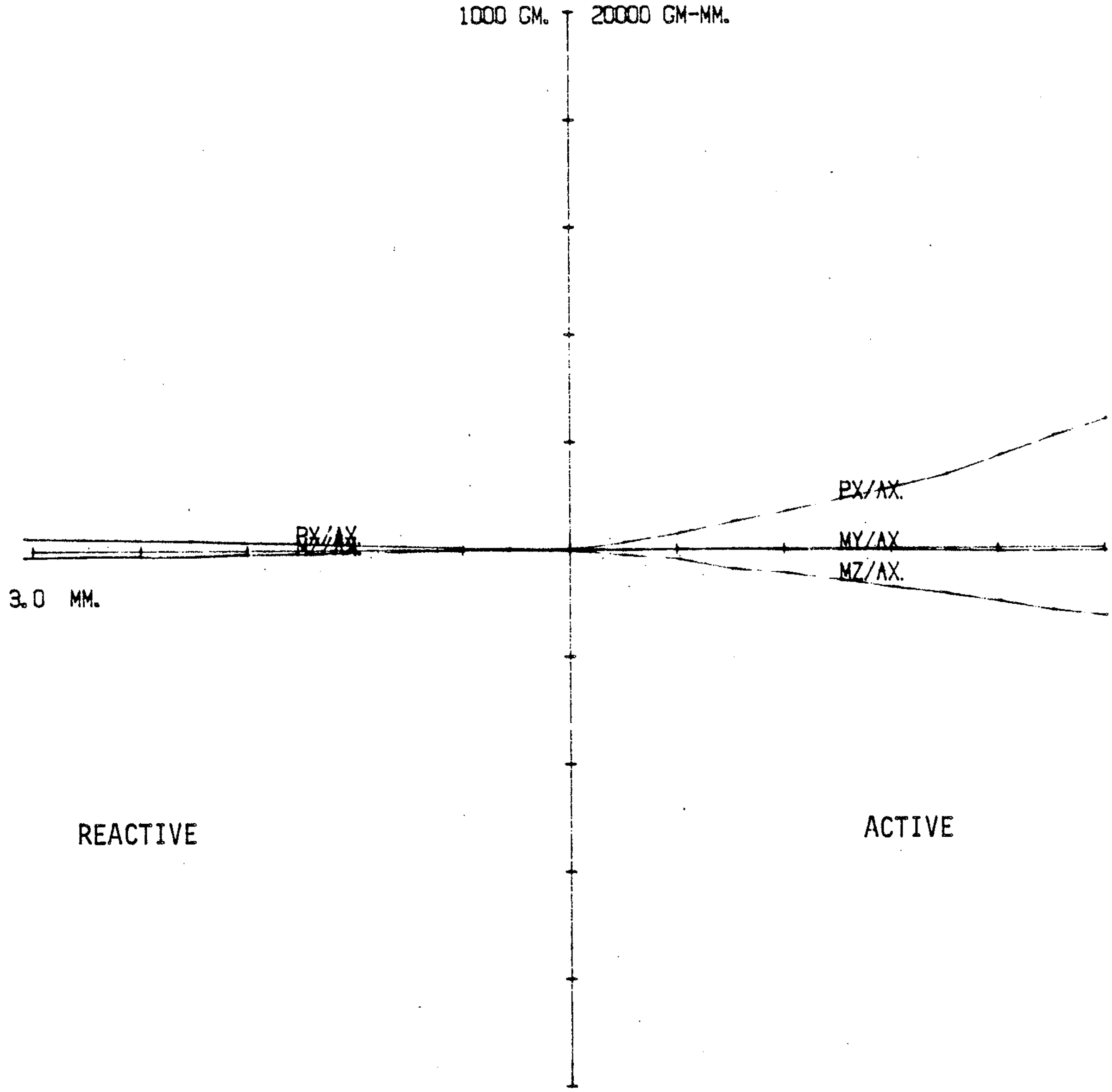


FIG. 10. L-LOOPS 10X7MM.
INTERBRACKET DISTANCE 10MM.
HORIZONTAL ACTIVATION.

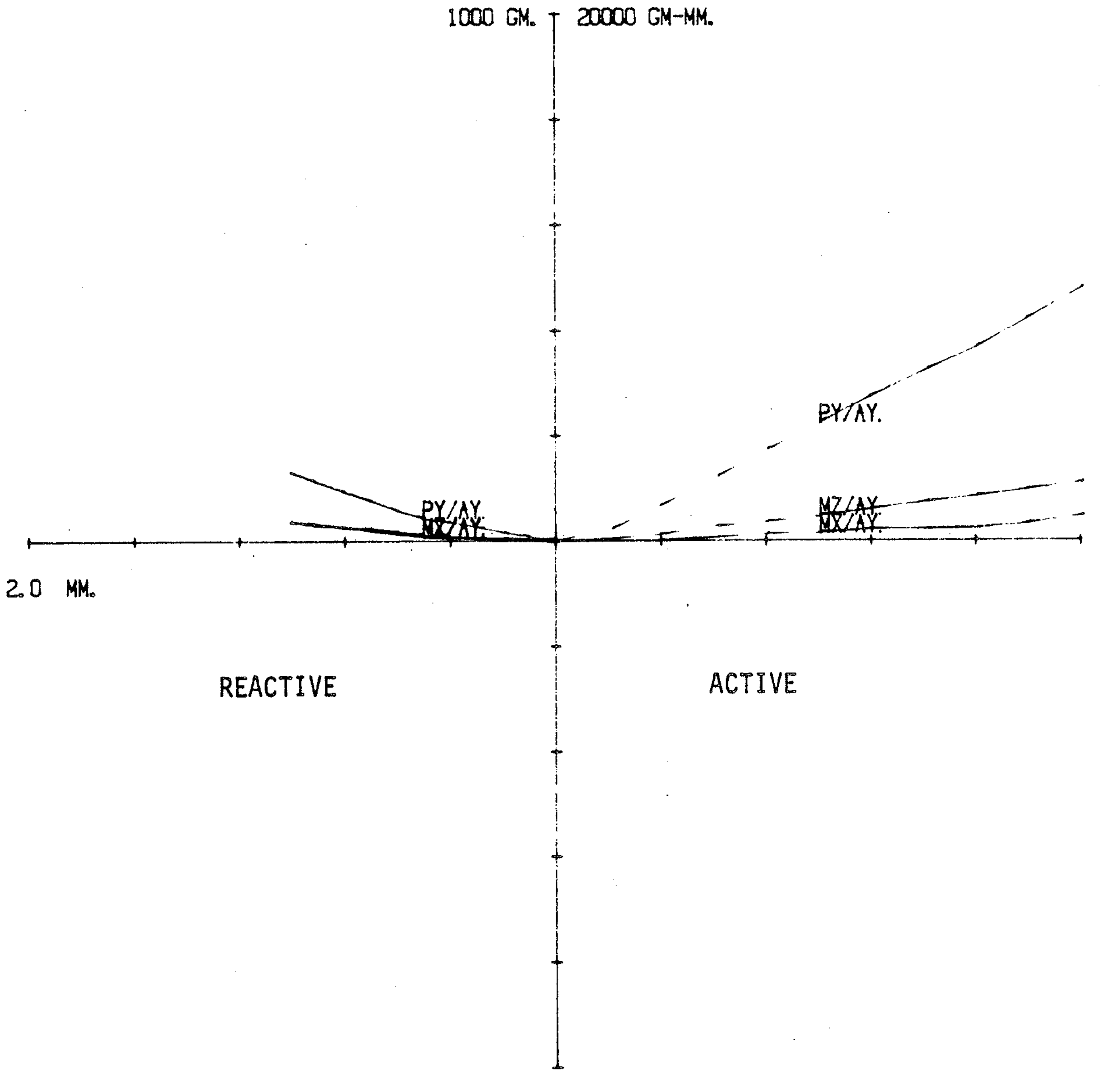


FIG. 11. L-LOOPS 10X7MM.
INTERBRACKET DISTANCE 10MM.
VERTICAL ACTIVATION.

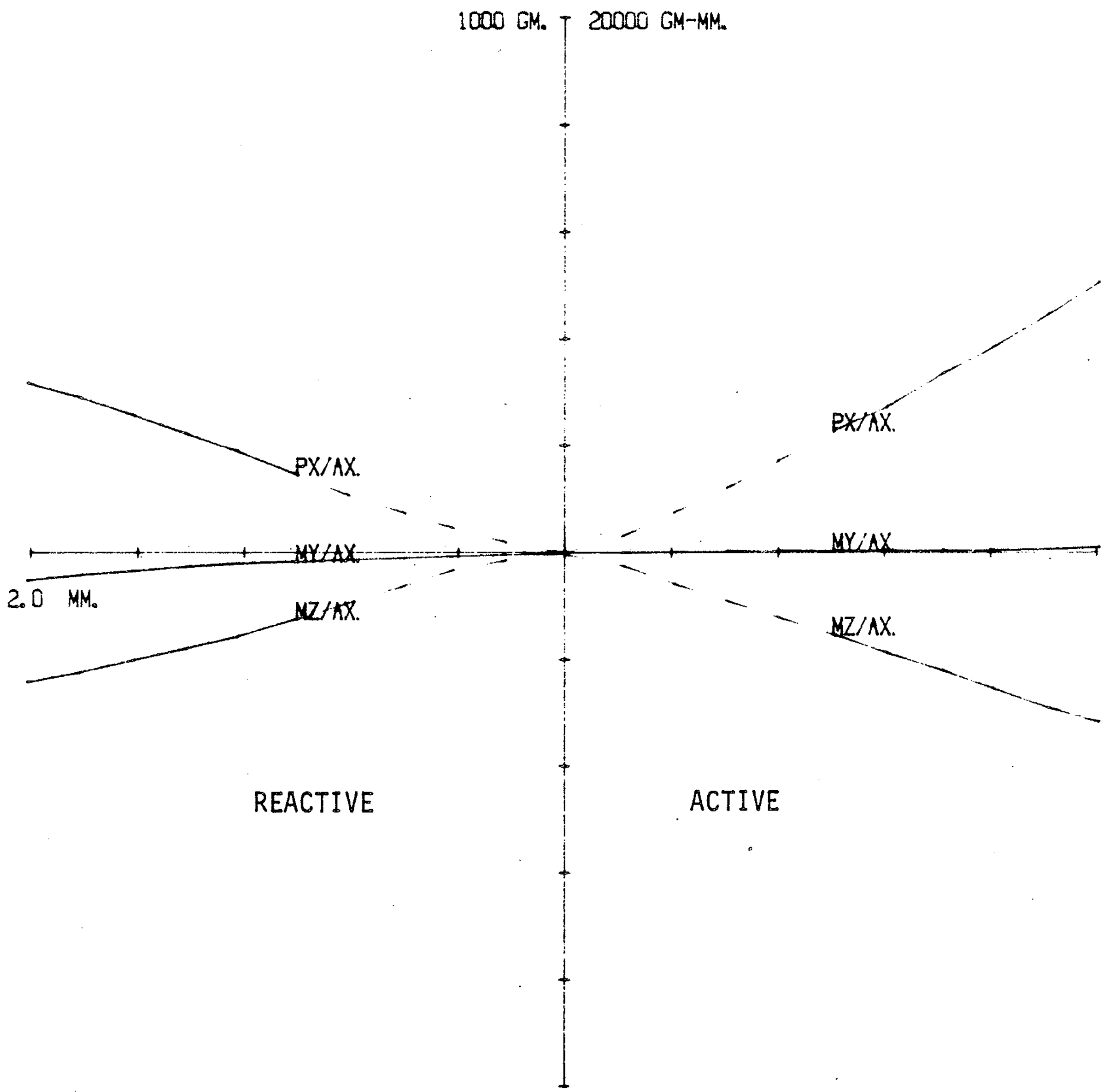


FIG. 12. VERTICAL LOOPS 10X1.5MM.
INTERBRACKET DISTANCE 5MM.
HORIZONTAL ACTIVATION.

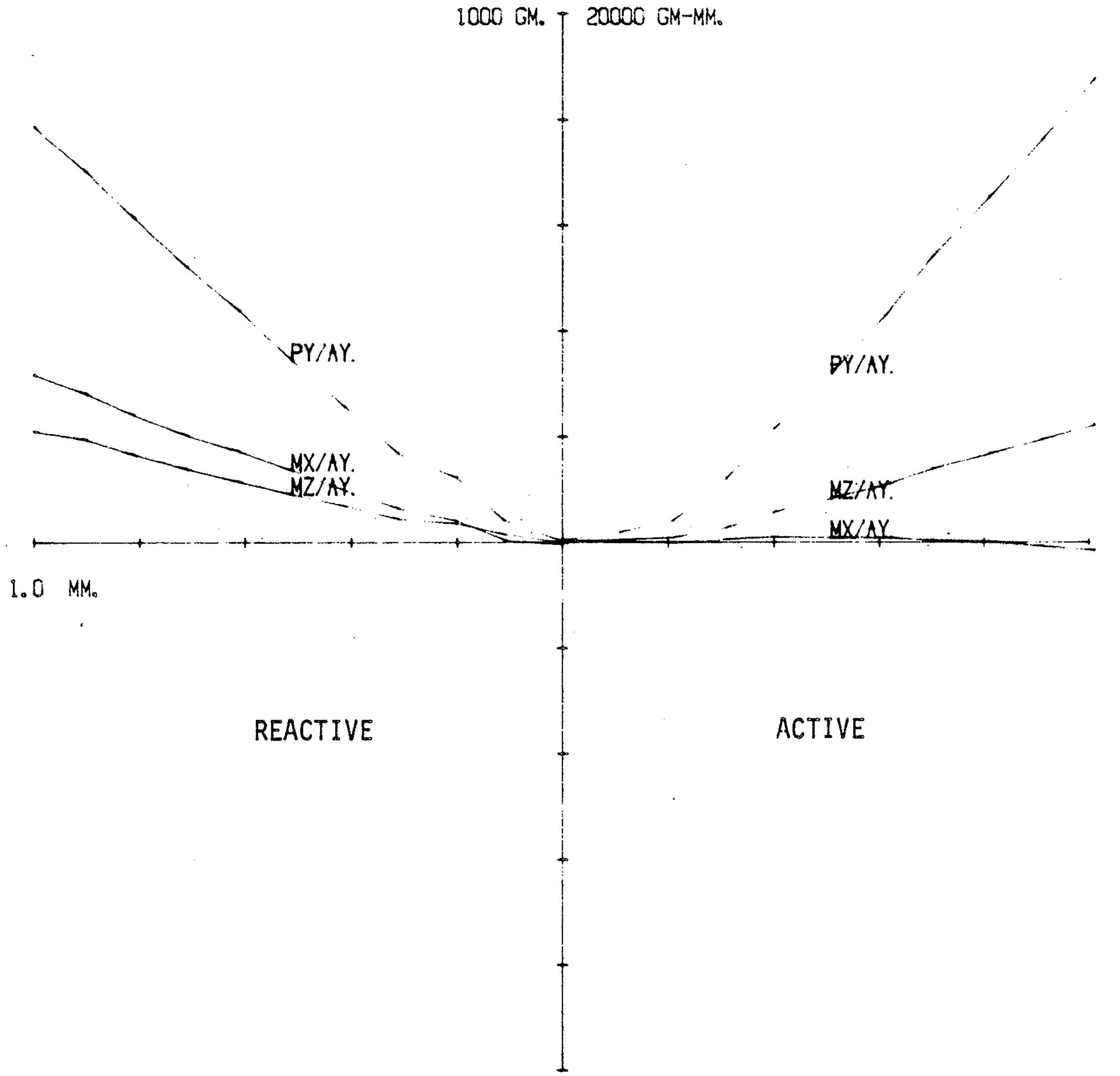


FIG. 13. VERTICAL LOOPS 10X1.5MM.
INTERBRACKET DISTANCE 5MM.
VERTICAL ACTIVATION.

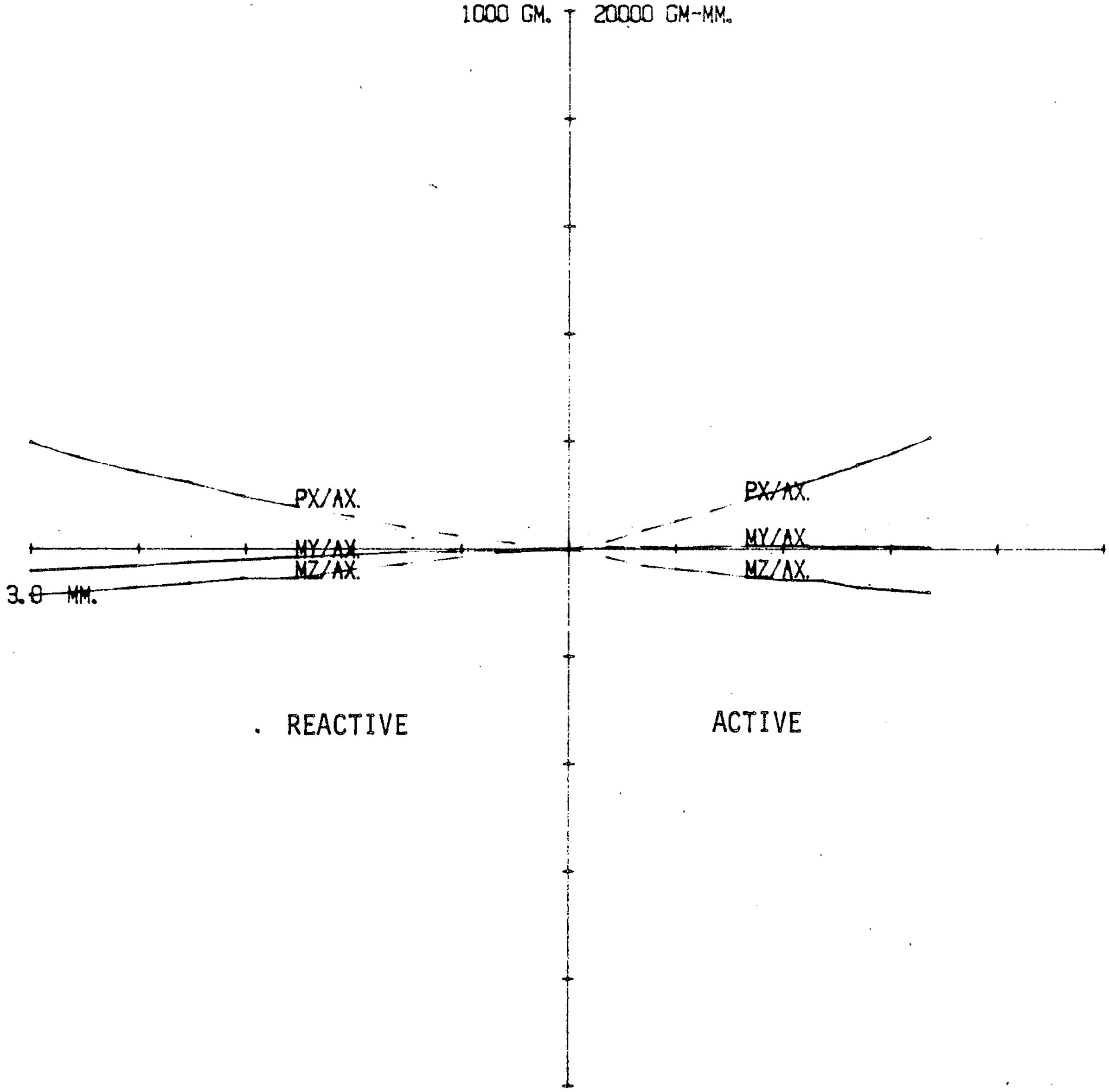


FIG. 14. L-LOOPS 10X7MM.
INTERBRACKET DISTANCE 5MM.
HORIZONTAL ACTIVATION.

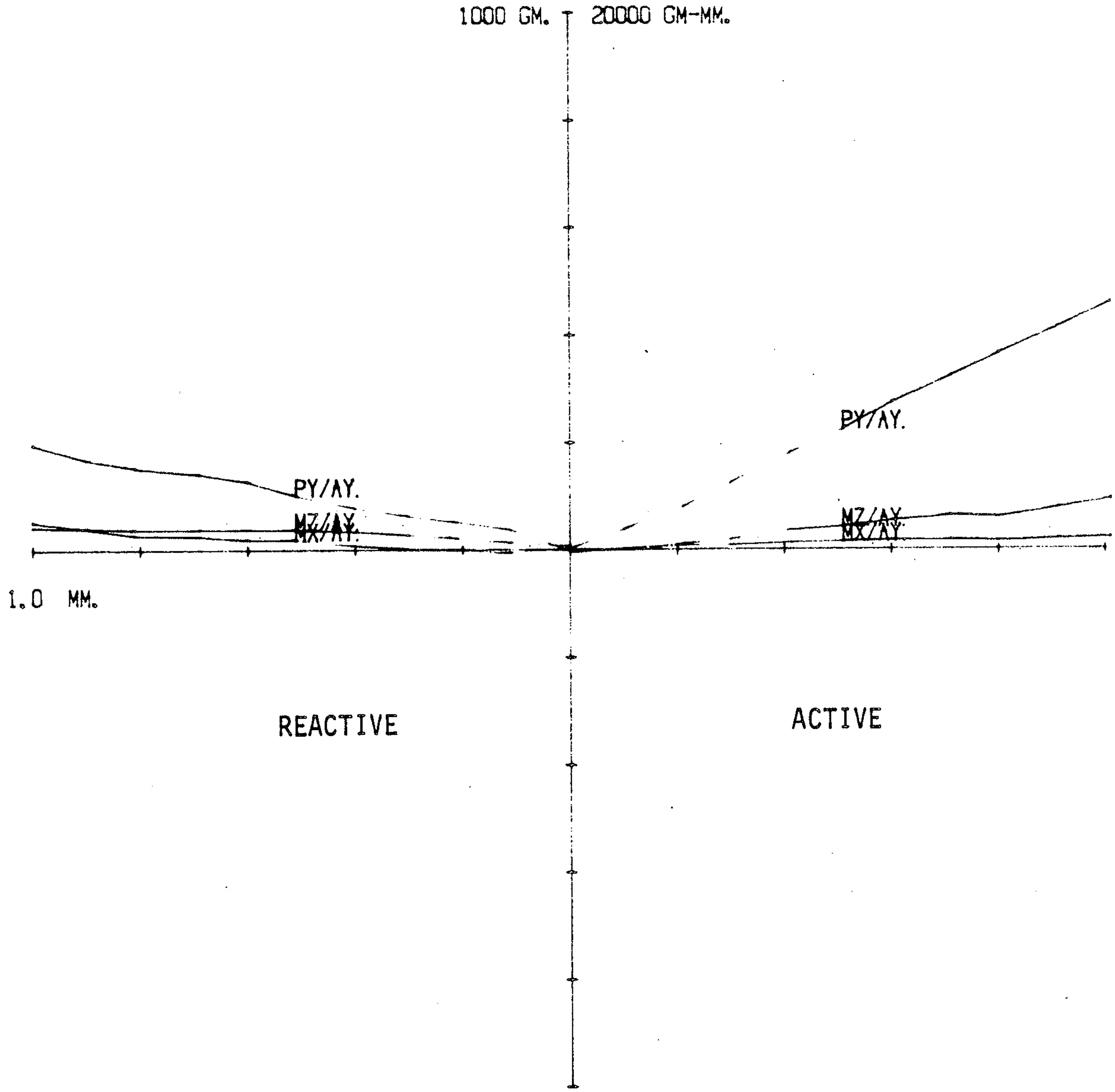


FIG. 15. L-LOOPS 10X7MM.
INTERBRACKET DISTANCE 5MM.
VERTICAL ACTIVATION.

