Testing the Constant Force Hypothesis: A Clinical Study of Cuspid Retraction Using Magnets

BY

JOHN DASKALOGIANNAKIS

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TESTING THE CONSTANT FORCE HYPOTHESIS:
A CLINICAL STUDY OF CUSPID RETRACTION USING MAGNETS

BY

JOHN DASKALOGIANNAKIS

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

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DEDICATION

To my parents, George and Effrossini Daskalogiannakis, and my sister Emmy, for their many sacrifices which made my education possible.

In loving memory of my grandmother Eumorphia Banagi.

ABSTRACT

During the clinical practice of orthodontics, movement of teeth is being performed under the influence of forces of variable magnitude, duration and rate of decay. The characteristics of an optimal force for tooth movement remain poorly understood. The objective of this study was to test the validity of the hypothesis that a constant force of long duration provides more effective tooth movement than a rapidly decaying impulsive force.

Volunteer patients were selected from the screening program of the University of Manitoba, Graduate Orthodontic Clinic on the basis of needing extraction of their maxillary first bicuspids. Cuspid retraction on these patients was performed on one side with the application of a force rapidly declining in magnitude, produced by a regular vertical loop, and on the contralateral side with the application of a relatively constant force. This type of force was achieved using a similar vertical loop which was constantly activated by three Parylene-coated Neodymium-Iron-Boron ($Nd_2Fe_{14}B$) block magnets, 2 x 3 x 5 mm in size, which were fixed on wire segments in the buccal vestibule. Re-activation of the vertical loop on the control side was performed six weeks after the initial activation. No re-activation was necessary on the experimental side for the duration of the experiment. Comparison of the velocity of tooth movement between the two sides, over a period of approximately three months, was undertaken on the basis of maxillary impressions which were taken at frequent intervals during the course of the study. Statistical analysis of the data was performed using the paired t-test.

Under the conditions of the study, the cuspids retracted with a constant force moved statistically significantly more than the control cuspids ($p<0.05$), during the experimental period. The average differences in the mean rates of tooth movement between the two sides were in the order of 2:1 in favour of the experimental side. There were no statistically significant differences in the changes of angulation (tipping) or rotation about the occluso-gingival (y-) axis, between the two sides. No appreciable movement was observed, on either side, until after an initial period of 20 to 30 days of force application. Instead, a series of changes in angulation of the teeth seemed to take place during this period.

The factors identified in this study as being critical in regulating rate of tooth movement were the duration of force application and the total energy available for tooth movement by the retraction mechanism. Conversely, magnitude of the applied force was not found to be of primary significance.

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SYMBOLS AND ABBREVIATIONS

alnico	Aluminum-Nickel-Cobalt
B	Flux Density of a magnetic field
ß-Ti	beta-Titanium
BH	Energy Product of magnetic field
CEJ	Cemento-Enamel Junction
CL1-3	Left Cuspid reference point 1-3
cm	centimetres
CR1-3	Right Cuspid reference point 1-3
CR	Centre of Resistance
°	degrees
F	Force
F_s	Static Friction
g	grams of force
g-mm	gram-millimetres
H	Magnetic Field Intensity
L	Three-dimensional Hypotenuse
LVDT	Linear Voltage Displacement Transducer
mm	millimetres
μm	micrometres
MRI	Magnetic Resonance Imaging
μ_s	Coefficient of Static Friction
N	Normal Force
$Nd_2Fe_{14}B$	Neodymium-Iron-Boron

Ni-Ti	Nickel-Titanium
PDL	Periodontal Ligament
Pt-Co	Platinum-Cobalt
Sm-Co	Samarium-Cobalt
SEM	Scanning Electron Microscopy
T	Tesla
TMA	Titanium-Molybdenum Alloy

CHAPTER 1

INTRODUCTION

INTRODUCTION

The entire specialty of Orthodontics is based upon the time-tested observation that teeth can be moved from one position in the dental arch to a new location by means of applied mechanical forces. Clinical experience has shown that movement will occur in response to forces of varying magnitudes, points of application and temporal patterns. The identification of an optimal orthodontic force remains elusive despite the investigations into numerous hypotheses generated within the specialty.

The desire to move teeth at a rapid rate, without patient discomfort or tissue damage, and with maximal directional control, has been a traditional one. Nowadays, fixed appliances are used widely, applying more or less continuous forces, in order to accomplish desirable tooth movement. This method is considered state-of-the-art, and a large industry has evolved around the production of various appliances and exotic wire alloys that facilitate the constant application of forces to teeth.

The disagreement between previous attempts to support this clinical trend with scientific evidence can be only partially explained by the large variability in biologic response to external stimuli. A significant contribution to the existing confusion was made by the fact that most studies were inadequately controlled for the characteristics of the applied force system (i.e. the magnitude of the force, the duration of time it is applied, its direction and the distance over which the force is active) and the type of tooth movement desired. In most cases, no effective methods of force calibration or measurement of tooth movement were available, resulting in inability to produce meaningful quantitative data. Knowledge of the characteristics of the force system applied to a tooth, at any given time as the tooth moves, is essential in order to determine what type of force system produces optimal tooth movement.

In the present study, the techniques utilized by Duff (1987), Sonya (1987) and Cohen (1991) have been modified to compare the rate of tooth movement in response to two dissimilar force systems: an "impulsive" force system (which exhibits a rapid rate of decay of the force with movement) and a "steady" force system (in which the force is maintained in relatively constant levels during movement of the tooth). In addition, methods of generating "impulsive" and "steady" force systems were introduced. An attempt was made to correlate the differences in the characteristics of the external force to the elicited response (evaluated by the mode and rate of tooth movement). The comparison was carried out between the right and left side of the same subject, in an effort to control for inter-individual biologic variability.

It is hoped that the study will enhance our understanding of the features of an "optimal force" for tooth movement.

CHAPTER 2

REVIEW OF THE LITERATURE

REVIEW OF THE LITERATURE

2.1 CUSPID RETRACTION

The need for retraction of cuspids, after extraction of the first bicuspids, is one of the most common tasks with which orthodontists are faced. In the vast majority of cases the desired type of movement is bodily movement (translation) of the cuspid. Translation implies a uniform distribution of stress over one entire aspect of the periodontal ligament, which in turn induces uniform resorption of bone (Hixon *et al.*, 1970).

A single force can produce translation only if it is applied through the Centre of Resistance (CR) of the tooth. For a healthy tooth the CR lies beneath the alveolar crest. However, forces can be applied only via an attachment on the crown of the tooth. This technical difficulty dictates the application of a counter-moment at the bracket, in addition to the required force, in order to achieve translation of a tooth (Burstone, 1985).

Since calibration of the forces and moments generated by an appliance is difficult, and thus is rarely performed in a clinical setting, the precise application of the required counter-moment for translation of a cuspid is often impossible. As a result, variable degrees of distal tipping and distolingual rotation can be expected during cuspid retraction (Nikolai, 1975).

Many previous investigations (Hixon and Klein, 1972; Duff, 1988; Cohen, 1991) have demonstrated that most retraction devices do not produce true bodily movement, but rather a series of many small distal tipping and uprighting movements.

Two general categories of cuspid retraction techniques exist: those that involve friction and those that are frictionless (Farrant, 1976). Friction occurs in the bracket-archwire interface when sliding mechanics are used for cuspid retraction. Sliding mechanics make use of a continuous archwire and a variety of auxiliaries (e.g. coil springs, elastics etc.) to generate the required force. Burstone and Koenig (1976) felt that two main disadvantages are involved with sliding mechanics for cuspid retraction: 1) the friction produced may prevent tooth movement, and 2) determination of force magnitude is difficult, since friction can only be measured indirectly.

In order to avoid friction, a loop has to be utilized, on a segmented archwire. The segmented archwire usually extends from the mesial of the cuspid bracket to the distal of the tube on the first molar. The loop formed on this wire can have various shapes and configurations. The retraction force comes from creating a separation between the legs of the loop, either by tying the distal extension of the loop to the molar attachment using hooks and ligature wire, or by cinching the loop wire distal to the molar tube.

There are three important characteristics that describe a frictionless retraction spring (Burstone and Koenig, 1976):

1) The ratio of the counter-moment to the retraction force is the most important characteristic, since it is this ratio that determines the position to which a tooth will move (i.e. whether the tooth will tip or translate),
2) The force developed by a specific activation of the spring,
3) The rate of decay of the generated force, i.e. the force/deflection ratio.

In summary, cuspid retraction is a routine procedure in many orthodontic treatment plans. Although most often bodily movement of the cuspids is desired, due to our inability to apply the required force system with precision, the cuspid usually experiences a series of tipping and uprighting movements during retraction. Retraction techniques can be divided in two large categories, namely those that involve friction (e.g. sliding mechanics) and those that are frictionless (e.g. segmental retraction loops).

2.2 FRICTION BETWEEN ORTHODONTIC BRACKETS AND WIRES

Friction is defined as a force that resists the relative motion of two bodies in contact. Its direction is tangential to the common boundary of the contacting surfaces. The classical physics model depicts the maximum value of the *static* (before motion) frictional force as the product of a coefficient of static friction and the resultant normal force ($F_s = \mu_s$ x N); the *kinetic* (during motion) frictional force is defined similarly but involves a kinetic coefficient. Normal force (N) is the component of the total contact force between the surfaces, which is perpendicular to the frictional component. The coefficients of static (μ_s) and kinetic (μ_k) friction, generally having magnitudes between zero and one, depend on the relative roughness of the contacting surfaces.

The classical laws of friction state that a frictional force is:

1) proportional to the normal force acting perpendicular to the area of contact,
2) independent of the area of contact and
3) independent of the sliding velocity.

Several studies have, however, challenged the validity of these laws, thus great controversy exists with regard to the variables that can significantly affect friction. The results from these studies must be interpreted with caution since the experimental conditions can play a critical role in the outcome of a study of this type.

Investigators seem to agree that frictional resistance is significantly higher in ceramic brackets (monocrystalline or polycrystalline alumina, or zirconia) than in stainless steel brackets for most wire size - alloy configurations, regardless of slot size (Angolkar *et al.*, 1990; Pratten *et al.*, 1990; Bednar *et al.*, 1991).

With respect to the frictional resistance with different bracket widths, Kamiyama *et al.* (1973), Tidy (1989), and Drescher *et al.* (1989) showed that friction is inversely proportional to bracket width. To the contrary, Frank (1979) and Kapila *et al.* (1988 and 1990) reported an increase in friction associated with wider brackets and Andreasen *et al.* (1970) described no relationship between these quantities.

Bracket-to-archwire angulation is considered by almost all investigators as a factor that substantially influences the magnitude of friction. As would be expected, increased tipping results in increased friction (as the normal forces produced by the wire on the mesial and distal wings of the bracket are higher) (Andreasen *et al.*, 1970; Echols, 1975; Tidy, 1989).

In a clinical situation, the effect of bracket/wire angulation on friction should be examined in correlation with bracket width, slot size, wire size, wire material and interbracket span, as all of these will determine the amount of tipping that the tooth will experience.

The order of coefficients for static and kinetic friction of archwires of different alloys against stainless steel or polycrystalline alumina bracket slots has been shown to be: stainless steel (lowest), Cobalt chromium, nickel-titanium, and ß-titanium (highest) (Kusy *et al.*, 1988; Drescher *et al.*, 1989; Angolkar *et al.*, 1990).

The findings by Kusy *et al.* (1990) may provide an explanation for the higher coefficients of friction of ß-Ti wires (such as TMA). By using SEM and energy dispersive x-ray analysis, these investigators demonstrated the occurrence of adhesive wear in the form of cold welding when ß-Ti wires slide through stainless steel bracket slots. Metal-metal bonds were being formed, broken and then reformed as the surface topography was undergoing modification, resulting in a "stick-slip" phenomenon. TMA wires are thus not recommended for sliding mechanics.

The potential role of saliva as a lubricant, is a subject of controversy. According to the findings of Andreasen *et al.* (1970) saliva plays an insignificant role in lubrication of the contacting surfaces. In contrast, Baker *et al.* (1987) showed artificial saliva to cause a statistically significant reduction in frictional force, of 15% to 19%. Other studies have found artificial saliva to actually increase the coefficient of friction (Stannard *et al.*, 1986; Kusy *et al.*, 1991).

Usually, frictional forces between wire and bracket are experimentally measured to be between 40 g to 300 g (Angolkar, 1990; Bednar *et al.*, 1991), but there have been reports of up to 1,500 g of friction (Andreasen *et al.*, 1970). These data would lead to the logical conclusion that it is impossible to move a tooth by sliding a bracket over an archwire because the forces needed to overcome friction are far beyond the limits of the forces created by orthodontic springs and elastics. Probably the saving grace in the clinical situation is the capacity of teeth to move to some degree within their alveolar sockets, which is enhanced by mastication and contact between upper and lower teeth. The resulting "jiggling" action may be clinically critical in significantly decreasing frictional forces and making tooth movement possible.

Friction between archwire and bracket depends upon the interrelation of many factors that cannot be separately evaluated. The considerable lack of agreement regarding the degree of influence of each one of these factors on the magnitude of friction is mostly due to different testing conditions. Anytime sliding mechanics are utilized for tooth movement, frictional forces will be implicated in a proportion of the motive force lost. This may be translated into lower tooth movement rates. However, probably due to agitation by mastication and contact of the teeth, friction is usually not clinically significant.

2.3 MAGNETS IN DENTISTRY

2.3.1 Applications

Magnets have been used for retention of prostheses since the early fifties (Freedman *et al.*, 1953) but it was the development of rare earth alloys in 1967 that launched their popularity. Over the past quarter century rare earth magnets have been the focus of numerous research efforts and clinical applications.

Magnets are routinely used in prosthetic dentistry, mainly for retention of overlay dentures, and are actually considered by some to be superior to precision and semiprecision attachments in that they provide adequate resistance to vertical displacement, yet they allow for slight lateral (horizontal) movement of the denture base. This favourable force distribution may be more beneficial periodontally for the long-term success of the prosthesis.

The use of magnetic force, as an alternative to traditional orthodontic force systems, was first demonstrated by Blechman and Smiley (1978) who bonded AlNiCo magnets to the teeth of adolescent cats to produce tooth movement.

The first *in vivo* experiments on orthodontic tooth movement produced by magnets showed decreased treatment time, good potential for force vector control, and no patient discomfort. These, coupled with the absence of friction and the fact that patient cooperation is not necessary, made magnets very appealing (Blechman *et al.*, 1978 and 1985; Muller, 1984; Kawata *et al.*, 1987; Dean, 1990). Magnets have been utilized in the treatment of orthodontic problems in all three planes of space.

In the vertical plane, rare earth magnets have been used in non-surgical treatment of patients with skeletal open-bites. Two pairs of magnets are embedded within separate maxillary and mandibular acrylic biteblocks, in the repulsion mode. The method has been credited with favourable clinical results, which have been attributed to intrusion of posterior teeth (Dellinger *et al.*, 1986; Kalra *et al.*, 1989; Kiliaridis *et al.*, 1990; Barbre *et al.*, 1991). Careful evaluations of long-term stability after treatment with these appliances are yet to be reported.

In a series of experiments on monkeys, Vardimon *et al.* (1989, 1990, 1994) have shown favourable response to functional appliances that use magnetic force to correct dentoskeletal malocclusions. The forces from rare earth magnets, embedded within separate maxillary and mandibular acrylic appliances in the attraction mode, were used to constrain the lower jaw in an advanced or retruded sagittal position.

Vardimon *et al.* (1987 and 1989) also utilized repelling rare earth magnets to achieve transverse expansion of the palate in monkeys.

For patients with borderline crowding, repelling magnets have also been used for distalization of molars as an alternative to extraction therapy (Gianelly *et al.*, 1988 and 1989; Itoh *et al.*, 1991; Bondemark *et al.*, 1992). In this case, advantages of magnets over the traditional techniques (i.e. headgear or springs) include easy insertion, good patient acceptance, rapid correction and little need for patient cooperation.

An attractive approach to the treatment of unerupted teeth has been described by Sandler *et al.* (1989, 1990 and 1991): A magnet is attached to the unerupted tooth, and a second, larger magnet is incorporated into a removable appliance. The system appears to be superior to the conventional technique (i.e. traction with an elastic or stainless steel wire) in that it is friction-free, requires very little adjustment and allows a better gingival contour. An apically repositioned flap is not necessary, since the magnetic forces can operate through the mucosa.

Springate *et al.* (1991) have reported the use of a pair of small neodymium-iron-boron magnets, bonded on the mesiopalatal aspect of the maxillary central incisors, to achieve fixed retention of midline diastema. This approach is considered preferable to the conventional fixed retention in terms of oral hygiene, because the patient is able to pass floss between the magnets and there are no wires passing close to the gingival margin. Moreover, the teeth are not physically splinted together, and thus differential loading of the crowns will not cause bond failure and subsequent dislodgement of the retainer.

2.3.2 Biological Effects

Early research with respect to intraoral applications of magnets showed an increased toxicity in the oral environment due to corrosion. This defect was overcome by isolating the magnets from the oral fluids. Methods of isolation include coating the magnets with a biologically inert material such as acrylic or embedding them in a non-magnetic, stainless steel case. Bondemark *et al.* (1993) recommended coating rare earth magnets with Parylene to ensure high biocompatibility for intraoral applications.

During the last few years interest has been focused upon cellular changes associated with the use of static magnetic fields. A depressed mitotic activity of sutural cells in rats (as revealed by a decrease of their thymidine uptake) has been registered, but with a tendency to rebound to normal (Camilleri *et al.*, 1990 and 1993). An increase in bone resorbing areas and a thinner epithelial cell layer in rats exposed to static magnetic fields have been demonstrated (Linder-Aronson, A. *et al.*, 1991). S. Linder-Aronson *et al.* (1991) also found increased bone resorbing areas in rat tibias with magnetic prostheses, in addition to a reduction in the number of epithelial cells under the areas where the magnets had been applied.

The significance of these findings, to this point, is rather obscure. It is questionable whether or not the reported cellular effects can be considered to be harmful and whether they are permanent or temporary. No conclusive evidence of harmful effects has been presented.

On the other hand, several studies have been released with no deleterious results being reported:

Cerny (1980) implanted Sm-Co magnets in the teeth, alveoli and perioral soft tissues of dogs and, after an exposure period of six months, found no clinical or microscopic evidence of tissue damage.

Esformes *et al.* (1981) conducted a series of experiments to determine the short-term biological effects caused by static magnetic fields that were created by Sm-Co magnets. Tissue culture studies of several cell lines showed no obvious effects on cell growth rate, morphology, or the ability to grow and remain confluent. In addition, no deleterious effects of the magnetic field were evident as a result of an *in vivo* study of wound and bone healing in rats.

Bruce *et al.* (1987) implanted Sm-Co magnets adjacent to induced radial fractures in adult rabbits. After 4 weeks of healing, the bone units were examined microscopically and mechanically. Significantly greater forces ($p<0.01$) were required to break those bone units exposed to magnetic fields. However, no significant difference was found between the longitudinal midcallus areas from magnetized and non-magnetized limbs.

Gonella *et al.* (1989) examined the biocompatibility of various coated and uncoated magnetic materials using cell cultures. Leaked lactic dehydrogenase was used as an indicator of membrane damage. They concluded that magnets in clinical use are not significantly cytotoxic.

Saygili *et al.* (1991) utilised two different microbiological methods to study the effect of magnets on the growth of pathogenic micro-organisms found in the oral flora. Neither of the two methods demonstrated magnetic effects upon colonization by the bacteria.

Altay *et al.* (1991) investigated the biological effects of magnets contained within titanium implants in the mandibles of dogs. After six months no evidence of pathology was seen upon microscopic examination.

Saygili *et al.* (1992) found no significant difference in human buccal mucosal blood flow between capillaries neighbouring dental magnets and controls, after 3, 7, 15, 30 and 45 days.

Sato *et al.* (1992) used SEM, microfluorometry of the nuclei and autoradiography to assess the effects of a non-homogeneous magnetic field (with max. magnetic flux density of 1.5 T) produced by Sm-Co magnets, on the growth of human cell cultures. No significant difference in the growth of HeLa cells or human fibroblasts (Gin-1 cells) was detected between experimental and control dishes.

Papadopoulos *et al.* (1992) studied the *in vitro* effect of static magnetic fields produced by attracting and repelling Neodymium magnets, on rat osteoblastic activity. Exposure of the cell cultures to the magnetic fields for 21 days did not affect the optical density, the orientation or the form of the osteoblasts.

In 1989, a World Health Organization (WHO) report suggested that static magnetic fields up to 2 T show no significant health effects. However, according to theoretical considerations, and some experimental study results, short-term exposure to static fields greater than 5 T may produce detrimental health effects.

A recent workshop on the effects of magnetic fields (*Ad Hoc* Working Group, International Agency for Research on Cancer, 1990) concluded that it is difficult to establish a link between magnetic fields and human cancer, since investigations at the basic molecular and cell structure level produce equivocal results.

In summary, an inherent problem associated with most rare earth magnets is their low resistance to corrosion, which makes them significantly cytotoxic in oral environment, unless they are coated with some inert material, such as Parylene. With regards to potential harmful effects of static magnetic fields on cells, no conclusive evidence has been presented so far. Some individual findings of this kind have been described, but their significance is unknown. On the other hand, most studies report no serious cellular changes caused by exposure to static magnetic fields.

2.4 TISSUE DAMAGE INCIDENT TO TOOTH MOVEMENT - ROOT RESORPTION

Root resorption occurs normally as part of the exfoliation of the primary dentition, but also may occur as an unpredictable sequela of pulpal or periodontal inflammation, or as an undesirable side-effect of orthodontic tooth movement. The biological mechanism for root resorption, remains poorly understood.

Previous studies on root resorption may be grouped under the following major areas (Farrell, 1990):

1) Resorption of primary teeth

2) Resorption of permanent teeth

   (i) Idiopathic root resorption

   (ii) Traumatic root resorption (including root resorption associated with endodontic, periodontic and orthodontic factors)

Several investigators have recognized root resorption as an inevitable and unwanted side-effect of orthodontic treatment (Reitan, 1947; Kvam, 1972; Rygh, 1974).

Kethcham (1927) was one of the first investigators to correlate root resorption with orthodontics. He reported that 21% of 500 cases treated orthodontically had radiographic evidence of root resorption. In another study where root resorption was assessed radiographically, Rudolph (1940), found 49% of 439 patients showing root resorption at the end of the first year of orthodontic treatment and 75% of the remaining 277 patients, at the end of the second year. The more recent studies on root resorption support a 50% to 100% incidence (Massler and Malone, 1954; DeShields, 1969; Hollander *et al.*, 1980).

Maxillary incisors seem to be most susceptible to root resorption, and maxillary and mandibular second molars least susceptible (Massler and Malone, 1954; Hollender *et al.*, 1980). A variety of orthodontic appliances have been implicated as predisposing a patient to root resorption. These include intrusive arches, Class II elastics, maxillary expansion, rectangular archwires (applying torquing forces on the teeth) and uprighting springs (Morse, 1970; Linge and Linge, 1983; Follin *et al.*, 1986).

Reitan (1951) claimed that interrupted forces caused less reduction in cellular activity than continuous forces, and thus caused less hyalinization and cementum resorption. The investigations of Harry and Sims (1982) and King *et al.* (1982) have suggested that the severity of root resorption increases as the magnitude and duration of orthodontic force application increases. Harry and Sims (1982) reported that the duration of a continuously applied force is more critical than the force magnitude in producing root resorption.

Once the resorption lacunae are formed they may follow one of several courses depending on ensuing events. If the orthodontic force is continuously applied, the resorptive process continues with the lacunae increasing in both size and number with time (Rygh, 1977; Williams, 1984). If, on the other hand, the orthodontic force is removed or decreased below a threshold level, the resorption lacunae repair and fill with cementum (Reitan, 1974). The exact mechanism through which this process is regulated remains unclear. Copeland and Green (1986) and Sharpe *et al.* (1987) determined that root resorption ceases following termination of active orthodontic treatment, provided that the retainers are not active upon insertion. Others observed, however, root resorptive activity continuing even after the removal of the stimulus (Barber and Sims, 1981; Harry and Sims, 1982).

Root resorption is a four-dimensional process, occurring in three dimensions of space, over time. The original studies of root resorption associated with orthodontic treatment suffer the disadvantage of being mainly radiographic investigations. In addition, most of them had limited standardization, and thus their results are more or less inconclusive. Even today true quantitative assessment of root resorption cannot be performed, but only crudely approximated. It is impossible to quantify accurately the initial three-dimensional shape or volume of the root once resorption has occurred.

2.5 TECHNIQUES OF MEASURING TOOTH MOVEMENT

Numerous *in vivo* and *in vitro* attempts to develop an accurate method of measuring changes in tooth position have been described in the literature.

Techniques of measuring tooth movement *in vivo,* usually involve the use of callipers (Burstone and Groves, 1961), a divider and a stone jig that fits over the anterior teeth (Andreasen and Johnson, 1967), or dial callipers, a T-bevel and an acrylic jig that caps the incisors and rests on the occlusal surface of the second premolars (Huffman and Way, 1983). These methods have inherent handicaps, including difficulty in selecting and locating accurate references for measurement. In addition, they require varying degrees of contact with the teeth, resulting in some risk of distortion, or displacement due to natural tooth mobility.

Simons (1924), was one of the first to develop a technique for three-dimensional measurement of tooth movement from study casts, using an instrument called the symmetrograph. Much later, van der Linden *et al.* (1972) described a similar technique using an instrument known as the Optocom. The Optocom consisted of a travelling microscope, mounted on a table that could be adjusted in two dimensions. Measurement in the third dimension required the re-orientation of the cast. The accuracy of the method was disputed by Moyers *et al.* (1976) who recognized a difficulty in accurate location of cusp tips due to the lack of depth perception resulting from a monocular eyepiece.

Stereophotogrammetry, a photographic technique which employs stereo-imaging, has been used to record contours from the casts of patients with cleft palate (Berkowitz and Pruzansky, 1968). The technique utilizes complex viewing techniques, which are not readily available (Speculand *et al.*, 1988).

