IN VIVO TESTING OF FRICTION ON ORTHODONTIC ARCH WIRE

BY

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- Pathology 0571
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A Thesis/Practicum submitted to the Faculty of Graduate Studies of The University

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MASTER OF SCIENCE

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DEDICATION

To my parents Sheila and Jerry Ziedenberg, who continually pushed me to be the best I could be.

To my wife Jill, whose support, hard work, and sacrifice made the realization of my dreams possible.

ABSTRACT

In clinical practice it has been assumed that during sliding mechanics, friction between orthodontic appliances and arch wire inhibits tooth movement. As a result, friction may be a major component affecting the duration of orthodontic treatment. Frictional resistance experienced by orthodontic appliances may be influenced by vibration arising from mastication and by the lubricating properties of saliva, which can only be properly assessed in the oral environment. To enable this investigation, a clinical technique was devised to simulate retraction of a bracket along an orthodontic arch wire which could be studied in both *in vivo* and *in vitro* environments. The objective of this study was to test the hypothesis that intraoral vibration during mastication does not reduce the *in vivo* coefficient of friction compared to *in vitro* measurements.

A clinical trial was undertaken using ten subjects that were orthodontic patients from the University of Manitoba. Subjects had alginate impressions made of their upper arches in order to produce a model on which a friction testing device could be accurately fabricated. An individual friction testing apparatus was fabricated for each subject and during experiments was inserted over existing orthodontic appliances. While wearing the friction testing device, subjects were asked to masticate chewing gum base in order to cause maximum vibration of the teeth and surrounding soft tissue. Each subject underwent two *in vivo* testing sessions with their individual friction testing apparatus. In addition, there were two *in vitro* testing sessions to obtain additional data. Comparison of the static coefficients of friction were made for two sliding tube lengths as well as for *in vitro* and *in vivo* testing environments.

Variables such as bracket width, the presence of saliva, and chewing vibration were incorporated into the investigation to determine their role in the reduction of the static coefficient of friction. Statistical analysis of the data was performed using a multiway analysis of variance (ANOVA).

Under the conditions of this study, intraoral vibration decreased the *in vivo* coefficient of friction as compared to *in vitro* results for both long and short tubes ($p < 0.001$). The decrease in the static coefficient of friction from the *in vitro* to *in vivo* testing environment was similar for both sliding tube lengths ($p<0.001$). Long and short tubes exhibited different coefficients of friction with long tubes having higher static coefficients of friction than short tubes for both testing environments. Intraoral vibration resulted in a decreased *in vivo* frictional resistance of the long and short sliding tubes, while the presence of saliva significantly increased the *in vitro* static coefficient friction for both long and short tubes ($p<0.001$). The increase in the coefficient of friction from dry to wet *in vitro* testing was greater for the long tubes ($p<0.02$)than for the short tubes ($p<0.05$). Frictional resistance was never completely eliminated with intraoral vibration.

This investigation identified that intraoral vibration dramatically reduced frictional resistance when compared to *in vitro* tests using the same apparatus. This confirms that, limiting orthodontic product evaluation to *in vitro* testing does not give accurate insight into clinical performance. Understanding of the factors influencing friction has the potential to advance the design of mechanics, making orthodontic treatment more efficient.

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

ABBREVIATIONS AND SYMBOLS

$<$	Less than
%	Percentage
β- Titanium	beta-Titanium
ANOVA	Analysis of Variance
C R	Center of Resistance
cc	cubic centimeter
d	Length of the sliding tube
df	Degrees of freedom
F	Force
F_N or N	Normal force
F_T	Tractional force or Motive force
Kg	Kilogram
L	Length of moment generating spring
M.S.	Mean of squares
mm	millimeters
M_S	Moment generated by moment spring
M_T	Moment of sliding tube
NCSS	Number cruncher statistical system
r	Spearman correlation coefficient
R^2	Spearman correlation coefficient squared

S.S.	Sum of squares
SEM	Scanning Electron Microscopy
TiN	Titanium Nitride
TiO	Titanium Oxide
TMA	Titanium-Molybdenum Alloy
T"	Paired T test value

CHAPTER 1

INTRODUCTION

INTRODUCTION

Movement of teeth into space created by tooth extraction is one of the most common types of tooth movement used in contemporary orthodontics. In cases where tooth retraction is preferred, bodily movement (translation) is the type of movement that is desired. Translation requires that a uniform distribution of stress occurs over one entire aspect of the periodontal ligament, which will induce uniform bone resorption. (Hixon *et al.*, 1970)

A single force will result in translation only if it passes through the "Center of Resistance" (CR) of a tooth. Periodontally healthy teeth normally have their CR located approximately just beneath the alveolar crest. However, contemporary orthodontic techniques limit the location of force application to the crowns of teeth. The distance between the point of force application (at the tooth's crown) and the tooth's CR requires the application of a counter-moment at the bracket, in addition to the required force , in order to ensure uniform stress distribution in the periodontal ligament leading to tooth translation. (Burstone, 1985)

Many previous investigations (Duff, 1988, Cohen, 1991) have demonstrated that most tooth movement devices do not produce true bodily movement, but rather a series of tipping and uprighting movements. Techniques for tooth movement can be divided into two major groups: those that involve friction and those that are frictionless. (Farrant, 1976)

Sliding mechanics commonly utilize a continuous arch wire or a segment of archwire to guide (or slide) orthodontic brackets in response to motive forces. Friction occurs at the bracket-arch wire interface as the tooth tips creating bracket binding on the guiding arch wire. The bracket binding that occurs is required in order to create the uprighting forces necessary to ensure tooth translation with sliding mechanics. The friction generated during sliding mechanics is thought to inhibit tooth movement, significantly affecting the duration of orthodontic treatment. In addition, the unpredictable magnitude of frictional resistance makes required motive forces difficult to determine.

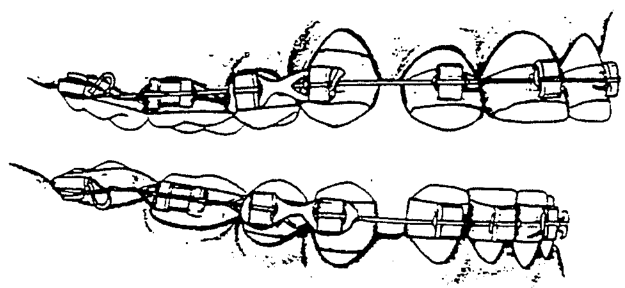


Figure 1.1 **Sliding Mechanics for Canine Retraction**
Graber,T.M. and Vanarsdall, R.L. (1994) Orthodontics; Current Principals and Techniques,: Figure 11-43A : 651

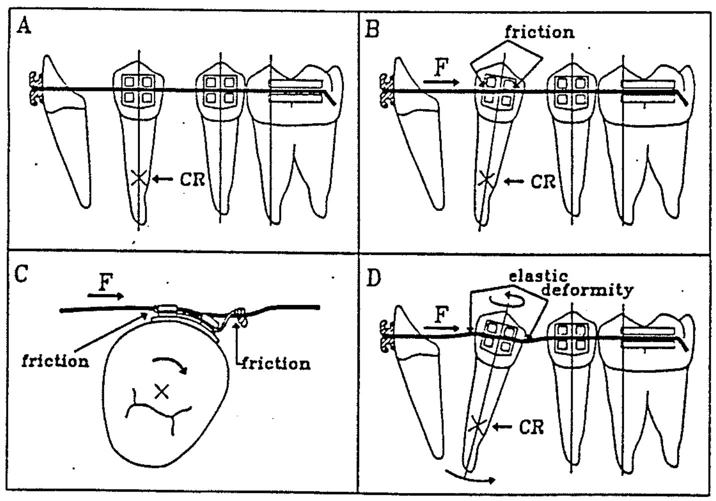


Figure 1.1b **Locations of Frictional Resistance**
B. Change of angulation of a tooth in response to a force applied at the crown
C. Rotation of a tooth in response to foce applied labial to the tooth's long axis
Drescher, D.; Bourauel, C. and Schumacher, H-A. (1989): Frictional forces between bracket and archwire, *Amer.J.Orthod.* 96:397-404.