Abreviations Used in Column Headings in Tables 1 to 8

Px	Force in the bucco-lingual direction in grams.
Py	Force in the occluso-gingival direction in grams.
Mz	Moment around the mesio-distal axis in gram-millimeters.
Mx	Moment around the bucco-lingual axis in gram-millimeters.
My	Moment around the occluso-gingival axis in gram-millimeters.
Mz/Px	Ratio between moment around bucco-lingual axis and bucco-lingual force in millimeters.
Mz/Py	Ratio between moment around bucco-lingual axis and occluso-gingival force in millimeters.
Tan θ	Tangent of angle over which cantilever x moves.
T/θ	Tortional stiffness of an appliance in gram-millimeters/radiant.
T'	Countermoment created by an appliance in gram-millimeters.
Mz_T	Total tipping moment around mesio-distal axis in gram-millimeters.
Mz_T/Px	Ratio between total tipping moment around mesio-distal axis and bucco-lingual force in millimeters.
$^{o}/_{o}$	Torsional stiffness of a looped wire or low-modulus wire in percentage of the stiffness of a straight stainless steel wire.

TYPE	SIZE	Px	Mz	Mz/Px	My	Stiffness
Vertical Loop	8 x 1.5 mm.	590	7000	11.9	0	300
		(280)	(3200)	(11.4)	(2200)	(140)
	10 x 1.5 mm.	380	4300	11.3	0	190
		(210)	(2200)	(10.2)	(1500)	(110)
	10 x 3 mm.	300	3300	11.1	0	150
		(170)	(1800)	(10.8)	(900)	(90)
	10 x 4.5	240	2600	10.8	0	120
		(130)	(1600)	(12.3)	(700)	(70)
L-loop	10 x 7 mm.	160	1600	10.0	0	80
		(80)	(900)	(11.3)	(0)	(40)
	10 x 12 mm.	100	1000	10.0	0	50
		(50)	(0)	(0)	(0)	(30)

TABLE 1. Horizontal activation Δx = 2 mm. Interbracket distance 10 mm. Numbers in brackets are forces and moments on the reactive tooth.

TYPE	SIZE	Py	Mz	Mz/Py	Mx	Stiffness
Vertical Loop	8 x 1.5 mm.	430	2300	5.4	0	430
		(300)	(1900)	(6.3)	(1500)	(300)
	10 x 1.5 mm.	420	2100	5.0	0	420
		(300)	(1600)	(5.3)	(1000)	(300)
	10 x 3 mm.	320	1800	5.6	0	320
		(280)	(1400)	(5.0)	(1700)	(280)
	10 x 4.5 mm.	280	1400	5.0	0	280
		(230)	(1400)	(6.1)	(1400)	(230)
L-loop	10 x 7 mm.	220	1100	5.0	0	220
		(120)	(800)	(6.7)	(0)	(120)
	10 x 12 mm.	120	900	7.5	0	120
		(100)	(0)	(0)	(0)	(100)

TABLE 2. Vertical activation Ay = 1 mm. Interbracket distance 10 mm.
Numbers in brackets are forces and moments on reactive tooth.

TYPE	SIZE	Px	Mz	Mz/Px	My	Stiffness
Vertical Loop	8 x 1.5 mm.	710	8700	12.3	0	360
		(350)	(3300)	(9.4)	(3000)	(180)
	10 x 1.5 mm.	500	6000	12.0	0	250
		(280)	(3500)	(12.5)	(1500)	(140)
	10 x 3 mm.	330	4400	13.3	0	170
		-	-	-	-	-
L-loop	10 x 7 mm.	210	1700	8.1	0	110
		(100)	(1000)	(10.0)	(1000)	(50)
	10 x 12 mm.	90	1000	11.1	0	50
		-	-	-	-	-

TABLE 3. Horizontal activation Ax = 2 mm. Interbracket distance 5 mm.
Numbers in brackets are forces and moments on the reactive tooth.

TYPE	SIZE	Py	Mz	Mz/Py	Mx	Stiffness
Vertical Loop	8 x 1.5 mm. *	440	2110	4.8	0	880
		(990)	(4700)	4.8	(3000)	(990)
	10 x 1.5 mm.	1000	5300	5.3	0	1000
		(900)	(5000)	5.6	(1600)	(900)
	10 x 3 mm.	830	4200	5.1	0	830
		-	-	-	-	-
L-loop	10 x 7 mm.	480	2200	4.8	0	480
		(260)	(1400)	5.4	(1000)	(260)
	10 x 12 mm.	230	1000	4.4	0	230
		-	-	-	-	-

TABLE 4. Vertical activation Ay = 1 mm.(* Activation only 0.5 millimeters). Interbracket distance 5 mm.

Numbers in brackets are forces and moments on the reactive tooth.

INTERBRACKET DISTANCE	TYPE	SIZE	TAN θ	T/θ	%
10 mm.	STRAIGHT		0.0385	8117	100
	Vertical Loop	8 x 1.5 mm.	0.0440	7102	87
		10 x 1.5 mm.	0.0453	6898	85
		10 x 3 mm.	0.0412	7585	93
	L-loop	10 x 1 mm.	0.0504	6200	76
		10 x 12 mm.	0.0592	5279	65
5 mm.	STRAIGHT		0.0226	13827	100
	Vertical Loop	8 x 1.5 mm.	0.0236	13243	96
		10 x 1.5 mm.	0.0279	11201	81
	L-loop	10 x 7 mm.	0.0349	8954	65

TABLE 5. Torsional stiffness of loops compared to that of straight wire.

INTERBRACKET DISTANCE	TYPE	TAN θ	T/θ	%
10 mm.	SS	0.0385	8117	100
	TMA	0.0886	3523	43
	NIT	0.1898	1636	20
5 mm.	SS	0.0226	13827	100
	TMA	0.0537	5809	42
	NIT	0.1111	2805	20

TABLE 6. Torsional stiffness of TMA and Nitinol wires compared to that of stainless steel wire.

INTERBRACKET DISTANCE	TYPE	SIZE	Mz	T/θ	T'	Mz_T	Px	Mz_T/Px
10 mm.	Vertical Loop	8 x 1.5 mm.	7000	7102	1207	5793	590	9.8
		10 x 1.5 mm.	4100	6898	1173	2927	380	7.7
		10 x 3 mm.	3100	7585	1289	1811	300	6.0
	L-loop	10 x 7 mm.	1600	6200	1054	546	150	3.6
		10 x 12 mm.	950	5279	897	53	100	0.5
5 mm.	Vertical Loop	8 x 1.5 mm.	8700	13242	2251	6449	710	9.1
		10 x 1.5 mm.	6000	11201	1904	4096	500	8.2
	L-loop	10 x 7 mm.	1700	8954	1522	178	210	0.9

TABLE 7. Total bucco-lingual tipping effect and moment to force ratio for looped wires.

INTERBRACKET DISTANCE	Mz	T/θ	T'	Mz_T	Px	Mz_T/Px
20 mm.	600	1000	85	515	50	10.3
15 mm.	1200	1300	110	1090	100	10.9
10 mm.	3300	1640	139	3161	260	12.2
5 mm.	10,000	2800	238	9762	750	13.0

TABLE 8. Total bucco-lingual tipping effect and moment to force ratio for straight 0.41x0.56 mm Nitinol wire.

Discussion

Introduction

Before the low-modulus archwires were introduced into orthodontics it was a commonly used technique to fabricate multilooped archwires from 0.336 millimeter (0.014 inch) or 0.406 millimeter (0.016 inch) stainless steel to achieve the initial alignment of malaligned teeth. As mentioned earlier this study is an extension on the work done by Sullivan (1982). The objective in this study was to compare the characteristics of the traditional multilooped archwires to those of the new low-modulus archwires in initial alignment which was the object of Sullivan's study.

The discussion of the results will deal with the following factors:

Load-Deflection Rate

Range

Force and Moment Control

Control of the Anchorage Area

Fabrication of Loops.

Load-Deflection Rate

When the initial alignment of severely malaligned teeth is attempted the archwire being engaged into the brackets has to be flexible enough so that the force levels are kept biologically acceptable and so that the elastic limit of the wire is not exceeded. The stiffness or the load deflection rate of an archwire between brackets on adjacent teeth can be reduced either by increasing the length of the wire or by using more flexible archwires. In the present investigation the length of the wire between the brackets was increased by increasing the interbracket distance and by incorporating loops into the archwire.

When a vertical loop is activated in the horizontal, bucco-lingual direction the vertical legs of the loops bend and torsion occurs at the apex of the loop. In an L-loop bending and torsion occur also in the horizontal legs and, hence, this part of the loop tends to act as a hinge. In both cases some bending occurs in the horizontal sections adjacent to the brackets. Increasing the length of any one of these components will reduce the stiffness. From Table 1 it can be seen that when the height of the vertical loops is increased from 8 millimeters to 10 millimeters the horizontal stiffness decreases 36 percent. It decreased from 300 to 190 grams/millimeter when interbracket distances were 10 millimeters. With the use of L-loops horizontal stiffness can be decreased even more. It can be seen in Table 1 that the horizontal stiffness of 10 x 12 millimeter L-loop (50 grams/millimeter) is only 27 percent of that of a 10 x 1.5 millimeter vertical loop (190 grams/millimeter). The length of the wire sections adjacent to the brackets was changed by using two different interbracket distances. It can be seen from Table 1 and 3 that when the interbracket distance increases from 5 millimeters to 10 millimeters the horizontal stiffness of the spans with 10 x 1.5 millimeter vertical loops decreases 24 percent (from 250 to 190 grams/millimeter).

When loops are activated in the vertical direction bending occurs in the horizontal and vertical components as well as the horizontal wire sections adjacent to the brackets (especially in the center part which acts directly on the malaligned tooth). The curvature of the loops is either opened or closed depending on the direction of the activation (occlusal or gingival). It can be seen from Table 2 that by widening the base of the 10 millimeter vertical loops from 1.5 to 3 millimeters the vertical stiffness can be decreased 23 percent (from 420 to 320 grams/millimeter). But only with the use of loops which have additional horizontal components can vertical stiffness be decreased to a clinically significantly extent. The stiffness of 10 x 12 millimeter L-loops (120 g/mm.) is only 28 percent of that of 10 x 1.5 millimeter vertical loops (420 g/mm.) when 10 millimeter interbracket distances are used. From Tables 2 and 4 it can be seen that when the interbracket distances are increased from 5 millimeters to 10 millimeters the vertical stiffness of the looped archwire is decreased 55 percent on average (e.g. from 1000 to 420 grams/millimeter for 10 x 1.5 millimeter vertical loops).

Due to the geometry of the loops one would expect them to behave differently when they are deflected into the occlusal or gingival direction and the curvature of the apex is either opened or closed. It is believed that because of residual stresses a loop is stiffer if it is activated into the direction of the original bending. In this work no differences could be shown in the force levels between the different directions of the activation.

The testing of straight lengths of 0.41 x 0.56 millimeter

Nitinol wire shows that, as expected from beam theory, the changes in interbracket distance has a large effect on the load-deflection rates in straight wire therapy. It can be seen from Table 8 that the 5 millimeter spans are more than three times as stiff as 10 millimeter spans and 15 times as stiff as 20 millimeter spans. (According to the beam theory the stiffness of a beam is inversely proportional to the cube of its length. Thus a 5 millimeter span should be 8 times as stiff as a 10 millimeter span. The difference here can be explained by the difference in supporting the beam: in these experiments wires were not firmly attached to the slots but they were free to slide through them.) The values of horizontal stiffness range from 50 to 750 grams/millimeter activation for different interbracket distances. It can be seen that with clinical interbracket distances straight rectangular 0.41 x 0.56 Nitinol wires would create excessive forces if they were used in initial alignment of the malaligned teeth. From Tables 3 and 8 it can be seen that with 5 millimeter interbracket distances the horizontal stiffness of the stiffest 8 x 1.5 millimeter vertical loops (360 grams/millimeter) was only one half of that of a straight Nitinol wire (750 grams/millimeter).

Range

Sullivan (1982) showed that when round low-modulus archwires are deflected through a distance of 2 millimeters the elastic limit of the material is exceeded in most cases. It is not an unusual clinical situation that a malaligned tooth is 2 millimeters or more out of alignment. Sullivan suggested that if straight low-modulus archwires are used to achieve the initial alignment a complete bracket engagement should not be attempted at the outset if the irregularity exceeds 1.5 millimeters.

It was shown in this investigation that the risk of exceeding the range is not very big when looped archwires are used. In spite of the use of the fairly rigid rectangular stainless steel wires no permanent measurable deformation occurred in the loops after they were deflected through a distance of 3 millimeters.

When the looped archwires are used the range is not as limited by the elastic limit as much as it is by the geometry of the loops. When the loops are activated the different parts of them should not interfere otherwise the force systems created are changed significantly. Obviously, the loops should not interfere with the oral tissues at any stage of activation or deactivation.

Force and Moment Control

When bucco-lingual or occluso-gingival forces are applied to the bracket a bucco-lingual tipping moment, Mz, is created at the center of resistance of the tooth. The magnitude of this moment is the product of the force applied and the perpendicular distance between the bracket and the center of resistance of the tooth. The ratio between the tipping moment and the force, the moment to force ratio, Mz/Px or Mz/Py, describes the effectiveness of the appliance to control the bucco-lingual tipping. By controlling the moment to force ratio the desired amount of tipping of the tooth can be be achieved. In order to achieve bodily movement of the tooth the moment to force ratio should be zero.

With the measuring arrangement used in this study one would expect the moment to force ratio to be 12.5 for horizontal and 4.2 for vertical activations if the appliance tested provided no control over bucco-lingual tipping. According to Sullivan's (1982) results and those of the present study, however, the moment to force ratios are even higher when continuous round archwires are used. The average moment to force ratio for bucco-lingual deflection using round wires was 15.7 (range from 14.9 to 16.7) (Sullivan, 1982) and for occluso-gingival deflection 5.5 (range from 4.4 to 6.3) (Tables B1, 2, 3 and 4 in Appendix B).

When full size rectangular archwires are used in rectangular brackets the torsional stiffness of the wire creates a countermoment which resists the bucco-lingual tipping of the tooth generated by the forces applied at the bracket. In order to keep the tooth upright the countermoment has to equal the tipping moment. Therefore an ideal archwire for initial alignment of malaligned teeth should have a relatively high torsional stiffness compared to its bending stiffness. The ratio between bending stiffness and torsional stiffness of a straight archwire span can be altered by changing the length of the span or the cross-section of the wire. The torsional stiffness of a straight wire is inversely proportional to its length whereas the bending stiffness is inversely proportional to the cube of its length. Therefore as the length of a span increases the torsional stiffness decreases less than the bending stiffness and so the control over bucco-lingual tipping improves.

To test the ability of a straight rectangular wire to control bucco-lingual tipping 0.41 x 0.56 millimeter Nitinol wire was used. Interbracket distance was increased from 5 millimeters to 20 millimeters in steps of 5 millimeters. The measurement of torsional stiffness, T/θ, was combined with the bucco-lingual tipping moment obtained with the measuring instrument using the method described in the Materials and Methods. The results in Table 8 show that when the interbracket distance increases from 5 millimeters to 20 millimeters the moment to force ratio decreases from 13.0 to 10.3.

It can be seen that with straight wires the control over bucco-lingual tipping cannot be significantly improved by increasing interbracket distances because the moment to force ratio does not reach zero until the length of the wire gets much longer than clinical interbracket distances. If the tipping angle of the tooth and the cross-section of the wire are known, it is possible to calculate the "interbracket distances" required to keep the tooth upright. The calculations for 0.41 x 0.56 millimeter stainless steel wire are shown in Appendix C.

According to the calculations the theoretical "interbracket distances" would have to be 107 millimeters if the wire is used edgewise and 74 millimeters if the wire is used ribbonwise. Hence, the increase in interbracket distances has no practical value for moment control in straight wire therapy. A small improvement can be achieved by using the wire ribbonwise instead of using it edgewise since its bending stiffness is reduced by 30 percent without changing its torsional stiffness.

Altering the wire material (for example from stainless steel to Nitinol) does not change the moment to force ratio because torsional stiffness and bending stiffness depend on the same material properties. (The ratio between elastic modulus and shear modulus is the same for most metals at 1.6). It can be seen in the last column of Table 6 where torsional stiffness of beta-titanium (TMA) and Nitinol wires are presented as percentages of that of stainless steel that their torsional stiffnesses are related in the same way as their elastic moduli.

When additional wire is incorporated between the brackets by using loops the countermoment created by the rectangular wire becomes clinically significant. The torsional stiffness of the looped wires was tested with 5 millimeter and 10 millimeter interbracket distances and the results are shown in Tables 5 and 7. The moment to force ratio for 8 x 1.5 millimeter vertical loops with 10 millimeter interbracket distances is 9.8 which is slightly lower than the 10.3 given by 20 millimeter straight Nitinol spans (Table 8). The moment to force ratio for 10 x 1.5 millimeter loops is 7.7 and it decreases as the length of wire in the loops increases reaching 0.5 for 10 x 12 millimeter L-loop.

The improvement of control over bucco-lingual tipping with looped archwires is primarily due to the geometry of the loops. When a vertical loop is deflected in the bucco-lingual direction the vertical components act as levers and the apex, which undergoes torsion, acts as a "hinge". Thus, as the length of the loop increases the perpendicular distance between the center of resistance of the tooth and the effective point of force application decreases. It can be seen from Table 7 that increase in the length of a vertical loop improves the moment to force ratio but that in this respect L-loops are superior to vertical loops. In an L-loop additional wire is incorporated into horizontal components which undergo torsion. The "hinge" in the apex of the loop is perpendicular to the horizontal component which acts directly on the bracket of the malaligned tooth. Therefore L-loops decrease torsional stiffness much less than bending stiffness.

It can be seen from Table 7 that if 10 x 7 millimeter L-loops are used instead of 10 x 1.5 millimeter vertical loops bending stiffness is reduced by 60 percent whereas torsional stiffness is reduced only by 20 percent. When the results with 10 millimeter interbracket distances are compared to those with 5 millimeter interbracket distances (Table 7), it can be seen that when looped archwires are used the effect of interbracket distances on moment to force ratios, Mz/Px, is small.

Due to the geometry of the loops small gingivally directed forces are created when they are activated in the bucco-lingual direction. Theoretically, these gingival forces increase the buccal tipping when the loops are deflected buccally and decrease the lingual tipping when loops are deflected lingually. The use of L-loops which provide flexibility in the vertical plane should decrease these unwanted vertical forces. This was not clearly shown in these experiments.

In order to align the malaligned teeth straight archwires have to slide through the brackets. Due to the friction between the archwires and the brackets unpredictable lateral, i.e. mesio-distal forces are created. It can be seen from Tables B3 and B4 in Appendix B that if the interbracket distances are not equal on either side lateral wedging forces, Pz, are created. The result of this study (Figures 8 and 15 and Appendix A) show that when looped archwires are used and the loops are bent symmetrically no measureable lateral forces are created because looped archwires, when activated and deactivated, do not slide through the brackets to a significant extent.

Control of the Anchorage Area

The results from Sullivan's study (1982) and those presented in Appendix B indicate that when straight low-modulus archwires are tied into malaligned teeth the magnitude of the forces and moments applied on the reactive teeth which are supposed to serve as anchorage are about one half of those applied on the active teeth which are to be moved. Obviously, deactivation of such a force system would cause not only the active teeth to move in the desired direction but the reactive teeth to move in the opposite direction.

In the present form, the measuring instrument does not allow movement of teeth in response to an applied force system. It represents the situation immediately after the activation of the wire. Consequently, the results obtained in this study show that, the use of looped archwires does not decrease the magnitudes of forces and moments acting on anchorage areas relative to those acting on active areas. It can be seen from Tables 1 and 3 that when the loops are deflected in the bucco-lingual direction forces and moments applied on reactive teeth are about one half of those applied on active teeth. Tables 2 and 4 show that when loops are deflected into an occluso-gingival direction forces and moments applied on reactive teeth are about three quarters of those applied on active teeth.

It is generally assumed that, within limits, the rate of tooth movement is a direct function of the stress applied to the periodontal ligament. It is also suggested that some forms of tooth movement require less force than others. (For example, tipping requires less force than translation.) The concept of differential force (Begg, 1956) is based on the idea that several teeth are used as an anchor unit against which forces are developed to move the malaligned tooth. For such a concept to be viable the teeth in the anchor unit must be mechanically linked so that the stresses applied to the periodontal ligaments are as equal as possible and that tipping and rotation of individual teeth in the unit are avoided. This means that inter-tooth coupling in the anchor unit should be made as rigid as possible.

The use of straight low-modulus archwires does not provide for controlled changes in stiffness around the arch. The use of multilooped archwires which are fabricated from full-dimension rectangular wires allows for an increase in flexibility in the active areas while maximum rigidity is maintained in the anchor areas. For example, the measured stiffness of a pair of 10 x 1.5 millimeter vertical loops fabricated from 0.41 x 0.56 millimeter stainless steel wire is 300 grams/millimeter when they are deflected into bucco-lingual direction, whereas the stiffness of a straight wire at the same 10 millimeter interbracket distance is 3030 grams/millimeter. (The stiffness of a straight stainless steel wire can be calculated from the results in the penultimate column of Table 8 and the modular ratio, 5, of steel to Nitinol.) Thus, it may be seen that a straight wire that connects the anchor teeth can be at least ten times as stiff as loops that apply force to a malaligned tooth.