The Reflex Metrograph is an optical plotter which was developed by Scott (1981) for three-dimensional measurements of dental casts. The instrument consists of an ordinary stereoscopic microscope, modified such that a small diameter light spot (5 - 20 μm) appears in the field of view. The cast is placed on a sliding table, underneath the microscope lens. The horizontal movement of the table allows measurement in two dimensions, while the vertical movement of the microscope gives the third coordinate. Measurement of the position of a marked point on the cast, is made by aligning the light spot, with the point in question. Coordinates in three planes of space are digitized and stored for analysis by a minicomputer. Reports on the accuracy of the instrument vary. Butcher and Stevens (1981) found the measurement error to be approximately 0.128 mm in the mesio-distal direction, 0.299 mm in the vertical direction and 0.353 mm in the bucco-lingual direction. On the other hand, Takada *et al.* (1983) concluded that even operators with no previous experience can determine specific points to an accuracy of $\pm$ 0.1 mm. Speculand *et al.* (1988) reported a standard error of less than 0.2 mm for linear distances. The same investigators evaluated the accuracy of a newer version of the same instrument, known as the Reflex Microscope, and found its standard error to be less than 0.15 for linear measurements (Speculand *et al.*, 1988).

Duff (1988) and Sonya (1988) developed a simple method for measuring dental casts, after orienting them in a standard way on a measuring table. The method involves that use of orientation splints that fit over the occlusal surfaces of the posterior teeth. Measurements are performed in three dimensions by use of a Vernier Height Gauge.

2.6 THE INFLUENCE OF VARIOUS FORCE CHARACTERISTICS ON THE RATE OF TOOTH MOVEMENT

The first recorded recommendation to use force for orthodontic reasons is credited to the Roman writer Aurelius Cornelius Celsus who suggested the application of finger pressure to erupting teeth for alignment purposes, around the year 1 A.D. (Weinberger, 1926). Since then, movement of teeth has been executed on the basis of clinical experience, using a wide range of appliances capable of producing various types of forces.

Oppenheim (1911) was one of the first to demonstrate the histological changes incident to tooth movement and to recommend the avoidance of high forces to limit destruction of the tissues. For the same reason, Schwarz (1932) suggested that orthodontic forces should be kept below the capillary blood pressure of the periodontal ligament (20 to 26 g/cm^2 of root surface). It was through such early experimental attempts that the concept of an optimal force for movement of teeth arose.

Two basic aspects of the "optimal force" will be discussed, namely its magnitude and its duration.

2.6.1 Magnitude of Applied Force

The orthodontic literature contains much speculation regarding the magnitude of an optimal force for tooth movement. A wide range of "ideal" magnitude values has been proposed. In addition, many investigators refer to "heavy" or "light" forces with no quantitative description, which makes evaluation of their findings and allegations even more difficult.

Storey and Smith (1952), compared cuspid retraction performed by light initial forces (175 - 300 g) to that under the influence of heavy forces (400 - 600 g). They found the optimum range of forces to be between 150 and 250 g. With forces below this range there was practically no movement of the cuspid tooth, whereas with heavier forces, much more anchor tooth movement occurred.

Begg (1956) based his differential force theory on the same observation. According to that theory, by varying the magnitude of the applied forces one could get increased movement of the anterior or posterior teeth. Thus, when retraction of cuspids was performed with light forces (in the 150 to 200 g range), maximum retraction and the most rapid rates of movement could be achieved, in addition to minimal anchorage loss. Conversely, relatively large forces caused the anterior teeth to resist the pressure and consequently act as anchor teeth, while rapid movement of the posterior teeth occurred.

Largely due to the meticulous experiments conducted by Reitan (1947, 1957, 1958, 1964) on human and animal material, we now have a comprehensive understanding of the histomorphological changes in the strained PDL. Reitan experimented with force levels and types of tooth movement for over four decades. He advocated the application of lighter initial forces during the first 5 or 6 weeks, when the compact bone lining the inner alveolar wall is being resorbed, since the tendency for formation of cell-free areas decreases after elimination of that dense part of the bone. During the secondary period of tooth movement, he claimed that a force within the range of 150 - 200 g is favourable for bodily movement of canines or premolars.

Burstone *et al.* (1961) used relatively continuous forces ranging from 25 to 150 g in a study designed to demonstrate the lower force thresholds for anterior tooth retraction by simple tipping. Optimal rates of tooth movement were observed in the groups that were subjected to forces of 50 - 75 g in magnitude.

Hixon *et al.* (1970) used latex elastics, changed three times per week on six patients for an experimental period of eight weeks, in an attempt to identify the optimum force magnitude for cuspid retraction. The elastics produced initial forces that ranged from 60 g to 1,037 g. Tantalum implants were placed in the jaws to provide fixed landmarks against which to measure tooth movement. Their conclusion was that within an individual patient heavier forces seemed to move teeth at a greater rate. However, variation in the individual response was too large to allow any universal conclusive statements to be made, with regard to force magnitude.

Gianelly (1971) recognized that the actual daily rate of tooth movement under the influence of heavy forces can be slower than that under the influence of light forces, reflecting a long delay period caused by hyalinization. However, the total amount of tooth movement produced by heavy forces over an extended period of time (2 - 3 months) may, in fact, be greater than that produced when lighter forces are used. He determined the optimal values of force to be 40 - 50 g for bodily movement of small teeth and 150 g for large teeth (such as cuspids).

Mitchell *et al.* (1973) conducted a study on fifteen mature cats on which they performed retraction of the maxillary cuspids using forces of 150 g and 400 g. Their results disagreed with the differential force theory, since there was more cuspid movement with less anchorage loss on the heavier force side. According to the same authors, the effects of individual variation, age, constancy of forces, and the type of tooth movement are just a few factors that need clarification before magnitudes of forces can be considered as a single factor.

Boester and Johnston (1974) in a clinical study on ten patients, used forces varying from 2 to 11 ounces (approximately 60 to 310 g) in order to retract the cuspids in all four quadrants. Their results also challenged Begg's differential force theory, since they found heavy forces as effective as light forces in space closure and anchorage preservation.

Nikolai (1975) reviewed the results of previous studies and different mechanical concepts. He proposed that the ratio of tipping force to bodily movement force for a maxillary canine is 1 : 3.5.

Proffit (1986) suggested that 50 - 75 g is the optimal range of forces if tipping is the desired type of tooth movement. For translation he recommended a force ranging from 100 to 150 g.

In a clinical study of cuspid retraction on four subjects, Cohen (1991) attempted to correlate tooth displacement, which was assessed three-dimensionally with the components of the applied force system, also measured in three dimensions. She observed a classic three-phase curve of tooth movement over time, and hypothesized that there exists not one optimal force for a given tooth movement, but rather an optimal force for each of the three phases of movement.

From the studies mentioned so far, it is evident that no agreement exists with regards to the magnitude of an optimum force for tooth movement. As Quinn and Yoshikawa (1985) pointed out, there are three major problems that complicate clinical studies of force magnitude and tooth movement: First, the type of tooth movement (i.e. tipping or translation) has not been adequately controlled. Second, because of the nonlinear, time-dependent course of tooth movement following appliance activation, measurements of tooth movement that are not coordinated with activations can systematically bias the data. Third, the large measurement errors, as well as the large variation in the rate of tooth movement both between patients and between quadrants in an individual patient, make results difficult to interpret in terms of significance.

2.6.2 Duration of Force Application

It has been shown (Weinstein, 1967) that a very low force, in the order of 2 g, which acts over a long period of time can move teeth. It is generally accepted, however, that there is a certain duration threshold for which a force must be sustained in order to produce movement. Experiments by Davidovitch *et al.* (1975) on cats showed that only after a force was maintained for approximately 4 hours did cyclic nucleotide levels in the PDL increase, indicating that this duration of stress is required to produce the "second messengers" needed to stimulate cellular differentiation. Proffit (1986) cited his clinical experience to support the idea that a force duration of approximately six hours is the lower threshold in humans. He suggested that increasingly effective tooth movement is produced if the force is maintained for longer durations.

Another aspect of force duration is related to how the force magnitude changes as the tooth responds by moving, namely the rate of decay of the force. According to Proffit (1986), on the basis of rate of decay, orthodontic force can be classified as follows:

- *Continuous* - a force whose magnitude is maintained at some appreciable fraction of the original, from one patient visit to the next.
- *Constant* - a continuous force of a constant magnitude (zero decay with tooth movement). This type of force remains theoretical and can only be approximated in clinical practice.
- *Interrupted* - a force whose magnitude declines to zero between activations.
- *Intermittent* - a force whose magnitude sporadically drops abruptly to zero (as is the case with forces generated by removable appliances).

Although the recommended magnitudes of force vary, most investigators contend that optimum tooth movement can be achieved if the forces used are light and continuous. Storey and Smith (1952) evaluated several cuspid retraction springs, with the working hypothesis that a light continuous force would be most effective in retracting a cuspid tooth. All of the springs examined were regarded to be unsatisfactory because their load/deflection ratio was too high.

Burstone (1961) claimed that the ideal orthodontic spring should have the ability to release a constant force throughout the entire range of its activation. He explained that this is not to imply that force levels should remain constant during a given type of tooth movement, but rather that sudden changes in force magnitude should be eliminated. In these terms a constant force spring would be one that approaches a zero load/deflection rate.

Pletcher (1958) stated that a light, continuous pressure will move the teeth farther, more easily and with less damage than any other method, provided that it can be properly controlled. He also devised a closed coil spring (referred to as a T-spring) that, due to its relatively low load-deflection rate, was alleged to produce higher rates of tooth movement for space closure.

Along the same lines, Gjessing (1985) designed a frictionless spring that was "capable of creating an optimal force for controlled canine retraction". The spring "fulfilled the biomechanical requirement of a low load-deflection rate, to secure the delivery of a constant tension in the periodontal membrane throughout an extended range of spring activation". The reported initial force of 160-180 g at activation would decline to zero after the tooth moved by approximately 3.5 mm (load/deflection rate: 45 g/mm of activation).

The efficacy of Gjessing's retraction spring was evaluated by Ziegler and Ingervall (1989) in a clinical study of bilateral cuspid retraction on 21 human subjects. The contralateral cuspid was retracted using sliding mechanics, by means of an AlastiK chain capable of producing an initial force of 380 g, which was changed every 3 to 4 weeks. The spring was activated according to the instructions of the manufacturer for an initial retractive force on the canine of 160 g, and was reactivated at the same time as the chain was changed. The comparison between the rates of retraction on the two sides favoured the side of the retraction spring, which demonstrated a rate of movement that was, on average, 0.5 mm per month faster than the side of the sliding mechanics. The authors attributed this result to the delivery of a more constant force by the Gjessing spring and/or the absence of friction.

Reitan (1985) recognized that intermittent forces or light continuous forces are necessary to minimize hyalinization.

Continuity of the applied force, however, is certainly not a prerequisite for tooth movement. In fact, numerous investigators have claimed that non-continuous forces should be preferred over continuous forces as more physiologic, since they give the PDL the chance to regenerate during the period of time that they are not active (Halderson *et al.*, 1953; Reitan, 1957).

More recently, there have been studies suggesting that pulsating forces of variable frequencies may be advantageous in inducing more physiologic tooth movement (Oates *et al.*, 1978; Shapiro *et al.*, 1979).

The desirability of forces of low magnitude that could be sustained over a long period of time during deactivation of the force-producing mechanism, led to the pursuit of new applications of the spectacular metallurgical advances which were achieved in the last decade. As a result, orthodontics was introduced to a variety of exotic wire alloys with low load-deflection rates, which can ensure that "the force magnitude is delivered more constantly, which allows the orthodontist to approach optimal forces and to negate excessive and subthreshold force zones" (Burstone, 1981).

Burstone (1985), in an article introducing the Chinese NiTi alloy, comments that "The unusual nonlinear loading curve builds into the Chinese NiTi wire a constant force mechanism in the middle range of deactivation. This is potentially a significant design feature for constant-force appliances."

One year later, Miura *et al.* (1986) presented the orthodontic world with the Japanese NiTi wire, possessing the extraordinary property of super-elasticity in addition to shape memory. They remarked "The clinical application of wires of this new alloy should be more likely to generate a physiologic tooth movement because of the relatively constant force delivered for a long period of time during the deactivation of the wire".

Evaluation of the forces generated by NiTi coil springs showed that they could deliver a relatively constant force over a range of 7 mm of tooth movement, with one activation (von Fraunhofer *et al.*, 1993). The authors also suggested that the forces produced by such coils are within the range of 75 - 100 g, proposed by Storey (1954) as optimal for tooth movement.

Several types of force systems have been evaluated in the literature in search of the one that moves teeth optimally. The examined forces have been applied intermittently or interruptedly, at various frequencies. However, the continuous application of a force with a low, constant magnitude is still considered ideal for tooth movement, and is consequently the objective of many treatment modalities. Nevertheless, the great majority of appliances and procedures used in contemporary orthodontics are only capable of delivering forces of impulsive nature.

CHAPTER 3

STATEMENT OF THE PROBLEM - RATIONALE FOR THE STUDY

STATEMENT OF THE PROBLEM - RATIONALE FOR THE STUDY

It has been repetitively acknowledged that mechanical strategies in orthodontics are largely drawn from clinical empiricism. This may well be a method for the attainment of successful results, but not for the acquisition of knowledge.

The important question of how the magnitude and the temporal distribution of force from orthodontic appliances influence the rate of tooth movement has received relatively little experimental study in humans. The data from a substantive portion of the studies which have been undertaken are difficult to interpret in three-dimensional space. As a result there is still no clear consensus on the most efficient way to move teeth.

Common to most of the studies reported in the previous review is a lack of three-dimensional determination of the force system applied and the type of tooth displacement achieved. This fact, coupled with the individual variability in the response that is expected from biological tissues, has resulted in significant divergence of opinion. With the existing evidence there are limitations in postulating any simple relationship between applied force and tooth movement. Nevertheless, calibrating the components of the applied force and reducing the measurement error by having an accurate technique of three-dimensional assessment of tooth position can only help in discerning patterns and correlations that would otherwise be overshadowed by variation.

The objective of this study was to test the hypothesis that a higher rate of orthodontic translation could be induced by delivering a constant force to a tooth than by using a system which delivers an interrupted force.

CHAPTER 4

MATERIALS AND METHODS

MATERIALS AND METHODS

4.1 INTRODUCTION

Most orthodontic appliances that are used in everyday practice are capable of producing forces of either intermittent, or interrupted nature. In order to compare tooth movement under the influence of a constant force to that produced by regular orthodontic appliances, it was necessary to design a mechanism that could deliver a force of constant magnitude.

A series of preliminary *in vitro* experiments was conducted, using a machine developed by Paquien (1978) which was capable of simultaneously measuring forces and moments in three dimensions. Forces from 0 to 1,000 grams and moments from 0 to 20,000 gram-millimetres could be measured by the machine which is described in more detail in Appendix B.

Assessment of the force-separation relationship for different sizes of Neodymium-Iron-Boron magnets was performed on the machine, to decide on the desired magnet size for the intended application.

Subsequently, the feasibility of utilizing three magnets in order to combine their force/separation curves was investigated. More specifically, it is well known that the force produced by the magnetic field between two poles increases, as the distance (air-gap) between these two poles decreases. The same is true whether the poles are in attraction or in repulsion mode. In fact, according to Coulomb's law, the generated force is approximately proportional to the square of the air-gap (Appendix A). Thus, the force/distance relationship between two magnetic poles that are separated along a linear path parallel to their magnetization axes is best represented by a hyperbolic curve.

The idea in this study was to arrange the magnets in such a fashion that the middle magnet would be repelled by the magnet on one side and attracted by the magnet on the other side. The rationale for this was to combine the two hyperbolic curves produced by each of the two pairs of magnetic poles (the two in attraction and the two in repulsion), in order to produce a region in which the force on the middle magnet would be reasonably constant.

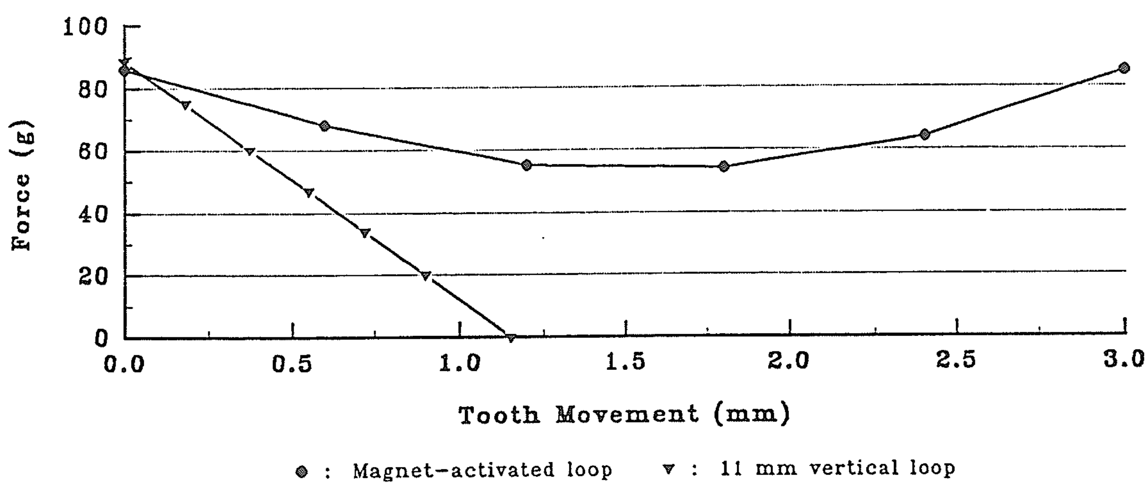

Figure 4.1 The force produced by a typical 11 mm TMA vertical loop decays linearly with deactivation. Conversely, the magnet-activated vertical loop offers the possibility of a relatively constant force over a range of 2.5 to 3 mm.

Evaluation of the forces generated on the middle magnet was performed on the force measuring machine at various separations. The results showed that, with the appropriate separation, this arrangement of magnets was capable of generating forces that varied by approximately ±20% of the mean (from 54 to 86 g) over a distance of 2.5 mm to 3 mm. This is a force with dramatically different characteristics from the forces developed by an ordinary vertical loop (see Fig 4.1), which typically shows a fairly rapid rate of decline after the initial activation of the loop (high load-deflection ratio). Based on this difference, a bilateral comparative study was designed. Retraction of the maxillary cuspids would be undertaken on the same subject using a conventional vertical loop on one side and a magnet-activated identical vertical loop on the other side. Comparison of the rate of tooth movement between the two sides could potentially give some insight into the "constant force theory".

Cuspid retraction was the tooth movement chosen for this study for the following reasons:

1) Cuspid retraction is a common procedure in orthodontics, which requires the tooth to move over relatively large distances. As a result, measurement of tooth movement can be performed more readily and with a smaller percent error.

2) Some data on cuspid retraction using similar retraction mechanisms are available from previous studies (Lack, 1980; Duff, 1987; Cohen, 1991)

Generally speaking, for magnets fabricated from a given material, the force produced is proportional to the volume of the magnet. Until recently, magnets had limited intraoral applications due to the size required to achieve desired forces. Neodymium-Iron-Boron magnets were chosen for this study mainly because they can provide the highest energy product per unit volume of any commercially available magnetic material. These magnets permit the delivery of forces appropriate for orthodontic practice, using appliances small enough for intraoral application.

The magnetic method was given preference over some other way of producing a constant force (such as super-elastic coil springs), because in such case, retraction of the cuspid would have to be done using sliding mechanics. In that case one would be relying on the archwire for the generation of the required counter-moment in order to avoid distal tipping of the teeth. In addition, friction between the bracket and archwire, which is difficult to assess, would complicate the evaluation of the results (see Section 2.2).

In the studies by Cohen (1991) and Duff (1988) vertical loops were used for retraction of cuspids and were found to be reliable in producing the intended movement with good control of tipping and rotation, as well as enough constraint to avoid all unwanted movements. The use of a vertical loop for cuspid retraction offers the advantage of ease and accuracy in recording the magnitude of the retraction force applied to the tooth at any time.

4.2 PATIENT SELECTION

Six subjects participated in this study. The subjects, selected from the screening program at the Graduate Orthodontic Clinic of the University of Manitoba, were chosen according to the following criteria:

1) A need for bilateral maxillary cuspid retraction (extraction of the maxillary first bicuspids) as part of the treatment plan.

2) The presence of the maxillary second molars, which also had to be completely erupted for use as anchorage reinforcement.

3) The absence of low-attached frena that would interfere with the vestibular placement of the appliance.

4) Relatively good arch alignment with no major rotations or cross-bites.

5) Willingness of the patient to tolerate the appliance and the numerous impression and intraoral photography procedures.

All the patients signed a consent form, approved by the Committee on Research Involving Human Subjects of the Faculty of Dentistry, University of Manitoba, which allowed the information that was to be gained from subsequent records to be used for research purposes (Appendix C).

The type of extraction undertaken on each patient participating in the study is shown in Table 4.2.

PATIENT	TYPE OF EXTRACTION
C.O. (♀)	4 first bicuspids
N.S. (♂)	2 maxillary first bicuspids
P.M. (♀)	4 first bicuspids
J.D. (♀)	4 first bicuspids
D.B. (♀)	maxillary first & mandibular second bicuspids
D.R. (♂)	4 first bicuspids

Table 4.2 The extraction patterns of the patients participating in the study.

In two of the cases with four bicuspid extractions (C.O. and D.R), retraction of the mandibular cuspids was undertaken at the same time as the experiment so that the patients would not suffer undue delay of their treatment as a result of participation in the study. In the case of J.D., the mandibular left second bicuspid was allowed to erupt further, after space was created for it with the extraction of the first bicuspid. In two cases (P.M. and D.B.), because bonding a bracket on the mandibular cuspids would cause an occlusal interference with the maxillary cuspids, maxillary cuspid retraction was initiated before mandibular cuspid retraction.

4.3 APPLIANCE DESIGN

The magnetic force delivery system consisted of three sintered Neodymium-Iron-Boron ($Nd_2Fe_{14}B$) block magnets (Magnet Developments Ltd., Swindon, England), 2 x 3 x 5 mm in dimensions, axially magnetized and pre-coated with Parylene C[1].

The mesial and distal magnets were attached to a vestibular wire segment whose distal end fitted into a vertical tube soldered on the buccal segment wire, between the first and second maxillary molars. The mesial extension of the vestibular wire was bent to fit passively in the bracket slots of the maxillary central and lateral incisors. The magnet-bearing portion of the wire lay parallel to the occlusal plane, at a distance

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1 Parylene C: A conformal polymer coating which is used to provide corrosion resistance and dielectric protection. Parylene is the generic name for members of a unique polymer series developed by Union Carbide Corporation. Nova Tran Corporation applies this coating uniformly by vacuum deposition at ambient temperature. The thickness of the coating in the present application was approximately 50 microns.

approximately 8 - 10 mm apically to the occlusal surfaces of the maxillary teeth, and 1 - 2 mm buccally to the buccal gingival wall (See Figure 4.3).

The distance (air-gap) between the mesial and distal magnets was 9.25 mm. This had been determined from the preliminary experiments on the force measuring machine to be the appropriate air-gap that would produce on the middle magnet a distally directed force with a mean magnitude of 65 - 75 g.

The middle magnet was attached on the distal extension of the loop that was intended to retract the maxillary right cuspid. The distal extension of the loop was bent so that the magnetic axis of the middle magnet would be approximately on the same line as the two fixed magnets. The middle magnet was oriented in such fashion that it would be repelled by the mesial and attracted by the distal fixed magnet. This distal force on the middle magnet would activate the loop for retraction of the cuspid. As the tooth moved distally, the constant force produced on the middle magnet by the field would maintain the opening of the loop. Therefore, no activation of the loop by cinching the wire distal to the auxiliary tube of the first molar was necessary on the experimental side. In fact, the extension of the loop wire distal to the middle magnet was free to slide through the auxiliary tube, under the influence of the magnetic force.

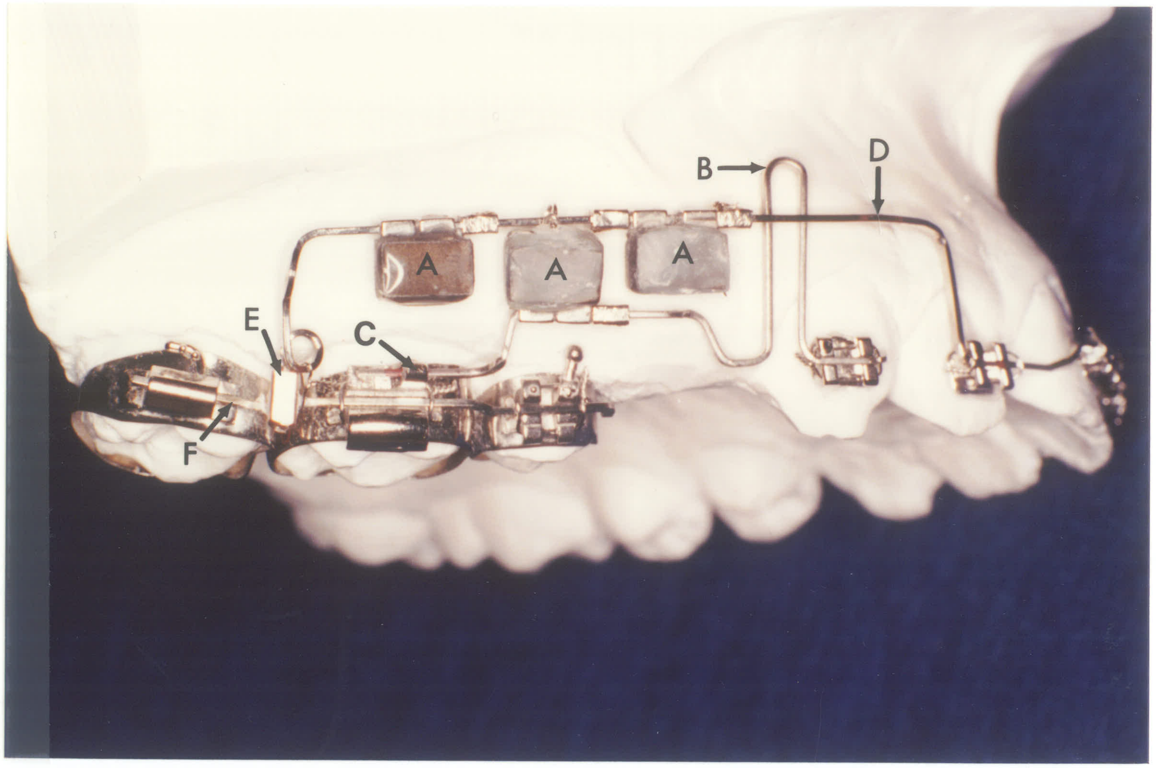

Figure 4.3 The constant force delivery system: The middle one of the three rare-earth block magnets (A) is attached on the vertical loop retraction spring (B), which is free to slide through the modified auxiliary tube (C) of the first molar band. The remaining two magnets are fixed on the vestibular wire (D). The distal end of the vestibular wire fits into a vertical tube (E) soldered onto the anchorage segment wire (F).