In order to avoid friction, a loop may be utilized on a continuous or segmented arch wire. The segmented arch wire usually extends from the mesial of the canine bracket to the distal of the tube on the first molar. Retraction loops may have various shapes and configurations, and are commonly utilized to retract groups of anterior teeth, or individual teeth. Closing loop arches are an example of loops used on continuous arch wires. Tooth traction forces are usually generated by creating separation between the legs of the loop, either by tying the distal extension of the loop to the posterior teeth or by cinching the loop wire distal to the molar tube. Use of a loop in an arch wire, permits storage of potential motive force, so that orthodontic brackets need not slide along a section of arch wire for tooth movement to occur. Design of the loop and its position on the segmental arch wire determine the uprighting forces distributed to the teeth being moved. The elimination of bracket movement along an arch wire creates a frictionless system, in which tooth movement is only limited by the periodontal biology and the decay of the generated motive force.

Figure 1.2 Vertical Loop Retraction for Canine Retraction
Proffit, W.R. (1986): Contemporary Orthodontics, Figure 20-14: 513

Vertical loop retraction is thought to minimize strain on posterior anchorage as the total generated motive force may be applied to tooth movement. However, clinically successful vertical loop retraction is technically more demanding of the orthodontist's abilities. The friction produced with the use of sliding mechanics requires generation of higher motive forces for tooth movement. The equal and opposite response to these higher motive forces may lead to greater stress on anchorage units.

Many *in vitro* studies have been undertaken to observe the effects of variables upon the frictional forces observed during sliding mechanics. Most of these studies make use of very accurate frictional force testing machines in finely controlled laboratory conditions. Most *in vitro* studies are, however, limited in usefullness as they ignore vibration which may play a significant role in reducing the coefficient of static friction *in vivo*. *In vivo* or *in vitro* studies which include vibration in the testing can give a better understanding of the influence of the oral environment on orthodontic frictional forces.

In the present study, a clinical model for retraction of a bracket along an orthodontic arch wire was devised to compare and contrast the quantitiy of friction in the *in vivo* and *in vitro* environments. An attempt was made to investigate the role of variables such as bracket width, saliva, pellicle, and intraoral vibration in reducing the clinical coefficient of friction compared to *in vitro* tests. The comparision of bracket width was carried out between the right and left side of the same subject in an effort to control for intra-individual variability.

This study will demonstrate that the effects of friction *in vivo* are significant in clinical orthodontic sliding mechanics. It is our aspiration that this study will facilitate a better understanding of the nature of friction in orthodontic mechanics, and will aid in the design and production of forces which, when applied, will result in efficient tooth movement.

CHAPTER 2

REVIEW OF THE LITERATURE

LITERATURE REVIEW

2.0 FRICTION

Friction is defined as a force that resists the relative motion of two contacting bodies in a direction tangent to the plane of contact. The frictional force between any two sliding surfaces is directly proportional to the normal force which presses the opposing surfaces together. The normal force (N) is the component of force which is perpendicular to the sliding surfaces, and thus is perpendicular to the frictional component. The static frictional force is the force that must be overcome before tooth movement can occur. This relationship allows the force of friction to be represented as a product of a coefficient of friction (specific to the situation) and the normal force ($F_T=\mu_s(N)$). The static coefficient of friction (μ_s) commonly has a magnitude between zero and one, which depends upon the electrostatic interaction of material properties and the relative roughness of the surfaces in contact. The classical laws of dry friction state that a frictional force is;

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

Contemporary studies have identified the phenomenon of friction to be multifactorial in nature. With respect to the orthodontic literature; numerous variables have been found to affect the observed levels of friction between bracket and arch wire. These variables may be considered of mechanical or biological origin; or due to the experimental design used to measure friction. Mechanical variables include bracket material, bracket width and slot size, and bracket design. Arch wire material, stiffness, surface roughness, and dimension, as well as ligature material and method of ligation, have also been investigated. Saliva, plaque, acquired pellicle, and corrosion have been implicated as some of the biologic factors that affect the magnitude of friction between arch wire and bracket. An overriding factor, which has profoundly affected the outcome of many of the previously reported studies is the design of the apparatus used to measure friction. An experimental protocol that was common to many studies measuring friction prevented angulation change between bracket and arch wire. Elimination of a change of angulation between the bracket slot and the arch wire, prevents the increase in normal force seen with tipping of teeth during sliding mechanics. It is this increase in normal force which produces an increase in frictional forces. Therefore, results from studies such as these must be interpreted with caution since the experimental conditions do not accurately represent the clinical situation.

2.1 BRACKET WIDTH

To undertake an accurate and clinically relevant study of the effect of bracket width on friction, the experimental design should allow for changes in angulation of the bracket with respect to the arch wire. Earlier theoretical studies (Kamiyama *et al.* ,1973) and *in vitro* experiments (Drescher *et al.* (1989); Tidy (1989); Sims *et al.* (1994)) reported that wider brackets produce less friction than narrow brackets by allowing less angulation change of the arch wire. Tidy (1989) concluded that friction is proportional to applied force and inversely proportional to bracket width by allowing an angulation change between bracket and arch wire. More recently, Bednar *et al.*(1991) and Yamaguchi *et al.* (1996) used ingenious methods for applying a change in angulation of a bracket to an arch wire. Yamaguchi *et al.*(1996) used a similar method to Bednar *et al.* (1991). Yamaguchi and co-workers investigated the relationship of point of application of retraction force, retarding force and bracket width during *in vitro* simulated sliding tooth movement along an arch wire. Their work reported that as the retardant force was increased, retraction force applied close to the crown tended to increase frictional resistance. In addition, they identified that retraction force at the level of the bracket produced significantly higher frictional resistance for narrow brackets than for wide brackets. It was concluded that retraction force located at the level of the bracket slot creates large moments, resulting in greater tipping of the tooth, causing bracket slot binding. The bracket binding that occurs on the arch wire, causes an increased normal force, thus leading to a greater frictional resistance. Berger (1992) criticised this approach firstly developed by Bednar *et al* (1991), for Berger thought that the use of a power arm to test the frictional resistance of a Speed® bracket was inappropriate, as the normal force it produced would deform the spring clip. Berger (1992) conducted *in vitro* studies evaluating the frictional resistance of the Speed® bracket without permitting a change of bracket angulation to arch wire, and found the self ligating bracket to produce dramatically less friction.