As more rigid wires are used between brackets the practical effects of the tightness of fit between the wire and bracket become more significant and therefore the calculation of the coupling stiffness between teeth is of limited value. Consequently, experiments are needed to quantify the major parameters affecting inter-tooth stiffness in anchor units. Such experiments were beyond the scope of the present work but some idea of the importance of this effect can be gained from elementary beam theory assuming a tight fit between the wire and bracket. If buckling of straight wires is ignored leaving bending and torsion as the only contributors to flexibility between teeth, it is clear that two factors are important in coupling stiffness: (1) the interbracket distance and (2) the wire stiffness number (Burstone, 1981). The wire stiffness number is calculated from the product of the elastic modulus and the second moment of the cross-sectional area of a wire.

Bending stiffness is inversely proportional to the cube of the interbracket distance which is determined by the dentition and bracket width. It is worth noting that this could lead to a situation in which large teeth form a poorer anchor unit than small teeth unless wider brackets are used.

The importance of wire stiffness number in maintaining unitary behaviour of anchor units can be demonstrated by assuming that the load is applied to the anchor tooth adjacent to the malaligned tooth. The amount of support this anchor tooth receives from the next anchor tooth is proportional to the relative movement between these two teeth. For good anchorage, the objective is to maximize the support force for any given displacement. Under the limiting conditions described earlier this support is proportional to the wire stiffness number, Ws, given in Table 9. The last column, "relative movement", indicates the amount of movement of the first anchor tooth which is required before the low modulus archwires create the same amount of support force as an 0.41 x 0.56 millimeter stainless steel wire creates after the tooth moves over a distance of one unit. From Table 9 it may be concluded that if support against tipping is needed an 0.41 millimeter round Nitinol wire allows the first anchor tooth to move lingually 15 times as much as an 0.41 x 0.56 millimeter rectangular stainless steel wire for a given level of support force from the next anchor tooth. If rotational support around the occluso-gingival axis is needed an 0.41 millimeter Nitinol wire again allows the first anchor tooth to rotate 15 times as much as an 0.41 x 0.56 millimeter rectangular stainless steel wire for a given level of support force from the next anchor tooth.

The exact amount of tooth movement in clinical situations

WIRE	SIZE (mm)	Wire Stiffness Number, Ws	Relative Movement
Tri-Flex	0.38	0.02	22
	0.44	0.03	15
Nitinol	0.41	0.03	15
	0.45	0.06	7
T M A	0.39	0.06	7
Steel	0.41x0.56	0.44	1

Table 9. The movement of the anchor teeth relative to the stiffness of the archwire.

cannot be predicted from these calculations but they do strongly indicate the importance of inter-tooth coupling in anchor units.

Fabrication of Loops

Significant unwanted effects showed up due to the small errors in loop construction. The loops were bent to a template drawn on millimeter paper and great care was taken to produce loops that were accurate in all three planes of space. In addition, a small sample of experienced clinicians were invited to construct sample loops but the results using these loops were not significantly different from those reported here.

These bending errors produced asymmetry in the forces generated by positive and negative activations. The undesirable side effects caused by "bending errors" turned out to be clinically significant especially when vertical loops were used. Since the vertical loops provide virtually no flexibility in the vertical plane very small misalignments of the bracket sections can cause significant unwanted vertical forces. These forces will extrude, intrude and tip the teeth when only bucco-lingual movement is required. The problem becomes less pronounced when L-loops are used. Due to the flexibility in the vertical plane the forces created by small misalignments of bracket sections are less significant. It is also much easier to adjust L-loops in the vertical plane after they are fabricated.

When the loops are bent into rectangular archwires unwanted torque is easily built in. With flat Tweed pliers it is easier to control the bending procedure in the third order direction than with conical bird beak pliers. When helical loops and sometimes L-loops are contoured the wire has to cross over itself and it is extremely difficult to avoid building torque into the apices of the loops. Unwanted torque can obviously cause significant tipping of the malaligned or of the anchorage teeth.

Conclusions and Recommendations

The purpose of this investigation was to compare the traditional multilooped archwires to the relatively new low-modulus archwires in the initial alignment of malaligned teeth. Instrumentation designed by Paquien (1978) was used.

From the data obtained and analyzed it was evident that incorporating loops into archwires is an effective means of decreasing the stiffness of an archwire span between the brackets on the adjacent teeth.

The use of multilooped archwires allows flexibility to be built into the archwire in the active areas where tooth movement is wanted while maintaining rigidity in the anchorage areas. Thus the reactive forces even though they are almost as high as the active forces are distributed over several teeth and better control over the final tooth position may be achieved.

In this investigation the looped archwires were fabricated of 0.41 x 0.56 millimeters (0.016 x 0.022 inch) stainless steel wire. By fabricating them of the same dimension beta titanium wire the stiffness could be reduced by an additional 57 percent. The stiffness of a Nitinol wire is only 20 percent of that of the same dimension stainless steel wire but the brittleness of Nitinol wires makes it impossible to contour loops in them. The disadvantage of the use of more flexible wires is, however, that they allow for more movement of the anchorage teeth.

It is suggested that by the use of rectangular archwires in initial alignment of malaligned teeth it is possible to move the teeth bodily without tipping them in a bucco-lingual direction. When straight lengths of 0.41 x 0.56 millimeter Nitinol wires were tested it was shown that as the interbracket distances increased the control over the bucco-lingual tipping improved but no significant improvement could be gained using clinical interbracket distances. When loops were incorporated into the rectangular archwire there was a remarkable improvement in the uprighting effect of the countermoment created by the appliance. By bending an L-loop on either side of the malaligned tooth the ratio between bucco-lingual tipping moment and bucco-lingual force could be reduced to 0.5 which can be considered almost ideal for bodily movement of the tooth. When looped arches are used interbracket distances are not very significant for force and moment control and wider brackets can be used to achieve better control over second and third order movements of teeth.

The accurate fabrication of loops was a problem. Small misalignments of the horizontal bracket sections can create significant unwanted forces and moments on both active and anchorage teeth. The problem is likely to be more real in a clinical situation when less time can be spent on contouring loops and when bracket positioning on teeth may not be ideal.

L-loops are superior to vertical loops for two reasons: (1) they provide more flexibility and better control over bucco-lingual tipping and (2) the forces they generate are less affected by small bending errors. The common belief that loops are stiffer when they are activated into the direction of original bending was not supported by this study.

The disadvantage of multilooped archwires is that they are fairly time consuming to fabricate. They cannot be preformed. Patient comfort sets limitations to the size and configuration of loops. When appliances are made more complicated they become more difficult to keep clean. Therefore multilooped arches may not be suitable for patients whose oral hygiene is poor.

It can be concluded from the present study and that of Sullivan (1982) that forces and moments to correct severely malaligned teeth should be delivered by looped archwires rather than using straight low-modulus archwires. Multilooped archwires provide a much better means of controlling the magnitude and the direction of forces placed on teeth especially if they are fabricated of rectangular wires. Much greater concern should be shown for the anchorage areas to gain better control over the final tooth position.

Future Research Recommendations

The following recommendations and suggestions can be made for future research:

(1) An investigation of more complex misalignments, for example, rotations of the malaligned teeth and the forces and moments generated when pre-torqued and pre-angulated brackets are used.

(2) An investigation into the effects of different ligation methods on wire performance.

(3) An investigation of other loop configurations, for example, box loops and helical loops.

(4) An investigation into the effects of friction between archwires and brackets.

(5) An investigation into the effects of inter-tooth coupling in the anchor units.

BIBLIOGRAPHY

Ackerman, J.L., Sager, C.R., Del Priore, R.C. and Bramante, M.A. (1969) A controlled light continuous force technique. Am. J. Orthod. 56:233-251.

Andreason, G. and Johnson, P. (1967) Experimental findings on tooth movements under two conditions of applied force. Angle Orthod. 37:9-12.

Begg, P.R. (1956) Differential force in orthodontic treatment. Am. J. Orthod. 42:482-510.

Burstone, C.J. (1961) The application of continuous forces to orthodontics. Angle Orthod. 31:1-14.

Burstone, C.J. (1962) Rationale of the segmented arch. Am. J. Orthod. 48:805-822.

Burstone, C.J. (1966) The mechanics of the segmented arch techniques. Angle Orthod. 36:99-120.

Burstone, C.J. (1981) Variable-modulus orthodontics. Am. J. Orthod. 80:1-16.

Burstone, C.J. and Koenig, H.A. (1974) Force systems from an ideal arch. Am. J. Orthod. 65:270-289.

Carey, C.W. (1956) Laminated arches - the double ribbon and double edgewise. Am. J. Orthod. 42:47-53.

Creekmore, T.D. (1976) The importance of interbracket width in orthodontic tooth movement. J. Clin. Orthod. 10:530-534.

Haack, D.C (1963) The geometry and mechanics as related to tooth movement studied by means of two dimensional models. J. Am. Dental Assoc. 66:157-164.

Halderson, H., Johns, E.E. and Moyers, R. (1953) The selection of forces for tooth movement. Am. J. Orthod. 39:25-35.

Hixon, E.H., Atikian, H., Callow, G.E., McDonald, H.W. and Tacy, R.J. (1969) Optimal force, differential force, and anchorage. Am. J. Orthod. 55:437-457.

Hixon, E.H., Aasen, T.O., Arango, J., Clark, R.A., Klostermean, R., Miller, S.S. and Odum, W.M. (1970) On force and tooth movement. Am. J. Orthod. 57:476-489.

Jarabak, J.R. and Fizzel, J.A. (1972) Technique and Treatment with Light-Wire Edgewise Appliances. Saint Louis, C.V. Mosby Co. Chapters 1, 2, 3, 7.

Johns, E. (1953) A study of forces inherent in orthodontic appliances and the selection of forces for tooth movement. J. Can. Dent. Assoc. 19: 423-436.

Lack, M.L. (1980) An investigation into the three-dimensional force and moment characteristics of selected cuspid retraction mechanisms. MSc. Thesis, University of Manitoba.

McLachlan, K.R. (1983) Personal communication.

Meador, D.K. (1982) An investigation of the three dimensional force and moment design parameters of selected orthodontic alignment loops. MSc. Thesis, University of Manitoba.

Moyers, R.E. and Bauer, J.L. (1950) The periodontal response to various tooth movements. Am. J. Orthod. 36:572-580.

Paquien, J.P. (1978) The measurement of forces and moments delivered by dental appliances. MSc. Thesis, University of Manitoba.

Reitan, K. (1957) Some factors determining the evaluation of forces in orthdontics. Am. J. Orthod. 43:32-45.

Schwartz, A.M. (1932) Tissue changes incidental to orthdontic tooth movement. Int. J. Orthod. 18:331-352.

Steiner, C.C. (1953) Power storage and delivery in orthodontic appliances. Am. J. Orthod. 39:859-890.

Stoner, M.M. (1960) Force control in clinical practice. Am. J. Orthod. 46:163-186.

Sullivan, J.S. (1982) An investigation into the three dimensional force and moment of selected low-modulus initial alignment archwires. MSc. Thesis, University of Manitoba.

Timoshenko, S. (1940) Elements of Strength of Materials. 2nd Edition, Lancaster, Pa., Van Nostrand Co., Chapter 9.

Thurow, R.C. (1982) Edgewise Orthodontics. 4th Edition, Saint Louis, C.V. Mosby Co., Chapter 1-12.

Warnock, F.V. and Benham, P.P. (1965) Mechanics of Solids and Strength of Materials. 3rd Edition, London, Pitman Co., Chapter 3.

Waters, N.E., Houston, W.J.B. and Stephens, C.D. (1981) The characterization of arch wires for the initial alignment of irregular teeth. Am. J. Orthod. 79:373-389.

Waters, N.E., Stephens, C.D. and Houston, W.J.B. (1975) Physical characteristics of orthodontic arch wires: Part II. Br. J. Orthod. 2:73-83.

Weinstein, S. (1967) Minimal forces in tooth movement. Am. J. Orthod.
58:881-903.

Appendix A

Additional Plots of Results of Tests on Loops at

10 Millimeters Interbracket Distances

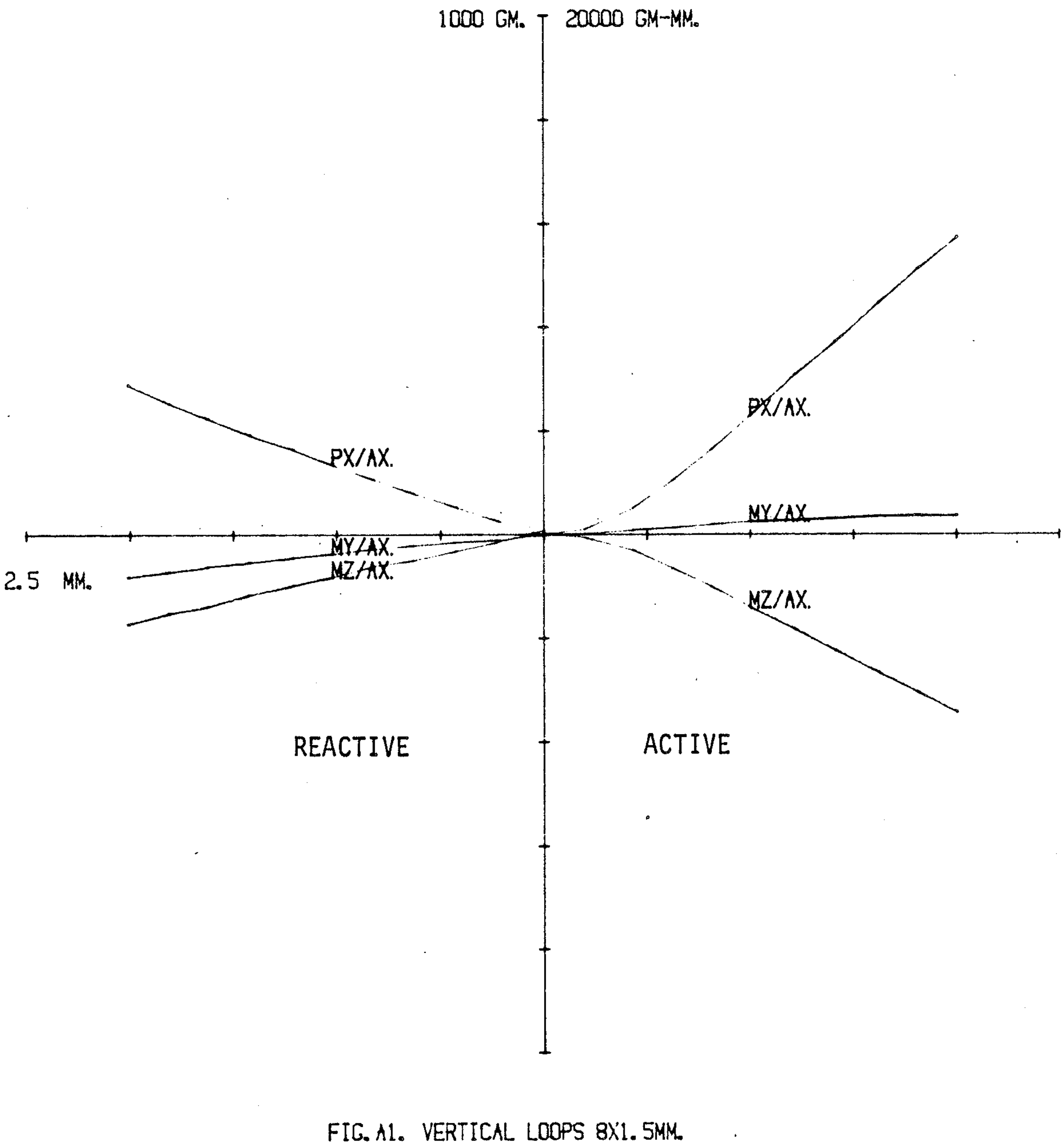


FIG. A1. VERTICAL LOOPS 8X1.5MM.
INTERBRACKET DISTANCE 10MM.
HORIZONTAL ACTIVATION.

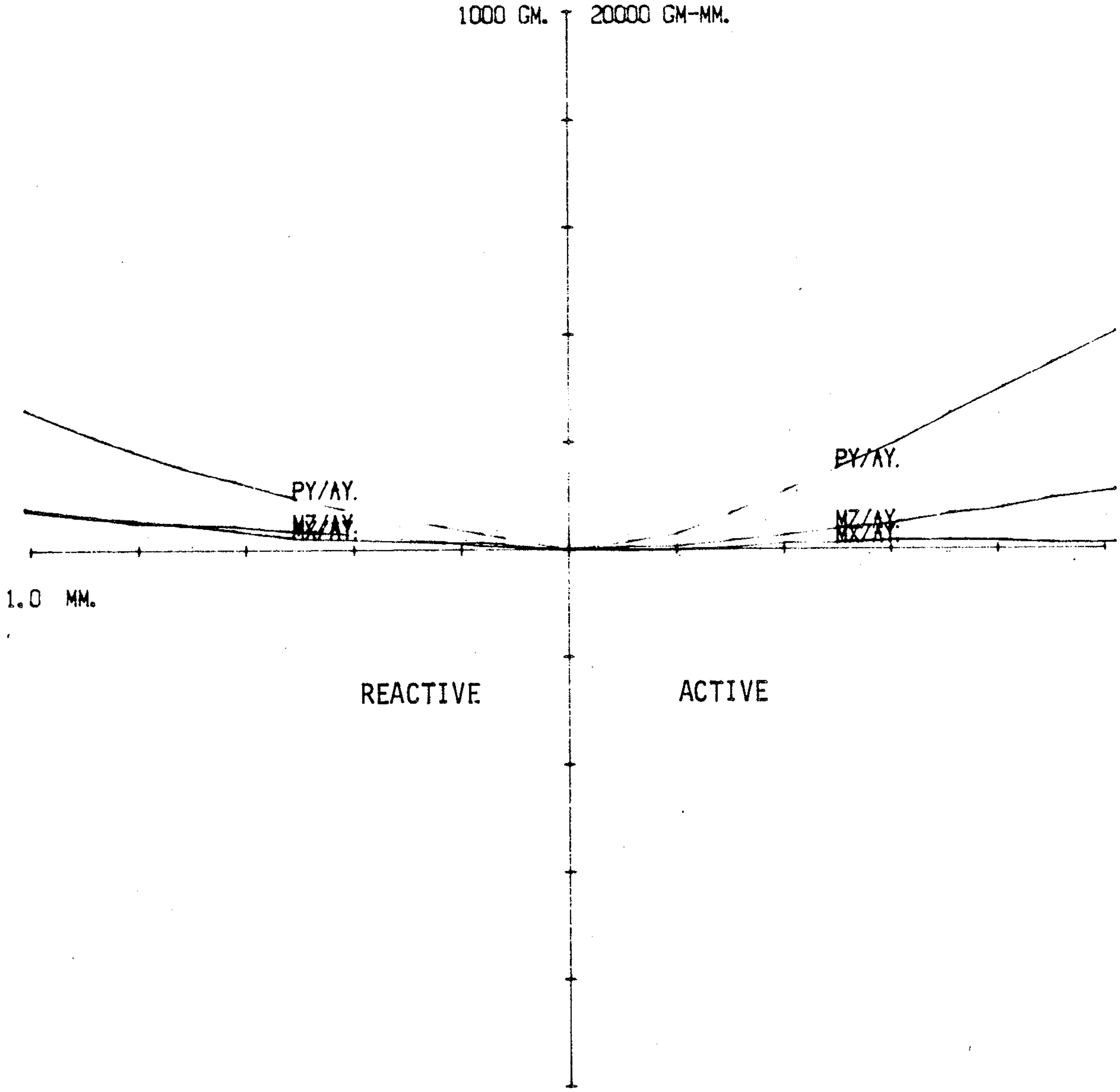


FIG. A2. VERTICAL LOOPS 8X1.5MM.
INTERBRACKET DISTANCE 10MM.
VERTICAL ACTIVATION.

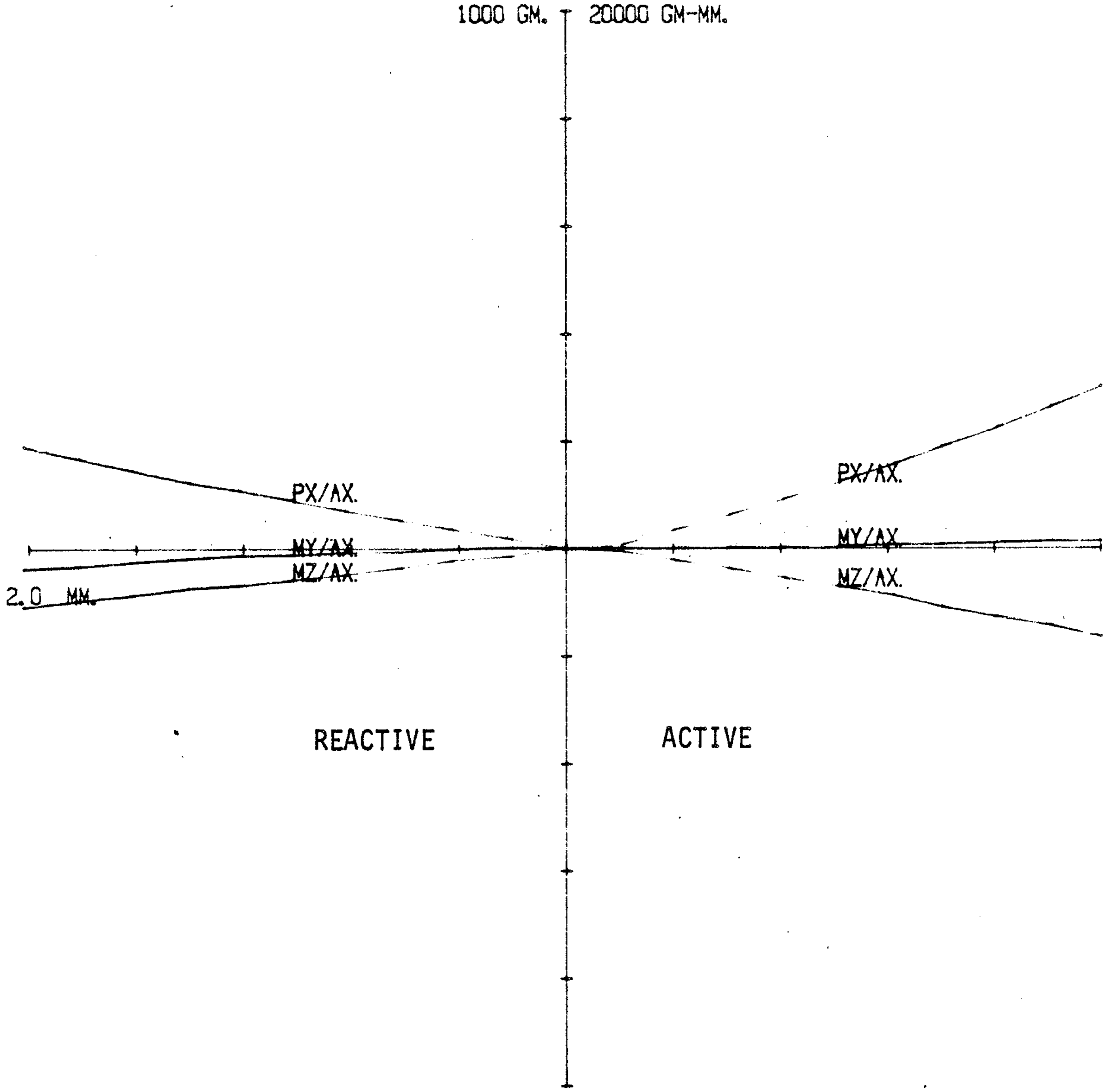


FIG. A3. VERTICAL LOOPS 10X3MM.
INTERBRACKET DISTANCE 10MM.
HORIZONTAL ACTIVATION.