Identical loops 11 mm in height, were fabricated for both sides for reasons of standardization. This height was determined as a reasonable compromise between a loop height that was likely to be tolerable to the patient, and one capable of generating an appropriate moment-to-force ratio to minimize distal tipping.

The appliance described above was designed to be capable of delivering a long, "gentle" force for cuspid retraction. In addition, the appliance had maximum flexibility along the direction that movement of the cuspid was intended, but at the same time it had maximum rigidity in all other directions (provided by the vertical loop). This property was critical in avoiding spurious movements.

4.4 *IN VIVO* EXPERIMENTAL PROTOCOL

In all patients maxillary second bicuspids and first and second molars were banded, using .018 inch slot size set-up (American Orthodontics[2]) on the left, and .022 inch set-up ("A"-Company[3]) on the right side. This was necessary to permit the distal extension of the loop on the magnet side (which was made of .017 x .025 inch wire) to slide more easily in an .022 inch molar auxiliary slot than in an .018 inch slot. The same slot size was used for the ipsilateral second bicuspid and second molar to provide maximum slot engagement and thus allow the passive segments to serve as rigid anchorage units. In addition, the auxiliary slot of the right first molar band was shortened to 1.0 - 1.5 mm long, to further facilitate sliding of the distal extension of the loop wire (see Figures 4.3.A and 4.4.A).

A combination of a Nance/TransPalatal Arch was soldered on the maxillary first molar bands of all patients for anchorage purposes. The five patients who had extractions in their mandibular arch had a fixed lingual arch attached on their mandibular

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[2] American Orthodontics: P.O. Box 1048, Sheboygan, WI 53082-1048

[3] "A"-Company: P.O. Box 81247, San Diego, CA 92138-1247

first molar bands, as well. The left maxillary posterior anchorage segment was established by inserting a stainless steel wire, .017 x .025 inch in dimensions, so that it would passively fit in the attachment slots of the second bicuspid, the first and the second molar. On the right (experimental) side a passive stainless steel wire .021 x .028 inch in dimensions was inserted. After the passive segments had been inserted, the extractions were performed one week to ten days before insertion of the magnetic appliance.

Two appointments were allowed for insertion of the appliance in each patient, due to the lengthy chairside procedures that had to be performed. In the first session the six maxillary anterior teeth were bonded, and a stainless steel tube with a slot size of .018 x .025 inch was soldered onto the right passive segment wire, between the first and second molar.

The vestibular segment was then fabricated out of stainless steel wire (.017 x .025 inch). As already mentioned, the distal end of this segment would fit into the soldered tube between the first and second molar. The mesial end of the latter segment was bent to passively fit the slots of the brackets of the anterior four teeth. Particular care was taken to adapt the wire segment to the existing anatomy of the vestibule, so that it would not aggravate the soft tissues of the cheek, nor impinge on the buccal gingiva.

Each magnet was attached to two serially placed, crimpable surgical hooks (American Orthodontics) by means of Jet cold cure acrylic[4], so that when a wire was inserted through the tubes of both hooks, the long axis of the magnet (which was also the

[4] Jet Acrylic: Lang Dental Mfg. Co. Inc., 175 Messner Dr., Wheeling, IL 60090

magnetic axis), would be parallel to the wire. The spherical parts of the surgical hooks had been previously bevelled to facilitate parallelism.

The mesial and the distal magnet were fixed on the vestibular segment in repulsion mode, with an air-gap of 9.25 mm. After crimping the hooks initially to fix the magnets on the wire, additional cold cure acrylic was used to encase the junction areas between the magnets and wires, to ensure maximum retention. Moreover, "corner-braces" with stainless steel wire were soldered in the mesial and distal 90° bend of the segment, to make it less flexible (see Figure 4.4.a).

A vertical loop, 11 mm in height, was fabricated from TMA[5] wire (.017 x .025 inch) and its distal leg was spot-welded onto a piece of stainless steel wire of the same dimensions, in order to increase its stiffness. The middle magnet was placed on this stainless steel part of the loop. The orientation of the middle magnet was such that it would be repelled by the mesial magnet and attracted by the distal one.

An identical 11 mm vertical loop, of .017 x .025 inch TMA wire, was fabricated at the same time to retract the contralateral (left) cuspid. The distal leg of the latter loop was also spot-welded to a piece of stainless steel wire of the same dimensions.

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[5] TMA: ORMCO Corp., 1332 South Lone Hill Avenue, Glendora, CA 91740-5339

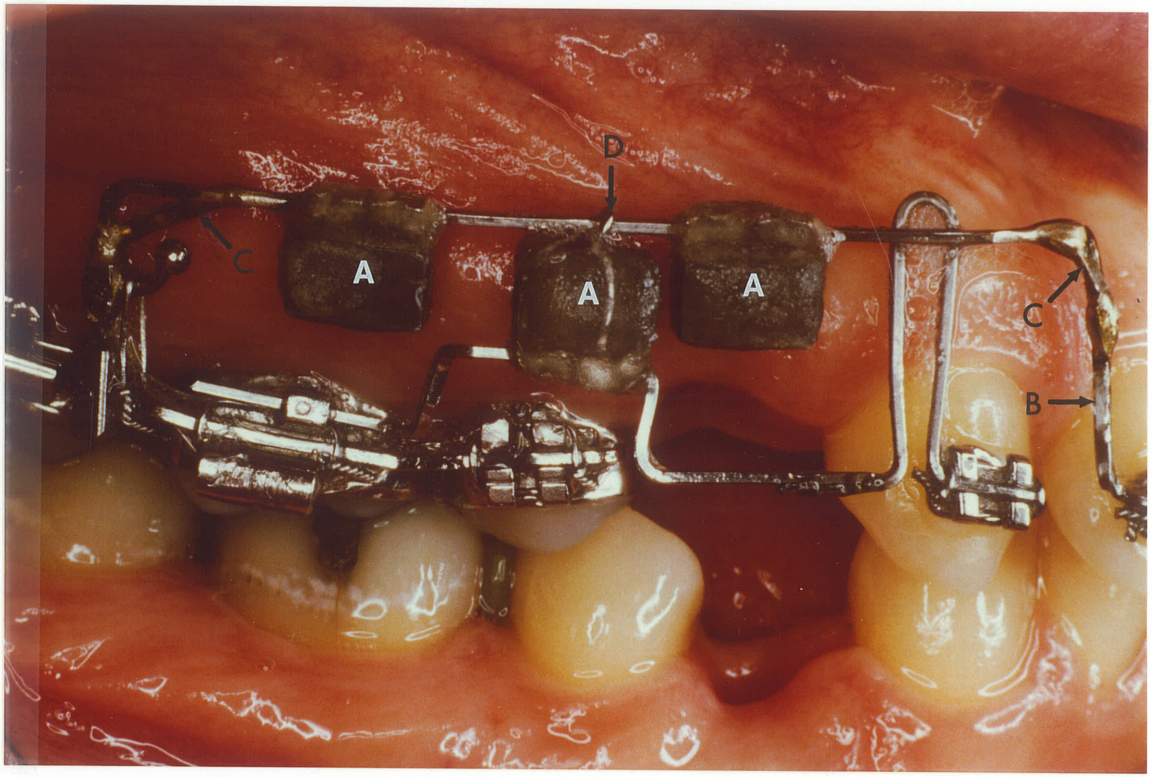

Figure 4.4.a The constant force delivery system *in vivo*. (A: rare earth block magnets; B: vestibular wire; C: "corner-braces" to increase stiffness; D: helix that allows the middle magnet to slide on the vestibular wire).

The forces that the loops were able to generate were calibrated using the previously described 3-dimensional force measuring machine. Once the loops were calibrated, the distance between the loop legs that corresponded to a force in the range of 65 - 75 g was noted. Custom made acrylic jigs were constructed, out of Jet cold cure acrylic, which were capable of maintaining that predetermined opening for each loop during assembly of the control and experimental retraction mechanisms.

In the second session the vestibular segment, with the mesial and distal magnets fixed on it, was inserted and tied into the bracket slots with stainless steel ligature wire, .010 inch in diameter.

The loop intended for the cuspid on the experimental side was then inserted and adapted so that it would fit passively between the cuspid bracket slot and the modified auxiliary slot of the first molar. Care was taken to ensure that the distal extension of the loop was free to slide through the auxiliary slot, with the mesial extension engaged in the cuspid bracket. After this was done, the necessary bends were placed mesial and distal to the middle magnet in order to achieve maximum interface between the interproximal surfaces of the juxtaposed magnets and parallelism of their magnetic axes. At that point, the freedom of the distal extension of the loop to slide through the auxiliary slot of the first molar band was reconfirmed.

The custom-made jig was then placed on the loop; this kept the legs of the loop apart by approximately 1 mm. The loop was then inserted in place again and the middle magnet was crimped on the loop wire (with the legs of the loop kept apart by the jig, as if it was active), with an air-gap between it and the mesial magnet of approx. 0.5 mm. A second acrylic jig of known thickness was placed between the mesial and the middle magnet before crimping it on the loop wire, to set the desired air-gap. After the middle magnet was crimped on the wire, the jigs were removed and cold cure acrylic was utilized to encase the junction area between the magnet and the wire, to reinforce retention.

At that stage, a small helix was made around the vestibular segment wire, between the mesial and distal magnet, with stainless steel ligature wire (.012 inch in diameter). The incisal extension of the helix was then embedded in the acrylic casing of the middle magnet with cold cure acrylic, so that the middle magnet was free to slide along the vestibular segment wire by means of the helix (Figure 4.4.a). This was done to provide some constraint for the distal extension of the loop, which would otherwise only be supported at the modified auxiliary tube of the first molar. The intention was to prevent any spurious movements of the cuspid that would be allowed by the increased flexibility of that wire. As well, the helix prevented the middle magnet from being deflected away from the magnetic axis of the field created by the remaining two magnets.

Following that stage, the control (left) loop was adapted so that it accurately fitted the cuspid bracket and the auxiliary tube on the molar band, passively. The experimental and control vertical loops were tied into their respective cuspid brackets with .010 inch diameter stainless steel ligatures.

The experimental (right) loop was automatically active after insertion, since it was activated by the magnetic force. The control loop was activated by using the custom-made acrylic jig to ensure the desired separation of the legs. Subsequently, the end of the loop wire distal to the auxiliary tube of the first molar was bent and the jig was removed.

In the case of the first patient (N.S.), both loops were made from .017 x .025 inch TMA wire, in an effort to avoid any materials that were even slightly paramagnetic, such as stainless steel. This idea was abandoned in the subsequent patients, as it was found that there was increased friction between the TMA wire and the auxiliary slot of the molar band, which was undesirable. Two reasons could account for this finding:

a) It has been demonstrated that when stainless steel brackets slide on TMA wire, a "stick-slip phenomenon" is observed, because of adhesive wear in the form of cold-welding. This temporary formation of metal-metal bonds can explain the dramatic increase in frictional forces (Kusy *et al.*, 1990).

b) Because of its substantially greater flexibility than stainless steel, TMA does not provide as good constraint characteristics, and consequently allows more angular displacement (tipping) to occur before a large enough uprighting moment can be generated, resulting in proportional increase on the perpendicular forces that cause friction.

The potential shunting effect of stainless steel wires on the magnets was further investigated by using the three-dimensional force measuring machine. The magnets were fixed on the machine using specially designed clamps, in an arrangement that would closely simulate the actual clinical application. The magnetic force generated was found to be virtually unaffected by the presence of the stainless steel wires in close proximity to the magnets. It was therefore decided for the remaining five patients, to weld a piece of stainless steel wire on the TMA loops, to increase their rigidity and improve their behaviour in sliding.

At the insertion appointment a maxillary impression was taken with fast-set alginate impression material. A plaster model was cast from this impression to serve as the reference model (Day 0). The same model was used in the fabrication of a master model, upon which custom trays for the subsequent progress impressions were constructed. For this purpose, a double sheet of pink wax was applied on the reference model as a spacer. The gingival three quarters of the vestibular and lingual surfaces of the teeth, as well as the entire palate, were also covered by wax, so that the resulting tray was only 2 - 3 mm deep and the impression material covered only the incisal and occlusal 2 - 3 mm of the teeth. This was deemed necessary to avoid any distortion of the loop wires or sticking of the impression material on the appliance during impression taking. A fast-set alginate[6] impression of the waxed-up reference model was taken and a master model was cast from that impression.

The custom trays for the subsequent progress impressions were fabricated on the master model, with Palatray[7] light-cured acrylic. Reprosil[8], an addition silicone impression material, was used for the progress impressions. This material was selected because of its ability to accurately reproduce surface detail, and because of its long-term dimensional stability which guaranteed accurate re-pour of a model in case the need arose.

Progress appointments were scheduled twice a week for the first month and once every two weeks for the second and third month. At each progress appointment, a maxillary impression was taken. Buccal intraoral photographs were also taken, to permit

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6 Alginate Jeltrate Plus (Type I) - Fast-Set: The L.D. Caulk Company, a Division of Dentsply International, Inc., Milford, DE 19963-0359

7 Palatray LC: Kulzer & Co. GmbH, Philipp-Reis-Str. 8, D-6393 Wehrheim/Ts

8 Reprosil: The L.D. Caulk Division, Dentsply International Inc., Milford DE 19963-0359

assessment of the separation of the legs of the loop throughout the study. This was necessary to calculate approximately the magnitude of the force applied on the tooth at every appointment. Prior to taking each impression, all the bands, brackets and wires were blocked out with rope wax to prevent distortion of the appliances or the impressions.

For the last four patients (P.M., J.D., D.R. and D.B.), two additional photographs were taken at each appointment to assess the changes in the angulation (tipping) of the cuspids as they were being retracted. For this purpose, custom made caps that fitted over the maxillary cuspid crowns, and around the brackets were fabricated from the reference model, using Jet cold cure acrylic. A straight piece of stainless steel wire, .018 inch in diameter and approximately 1 inch in length, was embedded in the acrylic cup, at the cuspid tip. These caps were fitted on the maxillary cuspids at each appointment before taking the photographs. A straight length of .036 inch stainless steel was inserted in the headgear tube, that extended mesially to the cuspids (Fig 4.4.b). At the distal end of the latter wire there was a small ball of solder that was used to stop it from sliding distally through the tube, and was appropriately trimmed to make the wire fit tightly in the headgear tube. The photographs were taken from a view approximately perpendicular to the plane of the two wires (the one on the cuspid cap and the one in the headgear tube), and with the patient in the dental chair, at a reclined position, to avoid any effects of gravity on the straight lengths of the wire.

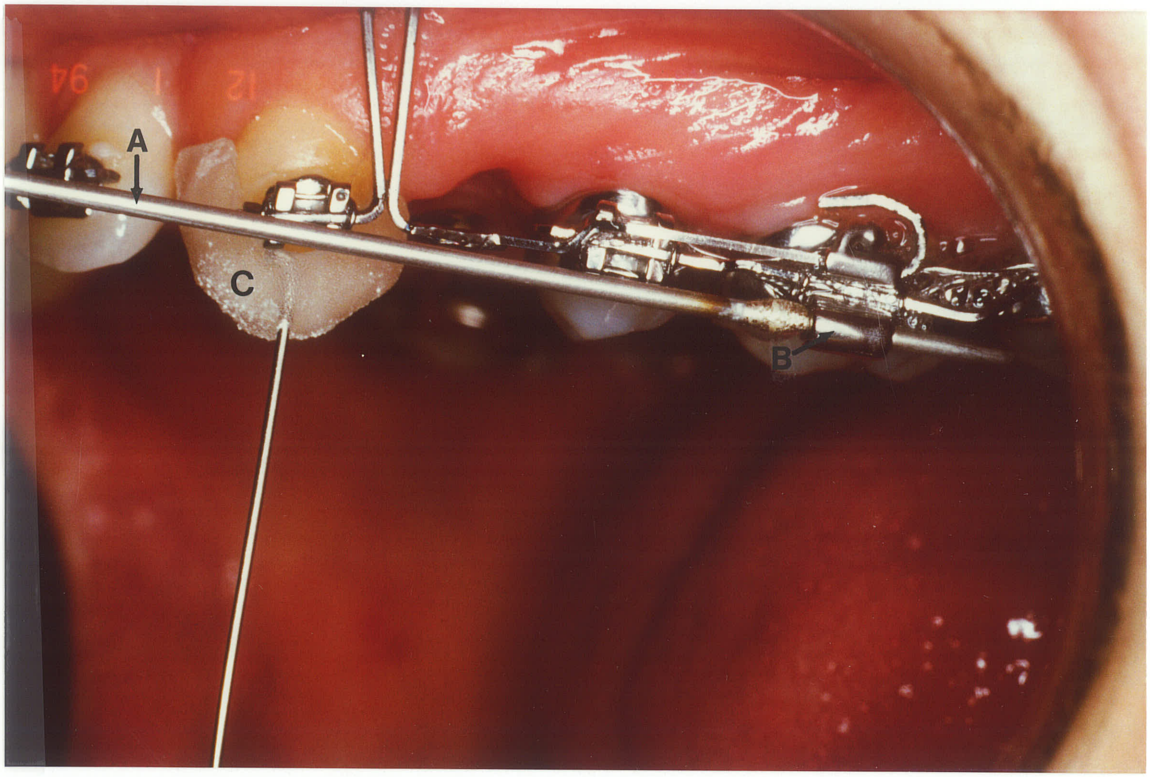

Figure 4.4.b Intraoral photograph for assessment of tipping. The angle between the .036 inch wire (A) which fits into the headgear tube (B) of the first molar band and the .018 inch wire embedded in the acrylic cap (C) on the cuspid, was measured on subsequent photographs.

The control side was re-activated 6 weeks after insertion, which is a typical re-activation period for a retraction spring of this type (Cohen, 1991; Duff, 1988). Re-activation was performed by further bending the wire extension distal to the first molar auxiliary tube, using the initial acrylic jig as an indicator of the correct loop opening for the activation. The loop on the experimental side was never activated in the same way, but the legs of the loop were kept apart by the magnetic force alone. The ligature wires holding the mesial extensions of the two loops in the brackets of the cuspids were cut periodically, and the loops were tested for any signs of plastic deformation. When any plastic deformation was found, the loop was "reformed" until its legs were again slightly touching in its non-active state.

Based on previous data by Cohen (1991), the cuspid movement on the control side was not expected to be greater than 2 - 3 mm over the experimental period of three months. A comparison between the total cuspid movement on the control and the experimental side was undertaken after the experiment was terminated.

After the end of the experimental period, the magnets were removed and the .022 inch set-up bands on the experimental side were changed to .018 inch ones. A new .017 x .025 inch stainless steel passive segment was constructed from the second molar to the second premolar on the right side, and new vertical loops were fabricated from stainless steel wire (.017 x .025 inch in dimensions). The loops were activated for further retraction of the cuspids, and the patients were referred to the first-year students at the Graduate Orthodontic Clinic in order for their treatment to be continued.

4.5 *IN VITRO* EXPERIMENTAL PROTOCOL

4.5.1 Force Measurements

Measurement of the forces that the magnets were able to generate in the X-axis, at various air-gaps was performed *in vitro,* using the force measurement machine developed by Paquien (1978), as was described earlier. This was done in order to determine the appropriate air-gap to generate the 65 - 75 g force desired for the study.

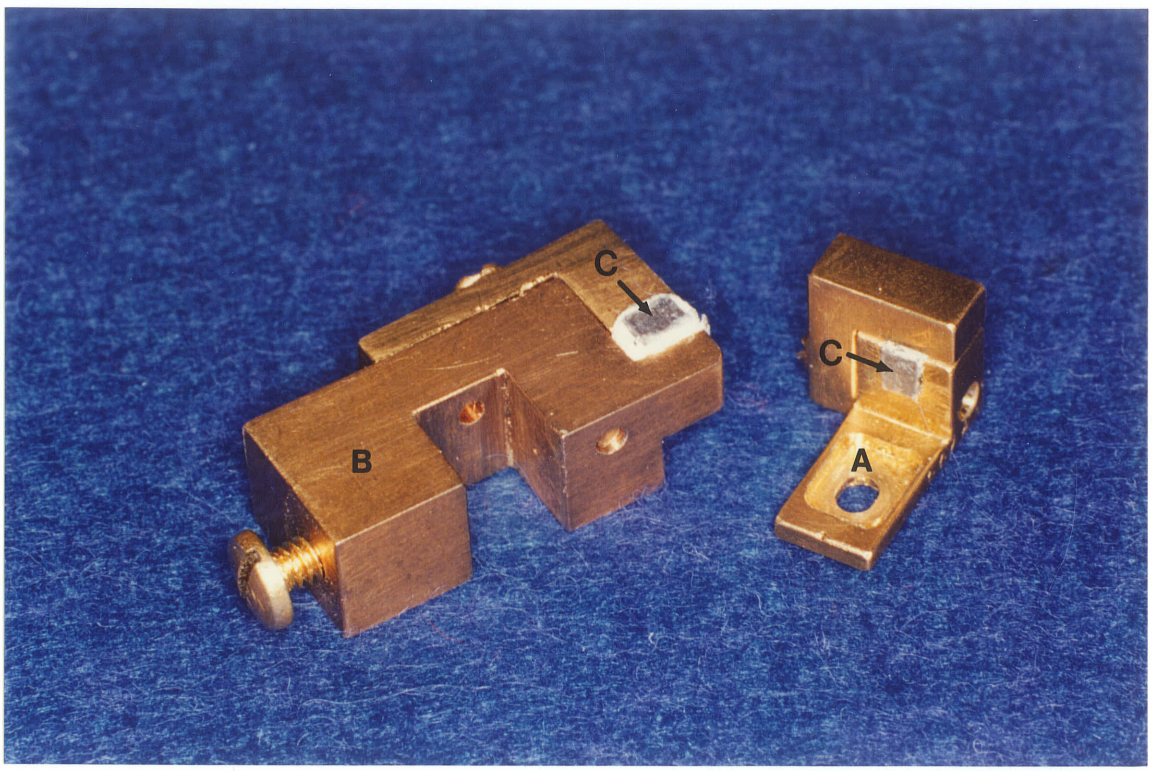

Figure 4.5.1.a The custom-made brass attachments used to fix the magnets on the force-measuring machine. (A: attachment for the middle magnet; B: attachment for the mesial and distal magnets; C: rare earth magnets in place).

The three magnets were placed on the machine simulating the arrangement that they would have in the actual appliance. Custom-made brass attachments were used to fix the magnets on the machine. Each attachment carried a slot that had been precisely machined to fit the thickness of one magnet (Fig 4.5.1.a). The attachments carrying the mesial and distal magnets were screwed onto a bar on the mobile component of the machine, at the desired corresponding air-gap to be tested (see Fig 4.5.1.b). The attachment carrying the middle magnet was fixed on the stationary component.

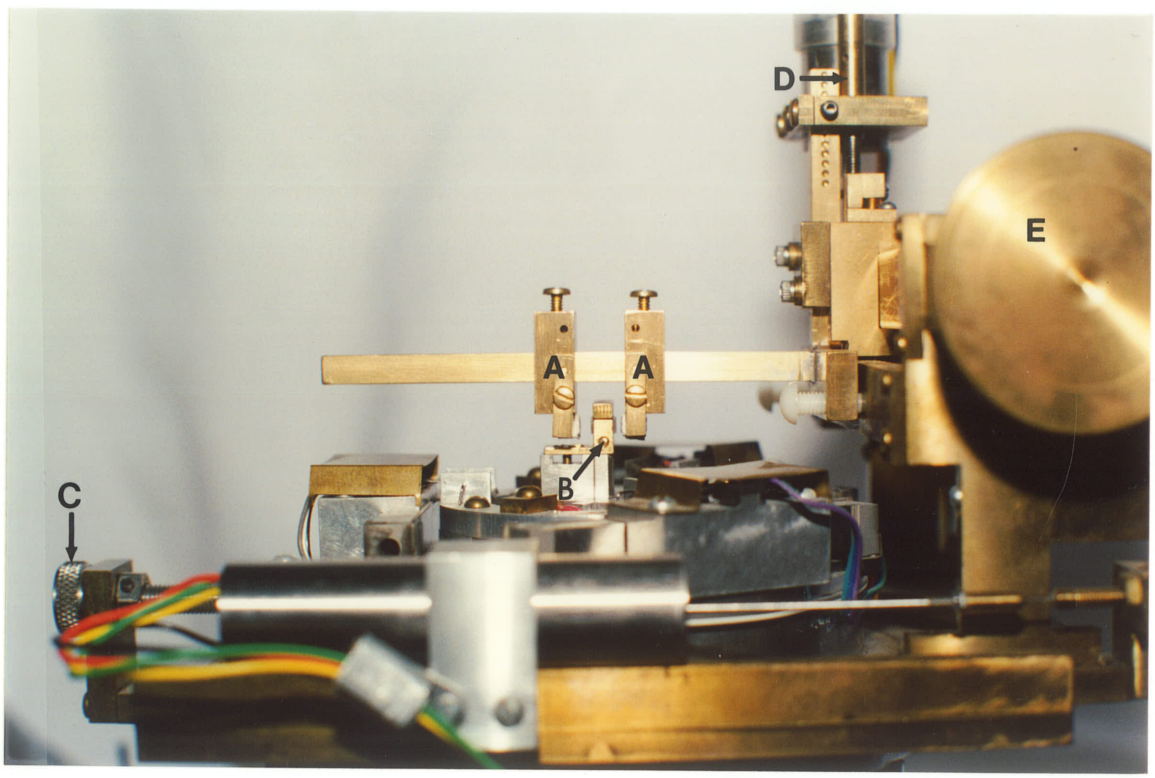

Figure 4.5.1.b The magnets on the force-measurement machine: The attachments carrying the mesial and distal magnets (A) are fixed onto a horizontal bar on the mobile part of the machine. The attachment carrying the middle magnet (B) is fixed on the stationary component of the machine. C,D,E: Adjustment screws for movement along the x, y, and z-axis, respectively.