Andreasen and Quevedo (1970), Frank and Nikolai (1980), Peterson *et al.* (1982), Kapila *et al.* (1989), and Omana *et al.* (1992) attempted to evaluate the frictional characteristics of brackets and arch wires utilizing experimental designs that did not permit a change in bracket angulation to arch wire. These investigators found that, generally, wide brackets produce greater friction than narrow brackets. Studies which restrict bracket angulation change to arch wire eliminate the generation of higher normal forces resulting from tipping when using sliding mechanics. Tooth tipping which is a result of tractional force in sliding mechanics creates the binding between the bracket slot and arch wire which generates friction. Results from studies such as these, must be interpreted with caution as they are not relevant to the clinical situation.

Andreasen and Quevedo (1970) identified that bracket width also influences interbracket distance which affects wire flexibility. Increased wire flexibility will decrease the normal force transmitted to a bracket as a result of change in angulation. Moore *et al.*(1993) used simple beam theory to demonstrate that the restoring couple of a bracket varies with the bracket width and the interbracket distance, leading to the conclusion that wide brackets have inherently less friction.

Nicholls (1968) took a different approach by fixing the angulation of a bracket and pulling archwire through it at different angles, concluding that narrow brackets allow a greater change in angulation for similar arch wire dimension. Sims *et al.*(1993) compared narrow self ligating Speed® brackets and wide self ligating Activa® brackets against wide edgewise brackets ligated using conventional elastomeric ligatures. No angulation change was permitted for any of the brackets tested. A 50-70% reduction in friction was identified between the Speed® bracket and the minitwin bracket. In addition, Shivapuja and Berger (1994) identified no statistically significant difference between the static frictional resistance of Speed® brackets of width 2.2mm with Activa® brackets of width 3.5mm because the arch wires were drawn through the brackets parallel to the bracket slots. (zero angulation change) Again, these studies bear little resemblance to the geometric changes that occur during the sliding of teeth on arch wires. Angulation changes occur, thus putting into question the applicability of the results of the above mentioned studies.

2.2 BRACKET MATERIAL

The development of orthodontic appliances which combine both acceptable aesthetics for the patient and adequate technical performance for the orthodontist has remained an elusive goal. Altering the appearance of, or reducing the size of stainless steel brackets has been one approach. Changing the material from which brackets are made to make them more asethetically pleasing is an approach that significantly affects friction at the bracket-arch wire interface. Early attempts to produce aesthetic brackets used polycarbonate. Ceramic brackets were developed as aesthetic appliances which could withstand most orthodontic forces and resist staining. Polycrystalline ceramic brackets are made of injection moulded sub micron sized particles of fused or sintered aluminum oxide, whereas monocrystalline brackets contain a single crystal of aluminum oxide. Monocrystalline brackets are machined from extrusions of synthetic sapphire, resulting in a harder bracket with more surface roughness (Birnie, 1990). It has been suggested that polycrystalline brackets were more suitable for orthodontic use than monocrystalline brackets because the polycrystalline strength did not drop dramatically following scratching from arch wire manipulation or ligation. Bednar *et al.* (1991) and Defranco *et al.* (1995) found that ceramic brackets exhibited higher frictional resistance than stainless steel brackets when bracket angulation change to arch wire was generated with a power arm. Bednar *et al.* presumed increased surface roughness of the ceramic bracket caused higher frictional resistance. Tanne *et al.* (1991) confirmed this hypothesis. SEM and a stereomicroscope investigation showed that the ceramic brackets had rougher and more porous surfaces which produced significant scratches in the arch wires. More recently, Tanne *et al.*(1994) used a metal tooth embedded in an elastic rubber form for a similar study contrasting the frictional characteristics of a new zirconium oxide bracket against an older zirconium bracket and a polycrystalline bracket. The newer zirconium bracket was found to generate less friction. SEM was used to evaluate the surface topography of the new zirconium brackets, which demonstrated smoother slot surfaces.

Ireland *et al.* (1991), Downing *et al.* (1994), and Keith *et al.* (1994) have all conducted experiments on the frictional resistance of different types of ceramic brackets. Unfortunately their experimental design prevented changes in bracket angulation relative to the arch wire. Various conclusions have been reported, yet these results must be interpreted with caution as the experimental designs utilized did not accuarately represent the clinical situation.

Tselepis *et al.* (1994) developed an approach similar to Nicholls (1968) by drawing arch wires through ligated brackets fixed at different angulations. Polycarbonate brackets were found to have the highest friction using this method. They also measured increasing friction in stainless steel, polycrystalline, monocrystalline and porcelain brackets, respectively, at zero angulation to the arch wire. Ireland *et* al. (1991) found that ceramic brackets did exhibit higher frictional resistance than stainless steel brackets when smaller arch wires (0.014 or 0.016 inch) were used. Perhaps this outcome is a function of the freedom of the arch wire to change angulation within the bracket slot during testing due to increased intra-bracket slot distance. Keith *et al.* (1993) conducted a study restricting change of 0.018 x 0.025 inch bracket slot angulation to 0.018x0.025 inch arch wire where polycrystalline and monocrystalline brackets were both found to produce higher frictional forces than stainless steel brackets. Ceramic brackets were evaluated by stereo light microscope and found to cause abrasive wear of arch wire surfaces, sometimes accumulating wear debris. This wear was found to be greatest with the highest ligation force. Rose and Zernick (1996) conducted a *in vitro* and *in vivo* comparison study of ceramic brackets with rounded slot corners. No results were published for the *in vivo* comparison study, yet Rose and Zernick concluded that the polished ceramic brackets were equally or more effective in moving teeth, compared to unpolished brackets. In addition, it was found that the control *in vivo* brackets showed a build-up of wire debris along the slot corners, whereas little to no build-up was evident on the polished brackets, when viewed using a SEM and a stereomicroscope.

Most investigators seem to agree that frictional resistance is significantly higher in ceramic brackets of any type than stainless steel brackets. This is also true for most wire sizes, alloy types, and slot sizes. (Angolkar et al., 1990; Pratten et al.,1990; Bednar et al.,1991). Ghafari (1992) and Kusy (1991) believed that the increased friction seen with ceramic brackets is due to the roughness of the ceramic bracket-arch wire interface which inhibits bracket movement over arch wire. The increased frictional resistance experienced using ceramic brackets can slow retraction of individual teeth, and facilitate loss of posterior anchorage. In addition, this increased frictional resistance may increase the anchorage value of upright canine teeth leading to anterior tooth extrusion with continuous arch mechanics.

The manufacturing and finishing process may also affect the frictional characteristics of brackets made of similar material. Sintered stainless steel brackets have been recommended as an alternative to individual cast brackets because of their increased slot smoothness. The stainless steel particles are compressed in a contoured, smooth rounded shape, as opposed to the older casting procedure in which the milling or cutting processes left sharp angular brackets that were rough. Matasa (1995) believed that injection moulded brackets produced a more accurate slot which did not require any additional milling. Vaughan *et al.* (1995) compared the *in vitro* frictional resistance of two types of sintered stainless steel brackets with conventional stainless steel brackets from a previous study by Kapila *et al.*(1990) where no bracket angulation change was permitted. There were no significant differences between manufacturers for the sintered stainless steel brackets. However, the friction of sintered stainless steel brackets was approximately 40-45% less than the friction of the conventional stainless steel brackets as measured with the same apparatus. Both of these experiments did not permit a change of bracket angulation to arch wire preventing the increase in normal force upon the arch wire necessary to produce frictional resistance. These results should be interpreted with caution, as the experimental design did not accurately simulate the clinical situation.