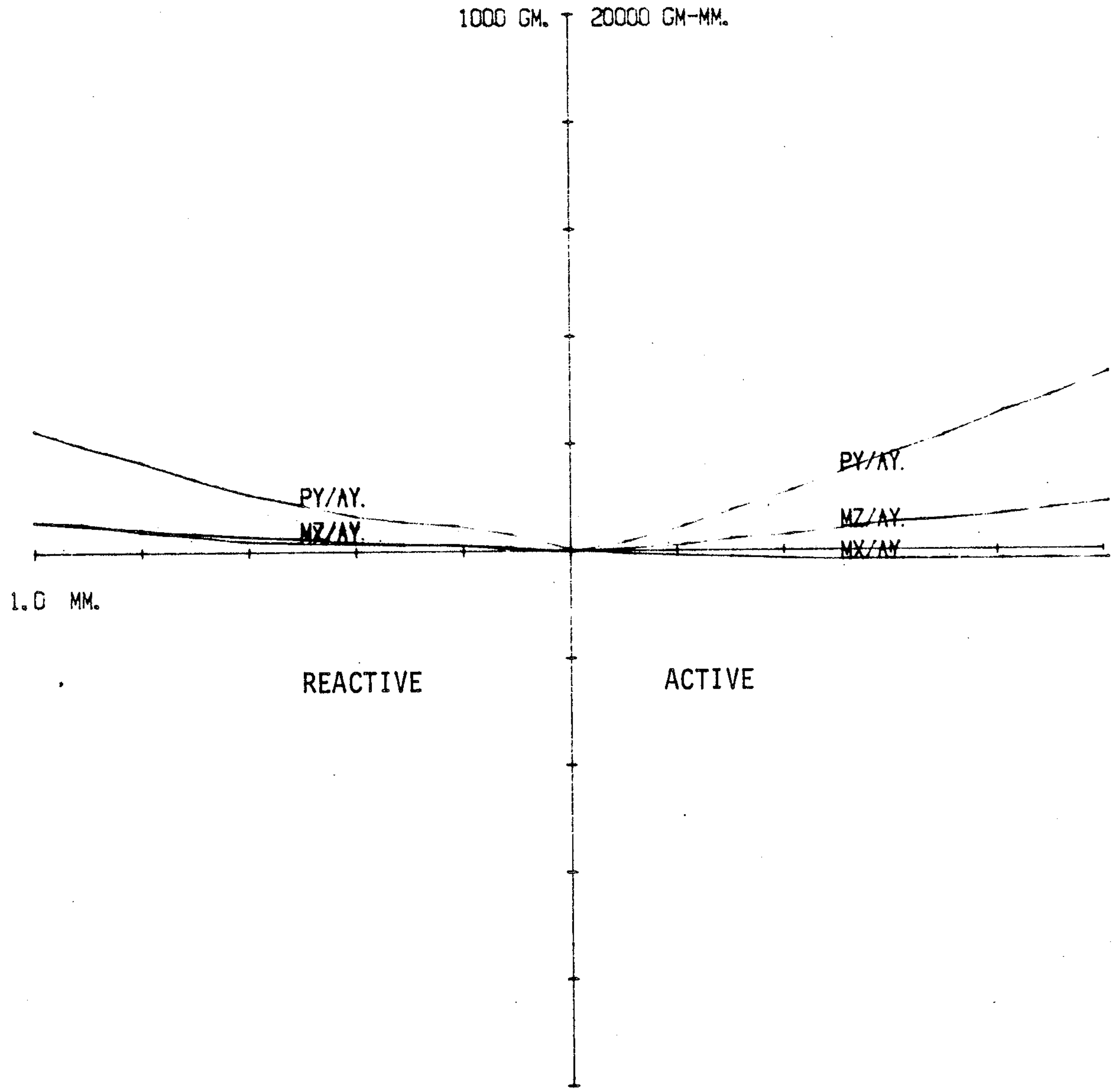


FIG. A4. VERTICAL LOOPS 10X3MM.
INTERBRACKET DISTANCE 10MM.
VERTICAL ACTIVATION.

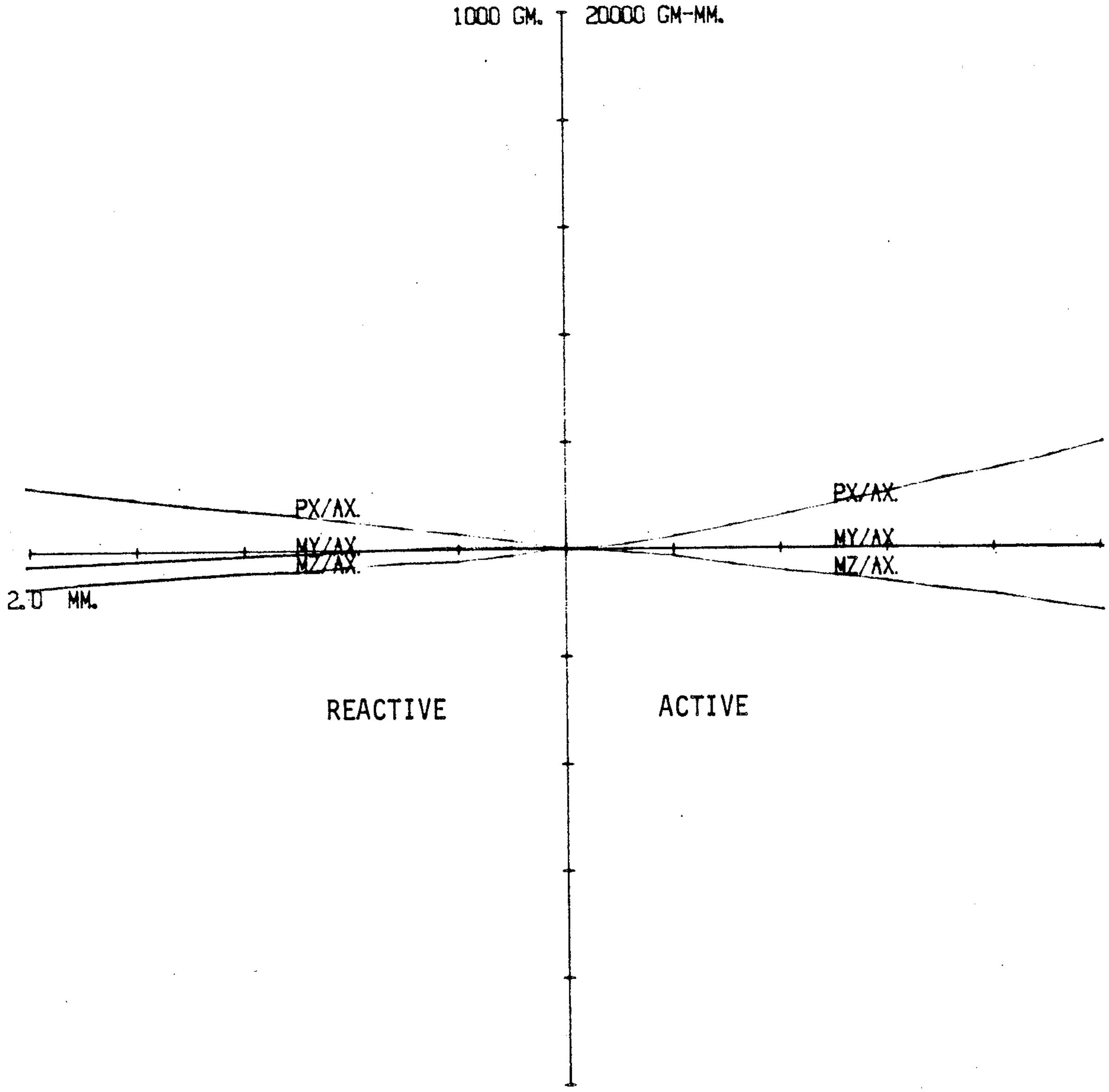


FIG. A5. VERTICAL LOOPS 10X4.5MM.
INTERBRACKET DISTANCE 10MM.
HORIZONTAL ACTIVATION.

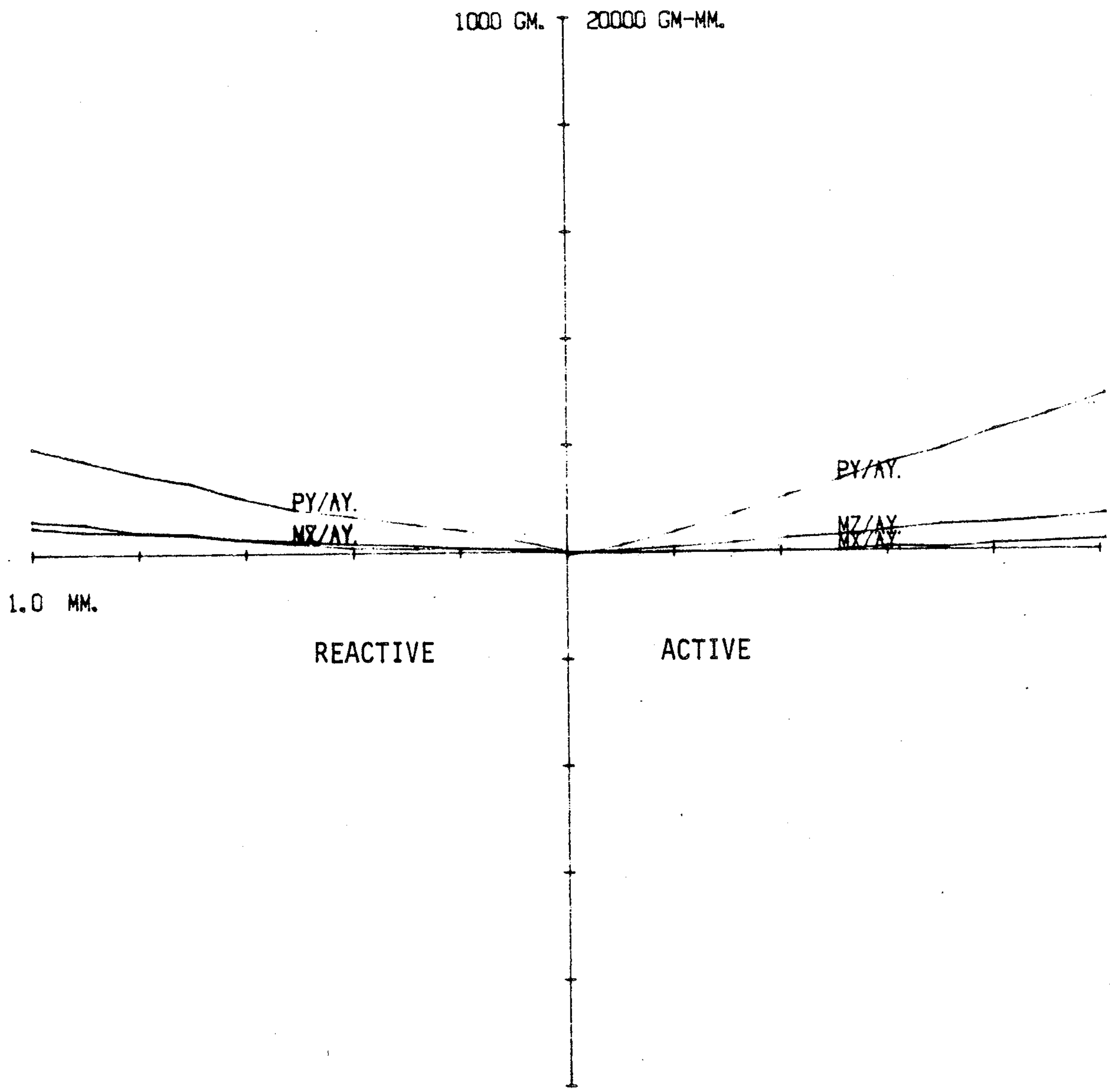


FIG. A6. VERTICAL LOOPS 10X4.5MM.
INTERBRACKET DISTANCE 10MM.
VERTICAL ACTIVATION.

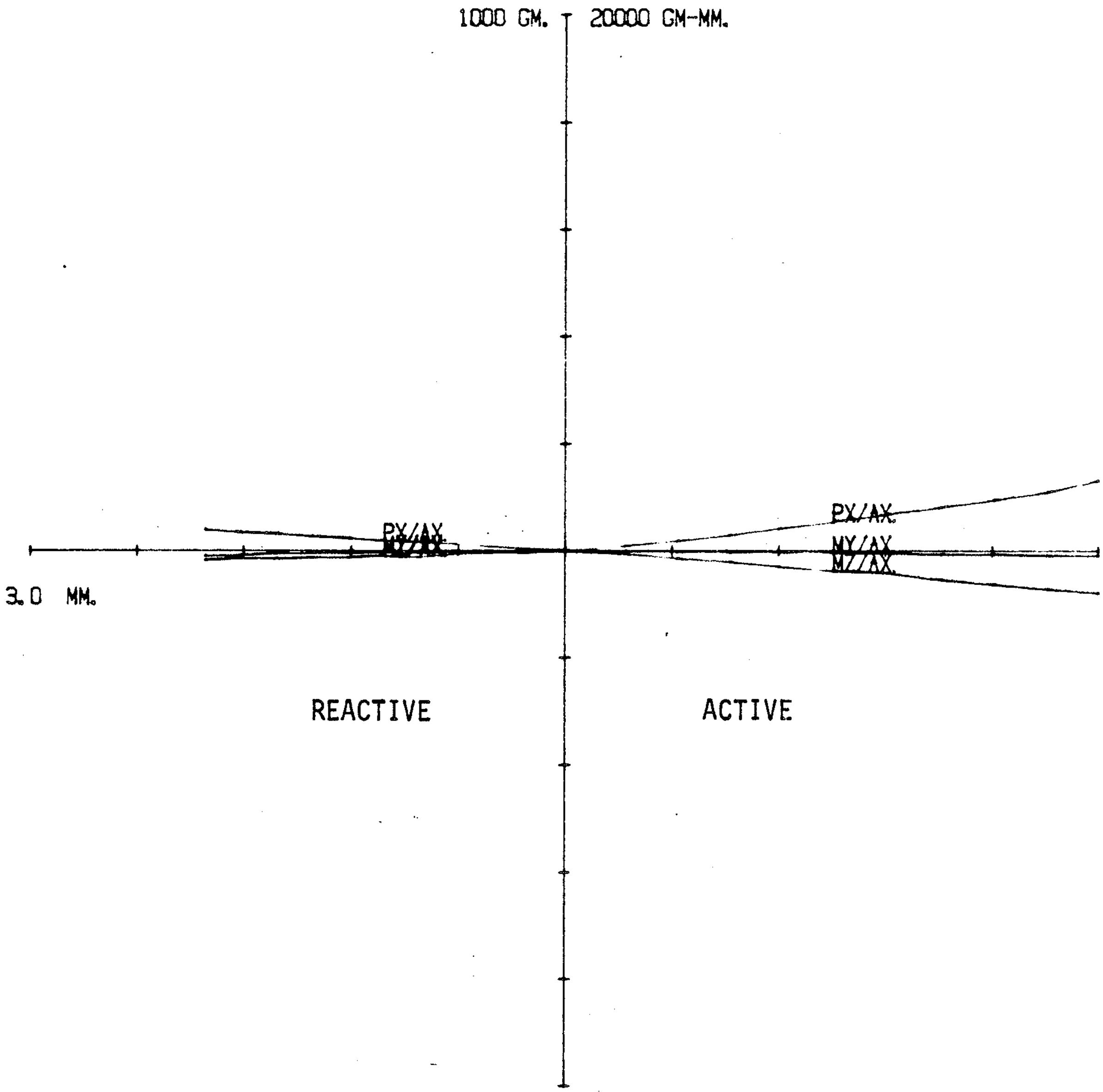


FIG. A7. L-LOOPS 10X12MM.
INTERBRACKET DISTANCE 10MM.
HORIZONTAL ACTIVATION.

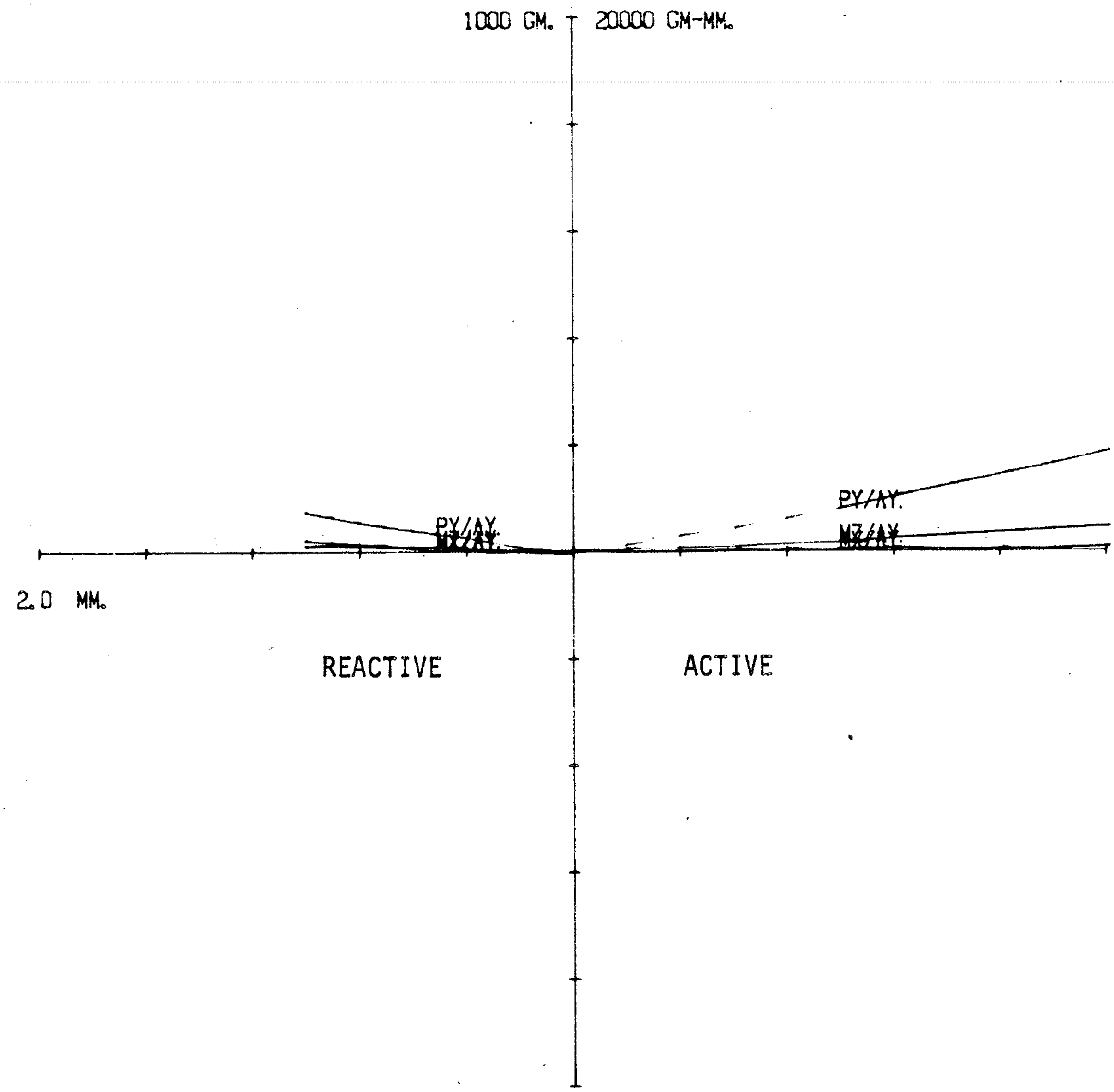


FIG. A8. L-LOOPS 10X12MM.
INTERBRACKET DISTANCE 10MM.
VERTICAL ACTIVATION.

Appendix B

Round Wires

Sullivan (1982) studied the effects of activated low modulus archwires on a bucco-lingually malaligned tooth and on the teeth adjacent to it. Subsequent to his work modifications to provide occluso-gingival activation were made to the measuring instrument. Consequently, the present study was started by extending Sullivan's work to include occluso-gingivally malaligned teeth.

The instrumentation is described in the section on materials and methods. Two sets of attachments were fabricated for the investigation of round archwires. They consisted of two identical stainless steel blocks which formed a slot where a tube of 0.5 millimeter (0.20 inch) diameter or 0.75 millimeter (0.030 inch) internal diameter could be clamped. Thus the "bracket" conditions were identical with those used by Sullivan to simulate the clinical situations of tighter or looser ligation. They will be referred to here as the SMALL tubes and the LARGE tubes.