After taking a zero reading, the middle magnet was brought between the mesial and distal magnets and aligned with them. The first force reading was taken at a distance of 0.5 mm between the mesial and middle magnets. Subsequent readings were collected for controlled 0.25 mm steps of the middle magnet, until the latter reached a distance of 0.5 mm to the distal magnet.

Using this procedure for various air-gaps between the mesial and distal magnets, it was determined that a 9.25 mm air-gap would be adequate in creating a mean force of 70 g that varied by $\pm$16 g (from 54 to 86 g), over a distance of 2.5 to 3 mm.

Calibration of the loops, before placement in the mouth, was performed on the same machine for activations ranging between 0 and 1.5 mm. All loops were 11 mm in height and, in the passive state, their legs were barely touching. In a few occasions in which the actual loops were not able to be calibrated before insertion, identical replicas of these loops were fabricated at the same time, and these were subsequently calibrated on the machine.

The loops were fixed on the machine with specially made attachments, similar to those used by Duff (1987). Each attachment carried a slot that had been precisely machined to fit an .017 x .025 inch wire. The distal extension of the loop was attached on the mobile part of the machine, and the mesial extension was attached on the fixed part. A zero reading was taken with the loop at its passive state. Readings of the force produced by opening the loop by controlled steps of 0.25 mm were collected, to a maximum activation of 1.5 mm. After removal of the loop from the machine, care was taken to ensure that the loop had not undergone permanent deformation, and the legs were still barely touching.

4.5.2 Measurement of Tooth Movement

An accurate technique for measuring tooth movement was necessary to evaluate the changes in tooth position under the influence of the two diverse force systems. At present, no satisfactory technique exists whereby tooth position changes can be measured intraorally. A technique previously utilized by Duff (1988), Sonya (1988), and Cohen (1991) was also used in the present study, due to the unavailability of costly instruments like the Reflex Metrograph.

Selected points on the teeth in the posterior anchorage segments were used as reference points on the cast. The points were spaced as widely as possible, to minimize the effect of measurement error. The duplication of the reference points on subsequent casts was performed by using sectional acrylic templates, that would fit on the occlusal surfaces of the posterior teeth (Figure 4.5.2.a). The templates were formed on the Biostar Pressure Moulding Machine[9], using 2 mm thick Biocryl[10] acrylic. Small holes were drilled on the templates at the appropriate reference points, and the reference points were marked on the subsequent casts with a fine lead pencil (Fig 4.5.2.b).

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[9] Biostar Pressure Moulding Machine: available from Great Lakes Orthodontic Products Inc., 199 Fire Tower Dr., Tonawanda, N.Y. 14150

[10] Biocryl II (clear) - 2 mm: Great Lakes Orthodontic Products Inc.

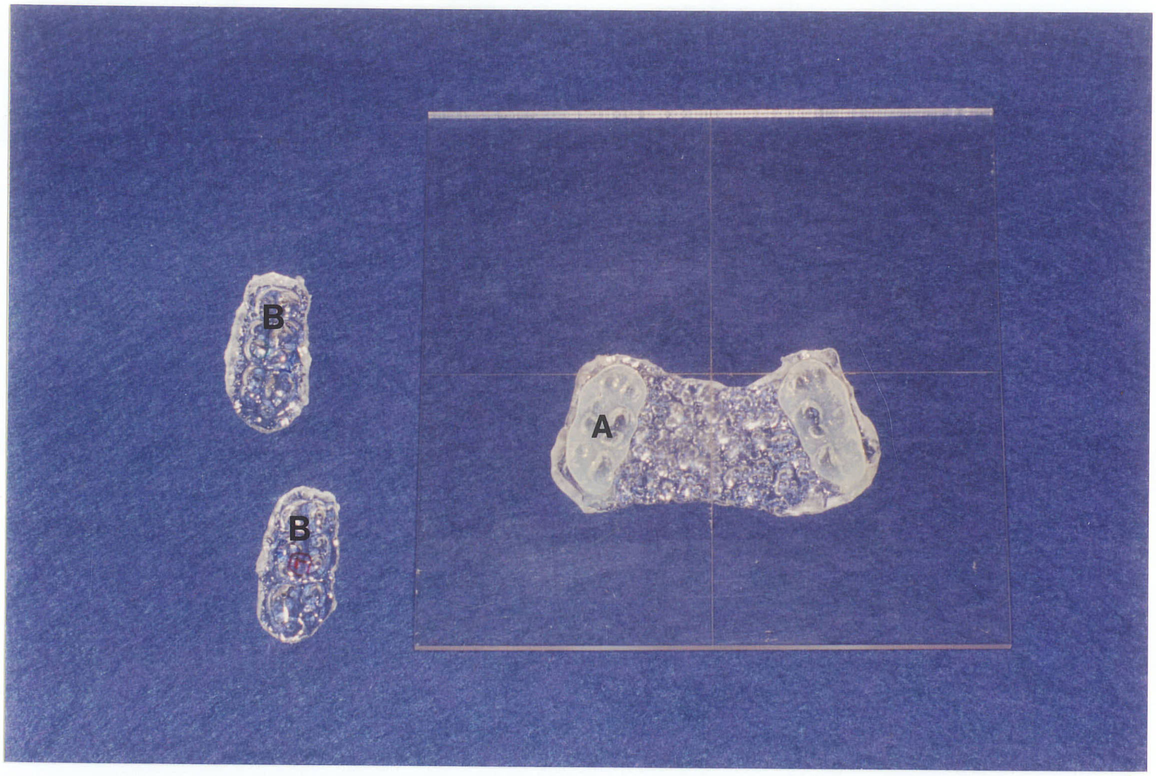

Figure 4.5.2.a Orientation splint (A) and perforated acrylic templates for duplication of points to subsequent casts (B).

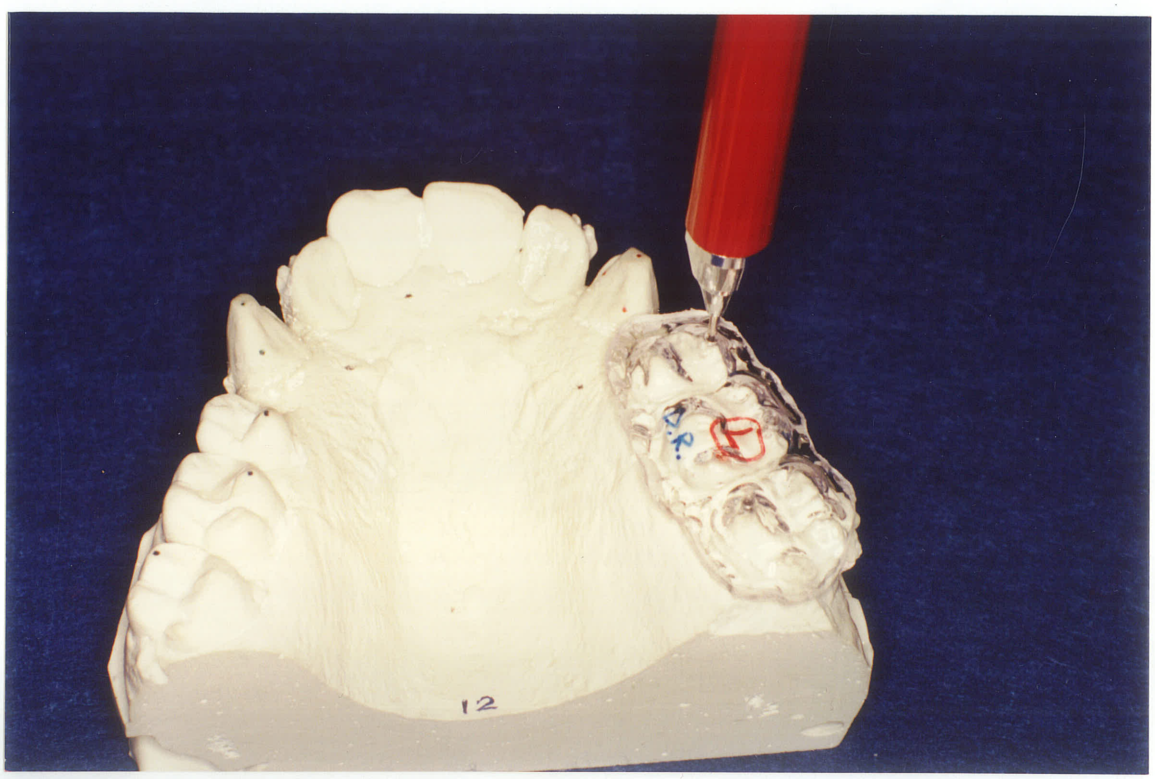

Figure 4.5.2.b Plaster cast with acrylic template and marking pencil in position.

Three reference points were used per anchorage segment (a total of six posterior reference points per cast), in addition to three reference points on each cuspid. Moreover, three points on the palatal rugae of each patient were used to assess possible anterior movement of the selected reference points on the anchor teeth.

The cast-measuring apparatus consisted of three machined metal baseplates, that were articulated to allow full adjustment of the position of the cast in three dimensions. The casts to be measured were fixed firmly on the top baseplate by means of three securing screws. The entire assembly was then placed on a precisely machined horizontal metal table, which also carried a fixed vertical backplate (Figure 4.5.2.d).

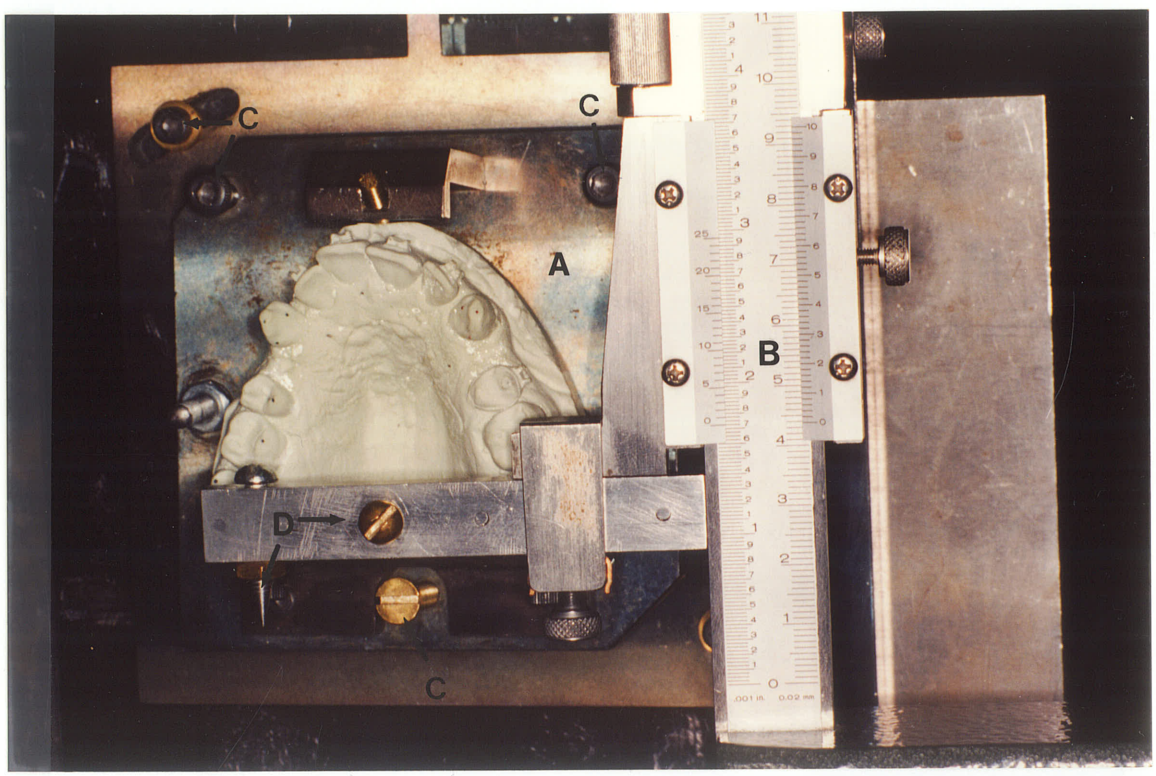

Figure 4.5.2.c Plaster cast positioned on the measurement apparatus for assessment of antero-posterior distances. (A: Baseplate assembly; B: Vernier Height Gauge; C: Adjustment screws; D: Pointers).

Standardized orientation of the subsequent casts on the measurement apparatus was necessary to collect meaningful results. This was performed by means of an orientation splint for each patient. The orientation splint was formed on the Biostar Pressure Moulding Machine using 2 mm thick Biocryl acrylic, using the initial (Day 0) cast. The splint covered only the occlusal surfaces of the anchor teeth. This splint was trimmed and fixed to a machined square piece of rigid acrylic, 125 x 125 x 3 mm in dimensions (Fig 4.5.2.a). Mutually perpendicular lines had been scribed on the latter piece of acrylic, to aid in orientation.

After the adjustable baseplate assembly carrying the cast to be measured was placed flat on the horizontal table, the orientation splint was placed on the cast. A spirit level was used to ensure that the rigid acrylic plate of the orientation splint was parallel to the horizontal table. Adjustments of the assembly were possible by tightening three screws which were capable of re-orienting the cast, at the same time keeping it securely fixed on the top baseplate.

The assembly was subsequently turned through 90° so that it was now fixed against the metal backplate, which was perpendicular to the horizontal table (Fig 4.5.2.c). The spirit level was then used again, to ensure that the superior edge of the orientation splint was completely horizontal. Distances between the reference points in the antero-posterior direction were made with the cast in this position. The instrument used for these measurements was a Vernier Height Gauge, capable of measuring to the nearest 0.01 of a millimetre.

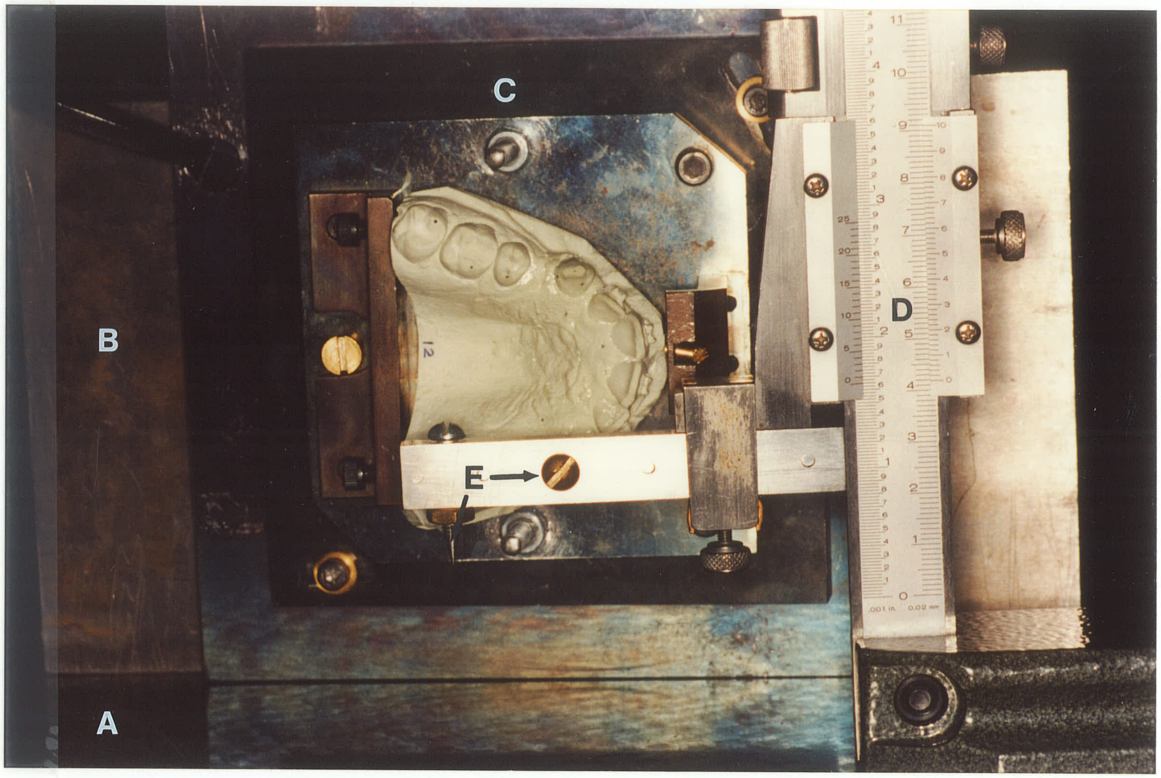

Figure 4.5.2.d Plaster cast positioned on the measurement apparatus for assessment of medio-lateral (transverse) distances. (A: Horizontal table; B: Vertical backplate; C: Baseplate assembly; D: Vernier Height Gauge; E: Pointers).

Since the Vernier Height Gauge could only measure distances in the vertical direction, the measurements in the bucco-lingual direction were performed with the baseplate assembly still fixed on the vertical backplate, but rotated by 90° relative to the previous position (Fig 4.5.2.d). Similarly, in order for the measurements in the occluso-gingival direction to be carried out, the assembly had to be re-oriented by 90°, so that it would rest on the horizontal table, with the orientation splint parallel to it (Fig 4.5.2.e).

Collection of data in the above manner, permitted a reproducible, quantitative three-dimensional description of the position of each point on each tooth. By evaluating the coordinates of the same points on subsequent casts, a detailed assessment of tooth movement in three dimensions was possible.

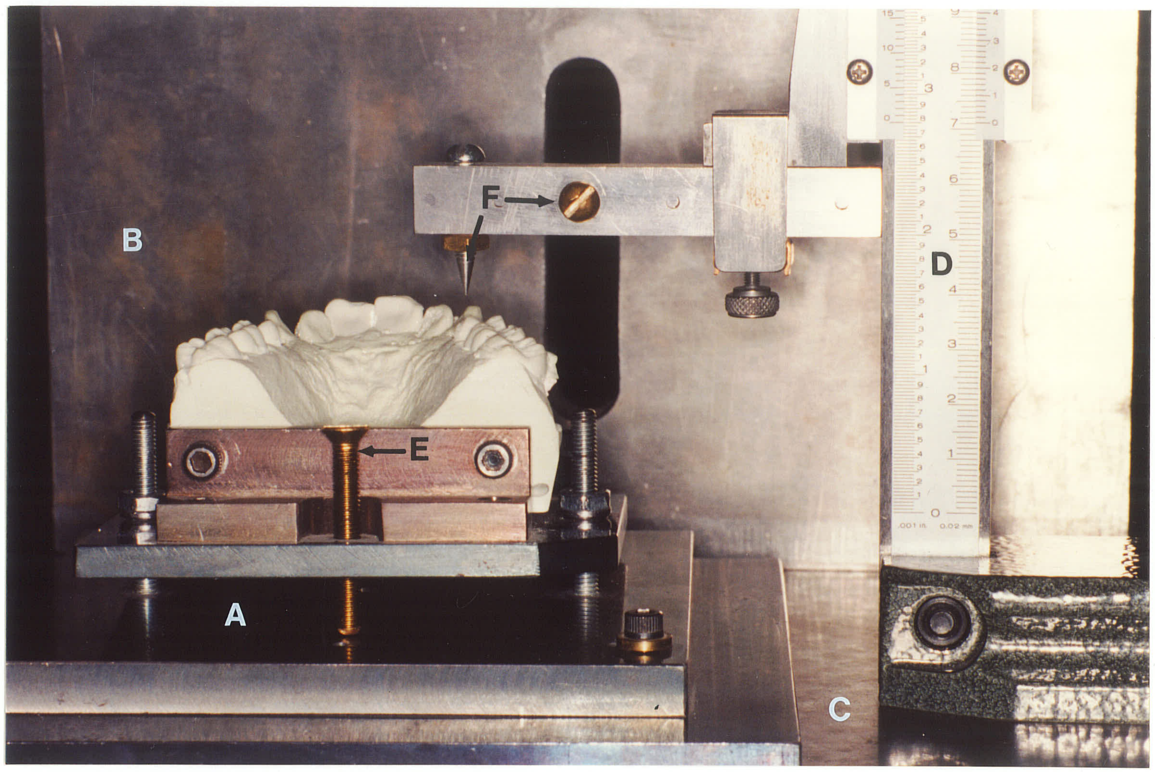

Figure 4.5.2.e Plaster cast positioned for vertical (occluso-gingival) measurements. (A: Baseplate assembly; B: Vertical backplate; C: Horizontal table; D: Vernier Height Gauge; E: Screw for occluso-gingival adjustment; F: Pointers).

4.5.3 Data Analysis

The data from the cast measurements was analyzed as follows:

A. ASSESSMENT OF NET A/P MOVEMENT OF THE CUSPID

The linear distances of the middle reference point on the cuspid to each of the posterior reference points were averaged for each cast. To determine the antero-posterior change in tooth position, the average distance mentioned above for each cast, was compared with the respective average distance on the initial (Day 0) cast for each patient. Plotting the differences in the average A/P distance of the cuspid to the posterior reference points between the Day 0 and every subsequent cast, as a function of time, provided a meaningful measure of net antero-posterior movement of the cuspid through the duration of the experiment.

B. ASSESSMENT OF 3-DIMENSIONAL POSITION OF THE CUSPID

For each cuspid, the differences x, y, and z in the coordinates of the mesial (CR1) and distal (CR3) reference points were calculated, in the three dimensions. Because of the way that the initial measurements were performed, these three mutually perpendicular linear segments formed three right triangles, with a common hypotenuse (three-dimensional hypotenuse, L) (Figure 4.5.3). L can be calculated by using the Pythagorean theorem:

$$L = \sqrt{x^2+y^2+z^2}$$

where:

$$x = CR1_x - CR3_x$$

$$y = CR1_y - CR3_y$$

$$z = CR1_z - CR3_z$$

Subsequently, the angle θ_y between the three-dimensional hypotenuse L and the true vertical, was calculated from the formula:

$$\theta_y = \sin^{-1}\left(\frac{y}{L}\right)$$

This angle on the Day 0 cast determined the original position of the tooth in this dimension. To evaluate the change in tooth position, the same procedure was followed for each subsequent cast and compared with the original angle on the Day 0 cast. The difference $\Delta\theta_y$ (in degrees), between the initial (Day 0) cast and any subsequent cast of a particular patient represents the angular change (tipping) in the xy plane. Please note that the above mentioned angle does not coincide completely with tipping as we define it clinically. This angle connotes change of angulation in the antero-posterior plane, whereas tipping is usually meant in the mesio-distal plane, which can be different than the former.

In addition, the angle θ_z between the three-dimensional hypotenuse L and the true horizontal, was calculated from the formula:

$$\theta_z = \sin^{-1}\left(\frac{z}{L}\right)$$

Using the same procedure as described above, the angle θ_z would be calculated for every cast and compared with the original one on the Day 0 cast. The difference $\Delta\theta_z$ (in degrees), between the Day 0 cast and any subsequent cast represents the angular change (rotation) in the zx plane of space.

Verification of the results from the three-dimensional analysis was performed using the following techniques:

Assessment of tipping was performed by using the buccal photographs taken at each appointment. As already mentioned, in four patients (P.M., J.D., D.R., and D.B.) special photographs for tipping assessment were taken. The angle between the stainless steel wire that was embedded in the acrylic cap on the cuspid, and the .036 inch stainless steel wire whose distal end was inserted in the headgear tube was measured at each appointment. In the remaining patients in whom these photographs were not available, assessment of tipping was made by directly measuring angles between the cuspid bracket and the anchorage segment wire on the regular buccal photographs.

Reconfirmation of the results of the three-dimensional analysis on the rotation of the cuspids around the y-axis, was performed by using a photocopying machine. Photocopies of the occlusal view of subsequent models were taken, and the angles were measured between constructed lines that were drawn through the reference points on the cuspids and on the anchorage segments.

C. STATISTICAL ANALYSIS

The paired t-test was used to compare the mean values for total cuspid movement and rate of movement between the control and experimental sides.

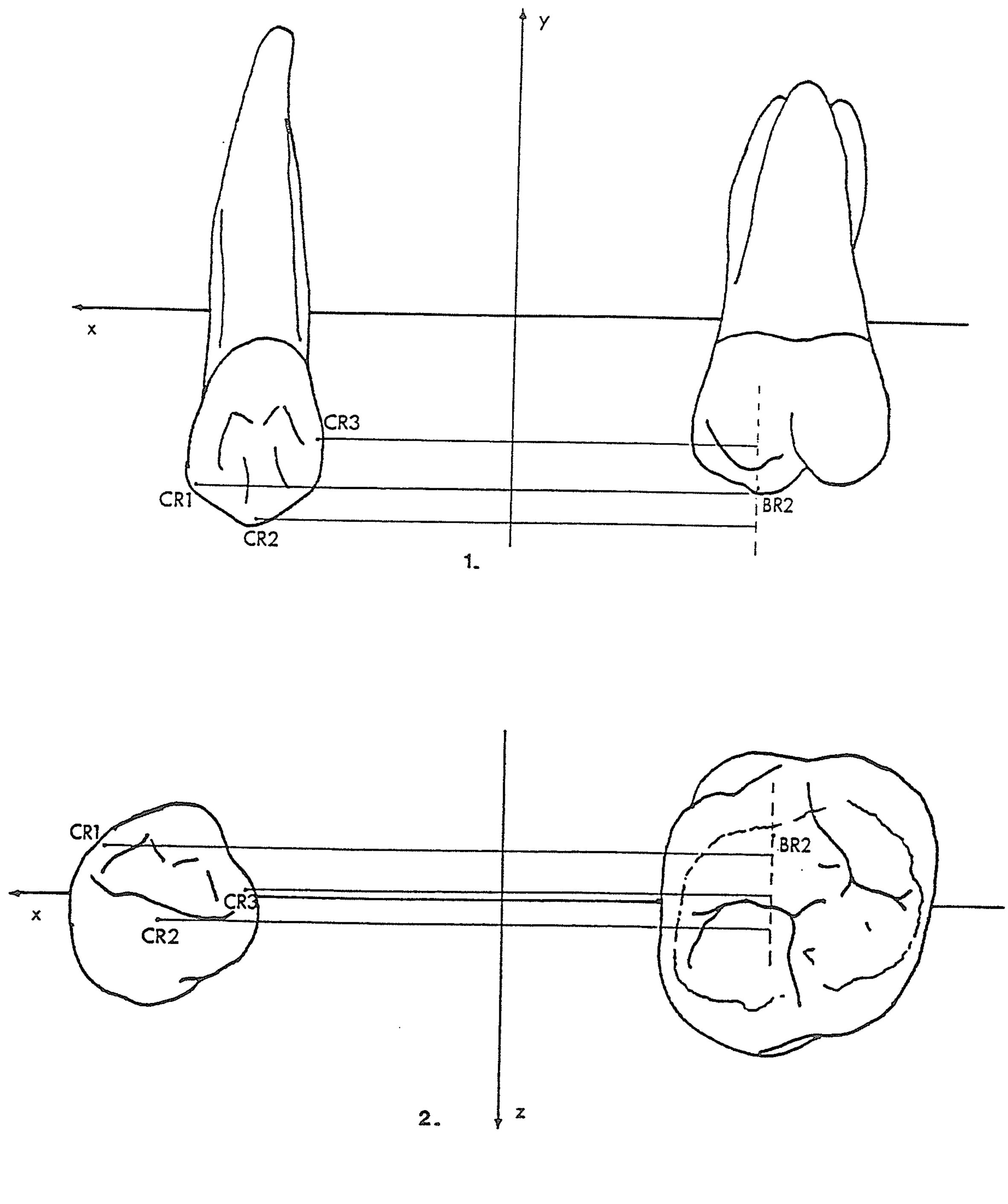

Figure 4.5.3 Palatal (1) and occlusal (2) view of a maxillary right cuspid and first molar with the reference points used in the three-dimensional analysis.