2.3 BRACKET DESIGN

Bracket design can also have a significant effect on the frictional resistance experienced with sliding mechanics. Hanson (1980,1986) designed the Speed® bracket with a rounded slot floor so that the arch wire would contact the base of the slot in the direct center of the bracket slot only if the tooth was perfectly aligned in three dimensions. The Speed® spring clip was designed to continuously seat an arch wire into the Speed® bracket slot until this occurred (Hanson, 1980). Bednar *et al.*(1993) conducted an *in vitro* study to evaluate the effect of bracket width and ligation on moment production of conventional and self ligating brackets during axial rotation. The study design simulated the clinical situation as the bracket slot was placed several millimetres labial to the center of rotation of an analog tooth. Both bracket width and ligation technique were found to affect moment production during axial rotation. Ligation technique had the greater influence. Bednar *et al.* (1993) found that the self ligating spring clip of the Speed® bracket delivered the least force over the greatest range of rotation. In addition, unlike rapidly decaying elastomerics, the steel spring clip continuously forced the arch wire into the Speed® bracket slot until the tooth was derotated and had achieved the correct second order expression of the bracket.

Berger (1990,1992) has conducted several investigations comparing and contrasting the Speed® bracket to conventional edgewise brackets in terms of frictional resistance to sliding mechanics. None of Berger's experiments had permitted a change of bracket angulation to the arch wire necessary for legitimate evaluation of clinically relevant frictional resistance. Berger (1990) correctly states that the use of the Speed®

bracket spring clip decreases the normal force placed upon the arch wire, thereby reducing an important component of frictional resistance.

Suyama *et al.* (1995) developed the Mini-Taurus Synergy® bracket to allow friction to be adjusted depending on the degree of ligation required during the course of treatment. Suyama *et al.* believed that the rounded corners of the bracket slot floor and walls would decrease arch wire binding when the bracket changed angulation to the arch wire, thereby decreasing frictional resistance. Suyama's Synergy bracket and Kuroe's Friction Free edgewise bracket both lift the force of ligation off the arch wire in order to reduce friction, yet this also reduces the constraint for the rotational tendency of sliding mechanics, creating a loss of rotational control. Suyama *et al.* (1995) completed an *in vitro* experiment comparing standard edgewise and Synergy® brackets that were placed out of vertical alignment without a change in bracket-slot angulation. Arch wires of two different sizes were drawn through a series of 4 maligned brackets of both types. Frictional resistance was lower using the minimal friction ligation option, yet when ligated conventionally, the Synergy® brackets showed higher frictional resistance than the standard edgewise brackets.

2.4 BRACKET DESIGN AND SLOT SIZE

Angolkar *et al.* (1990) evaluated the effect of slot size for ceramic and stainless steel brackets allowing minimal angulation change of the brackets to four different arch wire alloys. Angolkar *et al.* concluded that slot size made no difference to frictional resistance. Kusy and Whitely (1990) completed an evaluation of the coefficients of friction for arch wires in polycrystalline and stainless steel bracket slots with no angulation change to an arch wire and concluded that independently, slot size made no difference. Tidy (1989) used a suspended weight to create angulation change of 0.022 and 0.018 inch bracket slots to two arch wire sizes and determined that arch wire and slot dimension had relatively little influence on friction. Study designs allowing a change of angulation of bracket slot to arch wire are crucial to evaluate the effect of slot size on frictional resistance.

Schudy (1990) completed several experiments to investigate the effect of bracket design, including slot size, on interbracket distance and wire flexibility. Schudy (1990) used computer modelling to conclude that wire flexibility is greatest and force deflection is least with maximum interbracket distance, maximum intrabracket space (increased slot size) and minimum wire size. Schudy (1990) also evaluated alternative bracket designs which utilize smaller wires and greater intrabracket space. The Bi-Metric appliance uses two different sized bracket slots in one appliance, having the smaller brackets (0.016 x0.025 inches) on the six anterior teeth and a larger bracket (0.022 x 0.025 inches) on posterior teeth. Schudy thought that this design would facilitate early torque control on the anterior teeth by filling the bracket slot earlier in treatment, yet allow greater intrabracket space in the posterior teeth to reduce friction easing sliding mechanics. The increased intrabracket space would facilitate loss of second order control in posterior teeth during sliding mechanics allowing greater change of bracket angulation to archwire (tipping). Greater tipping of posterior teeth would increase the normal force upon the arch wire from the bracket, leading to greater binding frictional resistance. The relationship between increased flexibility of arch wires due to increased interbracket distance and the amount of normal force upon the bracket due to angulation change was never examined.

2.5 ARCH WIRE MATERIAL

Arch wire material greatly affects the frictional resistance experienced with sliding mechanics. The contribution of arch wire material to frictional resistance can be subdivided into the effect of arch wire stiffness and arch wire surface characteristics. These two distinct properties of arch wires present a challenge to investigators of frictional resistance as their friction consequences are difficult to destinguish. The order of increasing coefficients of static friction of arch wires of different alloys against stainless steel or polycrystalline alumina bracket slots has been shown to be: stainless steel (lowest), cobalt-chromium, nickel-titanium, and β-titanium (highest) (Angolkar *et al.*, 1990; Drescher *et al.*,1989; Garner *et al.*,1986; Ho and West, 1995; Kusy *et al.*,1988,1989,1990; Prosoki *et al.*, 1991; Kapila *et al.*, 1990; Saunders *et al.*, 1994; and Vaughan *et al.*, 1995).

Many investigators agree that the increased surface roughness of certain arch wire material creates higher frictional resistance. (Angolkar *et al.*, 1990; Garner *et al.*, 1986; Kusy *et al.*, 1989,1990; Prosoki *et al.*, 1991; Kapila *et al.*, 1990; Saunders *et al.*, 1994; and Vaughan *et al.* ,1995). Ho and West (1995) found no definite relationship between arch wire surface roughness and friction, but identified arch wire stiffness as a controlling factor of frictional resistance. Prososki *et al.* (1991) measured arch wire surface roughness with a profilometer and determined there was no correlation between surface roughness and frictional resistance. Kusy *et al.* (1988) used specular reflectance with laser spectroscopy to gauge the surface roughness of orthodontic arch wires. Kusy *et al.* (1988) showed that stainless steel wires had the smoothest surface followed by cobalt-chromium, β-titanium and nickel-titanium wires in order of increasing surface roughness. In addition, Kusy *et al* (1988) found that surface roughness was not always related to frictional forces, especially for the β-titanium wires. Nickel-titanium was found to be a rougher material than β-titanium, yet the softer β-titanium wires were found to have a greater mean frictional force than the nickel-titanium wires. Kusy *et al.*(1990) then repeated several experiments in which β-titanium and nickel-titanium arch wires were drawn through contact plates and brackets of stainless steel and polycrystalline alumina at different velocities. The velocity of arch wire movement through the contact flats was found to be significant. At the slowest testing speeds (equivalent to 21.6mm of tooth movement /month), corrosive wear of the arch wires was found to occur as oxide layers were removed. Arch wire movement equivalent to 21.6mm/month as well as normal forces as high as 5 Kg made the results of this study difficult to interpret. The classical laws of dry friction dictate that friction is independant of sliding velocity. Subsequent x-ray elemental analysis of the contact plates and brackets revealed that adhesion of β-titanium arch wire material to the contact flats and brackets was the cause for the high coefficients of friction. A cold welding effect was postulated to occur which caused particles of the β-titanium to adhere to the stainless steel brackets, while mechanical abrasion of the softer β-titanium was responsible for the adhesion to alumina brackets. The cold welding effect involved the formation, destruction, and reformation of metal to metal bonds as the surface topography of the β-titanium arch wire continually experienced modification during its movement within the stainless steel brackets. Recent engineering investigations have determined that friction is a material property phenonmenon which is dependant upon the electrostatic interaction between the materials in contact. Surface roughness of arch wires can be considered a mechanical impedement to movement, and not classical friction. In addition, engineering theory speculates that that when very smooth surfaces come into contact, friction may increase as a result of surface tension of fluid films or molecular attraction between surfaces in intimate contact. (American Society for Metals Handbook, 1986)