The same low modulus archwires as used by Sullivan were chosen for this study. They were:

Tri-flex	0.38 mm.	(0.015 inch)
	0.44 mm.	(0.0175 inch)
Nitinol	0.41 mm.	(0.016 inch)
	0.45 mm.	0(.018 inch)
TMA	0.39 mm.	(0.016 inch).

The experimental procedure was similar to that used for experiments on looped archwires and it is described in the section on materials and methods. The amount of activation for each wire was chosen according to its elastic properties so that it was activated as far as possible without exceeding its elastic limit and without overloading the measuring instrument. The archwires were activated over ten steps in the occlusal (positive) and the gingival (negative) directions. After each activation the archwire was deactivated back to zero to show the effects of hysteresis.

The force systems were first measured using the three tooth assembly (assembly I) where the measured tooth acts as an active tooth. In the first two sets of experiments the interbracket distances were 3 millimeters on either side of the measured tooth. The force system created when the interbracket distances were symmetrical are shown in Figure B1. The magnitudes of forces and moments at the activation level of 1 millimeter are shown in Table B1 for the LARGE tubes and in Table B2 for the SMALL tubes.

The vertical force, Py, can be considered excessive. Large moments, Mz, around the bucco-lingual axis are created indicating lack of control over the bucco-lingual tipping. When the results in Tables B1 and B2 are compared it can be seen that the magnitudes of these forces and moments are largely influenced by the size of the tubes (the tightness of ligation) and by the interbracket distances.

In the third set of experiments the same activation assembly I was used but the interbracket distance mesial to the malaligned tooth was 2 millimeters and distal to the malaligned tooth was 4 millimeters. The force systems created when interbracket distances were asymmetrical are shown in Figure B2 and in Table B3 for LARGE tubes. The results for SMALL tubes are not shown because, as it can be concluded from Tables B1 and B2, they would be similar to those for LARGE tubes except that all the forces and moments would be higher. It can be seen when the interbracket distances between the malaligned tooth and the teeth adjacent to it are not symmetrical mesio-distal tipping moments, Mx, are also created. These tipping moments are caused by the lateral wedging forces which are larger on the distal end where the interbracket distance is smaller and therefore result in a mesially directed (positive) force, Pz, and a mesial (positive) tipping moment. The occluso-gingival forces acting on either end of the tube are not equal but larger on the distal end. These so-called cantilever forces form a couple which creates a mesial (positive) tipping moment with occlusal activation and a distal (negative) tipping moment with gingival activation. The mesially directed wedging force also creates a small mesially rotating moment around the occluso-gingival axis.

In the last set of experiments force systems were measured on a reactive tooth using the four tooth assembly (assembly II). Only 0.41 millimeter Nitinol wire was tested using three different interbracket distances and LARGE tubes. In the first experiment all the interbracket distances were 3 millimeters as shown in Figure B3 which presents the force system on a reactive tooth. In additional experiments two asymmetric spacings were tested. These spacings are given in Table B4 in the same left to right order as that of the teeth in Figure B3. From Table B4 it can be seen that the magnitudes of occluso-gingival forces are approximately one half of those acting on the active tooth and they create significant bucco-lingual tipping moments. The asymmetrical loading creates lateral wedging forces and cantilever forces create mesio-distal tipping moments. The lateral wedging force also creates a small moment around the occluso-gingival axis. It can be seen that the magnitude of these undesirable forces and moments on the adjacent teeth largely depend on the interbracket distances. When the distance between the active and the reactive tooth is reduced by one half the force and moment levels increased approximately three times.

The conclusions from this investigation are in agreement with those of Sullivan. The forces created with these so-called flexible archwires are much higher than those that are normally accepted as ideal. Three dimensional control over final tooth positions is impossible with continuous round low-modulus archwires and significant movement of adjacent teeth should be anticipated.

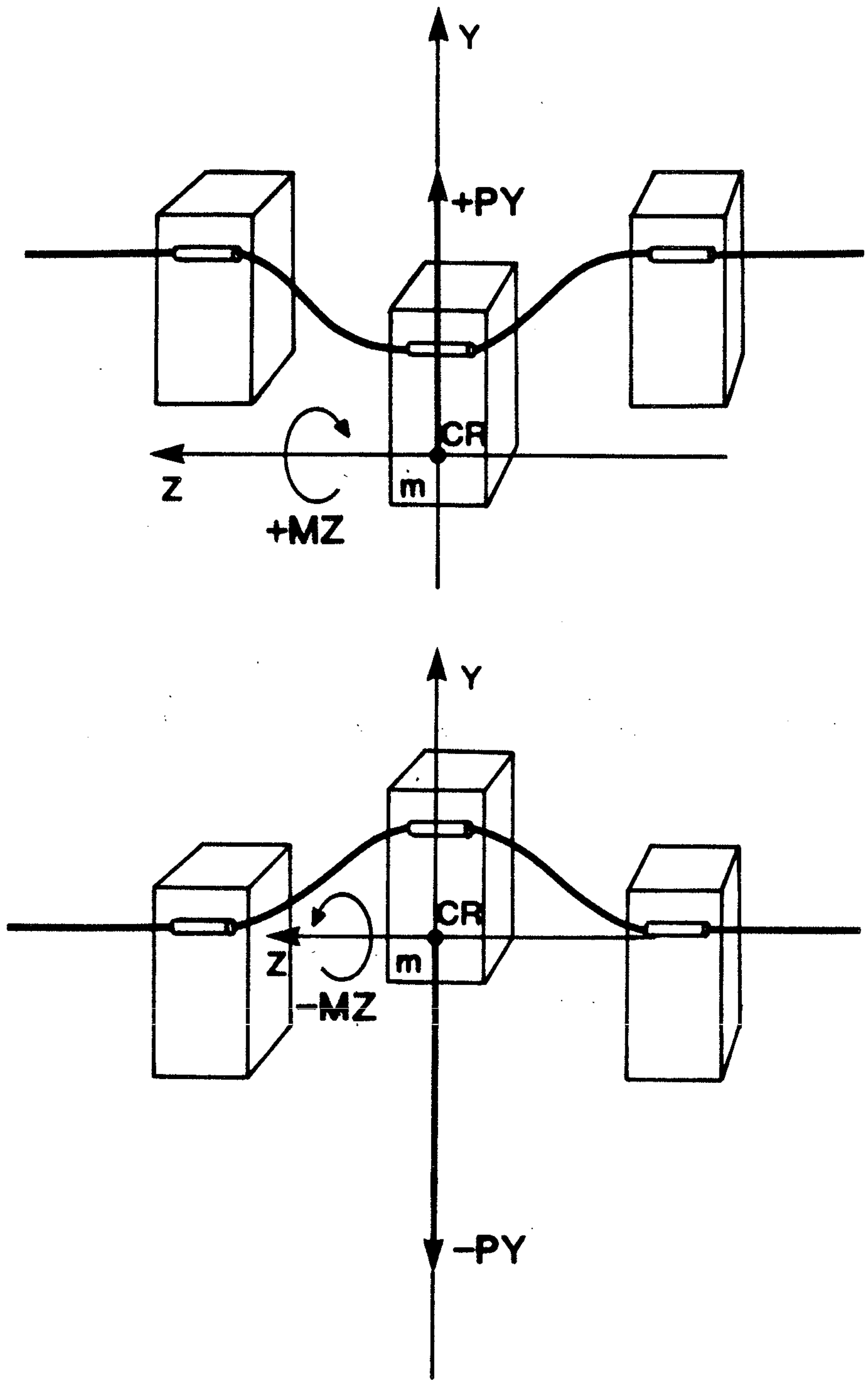


FIGURE B1 : FORCE SYSTEMS ON ACTIVE TOOTH. INTERBRACKET DISTANCES 3 mm and 3 mm.

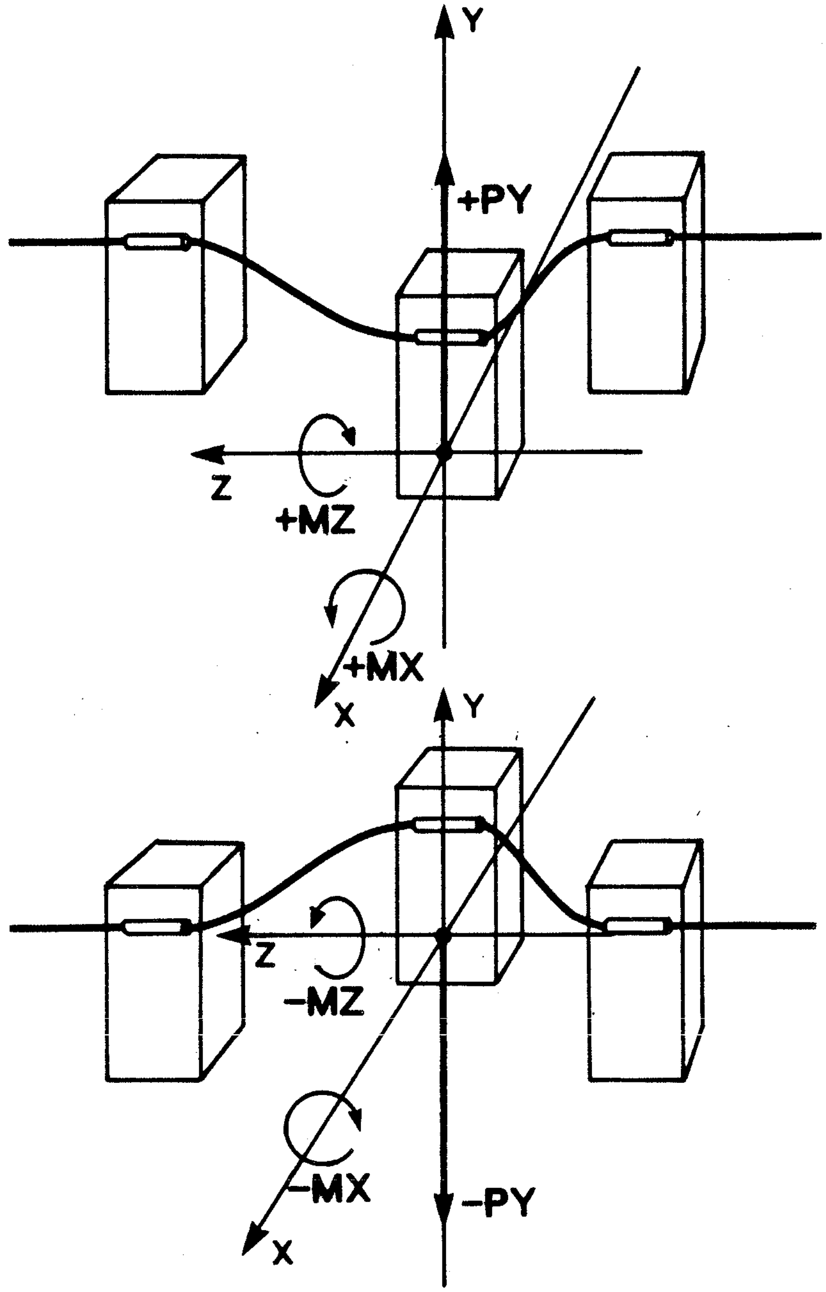


FIGURE B 2 : FORCE SYSTEMS ON ACTIVE TOOTH. INTERBRACKET DISTANCES 4 mm and 2 mm.

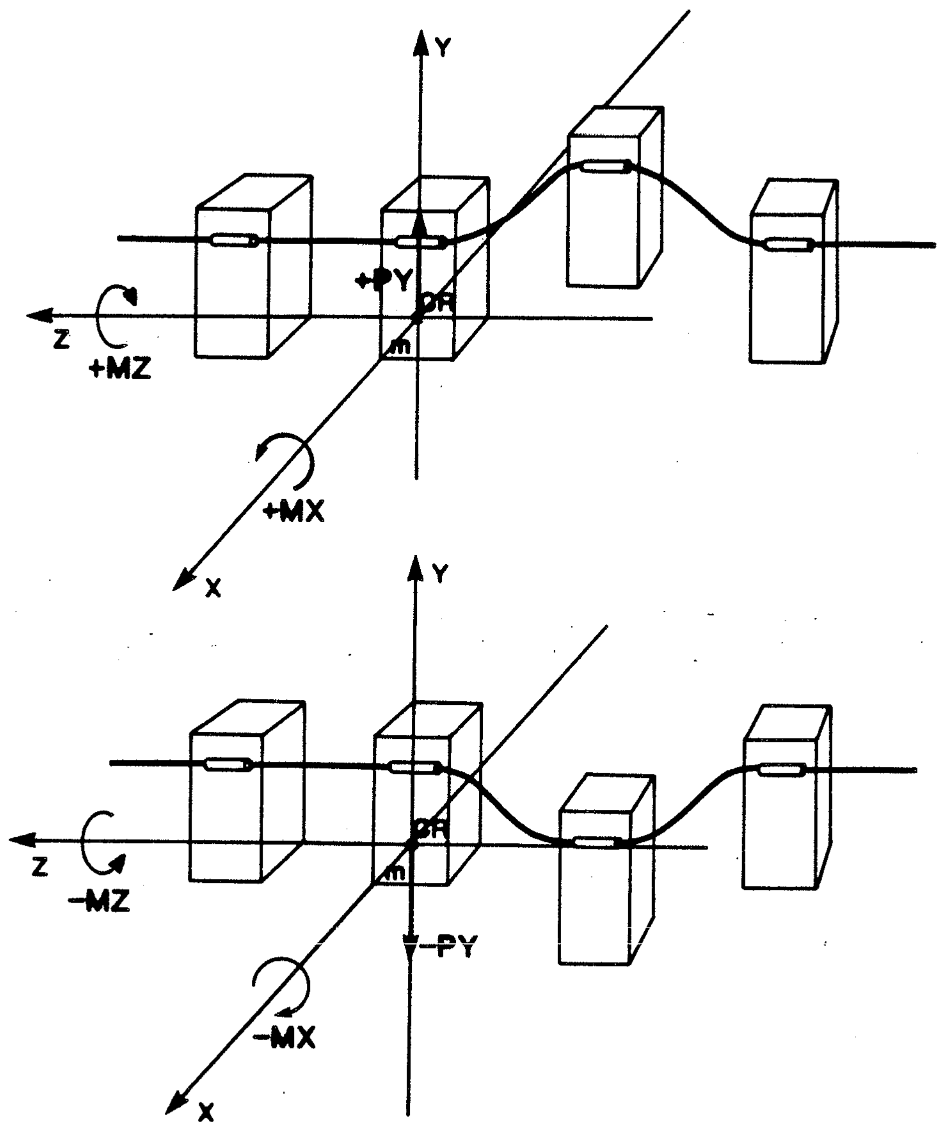


FIGURE B3 : FORCE SYSTEMS ON REACTIVE TOOTH.

TYPE	SIZE	Py	Mz	Mz/Py	Pz	Mx
Tri-Flex	0.38 mm.	150	700	4.5	0	0
	0.44 mm.	290	1200	4.2	0	0
Nitinol	0.41 mm.	470	2100	4.6	0	0
	0.45 mm.	550	2800	5.1	0	0
TMA	0.39 mm.	450	2300	5.0	0	0

TABLE B1. Force systems on active tooth with 3 mm. and 3 mm. interbracket distances. LARGE tubes. Vertical activation Ay = 1 mm.

TYPE	SIZE	Py	Mz	Mz/Py	Pz	Mx
Tri-Flex	0.38 mm.	220	1300	5.9	0	0
	0.44 mm.	470	2700	5.7	0	0
Nitinol	0.41 mm.	510	2700	5.3	0	0
	0.45 mm.	960	5100	5.3	0	0
TMA	0.39 mm.	850	4800	5.7	0	0

TABLE B2. Force systems on active tooth with 3 mm. and 3 mm. interbracket distances. SMALL tubes. Vertical activation Ay = 1 mm.

TYPE	SIZE	Py	Mz	Mz/Py	Pz	Mx
Tri-Flex	0.38 mm.	220	1300	6.0	50	1100
	0.44 mm.	390	2200	5.5	100	2000
Nitinol	0.41 mm.	400	2200	5.6	150	2800
	0.45 mm.	610	3600	5.9	130	2800
TMA	0.39 mm.	660	3700	5.5	50	1700

TABLE B3. Force systems on active tooth with 4 mm. and 2 mm. interbracket distances. LARGE tubes. Vertical activation Ay = 1 mm.

INTERBRACKET DISTANCES	Py	Mz	Mz/Py	Pz	Mx
3 mm., 4 mm., 2 mm.	130	800	5.9	0	0
3 mm., 3 mm., 3 mm.	230	1400	6.1	50	1000
3 mm., 2 mm., 4 mm.	470	3000	6.3	30	2000

TABLE B4. Force systems on reactive tooth. Nitinol 0.41 mm. LARGE tubes. Vertical activation Ay = 1 mm.

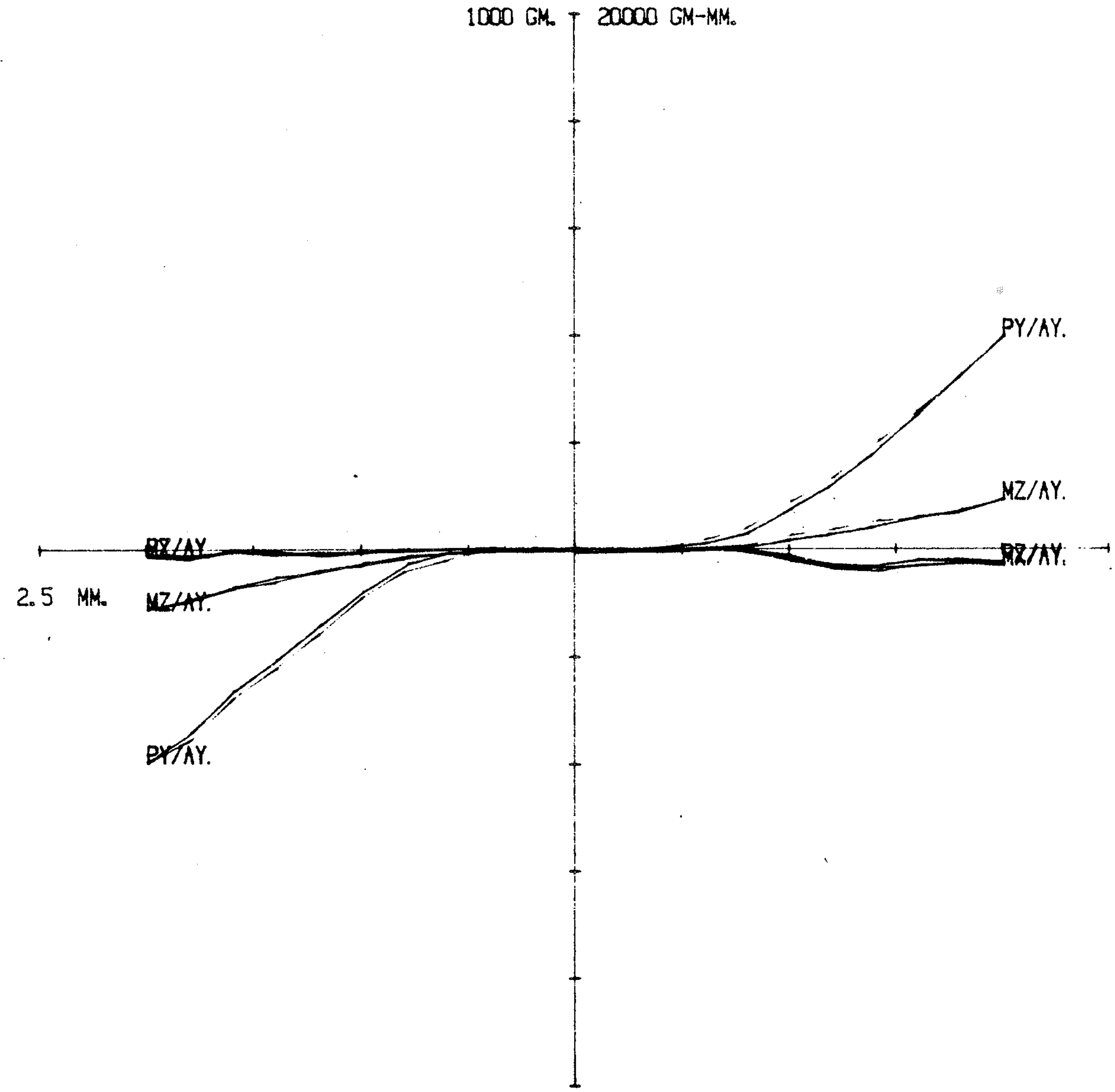


FIG. B4. TRI-FLEX 0.38MM. (0.015INCH).
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