4.5.4 Error of the Method

Determination of measurement error was performed by measuring the three dimensional coordinates of one single point on a cast twenty times, and by calculating the standard deviation of these measurements. In addition, in order to estimate the size of other sources of error with the method of measurement, the following trial was undertaken: A single cast was duplicated ten times, using Deguform[11], a high precision liquid silicone material, for replication of dental casts. The duplicates were subsequently placed on the measuring apparatus, and the coordinates of four widely spaced points on each cast were determined. The distances between these points in the x and the y dimension were then calculated, and the means and standard deviations were determined.

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[11] Deguform: Degussa AG, Geschäftsbereich Dental, D-6000 Frankfurt am Main 11.

CHAPTER 5

RESULTS

RESULTS

5.1 INTRODUCTION

For this project, data were collected and analyzed for twelve cuspid teeth (six experimental and six control). For every patient, a comparison of tooth movement between the control and the experimental cuspid is presented. The results are displayed in a graphical form.

Total (linear) movement of each cuspid has been plotted in millimetres as a function of time. These results were obtained by measuring the distances from the three points on the anchor segment to the middle point on the cuspids (C2), and then averaging these three values. Results on the stability of the anchor segment, are shown in section 5.4.

The results obtained for *posterior tipping* (about the z-axis) and *postero-medial rotation* (about the y-axis), have been plotted in degrees, also as a function of time. These results were obtained from the three-dimensional analysis, as explained in section 4.5.3.

As can be noted from the time scale, the data were not collected at equally spaced time intervals. Moreover, although the data points are connected by straight lines to indicate their temporal relationship, the plots obviously do not represent the actual path of the teeth during movement. Finally, all measurements were made in reference to a true antero-posterior, medio-lateral axis system. For this reason, the terms "distal" and "distolingual" have been replaced by the more appropriate terms: *posterior* and *postero-medial*.

In the three different graphs for each patient, which are labelled "total cuspid movement", "posterior tipping of cuspid" and "postero-medial rotation of cuspid", the following conventions are followed:

- Positive values represent net posterior movement, posterior tipping and postero-medial rotation, respectively.
- Negative values represent anterior movement, anterior tipping and postero-lateral rotation, respectively.
- The vertical error bars on the graphs depicting linear movement represent the standard deviation of linear measurements, which was ± 0.15 mm (see section 5.5).
- The vertical error bars on the graphs portraying angular changes represent the standard deviation for angular measurements, which was $\pm 1.7°$ (see section 5.5).

It should also be noted that the total antero-posterior movement of the cuspid was assessed relative to point C2, which in most cases, was positioned approximately on the cusp tip. For a typical tooth with an average distance from C2 to the centre of resistance (CR) of 8 to 10 mm along the y-axis, posterior tipping of the tooth by 6 to 8° would result in posterior movement of C2 of approximately 1 mm. Assuming that a tooth is rotating around an axis through its centre of resistance, with an estimated average distance from C2 to the centre of resistance of 3 mm along the z-axis, postero-medial rotation of 10° would result in posterior movement of C2 of approximately 0.5 mm.

Therefore, data from all three graphs should be taken into account simultaneously in order to evaluate the movement of the tooth in three dimensions.

Data from the photographic evaluation of the loop opening have been combined with those from the loop calibration to provide an estimation of the force acting on the tooth at each point in time during the trial. These results are displayed in Applied Force/ Time graphs. These graphs show only the forces that were produced by the two springs in the antero-posterior direction. When the loop analysis showed that there had been some plastic deformation of the loops, the force on the tooth was assumed to be zero until the next re-activation took place. This is a reasonable assumption since it is highly unlikely that there would be a negative (anterior) force on the cuspid as a result of plastic deformation. In the event that a loop became deformed in such way that its legs were now further apart than the initial activation, that would still not necessarily translate into an anterior force, as the distal leg of the loop would be free to slide distally through the tube of the first molar attachment, until equilibrium was achieved.

Results from tooth movement are combined with those from analysis of the applied force throughout the duration of the experiment to produce an integrated interpretation of the biomechanics of tooth movement.

5.2 COMBINED RESULTS FROM TOOTH MOVEMENT AND LOOP ANALYSIS

5.2.1 Patient N.S.

Tooth movement in this patient was followed for 73 days. The appliance was then removed, as the middle magnet on the experimental side was almost touching the distal magnet which permitted no further activation of the appliance. Moreover, some distal tipping of the cuspids was clinically discernible.

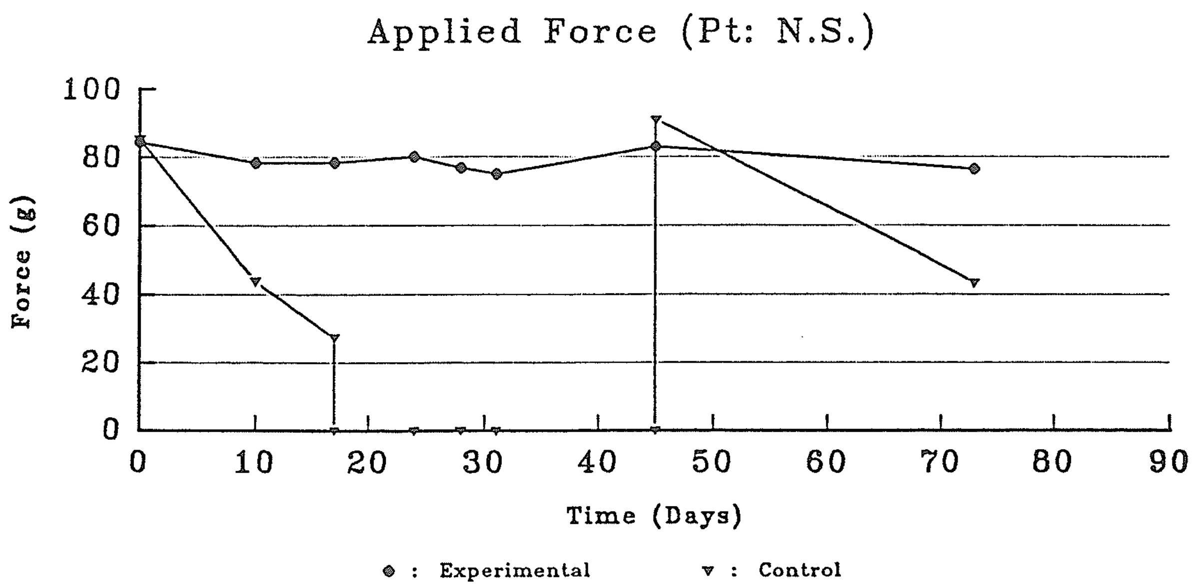


Figure 5.2.1.a The forces in the posterior direction applied on the experimental and control cuspids of patient N.S., throughout the duration of the study. The control loop was permanently deformed between days 17 and 45.

Analysis of the loop activation showed that the loop on the control side was not active due to permanent deformation from day 17 until day 45, when it was reactivated. Consequently, no force in the antero-posterior direction was exerted on the tooth during that period (Fig 5.2.1.a). However, the fact that the tooth exhibited considerable changes in angulation, shows that substantial moments were present. Before reactivation, the ligature holding the wire was cut and the loop was adjusted so that its legs touched again (see section 4.4). Plastic deformation of 0.39 mm was found on the control loop and 0.29 mm on the experimental loop, at removal.

The mean posterior movement of the cuspid was 1.9 mm on the experimental side, and 0.8 mm on the control side (Fig 5.2.1.b). However, the cuspids tipped posteriorly by 9.9° and 7.2° respectively. The total postero-medial rotations were 6.5° for the experimental and 10.2° for the control cuspid.

As mentioned in section 4.4, the appliance utilized on this patient was different from that used for the remaining patients; both the experimental and control loops were fabricated entirely from TMA wire. The above mentioned findings are discussed in greater detail in the next chapter.

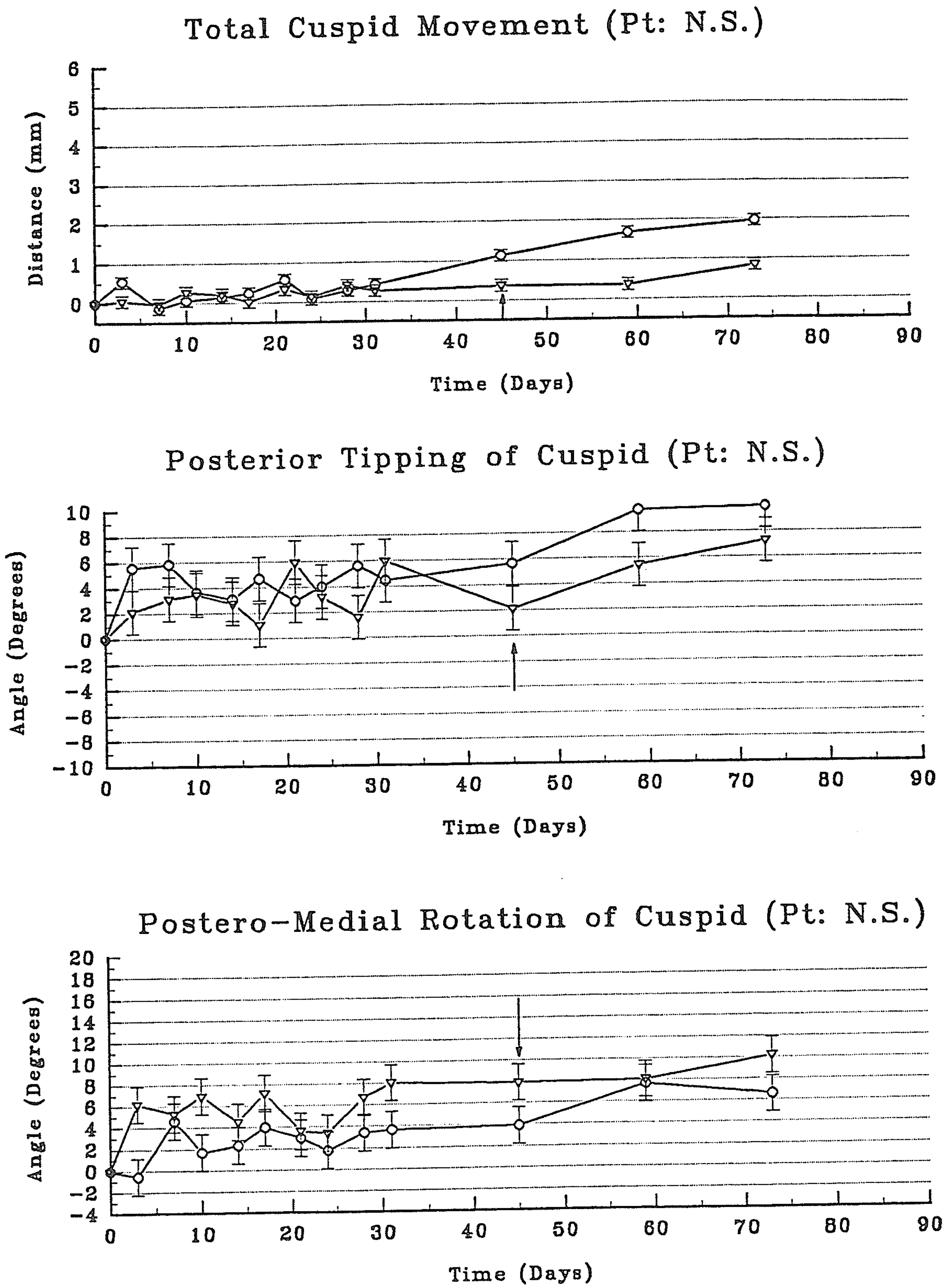


Figure 5.2.1.b Total distance travelled, tipping and rotation of both experimental (○), and control (▽) cuspids on patient N.S.. The arrow indicates the time or reactivation on the control side.

5.2.2 Patient C.O.

Tooth movement in this patient was followed for 70 days. The experiment was terminated at the time that there was no activation remaining on the experimental side, as there was no longer any separation between the middle and distal magnet. Although the loop on the experimental side was still active at that time, it was recognized that after that point the legs of the loop would begin to approximate and the force would no longer be "approximately constant". Therefore, the appliances were removed on day 70.

The results from measurement of tooth movement show a net antero-posterior retraction of 4.3 mm on the experimental and 1.8 mm on the control side (Figure 5.2.2.b). The experimental cuspid tipped posteriorly by a net 2.2° and the control cuspid by 3.8°. The corresponding numbers for rotation about the y-axis were 15.4° in the postero-medial direction for the experimental cuspid, and 3.6° for the control.

Figure 5.2.2.b shows a significant posterior movement of approximately 1 mm was noted between days 24 and 28 on the experimental side. The most likely explanation for this, other than error of observation, would be the fact that the cuspid also rotated postero-medially by approximately 4.5° during the same interval.

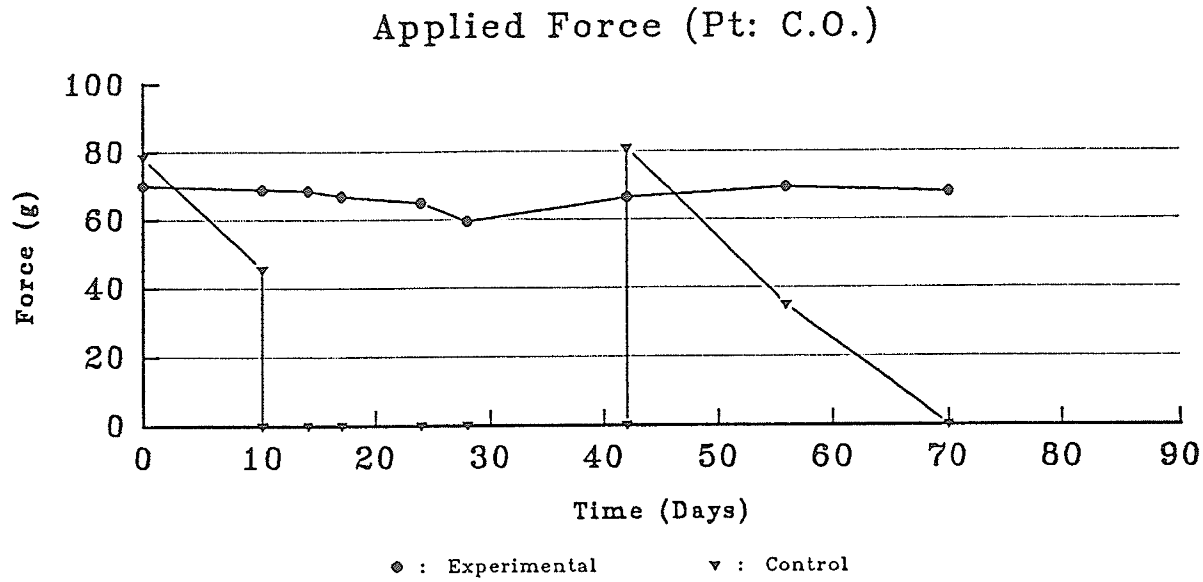


Figure 5.2.2.a The posteriorly directed forces acting on the experimental and control cuspids on patient C.O. during the study. Plastic deformation was noted on the control loop between days 14 and 42, and thus the applied force during that period was assumed to be zero.

As illustrated in Figure 5.2.2.a, no force in the posterior direction was acting on the control cuspid from day 14 until day 42, due to plastic deformation of the loop. The force levels were restored to approximately 80 g (same as the initial force) after reshaping and reactivation of the loop, on day 42. No permanent deformation of either loop was found upon removal of the appliances.

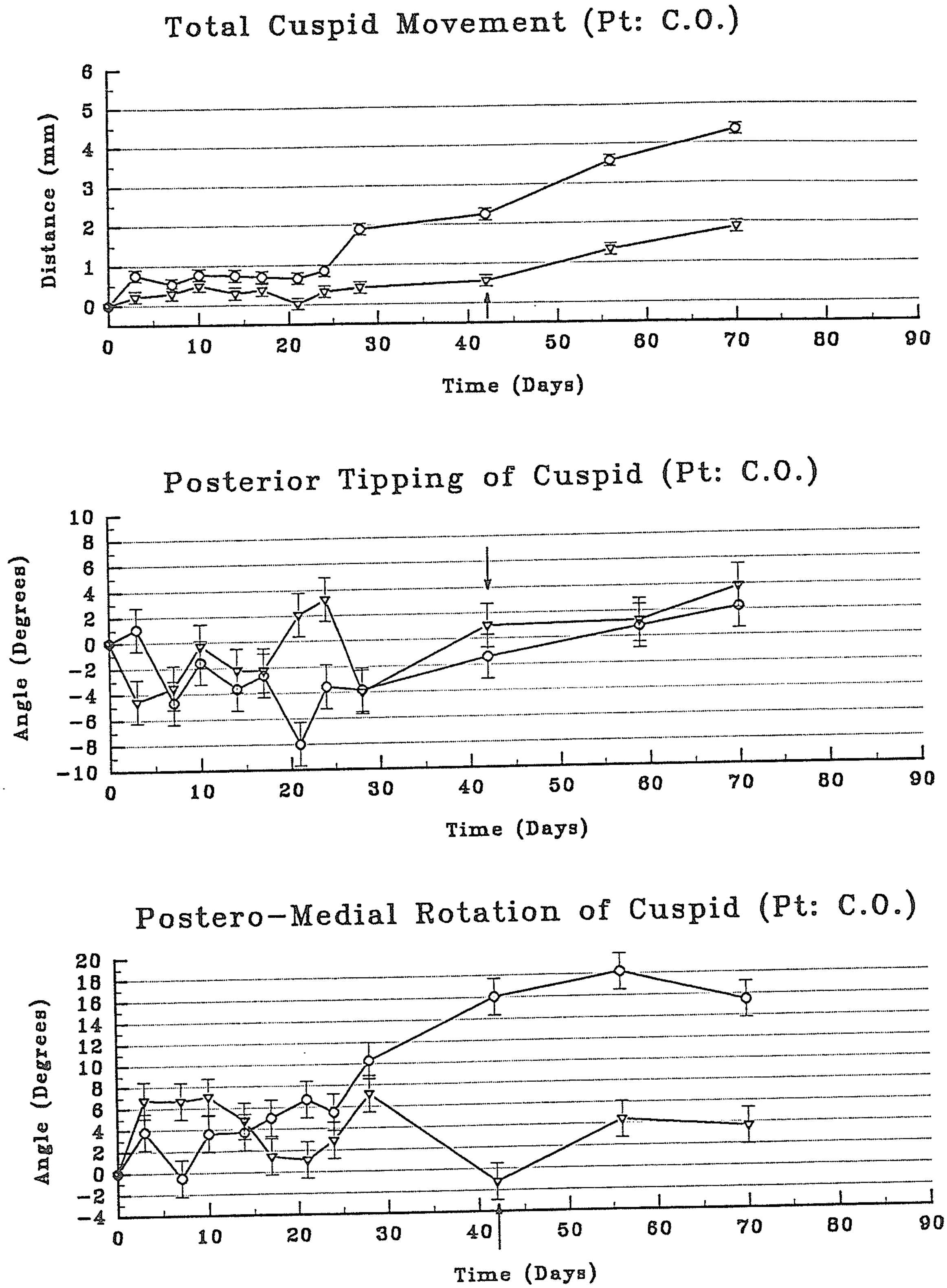


Figure 5.2.2.b Total distance travelled, tipping and rotation for both experimental (○) and control (▿) cuspids on patient C.O.. The arrow indicates the time of reactivation on the control side.

5.2.3 Patient P.M.

Cuspid retraction in this patient was monitored for 87 days. A total antero-posterior retraction of 2.16 mm was noted on the experimental side at the end of this period. During the same interval the control cuspid moved posteriorly by 0.5 mm (Fig 5.2.3.b).

No antero-posterior movement can be seen on the control side between day 31 and day 45 (Fig 5.2.3.b). The loop analysis verifies that the loop had completely worked out and its legs were touching by day 31 (Fig 5.2.3.a). During the same interval, some uprighting of the control tooth occurred (approx. 1°), in addition to some antero-medial rotation (approx. 0.9°). Antero-posterior movement resumed on day 45, when the loop on the control side was reactivated (Fig 5.2.3.a). No plastic deformation was noted on either side at the removal of the loops.

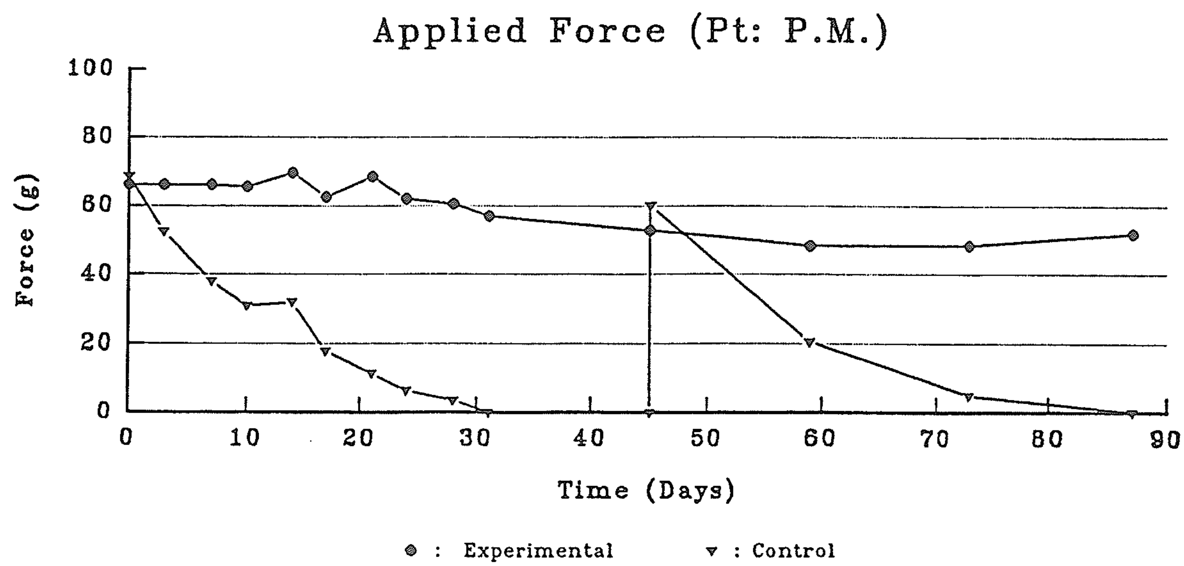
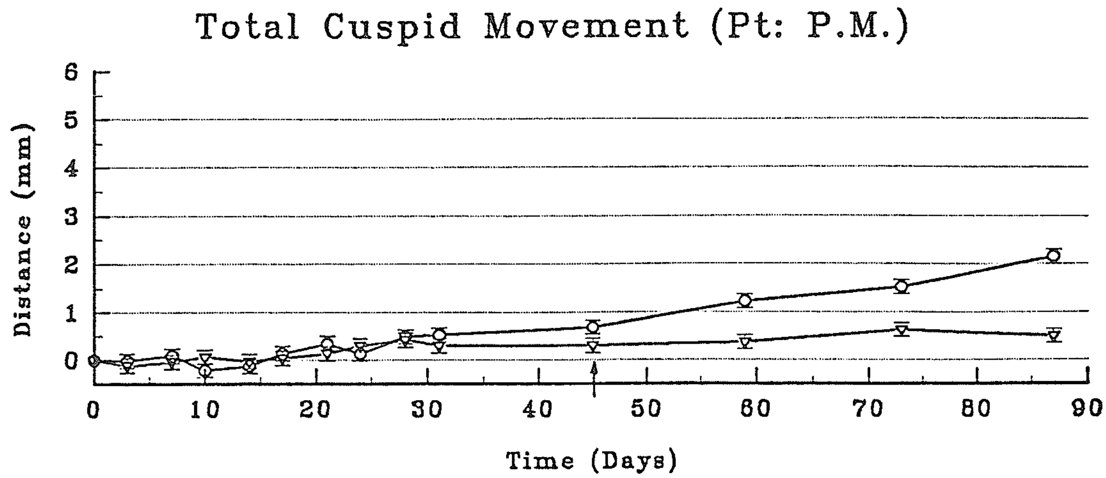
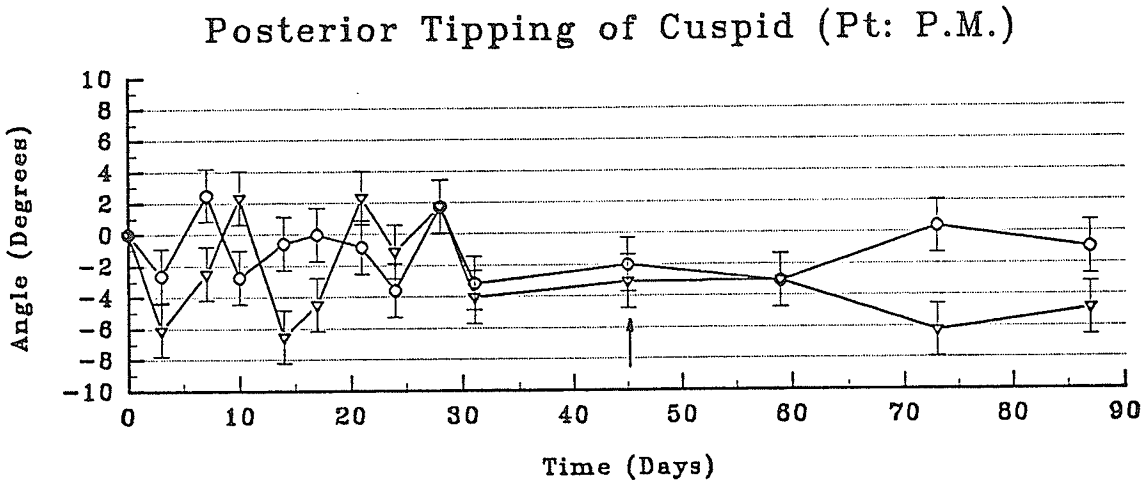


Figure 5.2.3.a The posteriorly directed force applied on the experimental and control cuspid on patient P.M. for the duration of the study.