The modulus of elasticity of orthodontic wires is another characteristic of arch wire material that is thought to affect frictional resistance. Investigators have agreed that stiffer arch wires of similar dimension place larger normal forces upon brackets as they change relative angulation. These larger normal forces can generate frictional resistance which tends to slow tooth movement. Thus, if a more flexible wire is used, less generation of normal forces upon the bracket will result in decreased frictional resistance. Drescher *et al.* (1989) demonstrated that elastic properties (flexibility) per se, can not explain the frictional behaviour of β-titanium, since this wire is stiffer than nickel-titanium, but exhibited significantly greater frictional forces. Drescher *et al.* (1989) stated that perhaps, the elastic properties of the arch wire are secondary in affecting friction. Drescher *et al.* explained that the forces resulting at the contact points between the bracket and arch wire were almost independent of the elastic properties of the wire. Wires with equal size and surface texture but with diverse elastic properties would generate nearly the same frictional forces for a given bracket width. Drescher stated that the only difference between these two wires would be the possible relative angulation change between the bracket slot and arch wire. Drescher believed that this is a result of the fact that a stiffer arch wire will produce a counter-moment earlier during the retraction process than a more flexible wire.

2.6 ARCH WIRE SURFACE TREATMENT

Investigators have determined that arch wires such as nickel-titanium and β-titanium have higher coefficients of friction due to their relative softness compared to harder stainless steel brackets. (Kusy and Whitely, 1990). Greenberg and Kusy (1979) completed a study to evaluate if arch wire coatings on orthodontic appliances could reduce friction or improve wear properties to permit more efficient and reproducible transmission of force to teeth. Metal-polymer coatings were added to arch wires which were then drawn between parallel stainless steel and polycrystalline alumina flats. The metal-polymer coatings reduced the static coefficient of friction by 55-83% for stainless steel and cobalt-chromium arch wires of different shape and dimension.

Ion implantation is a process by which various elements or compounds are ionized and then accelerated toward a target. Gas ions (usually nitrogen and oxygen) are simultaneously extracted from a plasma spray and directed toward a target (arch wire or bracket). The ions penetrate the surface and build a structure that consists of the original arch wire or bracket material and a layer of compounds (TiN and TiO) on the surface and immediate substrate. This surface is extremely hard. Ion implantation improves the surface characteristics with out degradation of the substrates mechanical properties. The thickness of the implanted surface can be precisely controlled and its properties engineered to affect hardness, friction, wear resistance, ductility and surface colour. (Burstone and Farzin, 1995)

Burstone and Farzin (1995) completed a study comparing untreated β-titanium with ion implanted β-titanium measuring maximum bending strength, tensile strength, modulus of elasticity, and coefficient of friction. Frictional testing by drawing the arch wires through parallel contact flats of stainless steel revealed that frictional forces of ion implanted β-titanium were approximately equal to that of stainless steel, with a reduction in the variance of the coefficient of friction, as the "stick-slip" phenomenon of β-titanium was eliminated. Surface treatment with ion implantation was found to maintain all the desirable properties of β-titanium and actually improve its ductility, and its resistance to fracture, wear and fatigue. Kusy *et al.* (1992) completed an investigation of the effect of ion implantation of β-titanium arch wires with polycrystalline alumina plates with nitrogen ion implantation. This combination of ion implantation reduced the static coefficient of friction to values similar to control couples of stainless steel, cobalt-chromium and nickel-titanium against stainless steel and polycrystalline alumina plates. Kusy *et al.* (1992) speculated that one day regions of arch wires could be ion implanted to permit sliding mechanics while other nontreated regions retain the properties which prevent anchorage loss.

Mendes *et al.*(1996) evaluated the effect of ion implantation of brackets and arch wires using SEM and *in vitro* Instron® testing. Ion implanted and non-implanted brackets, each fixed at zero angulation, were guided along ion implanted and non-implanted arch wires. Ion implanted brackets were found to produce comparable results to ion implanted arch wires with non implanted brackets. No investigators have conducted friction testing experiments with ion implantation that permitted bracket angulation change to arch wire. This change of angulation which causes a dramatic increase in normal forces at the arch wire-bracket interface is essential to effectively evaluate the clinical significance of ion implantation on frictional resistance.

2.7 ARCH WIRE SHAPE AND DIMENSION

Arch wire shape and dimension have also been thought to significantly affect the frictional resistance experienced with sliding mechanics. Investigators who designed studies that permitted bracket angulation change to arch wire in response to tractional forces agree that frictional resistance to movement increases with increasing wire size. (Andreasen and Quevedo, 1970; Frank and Nikolai, 1980; Huffman and Way, 1983; Tanne *et al.*, 1991; Bednar *et al.*, 1991; Drescher *et al.*,1989; Ogata *et al.*,1996) Other investigators who designed experiments which did not provide bracket angulation change to arch wire in response to tractional forces concluded opposing results, with smaller arch wires producing the largest frictional resistance. (Baker *et al.*,1987; Schudy, 1990; Ireland *et al.* 1991; Rilely *et al.*, 1979; Shivapuja and Berger, 1994). Smaller and more flexible arch wires bend and change angulation with greater intrabracket space. In addition, larger intra-bracket dimensions allow for a greater degree of tipping, relative to these small arch wires, before larger normal forces are producded. Baker *et al.*(1987) believed that decreased play between bracket slot and arch wire does not allow as large a degree of wire angulation within the slot, creating less potential for binding. In addition, stiffer arch wires are less likely to distort under occlusal load, which could also increase the potential for binding resistance. Peterson *et al.* (1982) showed no change in frictional resistance with increasing wire size at zero angulation if the guiding arch wire is kept in tension. Tension on the arch wire prevents arch wire bending within the bracket slots when brackets are fixed in angulation. Bednar *et al.* (1991) tested the significance of arch wire size upon steel, ceramic and self ligating brackets. All three bracket types exhibited increased friction with increasing wire dimension. Drescher *et al.* (1989) believed that the influence of wire dimension on friction is determined by the vertical dimension of the arch wire. They found that 0.016 inch round and 0.016 x 0.022 inch rectangular stainless steel wire showed virtually the same amount of friction, if the bracket was allowed to change relative angulation.