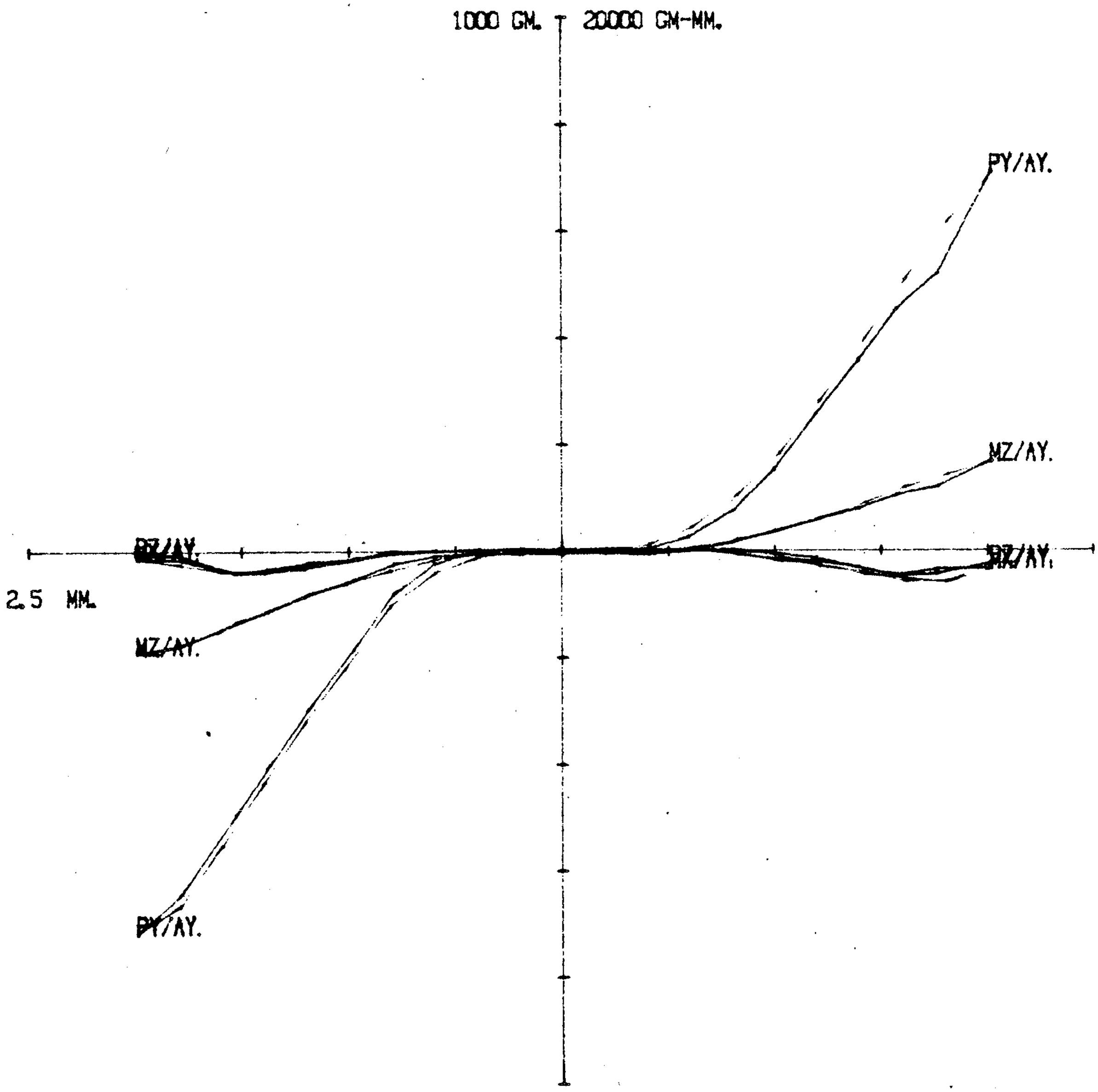


FIG. B5. TRI-FLEX 0.44MM. (0.0175INCH).
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

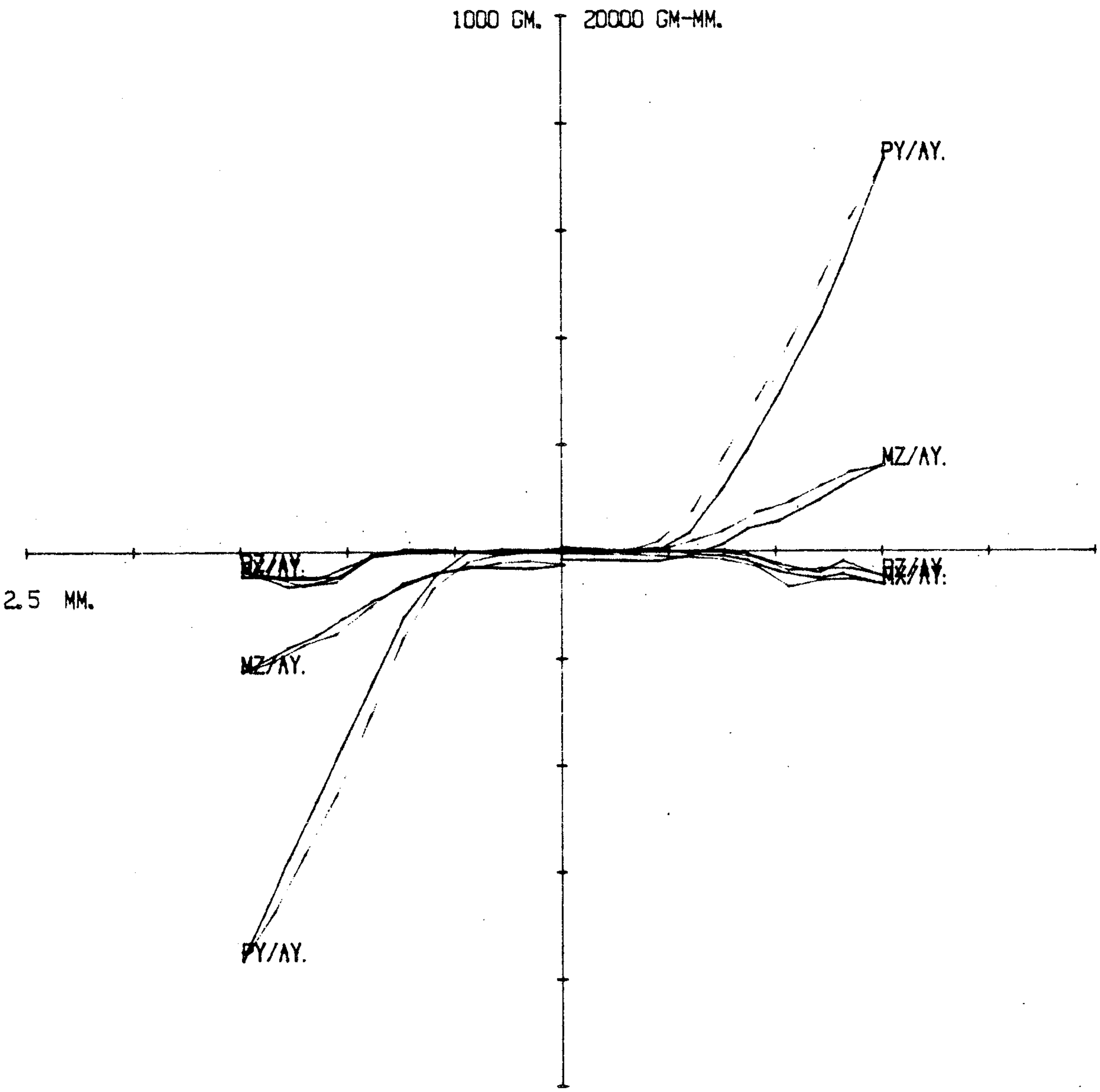


FIG. B6. NITINOL 0.41MM. (0.016INCH).
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

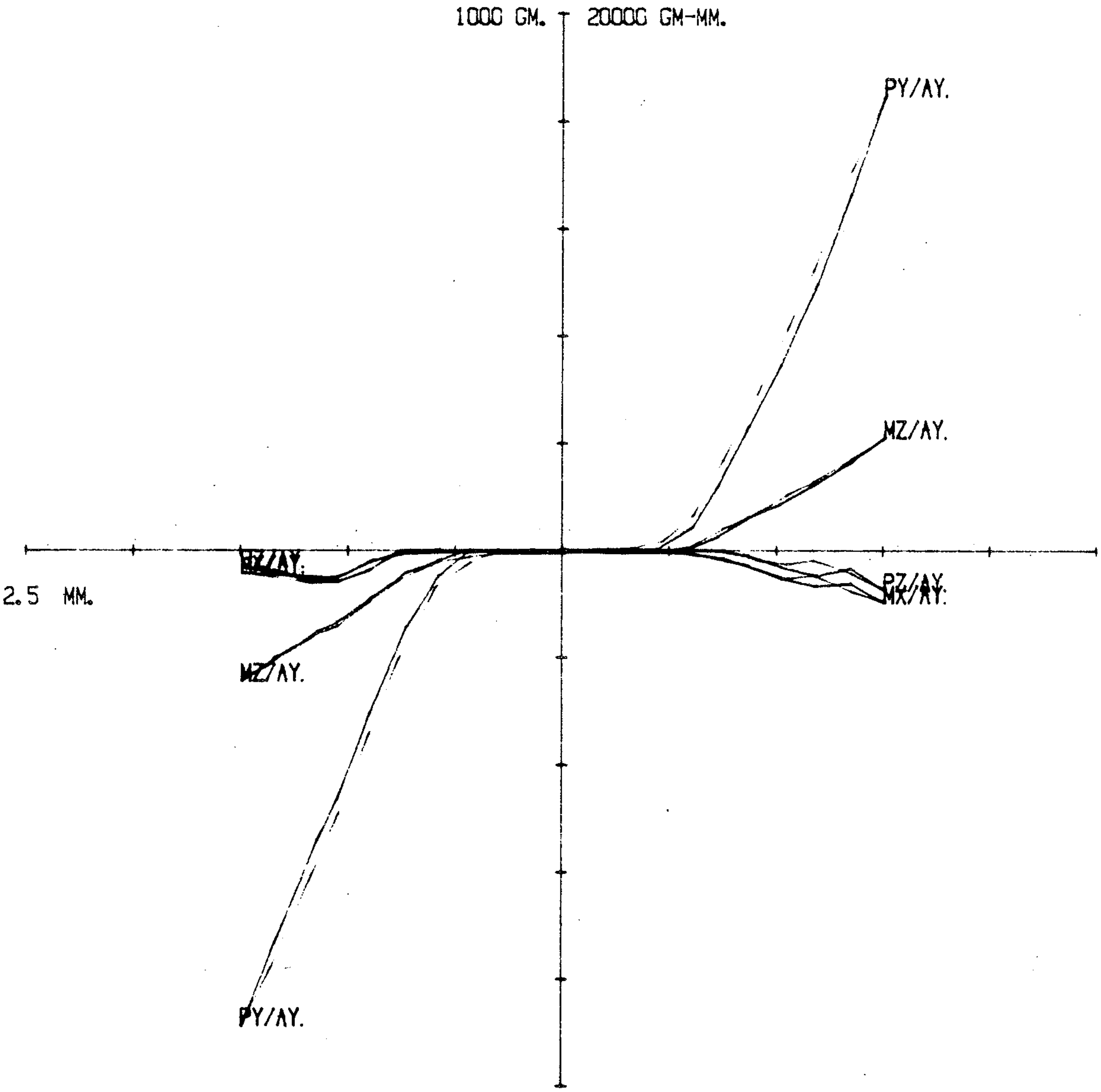


FIG. B7. NITINOL 0.45MM. (0.018INCH).
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

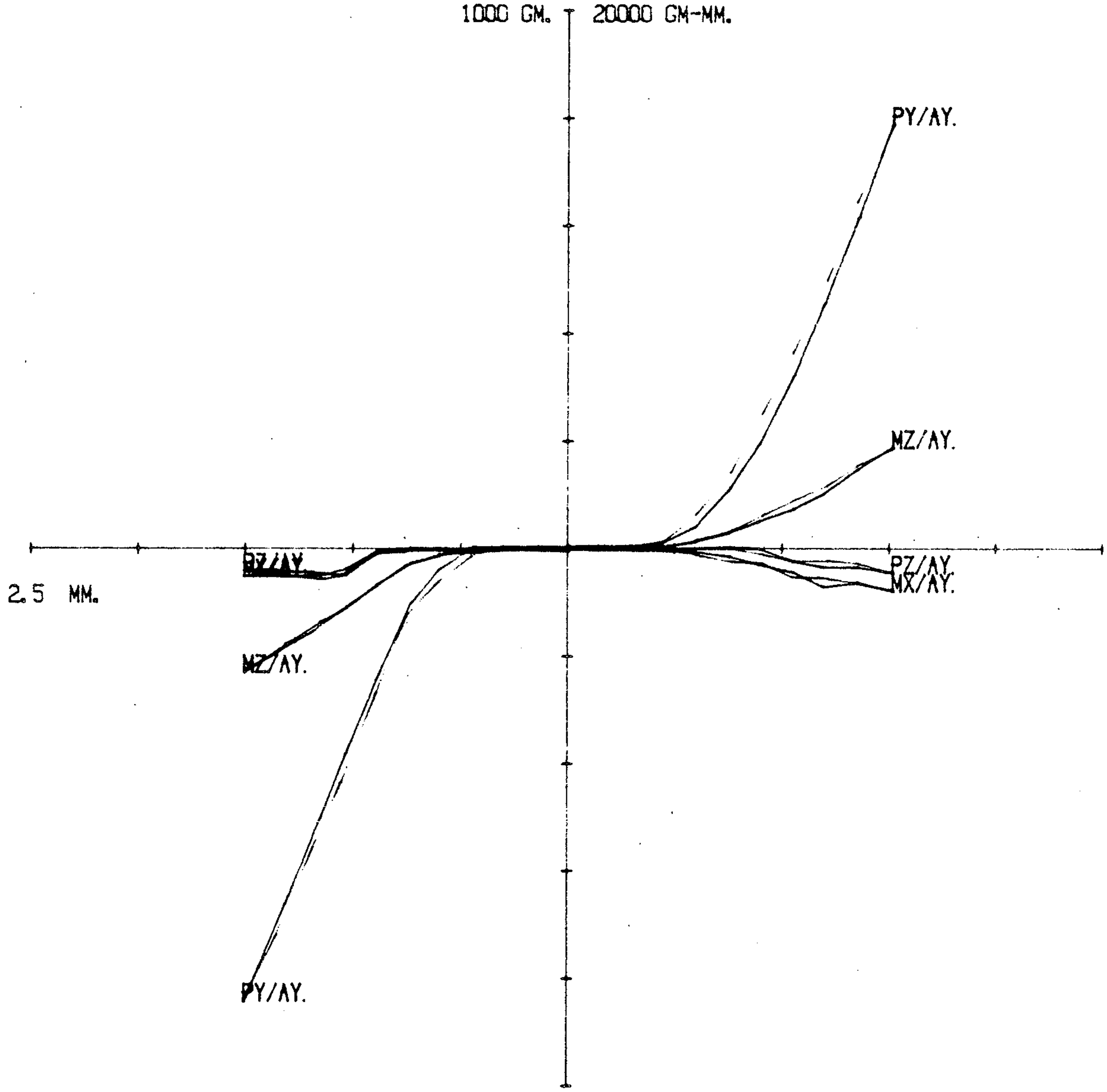


FIG. B8. TMA 0.39MM. (0.016INCH)
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

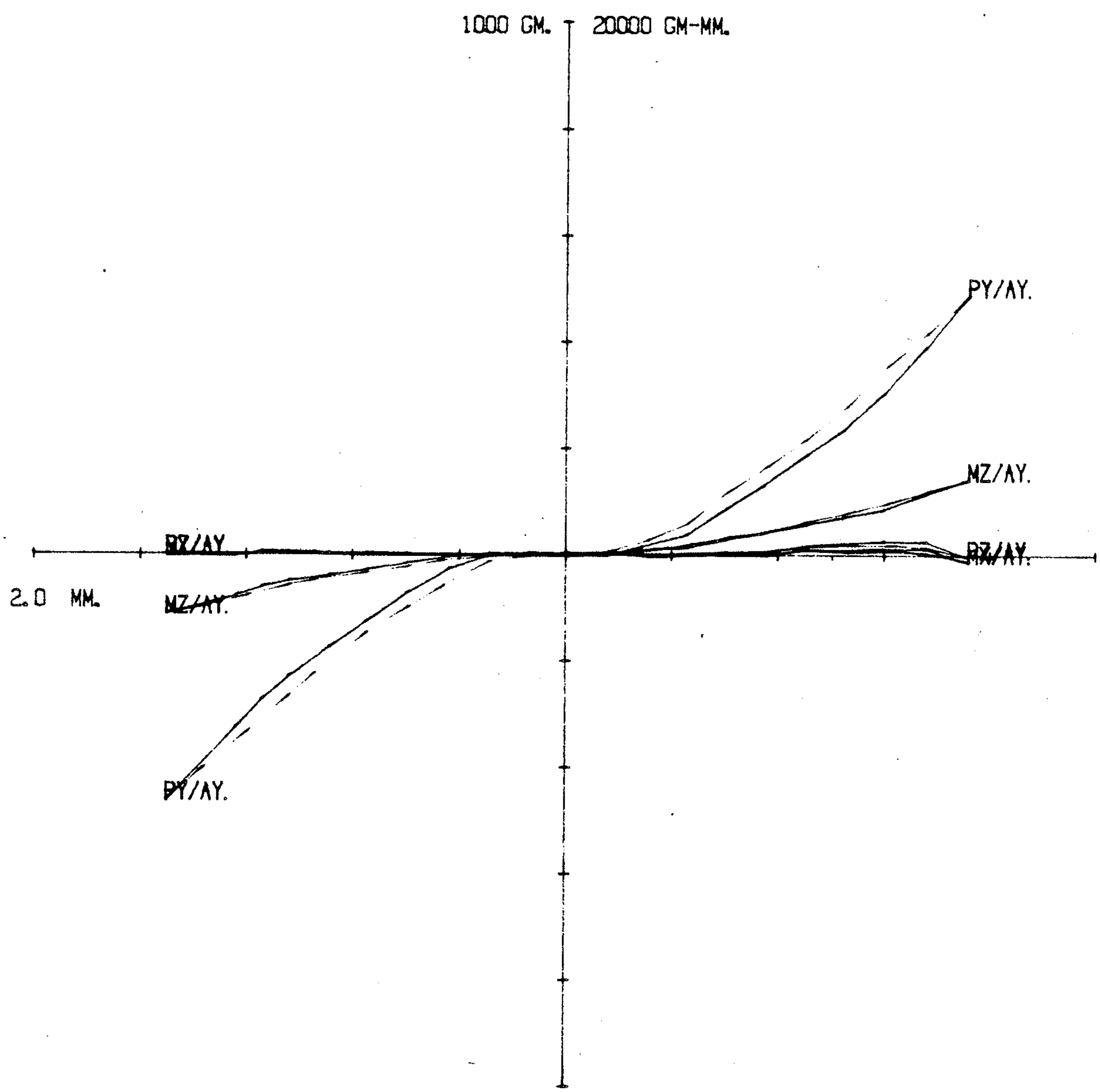


FIG. B9. TRI-FLEX 0.38MM. (0.015INCH).
SMALL TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

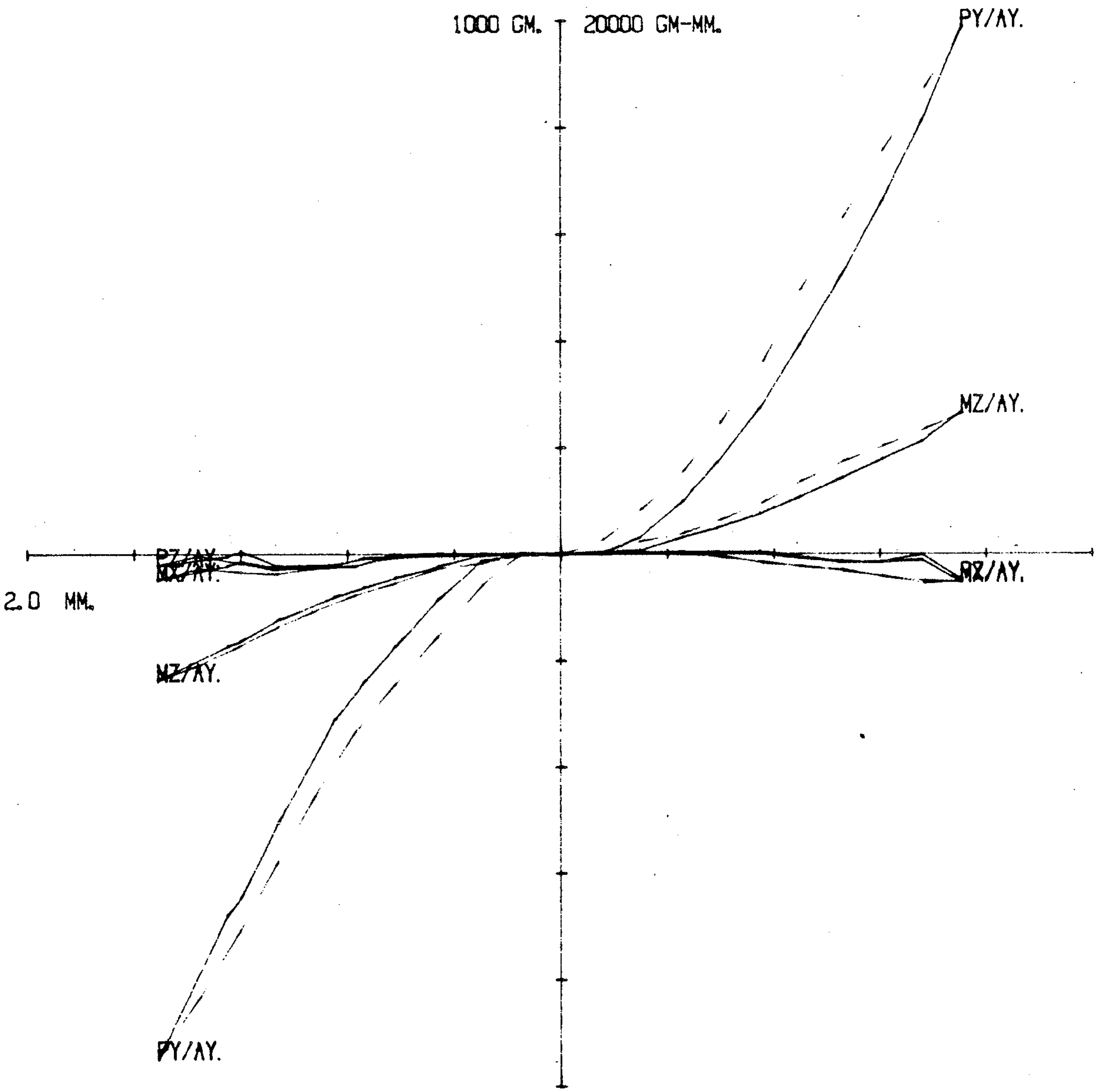


FIG. B10. TRI-FLEX 0.44MM. (0.0175INCH)
SMALL TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