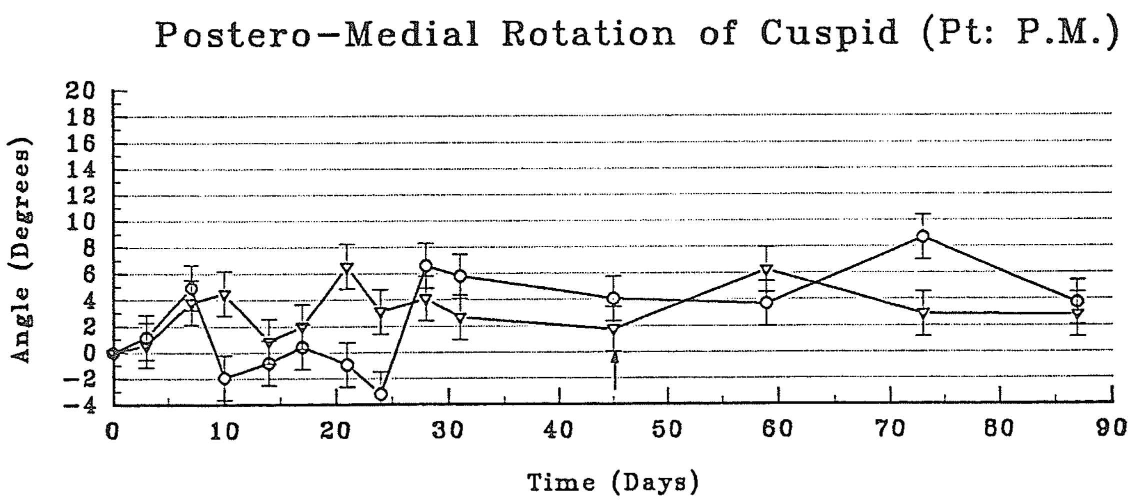


Figure 5.2.3.b Total distance travelled, tipping and rotation exhibited by the experimental (○) and the control (▽) cuspid on patient P.M. for the duration of the study. The arrow indicates the time of reactivation on the control side.

Marked changes in angular measurements were found on both sides during the first month of the experiment; changes which involved tipping (around the z-axis) and rotation (about the y-axis). Both teeth appeared to tip posteriorly during the first three days of force application (Fig 5.2.3.b), although they also seemed to move anteriorly during the same time. The net anterior movements of approximately 0.1 mm on both sides are within the range of error of measurement. The total posterior tipping on the experimental side was approximately 1°, whereas the control cuspid tipped posteriorly by 4.8°. The corresponding totals for postero-medial rotation were 3.7° for the experimental and 2.8° for the control cuspid.

During the last two weeks of the experiment (day 73 to day 87), the control cuspid appears to have been moved slightly anteriorly, presumably due to the uprighting (anterior tipping) movement that was occurring at the same time (Fig 5.2.3.b).

5.2.4 Patient J.D.

Tooth movement on patient J.D. was followed for 70 days. The appliances were removed on day 70, since the middle and distal magnets were in contact and there was no activation remaining. A total posterior movement of 3.2 mm was measured at the end of that period on the experimental side (Fig 5.2.4.b). The control cuspid moved posteriorly by approximately 2 mm during the same interval.

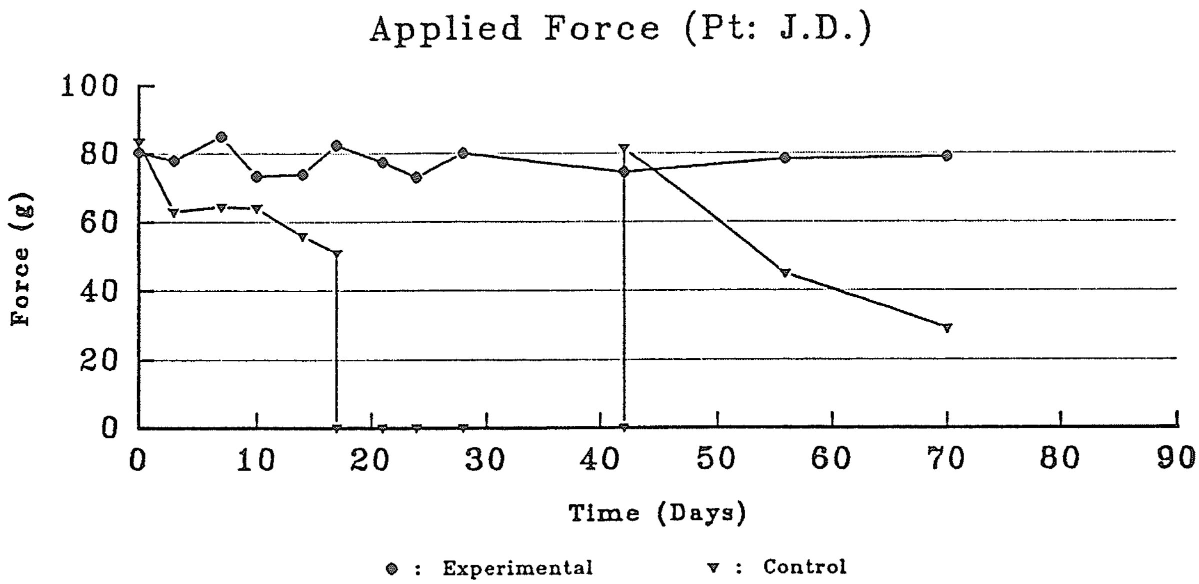


Figure 5.2.4.a Forces in the posterior direction acting on the experimental and control cuspids of patient J.D.. Analysis of the loop showed plastic deformation on the control side, between days 17 and 42.

Plastic deformation of the loop on the control side was noted between day 17 and day 42, at which time the loop was "reformed" prior to re-activation, as explained in section 4.4. Consequently, no posteriorly directed force was exerted on the control cuspid during the same period (Fig 5.2.4.a). The antero-posterior movement exhibited by the control cuspid during this interval (Fig 5.2.4.b), can be explained by the similar patterns of rotational movements about the z-axis and the y-axis, between day 17 and day 42.

The experimental cuspid had tipped posteriorly by a total of 2° at the end of the experiment, whereas the control cuspid had tipped posteriorly by 4.2°. The total amount of postero-medial rotation was 10.7° for the experimental, and 4.1° for the control cuspid.

Plastic deformation of 0.45 mm was found on the control loop and 0.25 mm on the experimental loop at removal of the appliances.

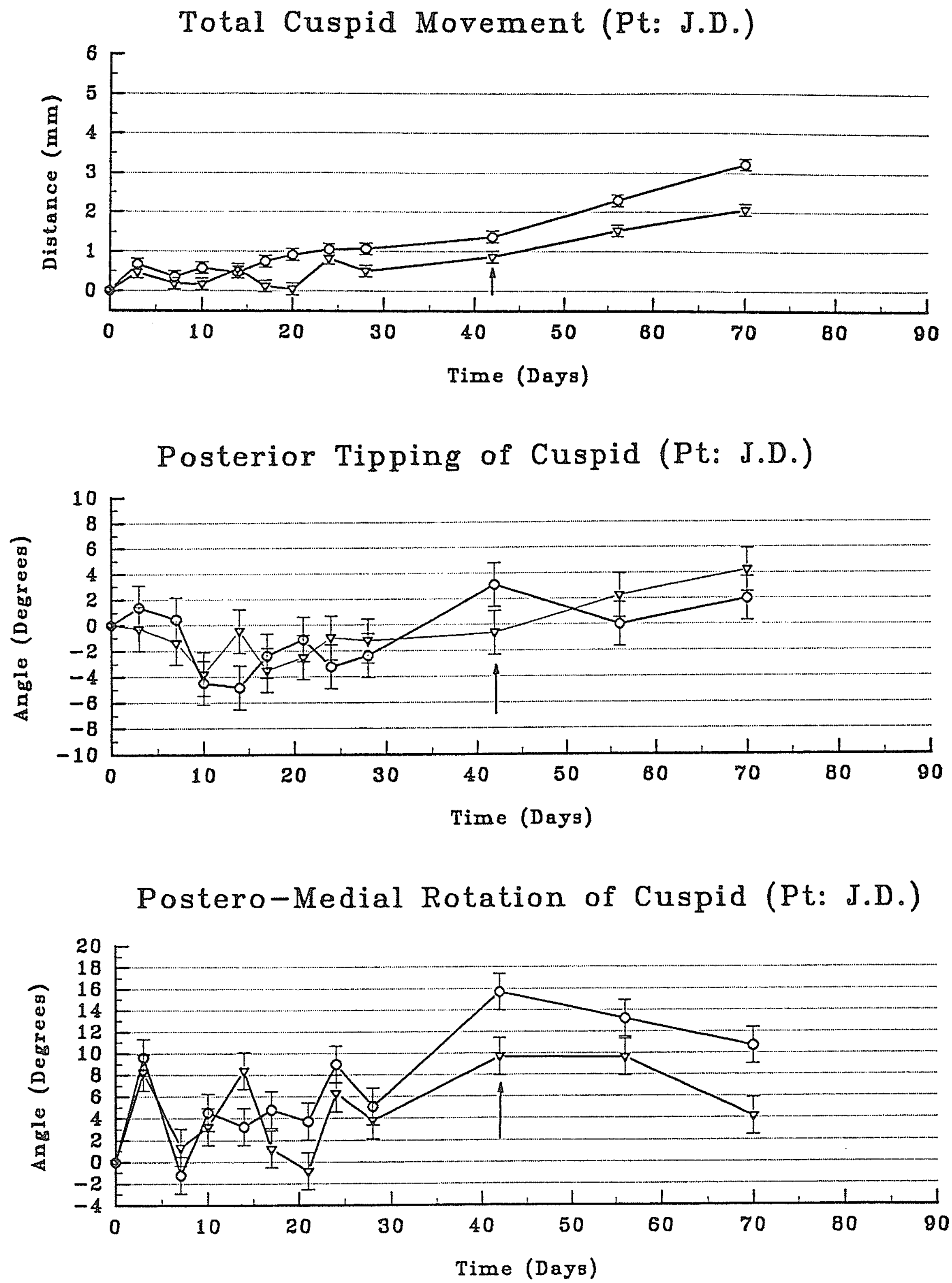


Figure 5.2.4.b Total distance travelled, tipping and rotation of the experimental (○) and control (▽) cuspids on patient J.D.. The arrow indicates the time of reactivation on the control side.

5.2.5 Patient D.R.

Cuspid retraction was monitored in patient D.R. for a total of 84 days. At the end of that period the experimental cuspid had been moved posteriorly by 2.2 mm, and the control cuspid by 1.6 mm (Figures 5.2.5.b and 5.2.5.c).

Surprisingly, the net posterior movement seems to have been greater on the control side until approximately day 65. This becomes difficult to explain, considering the fact that the loop analysis results showed plastic deformation of the control loop between days 10 and 42, when presumably very little, if any, antero-posterior force was acting on the control cuspid (Fig 5.2.5.a). Considerable rotational movements about the z-axis and the y-axis appear to have occurred during the same interval (Fig 5.2.5.b).

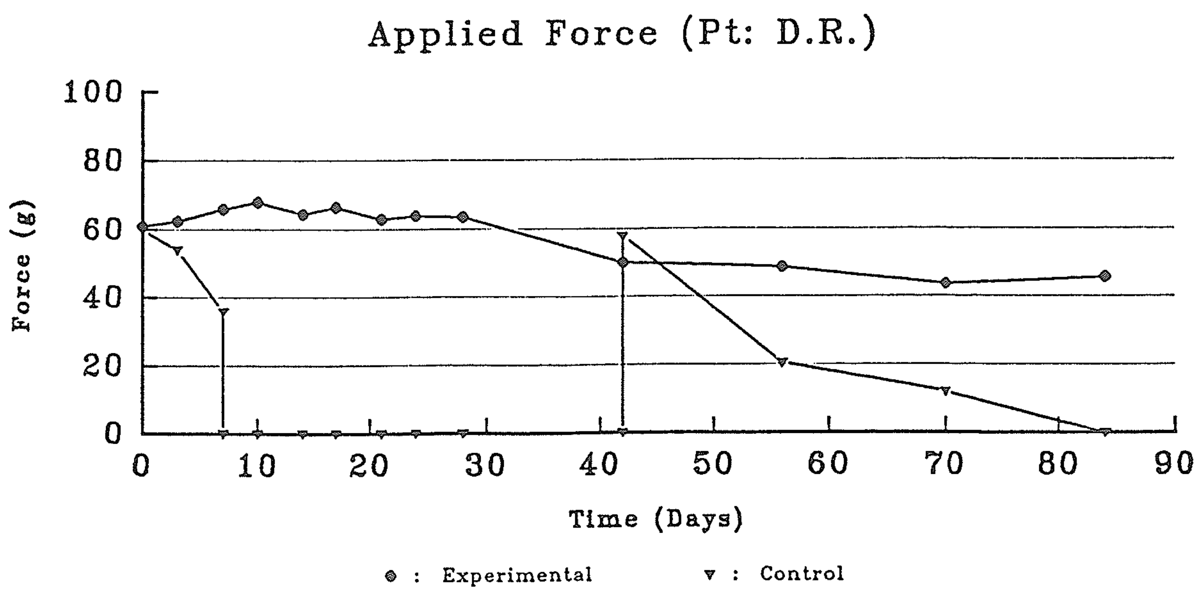
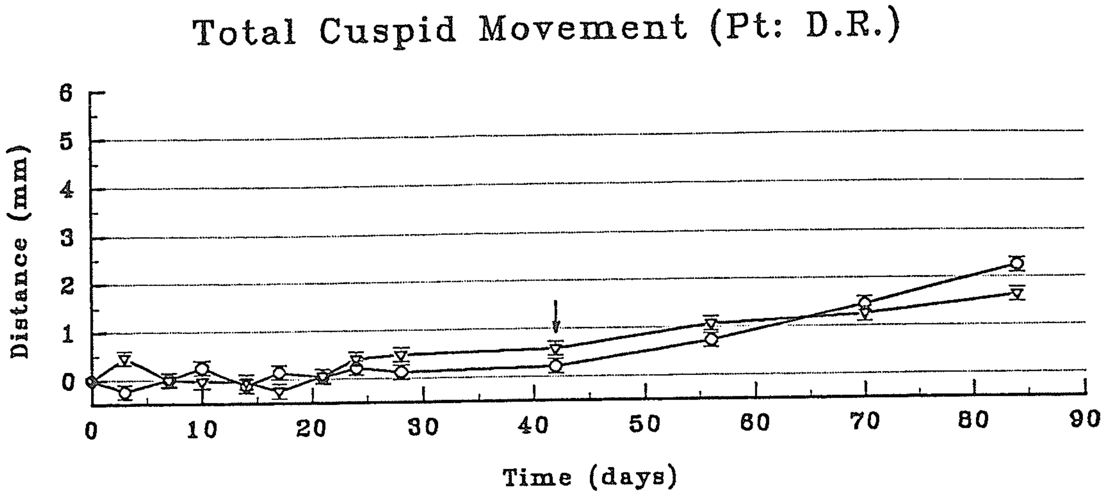


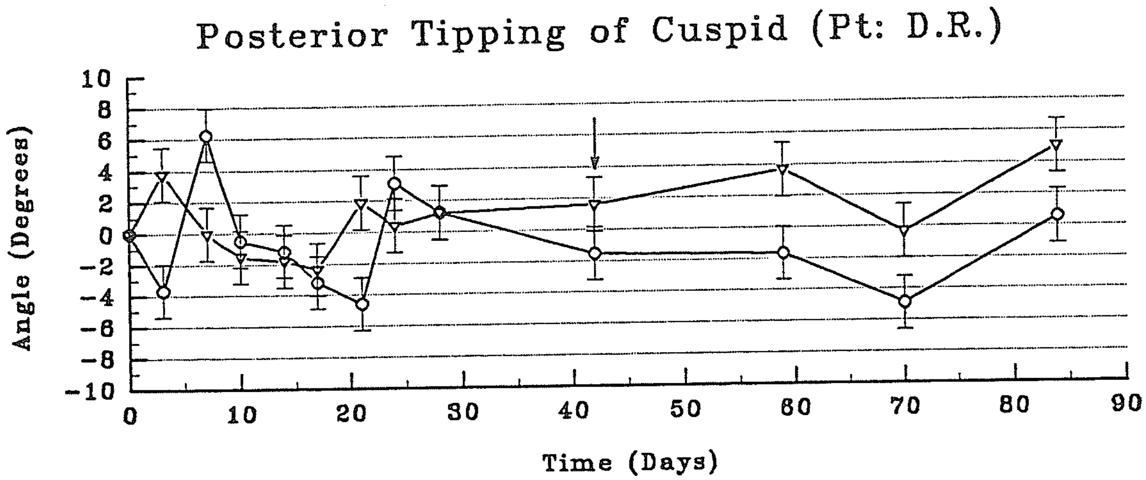
Figure 5.2.5.a Forces in the posterior direction acting on the experimental and control cuspids on patient D.R., for the duration of the study. The control loop was plastically deformed between days 10 and 42.

Both loops were found to have slight plastic deformation, and hence were "reformed" on day 42 (see section 4.4). At that time, the loop on the control side was also reactivated.

The amount of anterior movement and anterior tipping that was experienced by the experimental cuspid in the first three days from insertion of the appliances can be attributed to lack of passivity of the loop on that side at insertion. The remarkable posterior tipping (by 9.9°) of the same cuspid during the following 4 days (from day 3 to day 7) is probably exaggerated by error of observation.

The total amount of rotation about the z-axis was 5° (crown posterior) on the control, and 0.6° in the same direction for the experimental cuspid. The experimental cuspid was also rotated postero-medially by a total of 13.8°, in contrast to the control side which was only rotated postero-medially by 2.6°. No plastic deformation was evident in either of the loops at removal of the appliance, and the loop on the control side had been completely de-activated.





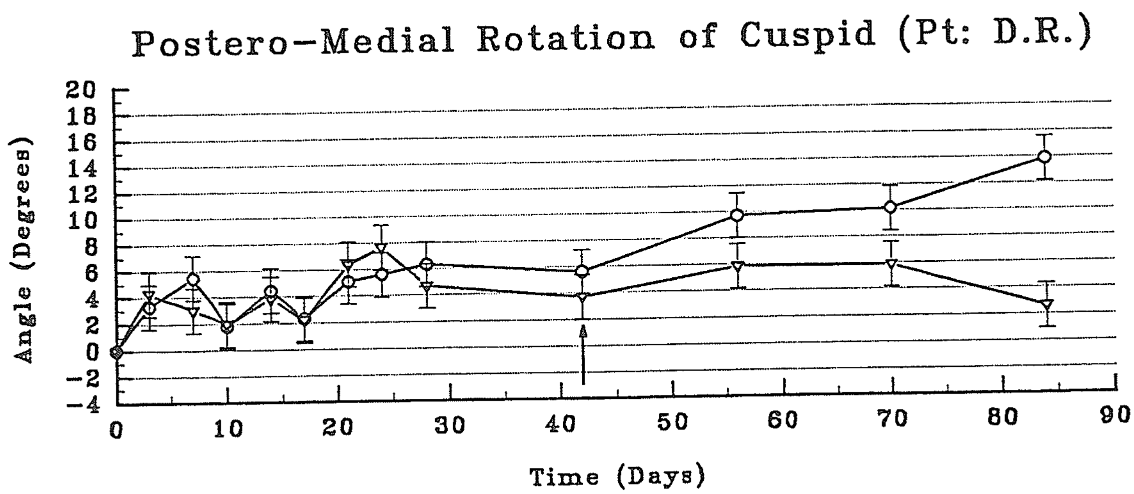


Figure 5.2.5.b Total distance travelled, tipping and rotation for both the experimental (○) and the control (▽) cuspid on patient D.R., throughout the duration of the study. The arrow indicates the time of reactivation on the control side.

5.2.6 Patient D.B.

Tooth movement was followed for a total of 70 days in patient D.B. The experimental cuspid, at the end of the experiment, was found 5.2 mm posterior to its initial position whereas the total posterior movement on the control side was 3 mm (Fig 5.2.6.b).

This distance for the experimental cuspid appears unreasonable at first. Because of the design of the appliance the maximum distance that the middle magnet is allowed to travel from the beginning to the end of the experiment is 4.25 mm (equal to the maximum separation between the middle and the distal magnet, at insertion). However, in D.B., manipulation of the loop on the experimental side while checking for permanent deformation on day 42, resulted in a failure in the welded junction between the stainless steel extension carrying the middle magnet, and the TMA loop. The loop was subsequently re-welded further distally on the stainless steel piece. This was done for two reasons: a) It was impossible to re-weld at exactly the same point on the wire, since the surface of the wire was no longer intact at that point. b) It was recognized that little activation of the appliance was remaining at the time, and the experiment would have to be terminated prematurely. The loop was spot-welded in such a way that, after re-insertion, the middle magnet lay mesially to its previous position. In this way, the total effective separation between the middle and distal magnets was increased, and some further activation of the appliance became available.

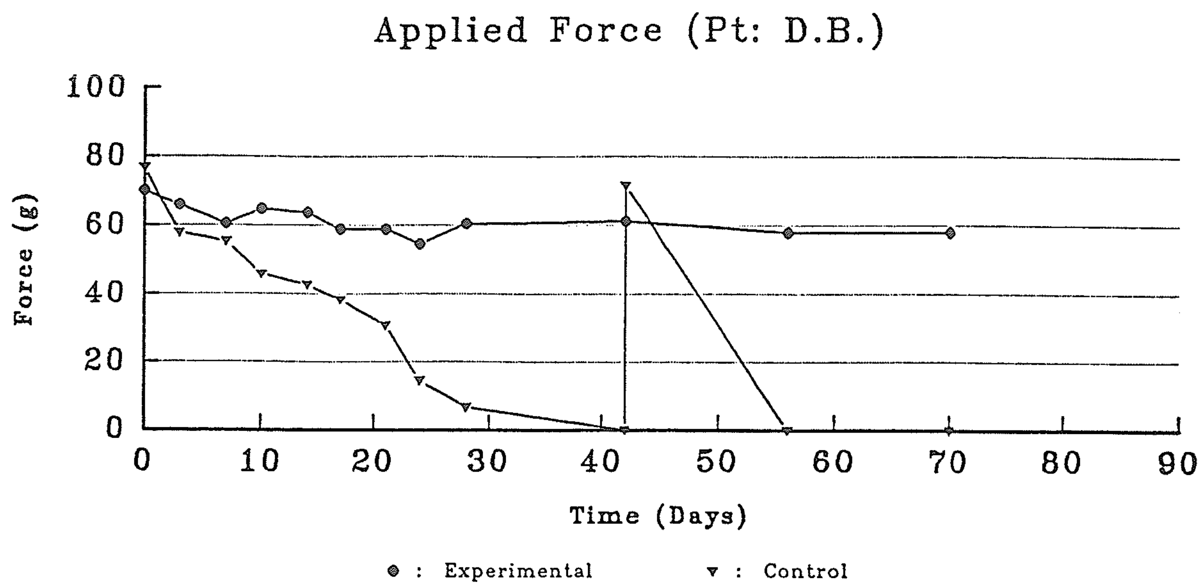


Figure 5.2.6.a The posteriorly directed forces applied on the experimental and control cuspids of patient D.B. during the study.

The activation on the control side had completely worked out by day 42, when the loop was re-activated, and no plastic deformation was found on either side. After re-activation, the loop on the control side was closed again in the next 14 days (between day 42 and day 56). The resulting forces can be seen in Figure 5.2.6.a.

The experimental cuspid had tipped anteriorly by 1° at the end of the experimental period, whereas the control cuspid had tipped posteriorly by 6.7° (Fig 5.2.6.b). The total postero-medial rotation at removal of the appliances was 7.9° for the experimental, and 12.2° for the control cuspid.