Tidy (1989), concluded that arch wire dimension and slot size had little affect on friction. Tidy (1989) stated that "the clearance between arch wire and slot is not itself important in controlling friction. However, kinks or deposits along a closely fitting arch wire are more likely to lead to binding in the slot and clearance is therefore of some secondary importance." In addition, Tidy (1989) points out that the component of friction caused by active "torque" may also be greater for a closely fitting wire because of its greater torsional stiffness and the reduced play between wire and slot. Many practitioners prefer the use of round wire for sliding mechanics because of anecdotal references that this technique eliminates friction caused by active "torque". Round wires generally produce less friction than rectangular wire when engaged in brackets out of alignment because of their greater flexibility (Tidy,1989). Tidy (1989) believed that as brackets align, and any torque becomes passive, the differences between round and rectangular wire of similar vertical dimension will become small. Friction, as a result of two point contact, is largely independent of wire stiffness or cross section and is likely to be little affected by the choice of round or rectangular wire. Frank and Nikolai (1980) state that because arch wire stiffness is proportional to the third or fourth power of the occluso-gingival dimension of the rectangular or round wire, respectively, smaller rectangular wire is stiffer in bending than the larger round wire of the identical material. Frank and Nikolai (1980) concluded that when binding, round wires make only point contact with a bracket slot edge, yet rectangular wires make line contact. The intensity of normal force (pressure) of wire against bracket is thought to be higher with round than rectangular wire because of the difference in contact area size. Frank and Nikolai (1980) believed that the higher pressure of the small contact area found with round wire results in a higher potential for arch wire indentation, which can adversely affect sliding mechanics. The classical laws of friction stipulate that it is independent of the area of contact. Therefore, Frank and Nikolai's assumption is not strictly speaking, related to friction, but mechanical impedements to sliding mechanics, due to the destruction of the surface integrity of the arch wires.

Huffman and Way (1983) completed an *in vivo* experiment to examine the effects of arch wire dimension on canine retraction. Their clinical study included 16 subjects with first premolar extraction using identical Pletcher springs bilaterally for retraction force . Each subject had 0.020 inch stainless steel arch wire on one side of their upper arch with 0.016 inch stainless steel arch wire on the other. Less tipping was found to occur on the 0.020 inch arch wire with similar amounts of canine retraction. No greater force was required to retract either side even though their arch wire sizes differed. Huffman and Way concluded that muscular pressure and the force of mastication must have displaced the teeth which released the frictional binding preventing canine retraction.

2.8 LIGATION

The significance of ligation to frictional resistance is known to depend on, the amount of normal force applied to the arch wire, the method of ligation, and the ligature material. Most investigators agree that increasing the normal force of ligation upon the arch wire increases frictional resistance. (Echols, 1975; Riley *et al.*, 1979; Frank and Nikolai, 1980; Keith *et al.*, 1993; Bednar *et al.* 1991,1993, Shivapuja and Berger, 1994). Paulson *et al.* (1970) was one of the earlier investigators to identify increases in friction commensurate with increasing forces of ligation. Keith *et al.*(1993) evaluated the significance of ligation force on frictional resistance of monocrystalline, polycrystalline and stainless steel brackets that were fixed in zero angulation to the arch wire. An increase in ligation force was found to increase frictional resistance in all cases. Various investigators have attempted to alter bracket designs to limit the normal force placed upon an arch wire in order to decrease frictional resistance. Ogata *et al.* (1996), Suyama *et al.*(1995), Berger (1990) and Kuroe *et al.* (1994) all agreed that bracket designs that minimize the normal force transmitted to the arch wire by ligation, minimized frictional resistance. However, Bednar and Gruendeman (1993) concluded that although both bracket width and ligation technique significantly affected the moment production during axial rotation, the ligation technique had the greater influence. Decreased normal forces placed upon arch wires from loose ligation will tend to decrease the moment necessary to offset the rotational tendency generated with sliding mechanics. This can result in tooth rotation when using sliding mechanics for tooth translation.

Self ligating brackets are thought to have inherently less frictional resistance due to the reduction of normal force from ligation. Bednar *et al.* (1993) concluded that self ligating brackets experienced significantly lower friction with larger wire sizes than conventional wide and narrow twin brackets. Shivapuja and Berger (1994) concluded that self ligating brackets transmit the lowest normal force to the arch wire of all ligation techniques, regardless of bracket width.

Numerous investigators have examined the influence of ligation material on frictional resistance. Interpretations of the results of these studies are difficult as reproduction of ligation technique is highly variable, and operator specific. This has lead many investigators to rely on the use of individual elastomeric ligation, which is thought to be most reproducible. Unfortunately elastomeric ligation is fraught with initial rapid force decay, moisture and temperature sensitivity, and wide physical material differences across manufactures. Specifics such as bracket perimeter distance, prestretching with or without relaxation, method of ligature placement, and temperature and humidity of testing environment may significantly affect the characteristics of ligation force utilized within any study. Echols (1975) used previously worn elastic ligatures in an *in vitro* study which concluded that the use of elastic ligatures over conventional stainless steel increased the binding force of the arch wire into the slot, thereby increasing friction. Echols comments that "elastic ligatures should be avoided where slippage of the arch wire through the bracket is desired." Kapila *et al.*(1989) identified the stretch of elastomeric ligatures around wider brackets as a cause for increased normal force of ligation. Rose and Zernick (1996) used, a ligature gun (Straight Shooter®, TP Orthodontics) to apply elastic ligatures in their study of fricitonal resistance. This is a method used to standardize the amount of force and prestretching of elastomeric ligatures, to aid in reducing experimental variables. Riley *et al.* (1979) compared stainless steel ligatures with elastic modules and found that stainless steel ligatures generated the highest frictional resistance. Adams *et al.* (1987) used engineering beam theory and *in vitro* testing to evaluate the affect of type of ligature tie, bracket width and inter-bracket distance on arch wire stiffness. Metal ligature ties were found to secure the arch wires tighter than elastomeric modules to increase the reactive force of the arch wire. For wide brackets, no significant difference was found between metal ties and elastomeric ligatures due to the increased stretch of the elastomeric ligatures around the wider brackets. Rock and Wilson (1989) investigated the influence of ligation on the stored energy within an arch wire. Rock and Wilson (1989) determined that elastic ligatures assisted the arch wire in greater deflection than stainless steel ligatures due to increased friction of the elastic against the arch wire at the adjacent bracket slots. Bednar *et al.*(1991) examined the effect of brackets lightly tied with stainless steel ligatures, and conventionally tied with elastomeric ties, and found the lightly tied steel to produce the lowest frictional resistance. Schumacher *et al.* (1990) completed a study with brackets fixed at zero angulation to arch wire where ligature force was found to have a greater influence on frictional resistance than arch wire size. Shivapuja and Berger (1994) completed a study where higher frictional forces were measured using elastic ligation instead of steel ligation on identical brackets. Sims *et al.*

(1993) concluded that ligation technique as well as material, can significantly affect frictional resistance. Using elastic ligatures in a figure eight pattern around identical brackets raised their frictional resistance from 70 to 220%.