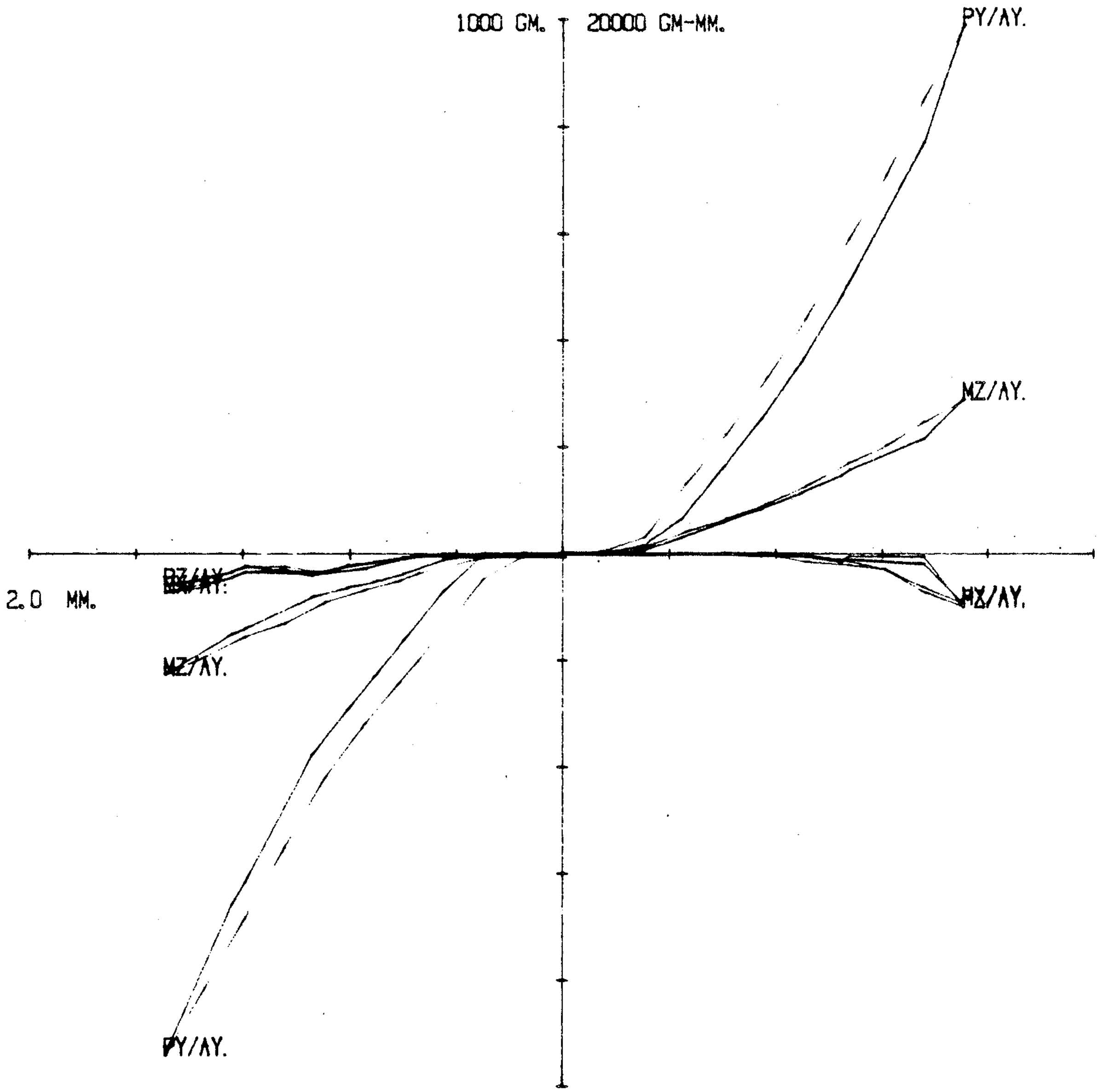


FIG. B11. NITINOL 0.41MM. (0.016INCH).
SMALL TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

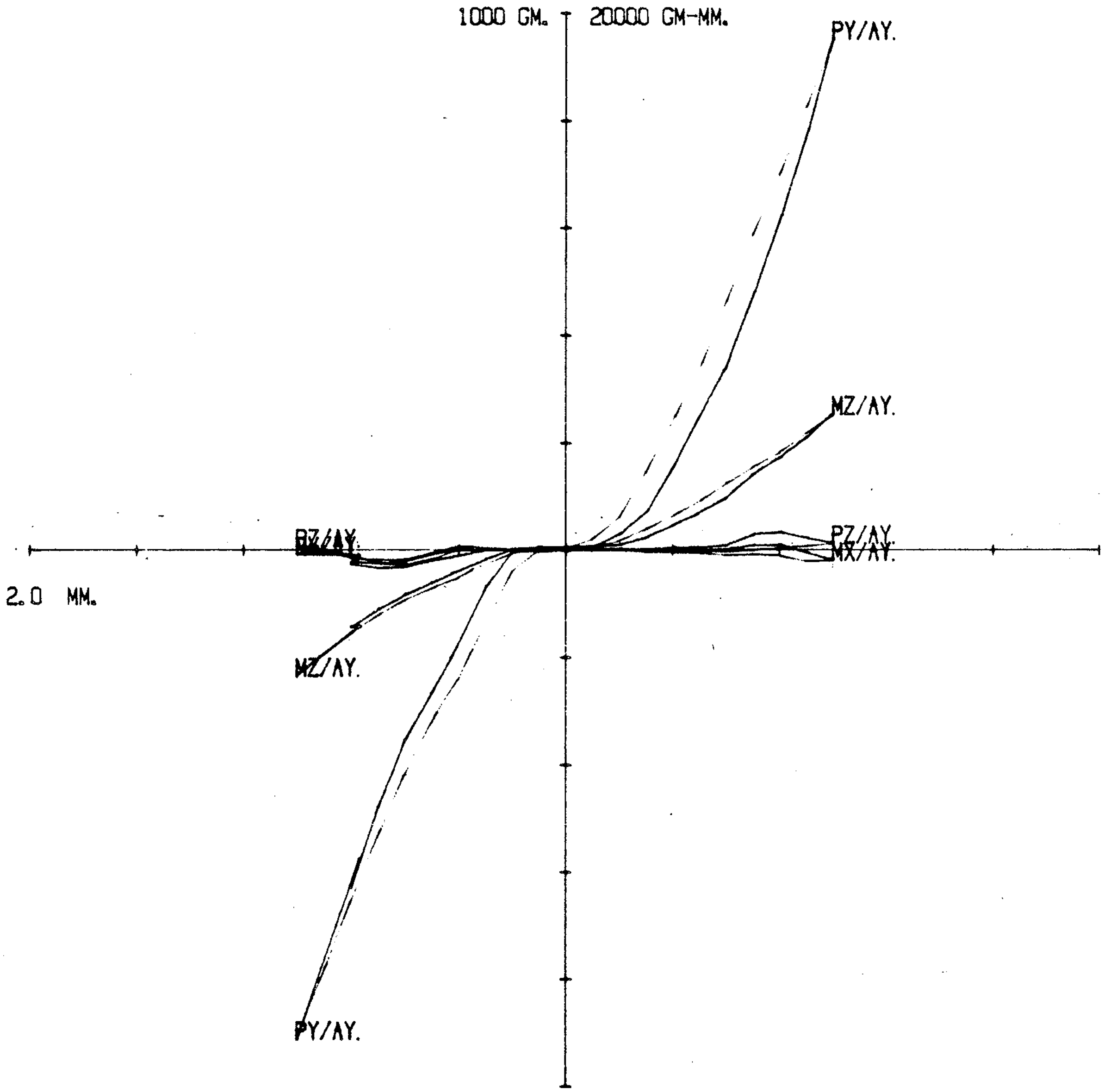


FIG. B12. NITINOL 0.45 MM. (0.018INCH).
SMALL TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

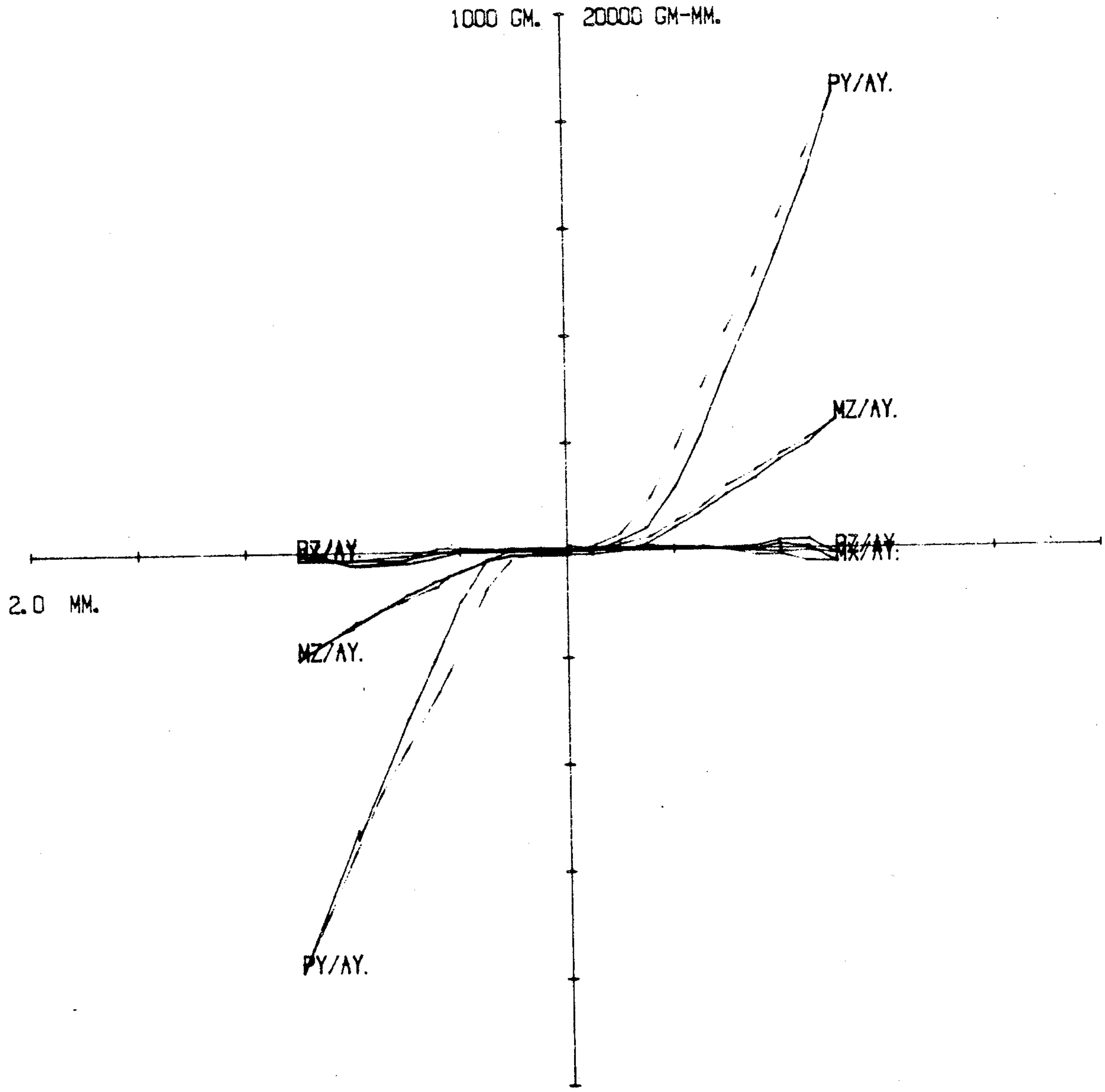


FIG. B13. TMA 0.39MM. (0.016INCH).
SMALL TUBES. ACTIVE TOOTH.
INTERBR. DIST. 3MM. AND 3MM.

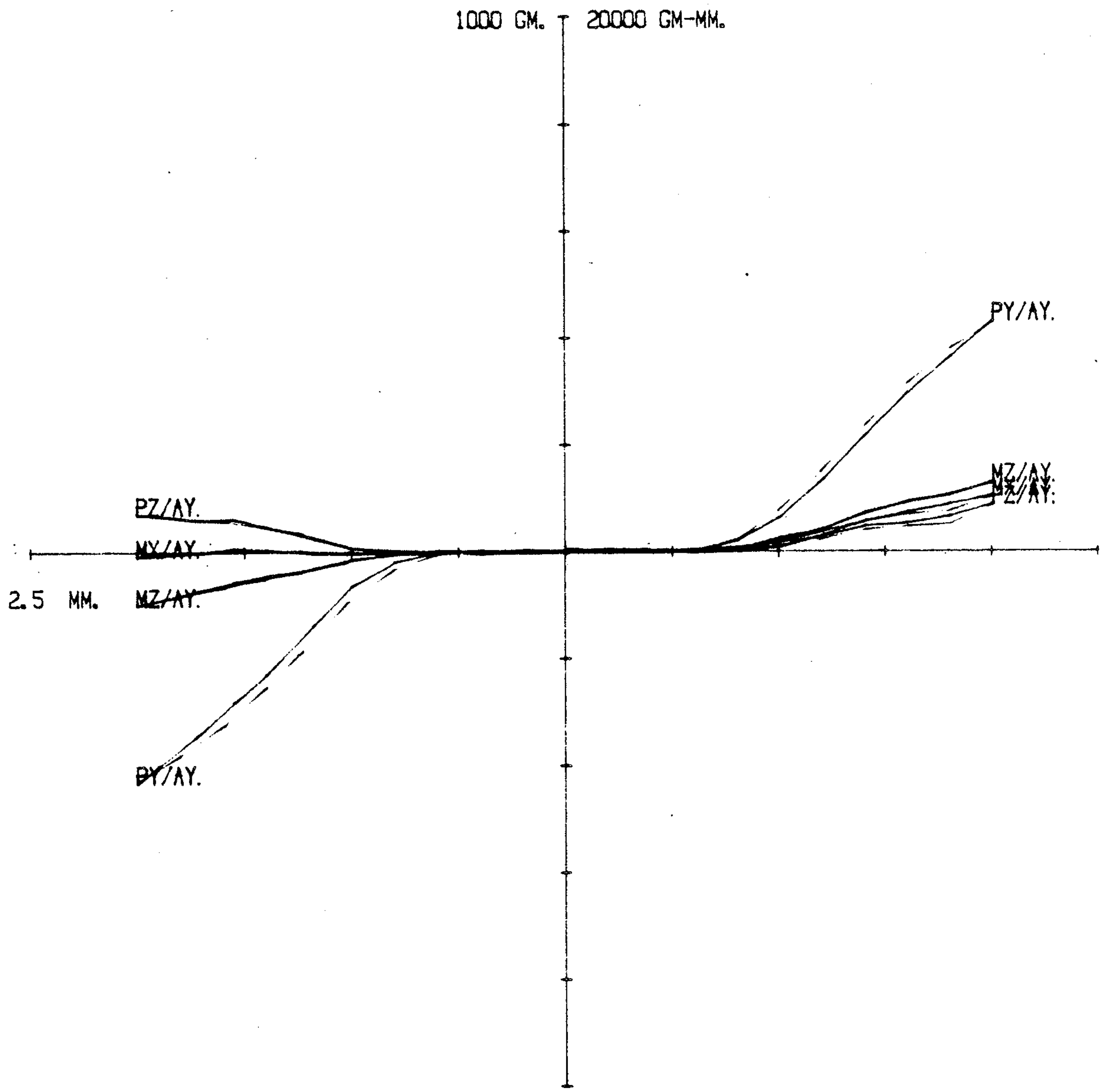


FIG. B14. TRI-FLEX 0.38MM. (0.015INCH)
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 4MM. AND 2MM.

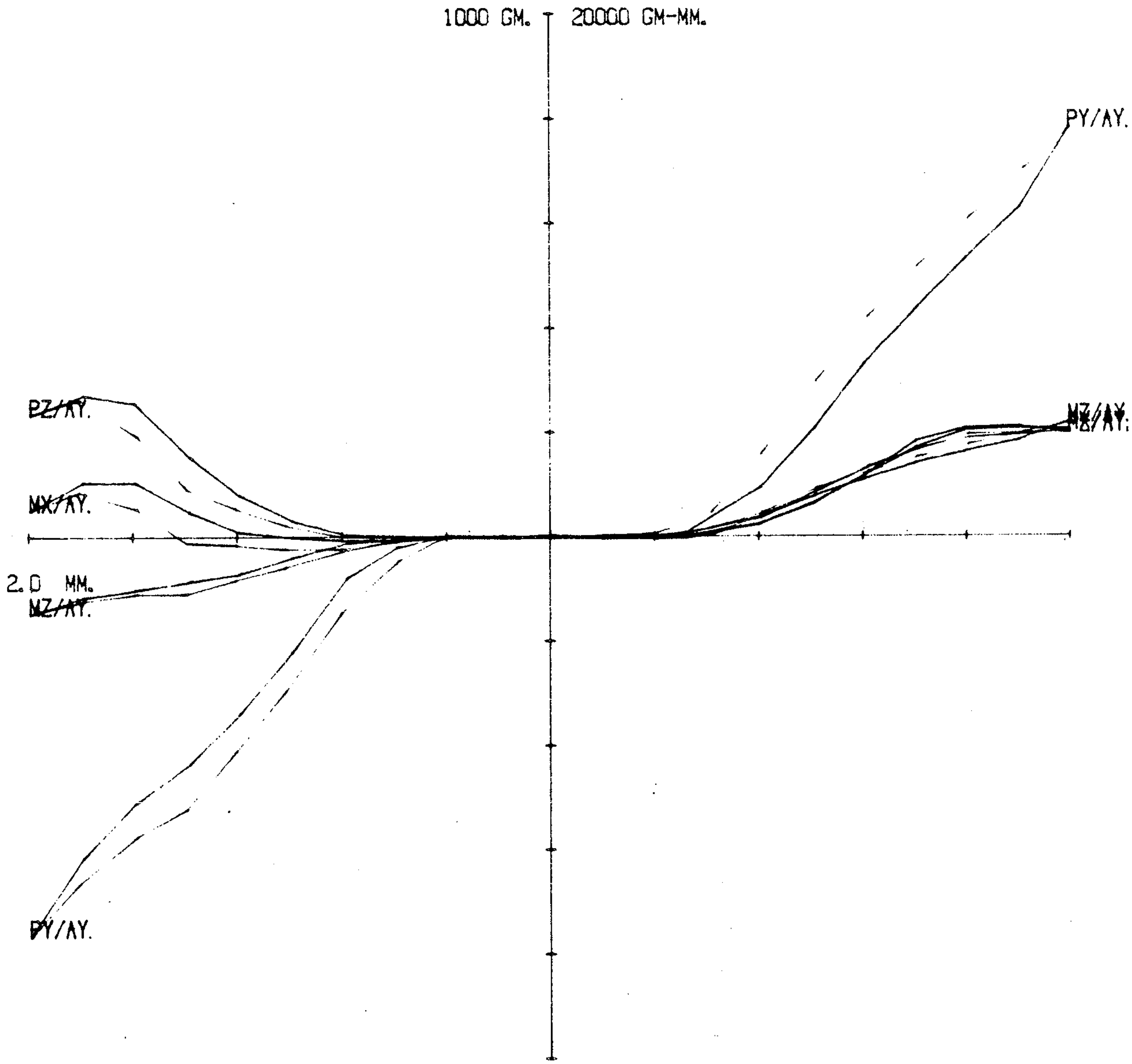


FIG. B15. TRI-FLEX 0.44MM. (0.0175INCH).
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 4MM. AND 2MM.

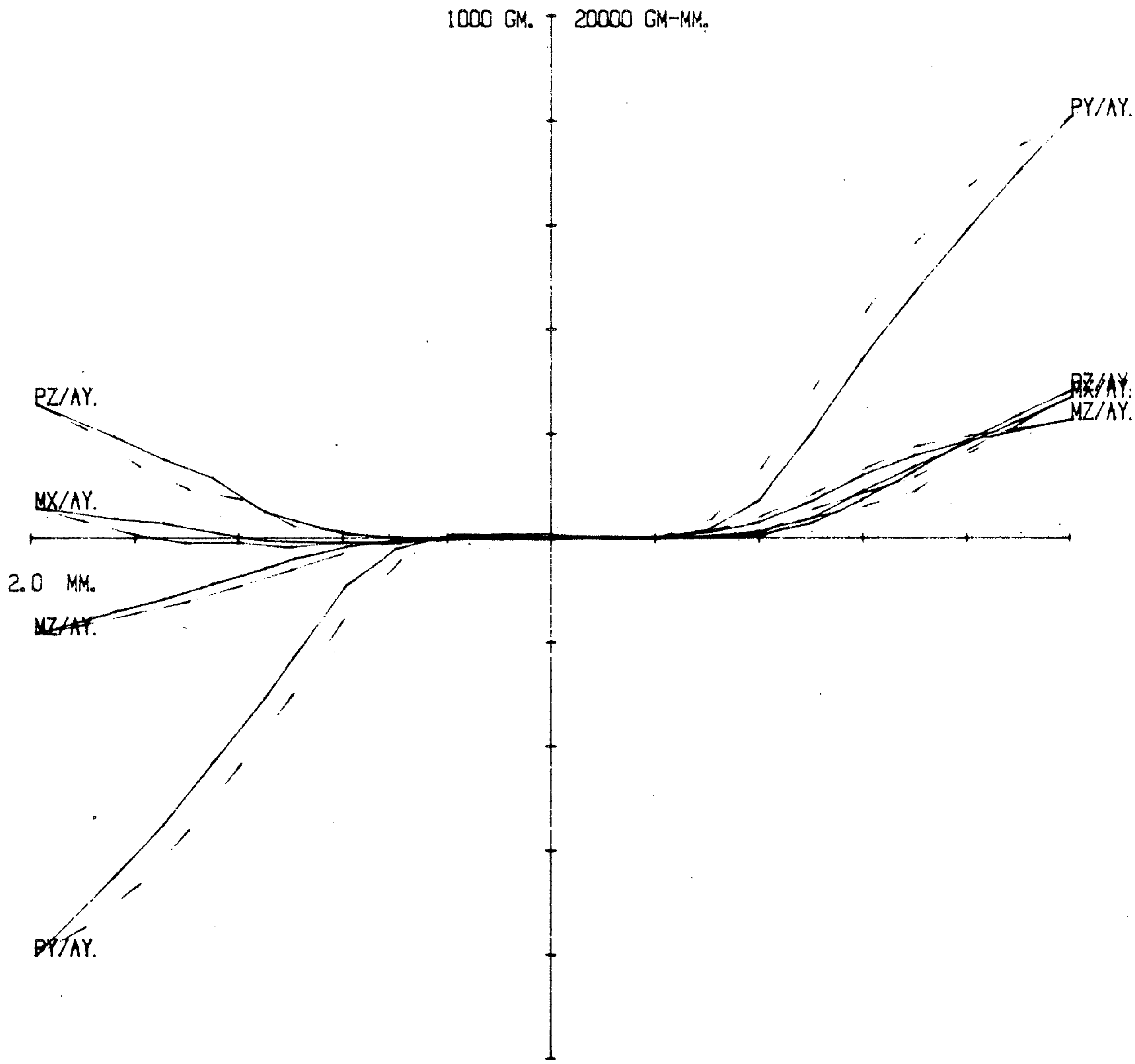


FIG. B16. NITINOL 0.41MM. (0.016INCH)
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 4MM. AND 2MM.