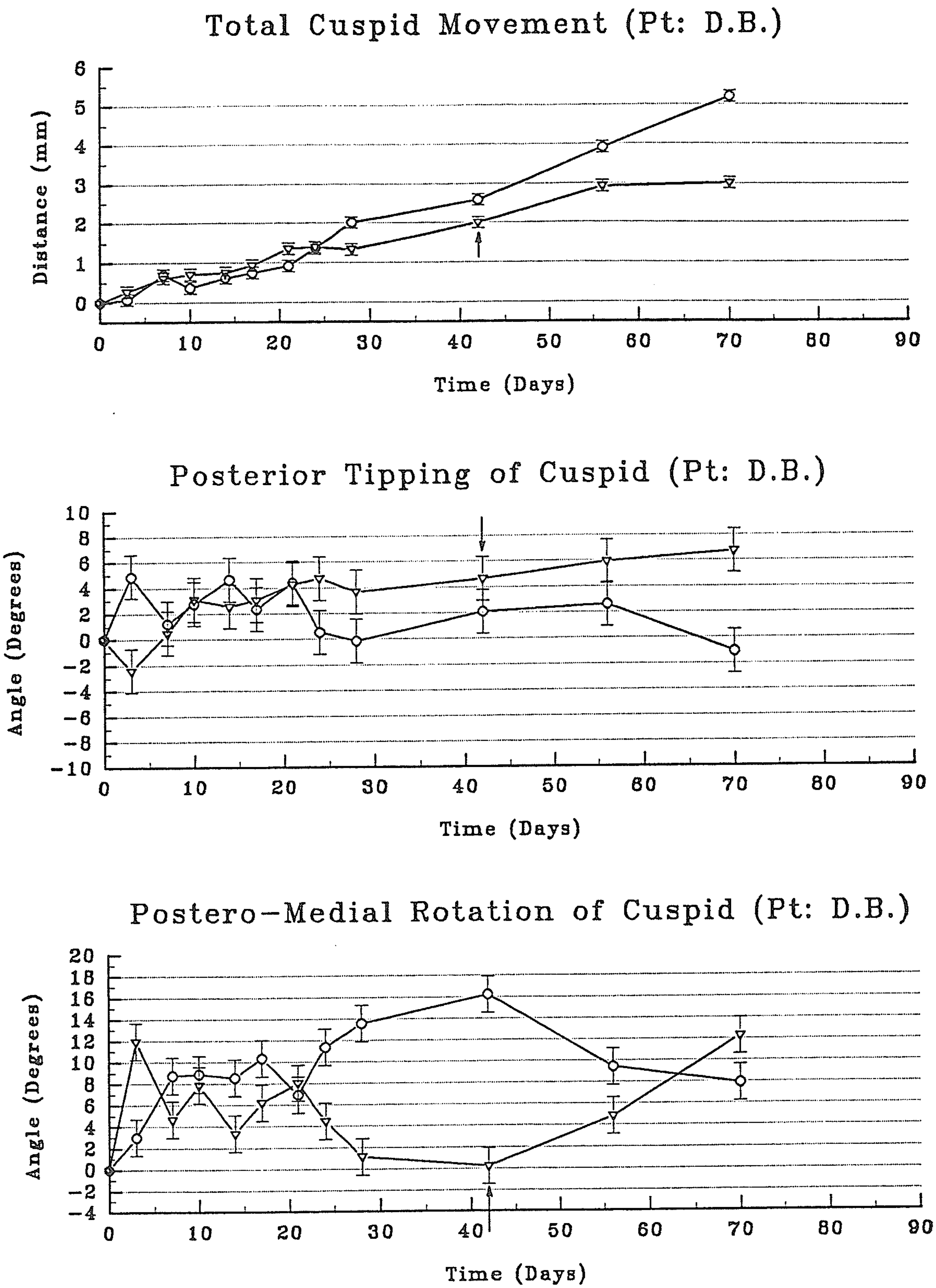


Figure 5.2.6.b Total distance travelled, tipping and rotation of both the experimental (○) and the control (▽) cuspid on patient D.B.. The arrow indicates the time of reactivation on the control side.

5.3 SUMMARY OF RESULTS

The mean cuspid movement for all patients at the end of the experimental period (approximately day 70) on the experimental side was 2.95 mm, as compared to 1.61 mm on the control side. The difference in cuspid movement between the experimental and control sides was statistically significant at the end of the experimental period ($p < 0.05$), but not at the end of the first activation (approximately day 42).

The cuspid on the experimental side moved at a significantly higher rate ($p < 0.05$) than the control cuspid, both overall and during the second phase of the experiment (after the re-activation on the control side). The average overall values were 0.63 mm/28 days on the control and 1.22 mm/28 days on the experimental side. During the second phase of the experiment, the control cuspids moved at an average rate of 0.8 mm/28 days and the experimental cuspids at an average rate of 1.62 mm/28 days.

During the first 20 to 30 days there was no appreciable posterior movement of either cuspid in most cases and, as mentioned, the differences between the two sides were not statistically significant. After that initial period, the posterior movement of the experimental cuspid was greater than the control, at all times, except in patient D.R.. The posterior movement of the experimental cuspid in this patient was smaller than the control cuspid until approximately day 65. However, a higher rate of movement was exhibited by the experimental cuspid both overall, and in the second phase of the study, which was a consistent finding in all patients.

The comprehensive results are summarized in Tables 5.3.a and 5.3.b.

	Total Cuspid Movement in ~ 42 days (mm)		Total Cuspid Movement in ~ 70 days* (mm)	
Patient	**Control**	**Experimental**	**Control**	**Experimental**
J.D.	0.87	1.38	2.07	3.22
D.R.	0.58	0.22	1.25	1.46
C.O.	0.57	2.25	1.88	4.35
N.S.	0.38	1.14	0.85	1.97
P.M.	0.30	0.69	0.64	1.53
D.B.	2.01	2.59	3.01	5.20
Mean	**0.78**	**1.38**	**1.61**	**2.95**
Exp/Cont	**1.77**		**1.83**	

Table 5.3.a Comparison of total cuspid movement between experimental and control sides, in the first part of the study (before reactivation on the control side), and at removal of the appliances. (* denotes statistically significant difference at the $p < 0.05$ level)

	Overall Rate of Movement* (mm/28 days)		Rate of Movement from ~ Day 42 to End of study* (mm/28 days)	
Patient	Control	Experimental	Control	Experimental
J.D.	0.83	1.29	1.20	1.84
D.R.	0.54	0.75	0.70	1.35
C.O.	0.75	1.74	1.31	2.10
N.S.	0.32	0.75	0.47	0.83
P.M.	0.16	0.69	0.13	0.98
D.B.	1.20	2.08	1.00	2.61
Mean	**0.63**	**1.22**	**0.80**	**1.62**
Exp/Cont	**1.93**		**2.02**	

Table 5.3.b Comparison of the rates of tooth movement, overall and for the second part of the study (after re-activation on the control side), between the control and experimental cuspids. (* denotes statistically significant difference at the $p < 0.05$ level)

CHAPTER 6

DISCUSSION

DISCUSSION

6.1 INTRODUCTION

The principal purpose of this study was to examine the differences between the rate of tooth movement produced by the constant (magnetic) force relative to the impulsive force of the conventional loop. For this purpose, data were collected and analyzed for a total of 12 teeth (6 experimental and 6 control). Although the sample size was small, the findings were generally significant in a statistical sense.

The cuspids on the experimental side were retracted over a statistically significantly greater distance than those on the control side, for the duration of the experiment (2.95 mm vs. 1.61 mm, on average).

The overall rates of tooth movement achieved on the experimental side were also statistically significantly higher than on the control side (1.22 mm/28 days vs. 0.63 mm/28 days, on average). As well, during the second phase of the study (after the control loop was re-activated) the experimental cuspid moved at a statistically significantly higher rate than the control (1.62 mm/28 days vs. 0.8 mm/28 days, on average).

The high clinical significance of these findings is evident considering that the differences between the experimental and control sides are in the order of 2:1. The rates of movement observed on the experimental side are surprisingly higher than the rates that have been reported so far for translation of teeth, with no evidence of tissue pathology.

The results from the loop analysis show that the force on the experimental side was maintained at reasonably constant levels compared to the force on the control side. The force diagrams demonstrate the dramatic differences between the two force systems. The forces typically used in orthodontic practice are very similar to the impulsive, rapidly decaying forces produced by the loops on the control side. On the other hand, forces of constant magnitude are very rarely used. Arguments such as the fact that "the periodontal ligament is given time for regeneration" have been used by orthodontists to support the desirability of such forces. The findings of this study suggest that there is practical wisdom in the quest for alternate ways of producing forces of a constant magnitude in everyday clinical practice. Any such mechanism of constant force delivery should maintain sufficient rigidity in all directions other than the intended direction of movement, to allow control over unwanted tooth movements.

Inter-individual variability has been a major impeding factor in tooth movement studies. In this study, bilateral comparisons and known, pre-calibrated force systems were utilized in an attempt to minimize sources of variation. The next step after these results seems to be the design of a study using constant forces of different magnitudes in order to define the optimum forces for tooth movement.

The intraoral occlusal photographs of patient D.R. before insertion of the appliances (I) and at the end of the experimental period (II), are shown in Figure 6.1 to give one a clinical feeling for the quantitative data that follow.

(I)

(II)

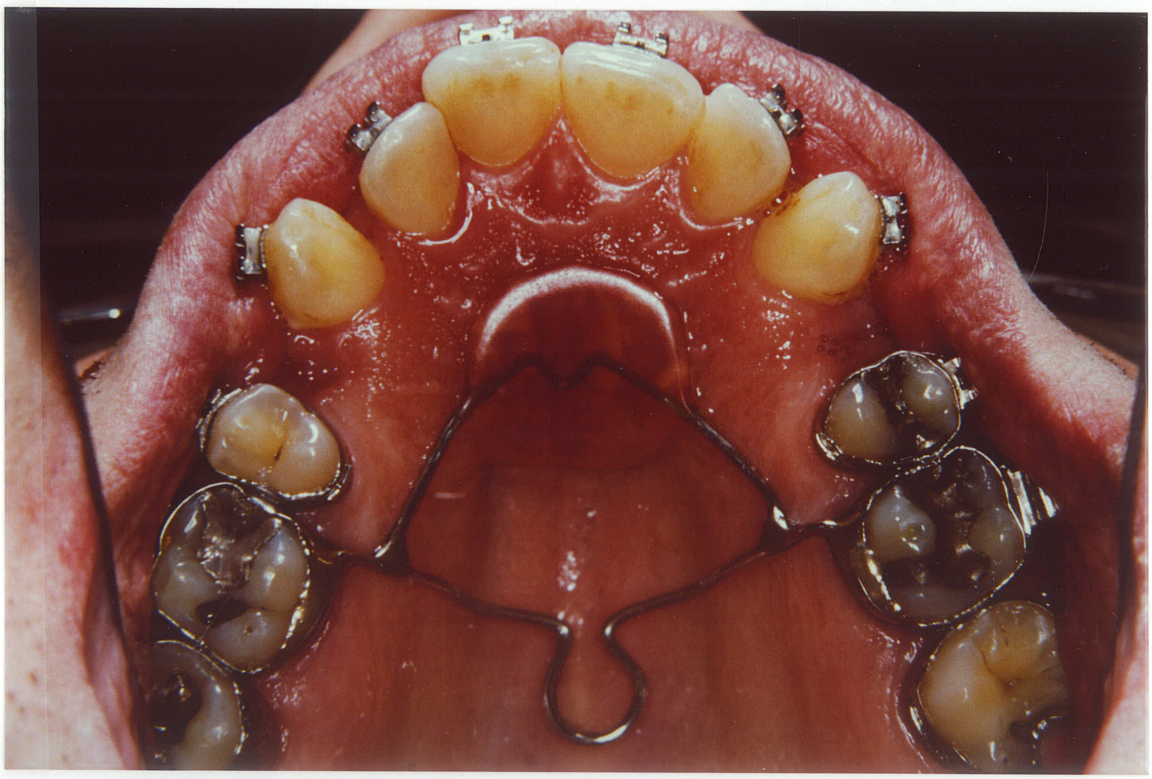


Figure 6.1 Maxillary occlusal view of patient D.R. before insertion of the appliances (I) and at the end of the experimental period (II).

6.2 STABILITY OF THE REFERENCE POINTS

The difficulty in identifying and establishing stable references intraorally is well recognized. The desirability of such references in studies such as this is obvious if accurate tooth movement measurements are to be made.

In this study measurement of cuspid movement was performed relative to selected points on the posterior anchor teeth. Therefore, the stability of the anchor segment was of critical importance in permitting accurate records of tooth movement.

The stability of the anchor segments was verified in two ways. First, for each patient, the same individualized orientation splint (constructed on the Day 0 cast) was utilized on all subsequent casts. For each patient, the splint fitted all respective sequential casts with negligible rocking and movement of the splint on the casts. This consistency for all patients over time implied that the anchor teeth on both sides remained stable in relation to each other throughout the duration of the study.

Second, three points on the rugae were selected on the initial (Day 0) and the final cast of each patient. The stability of rugae over reasonable time intervals, has been demonstrated by van der Linden (1978), and was confirmed by the findings of Cohen (1991). The distances between the reference points on the anchor segments and the selected rugae points were measured, and the mean distances were found for each cast. The mean distances before and after the experiment were then compared. At no time was the difference found to be greater than 0.5 mm, which is within the reported range of stability of the rugae. One should note that the differences were in the order of 0.5 mm only on two occasions, and these were on the experimental side of patients D.B and C.O. The total movement of these two cuspids was the greatest observed in the study, at 5.2 mm and 4.3 mm, respectively.

The above evidence supports the conclusion that the reference points on the anchorage segments were adequately stable to serve as reference points for assessment of the relative movement of the cuspids.

6.3 ACCURACY OF THE TOOTH MOVEMENT MEASUREMENT TECHNIQUE

In order to estimate the error of the method, two separate trials were undertaken (section 4.5.4). The first one addressed the error of measurement for a single point on the cast-measuring apparatus. The coordinates of the same point on a single cast were measured 20 times and the standard deviation of these measurements was calculated as an indicator of the consistency of the measurements (intra-operator variability). The standard deviations were ± 0.028 mm for the measurements along the x-axis, ± 0.014 mm for the measurements along the y-axis, and ± 0.037 mm for the measurements along the z-axis.

The second trial involved the estimation of the overall technique error, including the duplication of the reference points in subsequent casts using the perforated acrylic templates. For this purpose, a single cast was duplicated ten times, and the coordinates of the same four widely-spaced points, duplicated in all ten casts, were determined. The greatest standard deviation in this trial was ± 0.147 mm, and is a meaningful (worst case) indicator of the overall accuracy of the method for linear distance measurements. It is emphasized again that this was the greatest standard deviation encountered in only one cast and for only one dimension. It is conceivable that the reported error may be an over-estimation of the actual error, since the second trial incorporated a source of error that was not present in the actual study, namely that of duplication of the same cast ten times.

In order to estimate the degree of error for angular measurements, the three dimensional analysis described in Section 4.5.3. was performed for five casts of different patients, based on linear distance values that differed by ±0.15 mm (equal to the maximum S.D. for linear measurements as determined earlier) from the actual values that had been measured during collection of data. The greatest standard deviation found for angular measurements was ±1.69°.

The linear measurements were therefore determined to be accurate to approximately ±0.15 mm with a probability of 68%, or to ±0.30 mm with a probability of 96%. Similarly, the angular measurements were found to be accurate to approximately ±1.7° with a probability of 68%, or to ±3.4° with a probability of 96%.

The data from the error trials have been included in Appendix D.

6.4 ACCURACY OF THE FORCE MEASUREMENTS

As mentioned in Section 4.1, all loops were calibrated, prior to insertion, on the force-measuring machine which was developed by Paquien (1978). The machine was capable of measuring forces from 0 to 1,000 g and moments from 0 to 20,000 g-mm, with a total error of ±3% of the full scale. This included both the random error (scatter) and the systematic error of the instrument.

The data from loop calibration were combined with data from monitoring of the loop opening at each appointment, the latter being derived from a photographic method. It is recognized that the weakest part of the study was the technique for force assessment. The inaccuracy of the adopted method can be blamed on the fact that the optical differences between the camera and the measured loop were very difficult to control, granted that the focal distance used was quite small. However, the data obtained were used for a relative assessment, and in conjunction with the data from loop calibration. Therefore, although high accuracy cannot be claimed, meaningful conclusions can be drawn using reasonable approximations of applied force values.

6.5 PATIENTS' RESPONSE

Patient's tolerance of the experimental appliance was a matter of concern in designing the study. The attractive properties of the magnets, and the relative bulkiness of the appliance, were the factors which the principal investigator anticipated would create most potential problems. Consequently, the experimental design included a plan to fabricate tooth-borne removable appliances with buccal shields that would separate the magnets from the buccal musculature, had patient tolerance been low.

At no time were any such difficulties encountered. In fact, two of the subjects (P.M. and D.B.) commented on the fact that they had experienced slight transient discomfort on the control side after reactivation, as compared with the experimental cuspid, which had moved comfortably and uneventfully.

There were some occasions in which the loops suffered some degree of distortion, possibly because of chewing on something hard, a situation which is quite commonly encountered in orthodontic practice. As well, at times some loops suffered some plastic deformation, possibly as a result of stress relaxation of the loop wire. As a result, not all cuspids had the posteriorly directed force acting on them for the intended period of time. An effort was made to "reform" the loops that had undergone plastic deformation, at the time of reactivation of the control side in order for the desired retraction force to be restored (section 4.4).

6.6 BIOLOGICAL CONSEQUENCES

As discussed in Section 2.3.2, there is no conclusive evidence on the biological effects of magnetic fields. The existing studies seem to agree that the potential risk is proportional to the intensity of the field and the duration of the exposure. Some of the current, established applications of magnets in medicine and dentistry involve exposure of human patients to very powerful magnetic fields (e.g. MRI), or for a very prolonged period of time (e.g. for retention of overlay dentures in prosthodontics). In this experiment subjects were exposed to a weak field, for a relatively short time (approximately three months).

No discomfort or change in taste were reported by any subject. No signs of discolouration or corrosion of the magnets were observed.

Periapical radiographs of both control and experimental cuspids were taken on all subjects prior to the initiation of cuspid retraction, and at removal of the appliances. The radiographs were examined by an oral pathologist, as well as the principal investigator. In all cases there were changes in the periodontal ligament and lamina dura consistent with orthodontic tooth movement, namely widening of the periodontal ligament and attenuation of the lamina dura. There was no evidence of root resorption. The trabecular pattern of the alveolar bone was stable. In three cases there were alveolar crest abnormalities characterized as either an increased distance of the CEJ from the alveolar crest or a post-extraction defect. These changes in all three cases were located on the distal aspect of the right maxillary cuspids (D.R., P.M. and J.D.).

6.7 TOOTH MOVEMENT

In analyzing the data one of the most striking observations was that, in all cases, movement of the cuspids on the graphs could be divided in two regions. During the first 20 to 30 days of activation of the appliance minimal net movement was noted. Instead, a series of "jiggling" movements was observed. This finding can only partially be attributed to the fact that more frequent appointments and thus, more frequent collection of data was performed during the first month of the study for each patient. Evidence from the graphs suggests that a sequence of anterior and posterior tipping movements was exhibited during this period in both the experimental and the control teeth. This finding is consistent with the findings of Cohen (1991). One has to note, however, that these initial small movements are within the error of measurement as revealed by the overlapping of the error bars on the graphs.

Significantly greater movement of the experimental cuspid relative to the control was a consistent finding in all patients, at the end of the study. These results support the initial hypothesis and are in agreement with the view expressed by Storey and Smith (1952 and 1954), Burstone (1961), Pletcher (1958) and Gjessing (1985) that a light force of continuous nature can produce the most effective tooth movement.

The total movement on the experimental side at the end of the experiment, ranged from 1.9 mm (N.S.) to 5.2 mm (D.B.). The total movement on the control side ranged from 0.5 mm (P.M.) to 3 mm (D.B.). The overall rate of tooth movement on the experimental side ranged from 0.69 mm/28 days (P.M.), to 2.08 mm/28 days (D.B.). In contrast, the overall rate on the control side ranged from 0.16 mm/28 days (P.M.) to 1.2 mm/28 days (D.B.). The lowest rate in both experimental and control cuspids was exhibited by the same patient (P.M.).

In the second region on the graphs, after the initial period of minimal net movement (20 to 30 days), considerable movement was observed in both control and experimental cuspids. The curve tends to be steeper on the experimental side as compared to the control side, representing a greater velocity (higher rate of movement) on the former.

The same is true even for patient D.R., in whom the control cuspid had moved posteriorly more than the experimental one until approximately day 65. Due to the higher rate of movement of the experimental cuspid in the second region of the graph, this tooth had travelled a greater distance at the end of the study, compared to the control cuspid.

On some occasions, the initial movement of the cuspid was in the anterior direction, (e.g. patient D.R., experimental side). This result appears paradoxical, and is in contrast to the findings of Cohen (1991) who found that the initial movement was a fast posterior movement attributed to compression of the PDL. An explanation for this could be that the loop was not completely passive when inserted in the cuspid bracket (i.e. the part of the loop that was to be inserted in the cuspid bracket did not completely line up with the bracket slot), and some existing angular activation resulted in anterior tipping of the tooth.

The observed anterior movements in later stages during retraction may be explained by other simultaneous tooth movements, such as tipping and rotations. More specifically, the total antero-posterior movement of the cuspid was assessed relative to point C2 which, in most cases, was on the cusp tip. It was calculated that for a typical tooth, with an estimated average distance from C2 to the centre of resistance of 8 to 10 mm along the y-axis, posterior tipping of the tooth by 6° to 8°, would result in posterior movement of C2 by approximately 1 mm. It is conceivable that such a tipping movement occurred on the experimental side in patient P.M., between days 7 and 10. Similarly, assuming that the tooth is rotating around an axis through its centre of resistance, with an estimated average 3 mm distance from C2 to the centre of resistance along the z-axis, postero-medial rotation of 10° would result in posterior movement of C2 by 0.5 mm. A rotation of this nature may account for the posterior movement observed on the control side on patient J.D. between days 21 and 24.

The graphs representing total posterior movement of the cuspids could have been corrected to account for the simultaneous tipping and rotating movements. However such correction is not worthwhile, because the total movement of the cuspid was, in most cases, much greater than that contributed by other simultaneous movements and because the number of data points collected was not sufficient for an accurate compensation.

Evidently, not all individual antero-posterior movements of teeth between appointments can be explained by considering the simultaneous tipping and rotational movements. These cases can be accounted for by measurement error. As mentioned previously, the measurement error for linear measurements was ± 0.15 mm, and for angular measurements $\pm 1.7°$. However, combination of errors in more than one dimension could have been accumulated to produce larger errors in some instances.

The overall rate of 0.16 mm/28 days observed on the control side on patient P.M., is quite interesting, especially since that cuspid only moved a total of 0.5 mm in 87 days. A possible explanation for this finding may be that the tooth, according to the tipping diagram, was eventually tipped anteriorly by approximately 5°, which could mean that an additional 0.5 to 0.75 mm could be added to the final 0.5 mm, had it been uprighted. Corroborating evidence to this explanation is the finding that the initial movement of the tooth was anterior, which means that there was already some activation in that direction at the placement of the loop.

A similar explanation could account for the apparent slower movement D.R.'s experimental cuspid relative to the control. The initial movement of the tooth was in the anterior direction, possibly due to the presence of some degree of angular activation in the loop at insertion. The tipping diagram confirms that the tooth was for the most time during the study tipped anteriorly, as opposed to the control which was tipped posteriorly.

The total movements of 5.2 mm and 4.3 mm found on the experimental side in patients D.B and C.O., respectively, appear to be incredible, but should be evaluated taking into account the fact that these two cuspids underwent considerable postero-medial rotation (7.8° and 15.4°, respectively). In addition, as mentioned in Section 5.4, the greatest anterior movement of the anchorage segments (0.5 mm), was noted with respect to these two cuspids.

The overall rates of tooth movement observed on the control side were similar to those reported by Cohen (1991), (between 0.57 mm/28 days and 0.99 mm/28 days). Although vertical loops were also used for cuspid retraction in that study, the initial forces utilized were significantly larger (from 215 g to 285 g). Similarly, Duff (1988), reported rates between 0.6 and 0.8 mm/month, with initial forces between 100 and 250 g. Thus, the same rates of tooth movement were achieved on the control side, with forces 2.5 to 4 times lower than these previous studies.

The rates of tooth movement observed on the experimental side were significantly higher than the usual clinically derived rates. As already discussed the differences are not only statistically significant, but also clinically significant, since the values are twice as high on the experimental side (1.21 mm/28 days vs. 0.63 mm/28 days, overall, and 1.61 vs. 0.8 mm/28 days in the second phase). This finding can be justified by the fact that the forces used on the experimental side in this study are markedly different than the ones utilized in a standard clinical situation.

In most cases, similar amounts of tipping were found between the control and experimental teeth at the end of the experimental period. The greatest difference was seen in D.B. (6°). Tipping ranged from 5° anteriorly (P.M. experimental), to 10° posteriorly (N.S. experimental).

The considerably greater tipping values found at the end of the experimental period in N.S. (10° for the experimental and 7° for the control cuspid) were attributed to the lower stiffness of the appliance used exclusively on this patient. As mentioned in section 4.4 this was the first patient in whom the appliance was inserted, and the sole one in whom the loops were made entirely of TMA. The stiffness of TMA is approximately 1/2 of that of stainless steel for the same dimension (Burstone *et al.*, 1980). Evidently, the appliance was not rigid enough to constrain the teeth from spurious movement. In the remaining patients, TMA loops with spot-welded stainless steel extensions were used. This clearly illustrates that stiffness of an orthodontic appliance in the directions in which no tooth movement is intended is critical in order to control unwanted movements (section 6.1).

The net rotation around the y-axis in all twelve teeth was predictably in the postero-medial direction. Significant variability can be observed in this type of movement, with a range from 2.6° (D.R., control), to 15.4° (C.O., experimental). Undoubtedly, postero-medial (disto-lingual) rotation is clinically desirable during cuspid retraction to various degrees (depending on the initial position of the cuspid), as the tooth is moved "around the corner" of the arch.

The magnitude of the force on the experimental side should be approximately the same for all patients, since the magnets and the separations were the same, and the acrylic jigs described in section 4.4, were constructed to achieve a force of approximately 75 g in all cases. Probably due to weaknesses of the photographic technique which was used for assessment of the loop opening, in some cases the measured openings of the loops resulted in reporting initial forces somewhat higher than the expected 75 g.

6.8 ENERGY CONSIDERATIONS

At this point, the concept of *potential energy* should be introduced. This is the energy available from each retraction mechanism for movement of the tooth. On the control side, the energy available for tooth movement was highly dependent on the distance that the tooth moved under the influence of the retraction force. Therefore, the energy available from the spring, after some movement had already occurred, was less than the initial available energy. Conversely, on the experimental side, the force delivered by the spring was such that, no matter how much the tooth moved, the magnitude of the force remained constant and, consequently, the amount of energy available for tooth movement was also constant.

As already mentioned, the total movement and the overall rate of tooth movement on the experimental side was invariably higher than on the control side. In an attempt to explain this difference by means of differences in the applied force systems, a secondary hypothesis was put forth, namely that the total potential energy available from each retraction mechanism can be correlated to the rate of the resulting tooth movement. The total potential energy can be estimated on the force/time diagrams, by calculating the area under the curve, for each tooth. By dividing the resulting area by the number of days that the experiment lasted, an estimation of the *mean force* applied on the tooth for the duration of the experiment, can be obtained. Mean force is a measure of the potential energy available to the tooth over the experimental period. The area between the curve and the x-axis was measured for each individual tooth, using a digitizer and a computer program. From the data presented in Table 6.5.a, it is evident that the potential energy available to the teeth on the experimental side was consistently higher than that available to the control teeth. As a result, the mean forces on the experimental side were consistently higher than the control. However, when the ratios between the mean forces on the two sides for an individual patient are considered, there is some evidence of a relationship between those ratios and the corresponding rates of tooth movement.