DeFranco *et al.*(1995) completed an investigation of Teflon coated metal ligatures with stainless steel, polycrystalline and monocrystalline brackets sliding upon stainless steel and nickel-titanium arch wires. All static frictional forces were less with Teflon coated stainless steel ligatures than with elastomeric rings. DeFranco *et al.* did permit the brackets to change angulation relative to the arch wire, yet stated that the effect of friction from the ligature is independent of bracket-arch wire angulation. The consequence of axial tooth rotation on ligation and frictional resistance was not investigated. One major flaw in the study was the absence of uncoated stainless steel ligature wire as a control to determine the effect of Teflon coating. Comparison of Teflon coated wire and elastomeric rings does not elucidate the effect of Teflon coating; uncoated stainless steel wire may have produced similar, lower frictional forces.

2.9 TESTING ENVIRONMENT

Perhaps the most significant factor affecting frictional resistance is the testing environment. Most *in vitro* tests tend to ignore the effects of vibration and saliva, two of the factors thought to substantially affect *in vivo* frictional resistance.

2.9.1 Testing Environment (Saliva)

Baker *et al.* (1987) evaluated the effect of lubrication on friction with a saliva substitute (Oralube®) at 37 degrees. A 15-19% reduction in friction was noted, with fixed bracket angulation to arch wire. Additional studies also showed the use of glycerin was an ineffective lubricant. Ireland *et al.*(1991) tested frictional resistance in the dry and wet states by pouring water over brackets that were fixed in angulation relative to arch wires. Frictional resistance in the wet state was found to be identical with the dry state. Kusy *et al.* (1991) conducted a study with healthy operators using their own saliva for lubrication supplied from a peristaltic pump at 3cc/minute at 34 degrees. Brackets were not permitted to change angulation to the arch wire during the frictional resistance testing. Statistically different frictional resistance for dry and wet states were found, although the magnitude and direction of change depended upon the specific bracket-slot and arch wire couple. Couples comprised of β-titanium wires exhibited lubricious behaviour in the wet versus dry state, while couples comprised of stainless steel wires suggested some adhesive behavior in the wet versus dry state. Kusy *et al.*(1991) stipulated that saliva could be acting to chemically break down chromium oxide surface characteristics which render certain arch wire surfaces chemically inactive, or alternatively, saliva could be acting as an adhesive because of surface tension effects. Kusy suggested that clinically, high pressure contacts between arch wires and bracket slots could create more boundry type lubrication. As more boundry lubrication occurs, more solid-solid contacts would apear in conjunction with solid-liquid contacts, thereby creating more stick-slip phenomenon with the use of β-titanium arch wires *in vivo*. In addition, Kusy (1991) states that "experiments conducted in artificial saliva are invalid, because artificial saliva is not a satisfactory substitute for fresh human saliva. Most synthetic saliva is designed to reduce the effects of xerostomia, not reduce friction in sliding mechanics." Shivapuja and Berger (1994) conducted experiments using artificial saliva substitute, finding that it produced higher frictional resistance due to rapid desication leaving cellulose adhereing to arch wires being tested. Pratten *et al.* (1990) evaluted the effect of lubrication from artificial saliva, finding that it increased static friction, when brackets were not permitted to change angulation relative to arch wires. Pratten *et al.* (1990) explained that at high loads (undefined), saliva may be forced out from the bracket-arch wire contacts resulting in an increase in frictional resistance. Pratten *et al.* (1990) concludes that saliva may only act as a lubricant at low loads determined by ligation force. Saunders and Kusy (1994) evaluated the difference betweem the dry and wet states using real saliva. Friction was reduced in the presence of saliva for ceramic brackets against titanium alloys, but not for stainless steel or chromium-colbalt alloy when bracket angulation change was not permitted.

Many investigators agree that restricting bracket angulation change to arch wire during frictional resistance testing with saliva removes any similarity to the clinical situation (Andreasen and Quevedo,1970; Ho and West, 1995; Stannard *et al.*,1986 and Tselepis *et al.*, 1994). Andreasen and Quevedo (1970) completed an experiment comparing wet and dry states allowing a change in bracket-arch wire angulation. The differences in frictional resistance with saliva as a lubricant with those made with dry arch wires was found to be insignificant. Ho and West (1995) evaluated the lubrication affect of artificial saliva allowing bracket angulation change. Friction decreased with artificial saliva lubrication versus dry conditions generally, yet several β-titanium arch wires showed increased friction with lubricated conditions. Stannard *et al.* (1986) conducted an investigation comparing frictional resistance in artificial saliva and the dry state. Artificial saliva increased the static coefficient of friction for stainless steel, β-titanium, and nickel-titanium compared to dry conditions, but did not change the frictional coefficient for cobalt-chromium or teflon. Stannard *et al.* (1986) stated that water and other polar liquids are known to increase adhesion and attraction among polar materials and thus increase friction. This behavior has been observed for several different dental materials in the presence of saliva, and has been explained by the adhesion theory of friction. This theory assumes that strong adhesive bonds developed in areas of contact and had to be sheared when the surfaces moved relative to each other. (American Society for Metals Handbook, 1986)

2.9.2 Testing Environment (Vibration)

Most *in vitro* frictional resistance studies demonstrate that the movement of teeth with sliding mechanics is very inefficient due the loss of effective tractional force through frictional resistance. It is quite difficult to move a tooth by sliding a bracket over an arch wire because the forces needed to overcome friction are often beyond the limits of lesser forces created by rapidly decaying orthodontic elastics and springs. These conclusions result from the static nature of the *in vitro* friction testing conditions. Fortunately, intraorally, there is an initial degree of movement between tooth roots and their alveolar housing via the periodontal ligament. This movement is enhanced with mastication, random contact with the opposing dentition, and deglutition which theoretically would help to decrease static friction at the bracket-arch wire interface. Subesquent osteoclastic bone resorption resulting from orthodontic forces, occurs in the alveolar socket walls allowing teeth to become more mobile, which could decrease friction to an even greater extent. *In vivo* testing which takes vibration into account along with saliva, plaque, pellicle, temperature, and humidity gives the most accurate representation of the clincial situtation.

Paulson *et al.*(1970) conducted an *in vivo* study on 6 subjects evaluating canine retraction after maxillary frist premolar extraction. Serial cephalometic superimposition permitted evaluation of tooth movement ranging from 0.9mm/month to 2.4mm/month. Paulson *et al.*(1970) stated that friction reduced the force applied to the periodontal ligament and that friction increased with tightness of ligation. Huffman and Way (1983) conducted an *in vivo* study of canine retraction with different sized arch wires bilaterally in 16 subjects. They hypothesized that muscular pressure and forces from mastication displaced teeth and released binding between brackets and arch wire to allow similar canine retraction on large and small stainless steel arch wires. Brinkman and Miethke (1991) devised a method to compare the frictional forces which occur *in vivo* and *in vitro*. A testing device was used which permitted the identical evaluation of frictional resistance in the *in vitro* and *in vivo* environments carried out on the maxillary central incisors of volunteers. A cast upper partial denture framework was designed to hold maxillary central incisors with a bonded twin bracket immobile so frictional resistance testing could be completed. Subsequently, these same tests were repeated without the partial denture framework in place. This technique permitted an evaluation of the change in frictional resistance from physiological tooth mobility and occlusal load. Friction measured *in vitro* with immobile brackets and *in vivo* without occlusal load did not differ significantly. Additional tooth movement with phyisological tooth mobility and occlusal load resulted in a significant reduction in frictional resistance. Brinkman and Miethke (1991) stated that this effect may be magnified during conventional orthodontic treatment due to iatrogenically induced tooth mobility. Therefore, frictional forces occuring with orthodontic treatment are even smaller in comparison to *in vitro* experiments with immobile brackets. Kapila *et al.* (1990) commented that *in vivo* factors such as occlusion, mastication, wire resilience, and tooth movement may alter the second order bracket-arch wire relationship as the bracket moves along the arch wire. This freedom in second order orientation of the brackets to the wire alters the frictional and normal force components with time, so that forces required to cause *in vivo* bracket movement will vary at different bracket-wire angulations. Ziegler and Ingervall (1989) completed a clinical study comparing the frictionless Gjessing retraction spring with sliding mechanics for efficiency in bilateral canine retraction in 21 subjects. Canines were retracted faster with less tipping using the Gjessing spring than with sliding mechanics, with a difference of 0.5mm/month. However, the canines were more rotated with Gjessing spring retraction than with sliding mechanics on 0.018 inch stainless steel arch wire. There was a significant error in the method of sliding mechanics, as the canines were ligated as tight as possible with stainless steel ligation. Tight stainless steel ligation would increase the normal force upon the arch wire , increasing the frictional resistance to movement. Ideally loose stainless steel ligation should be used for translation during sliding mechanics.