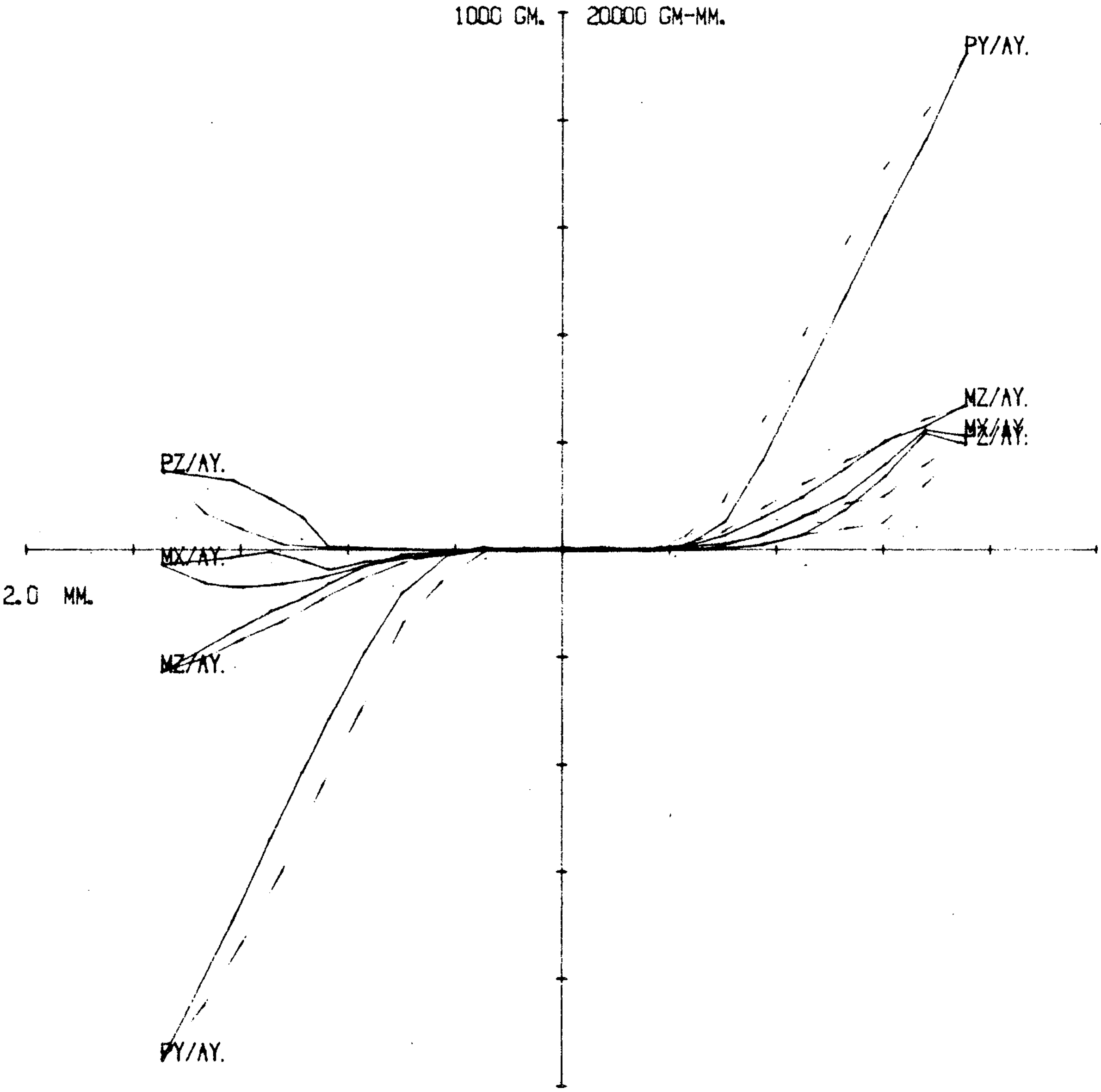


FIG. B17. NITINOL 0.45MM. (0.018INCH).
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 4MM. AND 2MM.

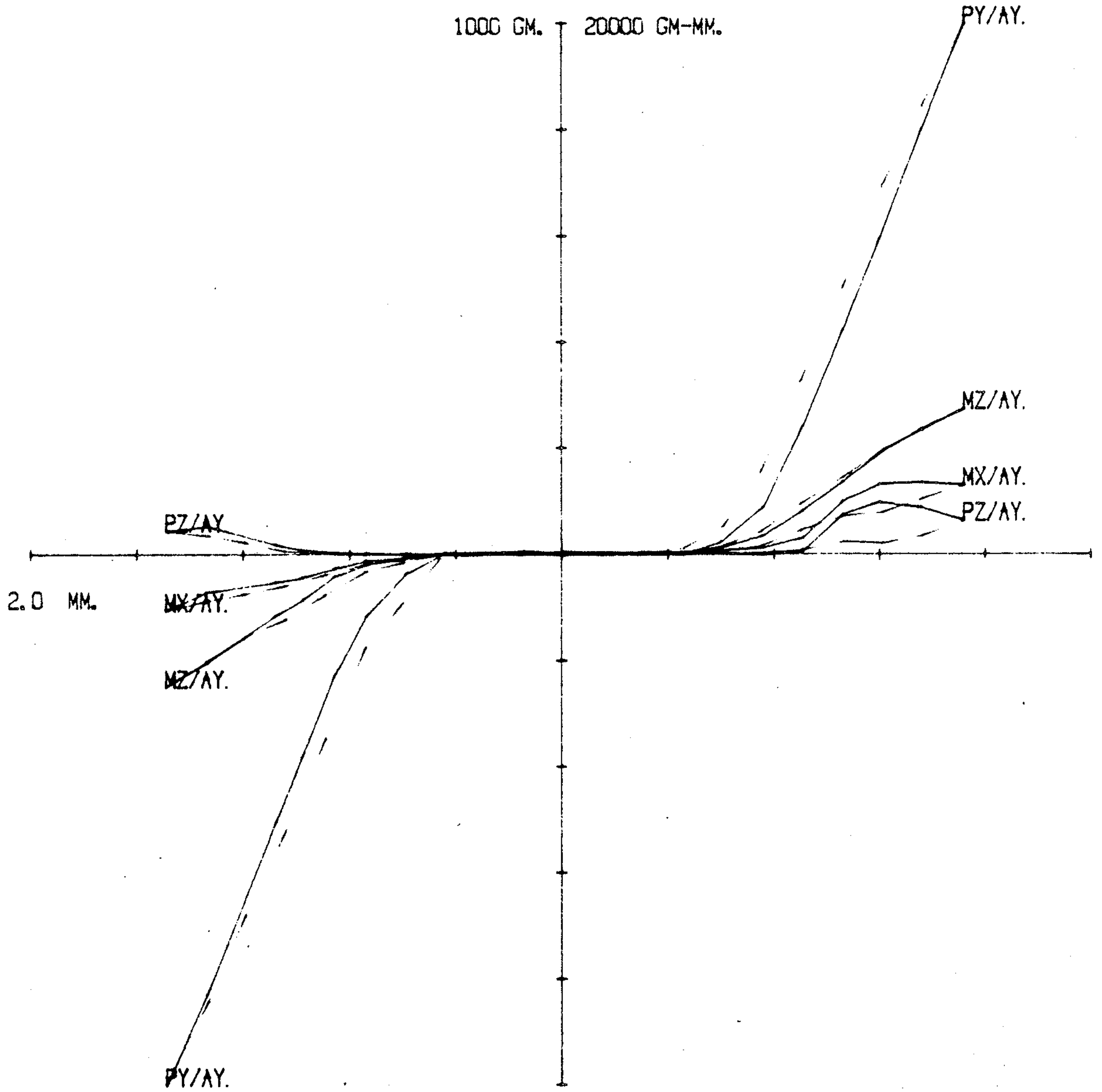


FIG. B18. TMA 0.39MM. (0.016INCH)
LARGE TUBES. ACTIVE TOOTH.
INTERBR. DIST. 4MM. AND 2MM.

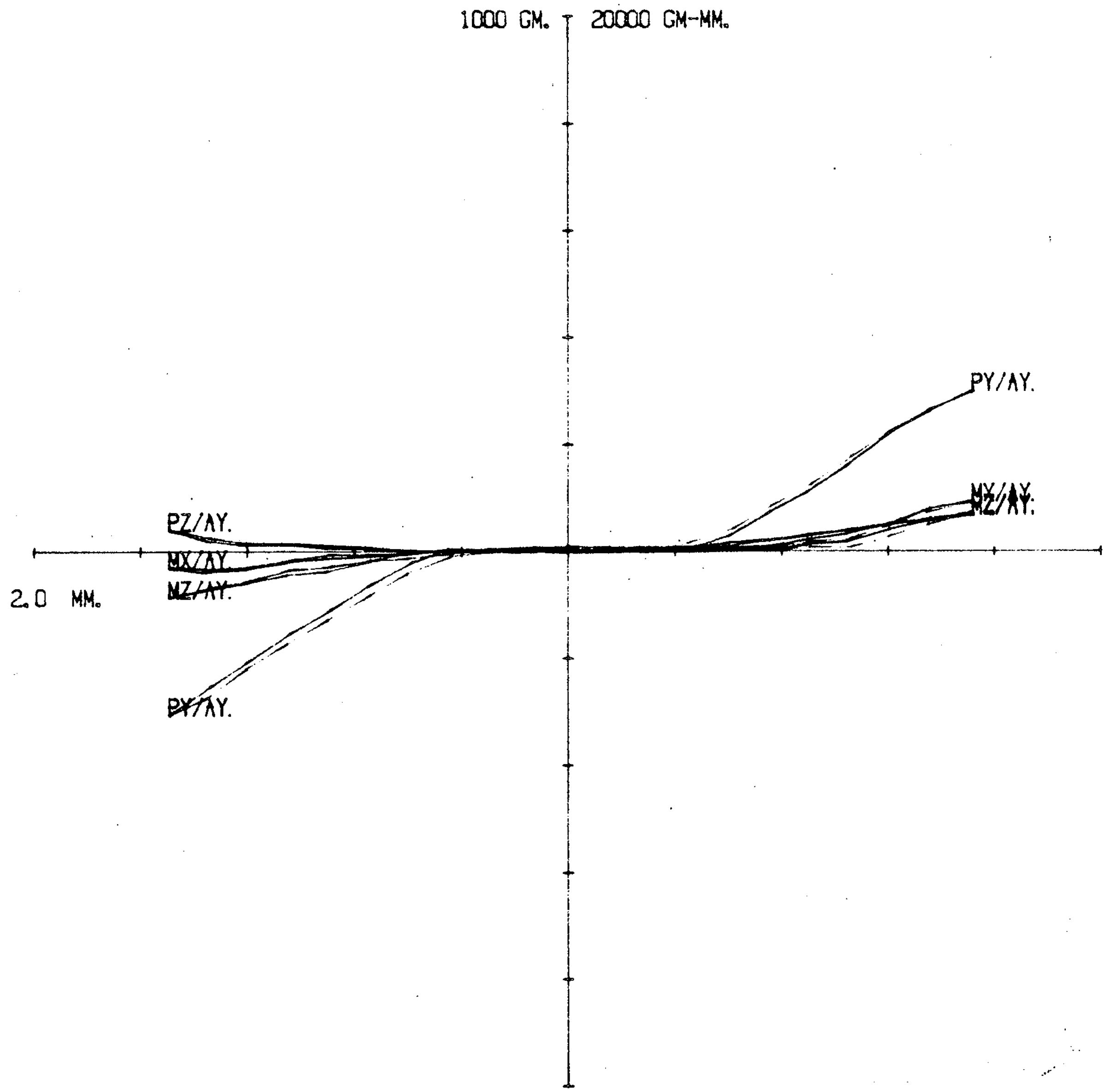


FIG. B19. NITINOL 0.41MM. (0.016INCH)
LARGE TUBES. REACTIVE TOOTH.
INTERBR. DIST. 3MM., 3MM., 3MM.

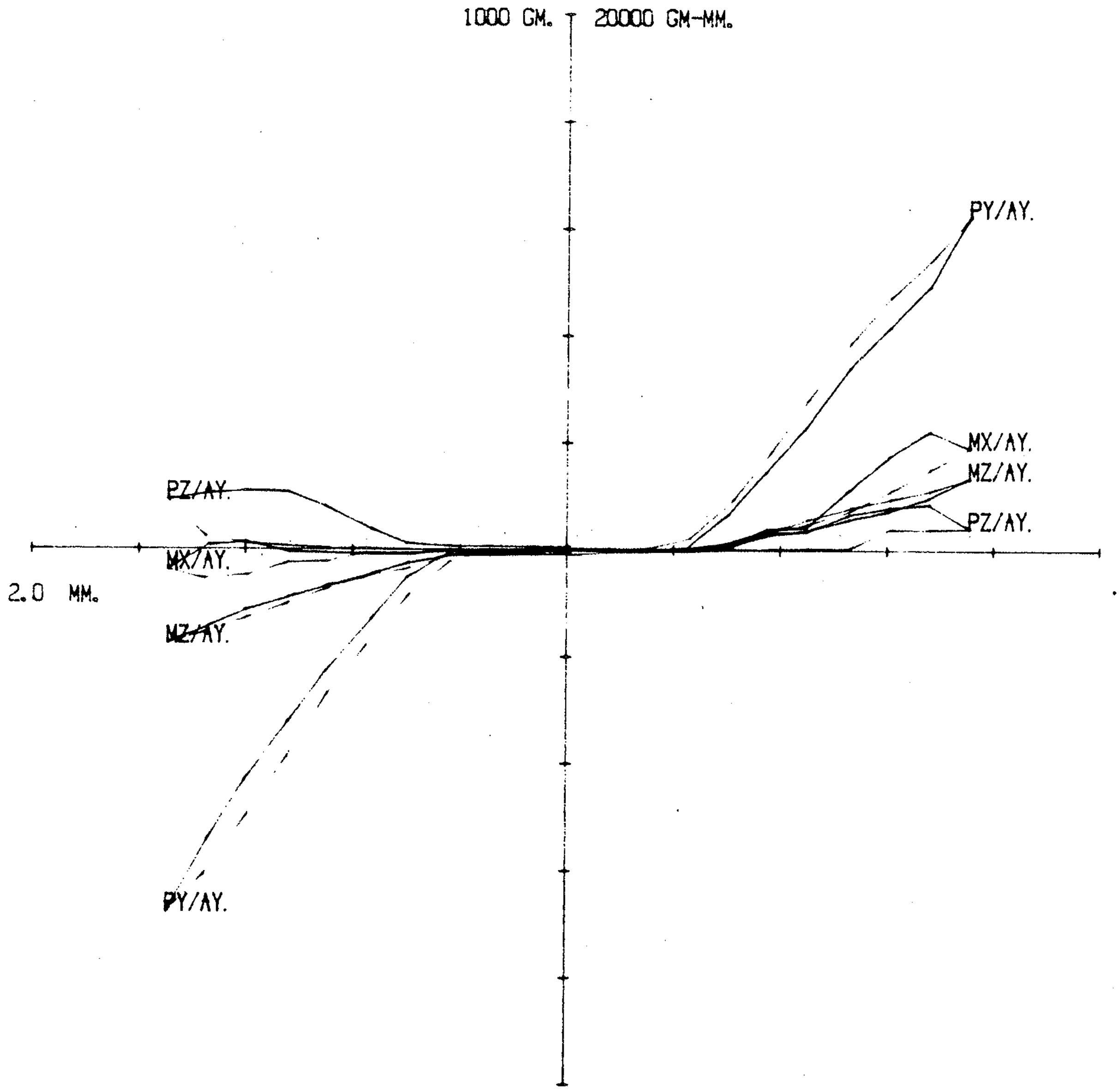


FIG. B20. NITINOL 0.41MM. (0.016INCH).
LARGE TUBES. REACTIVE TOOTH.
INTERBR. DIST. 3MM., 2MM., 4MM.

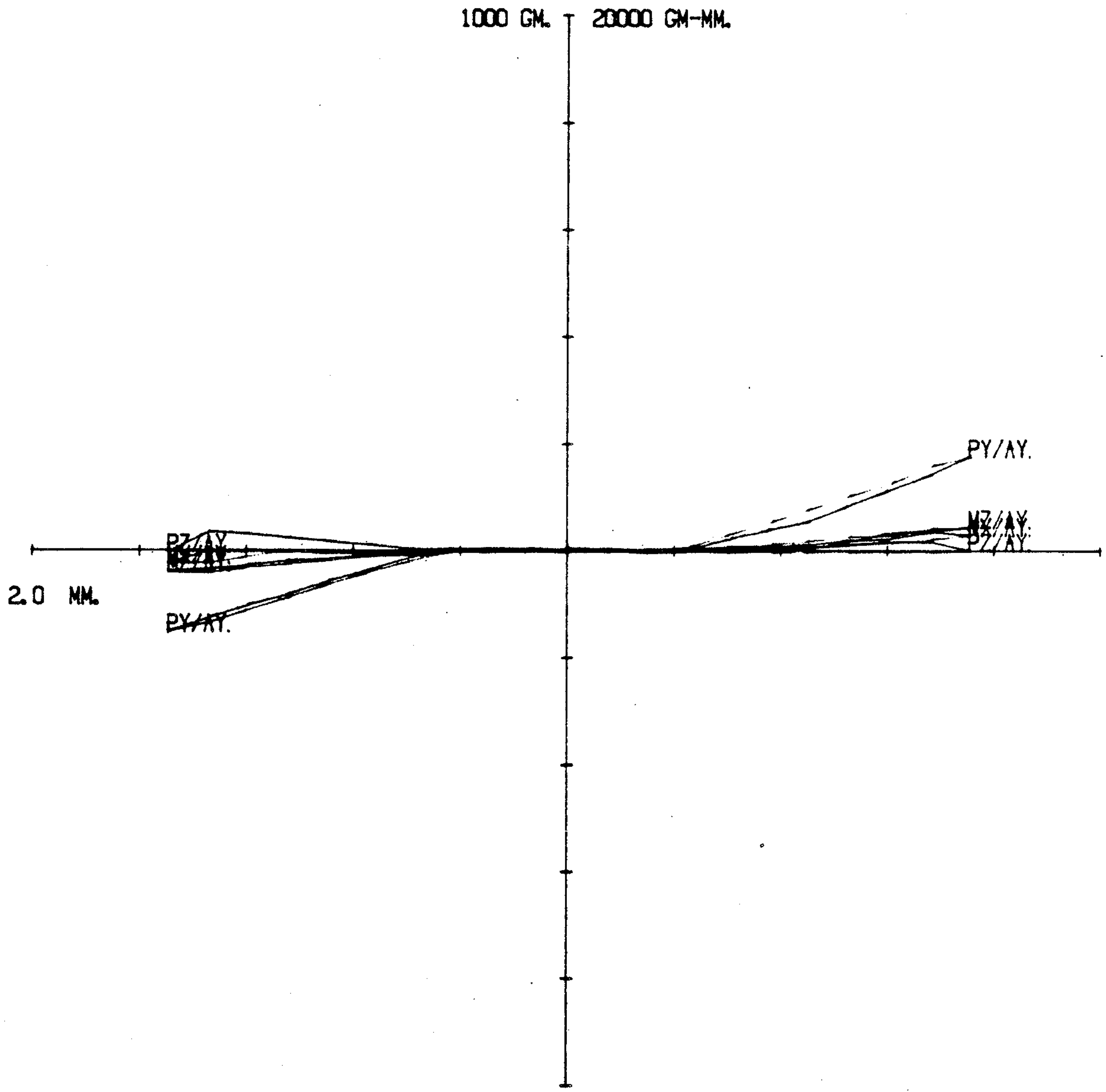


FIG. B21. NITINOL 0.41MM. (0.016INCH).
LARGE TUBES. REACTIVE TOOTH.
INTERBR. DIST. 3MM., 4MM., 2MM.

Appendix C

The Effect of Interbracket Distance on the Control of Bucco-lingual Tipping When Straight Wires Are Used.

The control over bucco-lingual tipping with straight rectangular archwires can theoretically be improved by increasing the interbracket distances. If the perpendicular distance between the center of resistance of the tooth and the point of force application (bracket position) and the type of wire used are known it is possible to calculate the interbracket distance required for bodily movement of the tooth.

A buccally directed force is generated at the bracket of a lingually malaligned tooth when a straight wire is tied in. This buccal force, W, creates a buccal tipping moment, M, at the center of resistance of the tooth:

$$M = Wd$$

where d is the perpendicular distance between the bracket and the center of resistance of the tooth.

If the material and cross-section and the length of wire are known the force W can be calculated:

$$W = \frac{\partial k E I}{D^3}$$

where k is a constant (which depends on the tightness of fit between the archwire and bracket); E is the modulus of elasticity of the wire; I is the second moment of the cross-sectional area of the wire; D is the interbracket distance; δ is the deflection at midspan under the load, W.

The archwire creates a countermoment T at the bracket on the tooth which, if we assume the angular deflection to be small, may be written as,

$$T = \frac{G\,J\,\delta}{D\,d}$$

where G is the shear modulus of the wire; and J is the polar second moment of area of the cross-section of the wire.

For the appliance to be able to move the tooth bodily the countermoment and the tipping moment it creates have to be equal in magnitude, or,

$$\frac{\delta\,k\,E\,I\,d}{D^3} = \frac{G\,J\,\delta}{D\,d}\ .$$

Thus, the wire length, D, required for the moment to force ratio to equal zero can be calculated from

$$D = \sqrt{\frac{d^2\,k\,E\,I}{G\,J}}\ .$$

If an 0.41 x 0.56 millimeter (0.016 x 0.022 inch) edgewise stainless steel archwire is used in tight brackets and the distance between the center of resistance of the tooth and the bracket is 12.5 millimeters and the bracket width is 4 mm. the interbracket distance required for bodily movement of the tooth is 107 mm. If the same wire is used ribbonwise the interbracket distance required is 74 mm.