Tooth		Total Time (days)	Avail. Energy	Mean Force (g)	Rate (mm/ 28 days	Mean Force exp/con	Rate exp/con
D.B.	con.	70	1772	25.3	1.20	2.42	1.73
	exp.	70	4292	61.3	2.08		
J.D.	con.	70	2701	38.5	0.83	2.05	1.55
	exp.	70	5524	78.9	1.29		
D.R.	con.	84	1345	16	0.54	3.47	1.38
	exp.	84	4673	55.6	0.75		
C.O.	con.	70	1829	26.1	0.75	2.57	2.32
	exp.	70	4699	67.1	1.74		
N.S.	con.	73	2880	39.4	0.32	2.02	2.34
	exp.	73	5833	79.9	0.75		
P.M.	con.	87	1657	19	0.16	2.97	4.31
	exp.	87	4916	56.5	0.69		
				Means		2.58	2.27

Table 6.5.a Overall values of available energy, mean force and rate of movement for experimental and control teeth. A relationship seems to exist between mean force applied to a tooth and rate of tooth movement.

These data offer some valuable insights. The force/time diagrams for patients P.M. and D.R. are very similar. In fact, the potential energy input on both sides of both patients is approximately the same, and the mean forces applied on the experimental teeth are reasonably similar (55.6 g in D.R, as compared to 56.5 g in P.M. on the experimental side, and 16 g in D.R as opposed to 19 g in P.M. on the control side). The rate of tooth movement on the experimental side is also very similar in the two patients (2.2 mm in 84 days, or 0.75 mm/28 days in D.R., as opposed to 2.1 mm in 87 days, or 0.69 mm/28 days in the case of P.M.). Startlingly however, the control cuspid in the case of D.R moved posteriorly by 1.6 mm in 84 days (at a rate of 0.54 mm/28 days), whereas the respective tooth on P.M. only moved by 0.5 mm in 87 days (a rate of 0.16 mm/28 days).

In the same context, patients J.D. and N.S. had very similar force/time patterns and comparable mean force values for both experimental (78.9 g and 79.9 g, respectively) and control cuspids (38.5 g and 39.4 g, respectively). Despite this fact, the rates of tooth movement observed are quite different (1.28 mm/28 days v. 0.75 mm/28 days, respectively, on the experimental, and 0.82 mm/28 days v. 0.32 mm/28 days, respectively, on the control side).

A rational explanation for these findings cannot be provided, other than that they could be yet another expression of inter-individual variability. However, it may be noted that the overall results include the period of "jiggling" in which minimal net movement was found and this period is variable.

Tooth		Net Time (days)	Avail. Energy	Mean Force (g)	Rate (mm/ 28 days)	Mean Force exp/con	Rate exp/con
D.B.	con.	28	552	19.7	1	3.05	2.61
	exp.	28	1683	60.1	2.61		
J.D.	con.	28	1419	50.6	1.2	1.55	1.53
	exp.	28	2176	77.7	1.84		
D.R.	con.	42	840	20	0.70	2.33	1.92
	exp.	42	1962	46.7	1.35		
C.O.	con.	28	1095	39	1.31	1.79	1.60
	exp.	28	1962	70	2.1		
N.S.	con.	28	1938	69.2	0.47	1.16	1.76
	exp.	28	2252	80.4	0.83		
P.M.	con.	42	782	18.6	0.13	2.77	7.53
	exp.	42	2097	49.9	0.98		
				Means		**2.10**	**2.82**

Table 6.5.b Values of available energy, mean force and rate of movement for control and experimental teeth during the second phase of the experimental period (after reactivation of the control side). There is evidence of a relationship between mean force applied on a tooth and rate of tooth movement for at least four out of six patients.

Table 6.5.b shows the corresponding values of potential energy, mean force and tooth movement rate for the second phase of the study (after the reactivation on the left side). In this case, there seems to be a relationship between the ratios of the mean force between the two sides, and the ratios of rate of tooth movement observed during the

same time period between the two sides. This is quite evident in at least four patients (D.B., J.D., D.R., and C.O.).

The results mentioned support the secondary hypothesis. The energy available from each mechanism appears to be a deciding factor in the rate of tooth movement produced. Since, on the experimental side, there was a continuously applied force the available energy on that side was consistently greater than that on the control side, hence the greater resulting rate of tooth movement on the former side.

If the total available energy is the main factor affecting the rate of tooth movement, the type of force (i.e. whether it is applied continuously or intermittently) may not be important. An intermittent force with a very high initial magnitude, could produce comparable total potential energy values (i.e. area under the force/time curve) with those observed in the present study using a continuous force. Another secondary hypothesis can therefore be proposed, namely that the type of force (i.e. whether it is intermittent or continuous) does not play as big a role in the regulation of the rate of tooth movement as the total potential energy available to the tooth. Thus, an intermittent force of a high initial magnitude can produce similar rates of tooth movement with the ones observed under the influence of the magnetic continuous force.

In order to test this additional secondary hypothesis, the results from two previous studies (Cohen, 1991; and Duff, 1988) were adapted and plotted similarly to the data reported here, and truncated to a two-month period to match the current study. Both studies also used vertical loops for cuspid retraction, but significantly higher force magnitudes were employed. The data from two patients from Duff (1988) and four patients from Cohen (1991) are presented. The total energy, mean force and rate of tooth movement for 6 teeth is depicted on Table 6.5.c.

Tooth	Total Time (Days)	Available Energy	Mean Force (g)	Rate of movement (mm/28 days)
J.H. (Duff)	60	5953	99	1.16
C.K. (Duff)	60	3773	62	0.32
D.R. (Cohen)	50	7549	150	0.71
B.J. #23 (Cohen)	59	9045	153	0.57
B.J. #13 (Cohen)	59	6255	106	0.67
J.B. (Cohen)	43	5688	132	0.73
		Means	**117**	**0.69**

Table 6.5.c Values of available energy, mean force and tooth movement rate for cuspids that were retracted with vertical loops under the influence of impulsive forces of significantly higher initial magnitudes than the ones utilized in the present study (adapted from Cohen, 1991 and from Duff, 1988).

It is recognized that the sample size is small, but the data from these previous studies negates the latter secondary hypothesis. Evidently, the potential energy available for tooth movement in these studies was significantly greater than that on the experimental side in this study. Consequently, the mean force values in table 6.5.c are up to twice as high as the mean force on the magnet side. Despite this fact, the rates of tooth movement produced by intermittent forces in the former studies are indisputably lower than these accomplished by the approximately constant force produced by the magnets.

It seems plausible, therefore, to assume that the duration of force application potentially plays a critical role in regulating the rate of tooth movement. It may be that maintenance of a force of higher magnitude than the critical "optimal" value over a considerable period of time, results in greater tooth movement. From the findings of this study, there is no indication that forces of higher magnitude produce faster tooth movement. Rather, two parameters that were identified as being important were total potential energy available and duration of force application.

CHAPTER 7

CONCLUSIONS

CONCLUSIONS

The following conclusions were drawn from the study, based on analysis of the collected data:

1) Under the conditions of this study, movement of the cuspid was appreciably greater, and the rate of movement was significantly higher, when the cuspid was retracted under the influence of an approximately constant force.

2) Detectable net tooth movement does not occur until after the first 20 or 30 days after the initiation of force application. During this initial period, a series of changes in the angulation of the tooth seems to take place, presumably due to the uneven pressure distribution along the compressed part of the PDL, with consequent creation of hyalinization areas.

3) In light of the results by Cohen (1991) and Duff (1988), impulsive forces of a much greater initial magnitude and substantially larger mean values than those used on the control side in the present study, do not appear to improve the rates of tooth movement.

4) The duration of force application and the total available energy are important factors in regulating the rate of tooth movement. Conversely, force magnitude does not appear to be of primary significance, once the threshold for tooth movement is exceeded.

5) The method of force generation used in the present study seems to offer constant potential energy availability, while permitting controlled movement of the tooth.

6) Selective stiffness of an orthodontic appliance (in the directions in which no tooth movement is desirable) is critical in creating the appropriate constraint forces to avoid spurious movement.

CHAPTER 8

RECOMMENDATIONS FOR FUTURE RESEARCH

RECOMMENDATIONS FOR FUTURE RESEARCH

Based on the results of this investigation, recommendations for future studies in the area of tooth movement include:

1) The study of tooth movement under the influence of constant forces of different magnitudes, in order to identify the range of optimal forces.

2) The development of more accurate ways of clinical assessment of forces applied to a tooth. The use of a sophisticated force gauge, or a means to precisely measure the separation of the loop in a clinical setting, could be considered as solutions.

3) The investigation of alternative ways of delivering forces of a constant magnitude.

4) Making the assessment of tooth position simpler and more accurate, would greatly facilitate clinical studies with larger samples. The use of a computer and a video camera capable of scanning a cast and accurately recording three-dimensional position information for individual teeth, is an enticing possibility.

5) The investigation of tooth movement under the influence of forces of constant magnitude applied for varying durations could offer some insight into the initiation of tooth movement and possibly identify the ideal force characteristics for this intriguing period.

6) The comparison of rates of movement achieved with constant forces to those produced with interrupted forces of pulsating magnitude could further clarify the temporal distribution of the optimal force.

7) The study on non-human subjects of tooth movement under carefully controlled impulsive and constant force systems, from both a biomechanical and a biological standpoint, may truly reveal the long-sought answers to questions such as the relationship between the rate of tooth movement and potential biological effects, optimal force range, and factors predisposing to root resorption.

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APPENDIX A

MAGNETS, MAGNETIC FIELDS

Magnetic fields have the unique property to generate forces that increase in magnitude as the distance between the two juxtaposed magnets is diminished. This is true for both repulsion and attraction. For a given pair of magnets, the only determinant of the force produced by the combination of the fields is the separation between the magnetic poles, namely the *air-gap*. The maximum force that can be produced by two given magnets in attraction is termed the *breakaway*. The breakaway is equal to the force required to separate the two magnets when they are in contact.

A generally accepted principle with respect to the use of a single magnet is that the force generated by the field at any point, is inversely proportional in magnitude to the square of the distance of that point from the magnetic pole (Coulomb's law, $F \sim 1/d^2$). Theoretically, when two magnets are employed, the field interaction is such that the rate of decrease of the force is somewhat less than the square of the distance between the magnetic poles, but for practical reasons, the latter is considered a very close approximation (Blechman, 1985). Therefore, the force/distance relationship for two magnets that are separated along a linear path parallel to their magnetization axes is hyperbolic.

The Energy Product BH, which is calculated by multiplying the field flux density B (Gauss) and the magnetomotive force (or intensity of the field) H (Öersted), is an indication of stored energy and potential force generated (Blechman, 1985). The higher the BH the stronger the magnet, and ultimately the greater the force that is available.

When one line of force per unit area emanates from a pole of a magnetic field, the flux density is defined as 1 Gauss.

The flux density (Gauss) is equal to the field intensity (Öersted) times the permeability of the material. Since the permeability of air is about 1, the flux density is essentially equal to the field intensity (Ketchen *et al.*, 1978).

The flux is analogous to a current in an electric circuit. If the two poles of a magnet can be connected by a plate made of any ferromagnetic material such as iron, the external flux field is shunted through the ferromagnetic plate (the *keeper*) because this is the path of least resistance. Thus, if other magnets and/or keepers are used, any magnet can be set up in an assembly forming a complete circuit (*closed-field assembly*).

Conventional permanent magnetic alloys include Pt-Co, ferrite and Al-Ni-Co (alnico). These alloys, when cut down to very small proportions (3 to 5 mm), lose almost all of their force. This is because, at small dimensions, the north pole comes in close proximity to the south pole and the opposite polarities cancel one another. If conventional ferrite or alnico permanent magnets are machined down to proportions of 3 to 5 mm in diameter, their breakaway is usually in the range of 5 to 10 g (Jackson *et al.*, 1987).

In 1967, a new class of permanent magnets was discovered by Joseph Becker, of General Electric Research laboratory and Gary Hoffer of the Air Force Materials Laboratory (Becker, 1970). They found that when a transitional element (cobalt or iron) was alloyed with a class of elements known as the lanthanum series (the rare earth elements), permanent magnets with very remarkable properties could be produced.

The first rare earth magnets to be commercially produced were fabricated from the transitional element cobalt and the rare earth element samarium (Sm Co_5, Sm_2Co_{17}). Pt-Co magnets would have to be 4 times greater in volume, ferrite magnets 200 times greater, and alnico magnets even larger in order to exert the same force as a Sm-Co magnet (Hayes *et al.*, 1990).

The second generation rare earth magnets were produced from the transitional element iron and the rare earth element neodymium ($Nd_2Fe_{14}B$). The addition of a third element, boron, has increased the fundamental stability of the crystalline structure. The newer Nd-Fe-B magnets are 70% more powerful than Sm-Co magnets of the same size (Sandler *et al.* (1989). However, they are brittle and, therefore, need to be handled with care.

Rare earth elements, despite their name, are not particularly rare and are not extremely expensive. They are found in such industrial items as misch metal, which is used to make flints for cigarette lighters. This family of elements has historically been called the rare earth elements because, although fairly abundant in nature, they are hard to separate and isolate in pure form. Only with the recent development of ionization column separation techniques has it been possible to economically produce rare earth elements. The rare earth element samarium is a nontoxic element which is also used in certain medicinal salts to alleviate sea sickness.

Rare earth magnets are produced by making a single-phase alloy in an alumina crucible, using a radio frequency induction furnace with a helium protective atmosphere. The ingots cast from this single-phase alloy are ground into particles less than 10 μm in size. This powder is pressed into shape in a very strong magnetic field and then sintered in an inert atmosphere. These magnets are extremely resistant to demagnetization. For this reason, the material allows magnetic components to be miniaturized without losing field strength (Jackson *et al.*, 1987).

APPENDIX B

THE FORCE-MEASURING MACHINE

The machine used for calibration of the loops and for assessment of the forces produced by the magnets in this study was essentially the same as that used in the studies by Duff (1988), Sonya (1988) and Cohen (1991). The instrument, which was developed by Paquien (1978), is capable of measuring simultaneously the three-dimensional components of the forces and moments produced on a tooth by a regular orthodontic spring. The essential components of the machine are:

a) a measuring system,

b) a data acquisition system, and

c) a minicomputer.

The above mentioned instrumentation was linked to an IBM computer with multiple display functions.

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

The five major parts of the instrument are the frame, the triangular shaped mounting block in the centre of the stationary component of the instrument, the internal suspended ring with Type A transducers, the external ring with Type B transducers, and an electromagnetic vibrator (Fig B.1). The vibrator permits electromechanical excitation of the mounting block, in order to eliminate the effects of stiction of the sliding contacts. The Type A transducers are attached on the internal suspended ring and measure horizontal forces and pivoting moments. The Type B transducers are attached to both the external ring (which is part of the frame) and the internal suspended ring and measure vertical forces and tipping (azimuthal) moments.

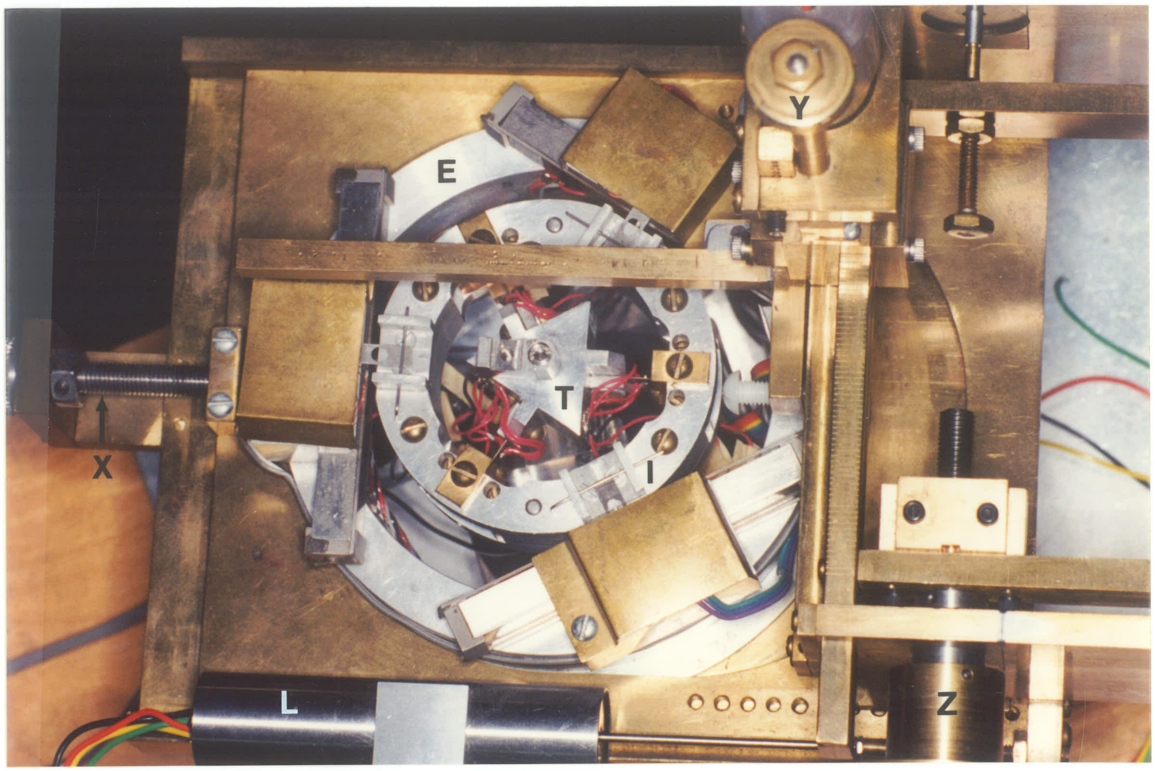

Figure B.1 Detailed view of force-measuring machine. (T: Triangular Mounting Block; I: Internal ring with Type A transducers; E: External ring with Type B transducers; F: Frame; L: Linear Voltage Displacement Transducer (LVDT); X,Y,Z: Adjustment screws for movement along the x, y, and z-axis, respectively).

The end of the spring to be calibrated is attached on the mounting block in the centre of the stationary component of the machine. The other end is attached on the mobile component, which is capable to move in all three dimensions of space by means of three adjustment screws (Fig B.1).

The instrument has a maximum force range of 1,000 g and a maximum moment range of 20,000 g-mm. The total error, including both the random (scatter) and systematic errors is ±3% of the maximum force and moment values.

For purposes of this investigation, specially machined brass attachments were used to fix the magnets on the measuring system (Fig 4.5.1.a). The attachments carrying the mesial and distal magnets were screwed onto a horizontal bar on the mobile component of the machine. The attachment carrying the middle magnet was fixed on the mounting block, in the centre of the stationary component. Readings of the forces produced on the middle magnet were taken at various air-gaps, monitored by the minicomputer. The posterior movement of the middle magnet as a result of cuspid retraction in the clinical situation was simulated by moving the mobile component of the instrument (and thus the mesial and distal magnets) relative to the middle magnet, by means of the adjustment screws (Fig 4.5.1.b).

APPENDIX C

UNIVERSITY OF MANITOBA - FACULTY OF GRADUATE STUDIES

TOOTH MOVEMENT STUDY: CONSENT FORM

I have agreed to participate in a study of tooth movement, to be conducted by John Daskalogiannakis, graduate student at the M.Sc. level. I have read the information sheets about the study, and it has also been explained to me by Dr. Daskalogiannakis. I understand that the study will take place as part of my orthodontic treatment, and that in excess of the regular dental records, I will be required to have impressions of my teeth made, twice a week during the first month and once every two weeks during the remaining two months of the study. I understand that special small magnets will be placed in my mouth, which may make my braces more bulky and bothersome.

I understand that there are no specific, personal benefits to me as a result of my participation in this study, except for a $200 discount on the standard fee, but that the results of the research are expected to contribute to scientific knowledge in the area of orthodontic treatment. I understand that the risks of personal harm or discomfort involved in this study are very small. It has been explained to me that although the possibilities are remote, should any problems associated with my participation in this study occur and persist, I will be seen for advice and\or appropriate treatment at the University of Manitoba, Graduate Orthodontic Clinic.

I consent to having special magnets placed in my mouth as part of my braces, for the duration of the study.

I consent to having impressions of my teeth taken repetitively for dental models, as well as having intraoral photographs of my teeth taken regularly during the course of the study.

I understand and consent to the fact that any such records will become the property of the University of Manitoba, and will be held in confidence, but may be used for research publication and presentation purposes.

I have volunteered to participate in this study on my own accord, and I realize that I am free to withdraw from participation at any time.

Signature of subject: ______________________________

Signature of Witness: ______________________________

Date: ______________________________

APPENDIX D

ERROR STUDY DATA

A separate study was undertaken to determine the possible errors of the method. Two trials were carried out as part of this error study. In the first trial the reproducibility of a single point (intra-operator variability) was determined. The second trial involved the estimation of the overall technique error.

D.1 Intra-operator variability

In the first trial the intra-operator variability was evaluated (i.e. the ability to accurately identify and measure the same point on successive casts) in order to assess the precision of the measurement technique. The three-dimensional coordinates of point BR1 (located on the maxillary right second bicuspid) on a single cast were measured 20 times using the measurement apparatus described in section 4.5.2. The standard deviation of these measurements was then calculated, as an indicator of the consistency of the measurements. The standard deviations were found to be ± 0.028 for measurements along the x-axis, ± 0.014 for measurements along the y-axis, and ± 0.037 for measurements along the z-axis. The data from the first trial can be seen in Table D.1.

Measurement Round No	x (anteroposterior or mesiodistal) (mm)	y (inferosuperior or occlusogingival) (mm)	z (mediolateral or buccolingual) (mm)
1	58.12	66.66	35.32
2	58.12	66.68	35.42
3	58.14	66.66	35.30
4	58.14	66.68	35.34
5	58.16	66.68	35.38
6	58.10	66.66	35.34
7	58.08	66.68	35.42
8	58.12	66.68	35.34
9	58.06	66.66	35.34
10	58.10	66.68	35.30
11	58.08	66.64	35.42
12	58.12	66.66	35.38
13	58.08	66.66	35.36
14	58.10	66.64	35.38
15	58.06	66.66	35.40
16	58.12	66.64	35.36
17	58.10	66.66	35.38
18	58.08	66.68	35.40
19	58.10	66.68	35.36
20	58.06	66.66	35.40
Mean	**58.10**	**66.66**	**35.36**
S.D.	**0.028**	**0.014**	**0.037**

Table D.1 Intra-operator variability.

D.2 Technique Error

The aim of the second trial was to estimate the size of the error introduced by other factors, such as the impression technique, and the duplication of the reference points in subsequent casts using the perforated acrylic templates.

For this purpose, a single cast was duplicated ten times using a high precision liquid silicone material (see section 4.5.4). The duplicate models were placed on the measurement apparatus and the anteroposterior and mediolateral coordinates of four widely spaced points (BR1, BR3, BL1 and BL3), transferred in all casts by means of the perforated acrylic templates, were determined. The linear distances between these points in the x and z dimensions were then calculated for all duplicate casts, and the means and standard deviations of these calculations were determined. These data can be seen in Table D.2.a.

The greatest standard deviation encountered in this trial was approximately ± 0.15 mm (± 0.147 mm) and is reported as a meaningful (worst case) indicator of the overall accuracy of the method for linear distance measurements. It is conceivable that this error may be an over-estimation of the potential error, since the second trial incorporated a source of error that was not part of the actual method (i.e. the duplication of the same cast ten times).

Duplicate Cast Code	Anteroposterior (mesiodistal)		Mediolateral (buccolingual)	
	Distance BR1-BR3 (mm)	Distance BL1-BL3 (mm)	Distance BR1-BR3 (mm)	Distance BL1-BL3 (mm)
A	14.18	19.68	10.74	0.60
B	14.28	19.64	10.84	0.64
C	14.24	19.66	11.14	0.56
D	14.26	19.60	11.08	0.54
E	14.24	19.62	10.84	0.60
F	14.16	19.62	11.14	0.62
G	14.30	19.60	11.10	0.58
H	14.06	19.66	11.08	0.62
I	14.10	19.54	11.12	0.56
J	14.12	19.56	11.04	0.66
Mean	**14.20**	**19.61**	**11.01**	**0.598**
S.D.	**0.069**	**0.044**	**0.147**	**0.039**

Table D.2.a Technique error for linear measurements.

In order to estimate the degree of error for angular measurements, the three-dimensional analysis was performed on five casts of different patients, based on linear distance values that differed by ± 0.15 mm (equal to the maximum S.D. for linear measurements, as determined previously) from the actual values that had been measured from the collection of data. In this way, the variation of the measured angle was estimated, when the linear distances on which its calculation was based, changed by $\pm$ 1 S.D.. The greatest standard deviation found for angular measurements was $\pm$ 1.69°. The data on the accuracy of angular measurements are shown in Table D.2.b.

Angle		Actual angular value found using 3D analysis (degrees)	Value from 3D analysis based on linear x,y,z distances increased by 0.15 mm (degrees)	Value from 3D analysis based on linear x,y,z distances decreased by 0.15 mm (degrees)
Cast #1	θ_x	32.34	33.08 (0.74)	31.86 (0.48)
	θ_y	3.64	5.01 (1.37)	2.14 (1.5)
	θ_z	57.68	57.32 (0.36)	58.20 (0.52)
Cast #2	θ_x	43.90	44.19 (0.29)	43.65 (0.25)
	θ_y	83.10	81.58 (1.52)	84.79 (1.69)
	θ_z	46.87	47.01 (0.14)	46.79 (0.08)
Cast #3	θ_x	27.32	26.86 (0.46)	27.64 (0.32)
	θ_y	85.98	86.11 (0.13)	85.84 (0.14)
	θ_z	62.92	63.13 (0.21)	62.69 (0.23)
Cast #4	θ_x	39.56	39.61 (0.05)	39.34 (0.22)
	θ_y	87.40	87.24 (0.16)	87.55 (0.15)
	θ_z	50.58	50.39 (0.19)	50.67 (0.09)
Cast #5	θ_x	40.17	40.12 (0.05)	40.36 (0.19)
	θ_y	88.80	88.96 (0.16)	88.65 (0.15)
	θ_z	49.67	49.84 (0.17)	49.59 (0.08)

Table D.2.b Technique error for angular measurements. (The differences between the derived angular values and the values calculated by the 3D analysis are shown in parentheses).