2.10 SUMMARY

The previous review of the literature identified the phenomenon of friction to be multifactorial in nature. The variables that have been found to affect the observed levels of friction between bracket and arch wire may be considered as mechanical or biological in origin; or due to the experimental design used to measure friction. Mechanical variables considered the type and nature of bracket, type and nature of arch wire and the type and nature of ligation. Saliva and plaque have also been implicated as some of the biologic factors affecting the magnitude of frictional resistance. The review of the literature indicates that the most significant consideration in the phenomenon of friction is the method by which the frictional resistance is evaluated. Bracket angulation change and vibration have been identified as the two factors responsible for the major differences in conclusions between *in vitro and in vivo* investigations. Thus, any relevant clinical simulation of the frictional resistance experienced with orthodontic sliding mechanics must take these two primary factors into account. The objective of this study was to design and utilize an apparatus that would permit a clinically relevant investigation of frictional resistance considering and including the *in vivo* contribution of bracket angulation change and vibration.

CHAPTER 3

MATERIALS AND METHODS

MATERIALS AND METHODS

3.1 INTRODUCTION

Sliding mechanics are commonly used for translation of teeth. Mesiodistal tooth movement is accomplished by guiding a tooth along a continuous archwire using an orthodontic bracket. A disadvantage of this technique is that friction is generated between the bracket and the arch wire, which tends to resist movement of the tooth. Friction is generated when the tooth, attached to the bracket, changes angulation in relation to the arch wire in response to a retraction force applied at a distance from the tooth's center of resistance. When a bracket slides over an arch wire the angulation of the bracket to the arch wire depends on a combination of the location of force application and the biological retarding force. The center of resistance of most teeth is usually located just beneith the alveolar crest, whereas the most common point of force application is buccal or labial to the center of resistance and at considerable distance coronal to the center of resistance. Consequently, force application labial and coronal to the tooth's center of resistance generates rotation along the vertical axis, and second order tipping along the buccal-lingual axis, which generates friction between the ligation and the arch wire as well as at the bracket-arch wire interface.

Many *in vitro* studies have been undertaken to observe the effect of possible variables on friction. Most of these studies are inadequate, however, as they ignore vibration, and fail to allow for a change of angulation of the bracket, which may play a significant role in determining the coefficient of static friction *in vivo*. In order to compare frictional forces consistently in the *in vitro* and *in vivo* environments, it was necessary to design a mechanism that would control angulation of the bracket slot to an arch wire, and thus control the normal force upon the arch wire, thus creating a model of the clinical situation. The friction testing apparatus, designed for this project, regulated the normal force between a bracket and arch wire to generate frictional resistance.

When in clipped on top of the utility arch wire, the moment generating spring (see Figure 3.2.1) creates an angulation change of the sliding tube upon the utility arch wire. This results in increased normal forces causing binding friction to occur. The activated moment generating spring and the sliding tube, as a single unit, were then pushed mesially , causing elongation of the retraction spring. The elongated retraction spring acts to pull the sliding tube and moment generating spring distally until the retraction force is equivalent to the force of friction, which then inhibits further movement of the sliding tube. In this position of equilibrium, the static coefficient of friction can be calculated.

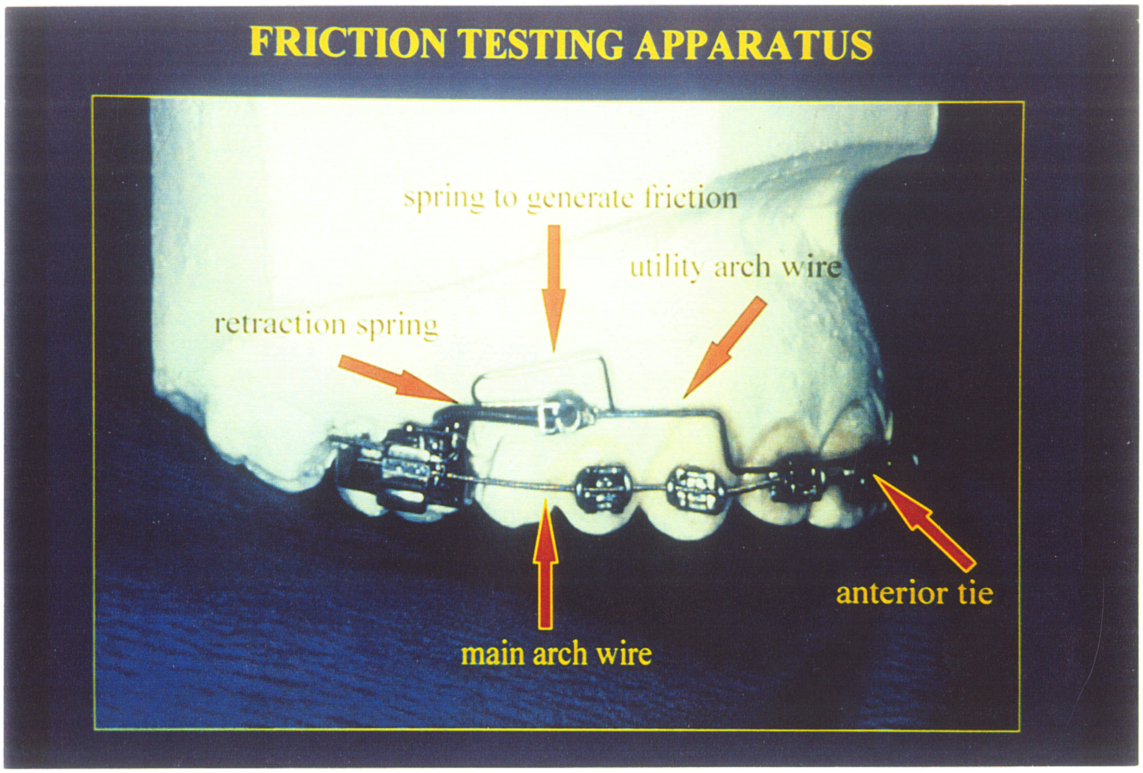
The experimental technique was designed to standardize variables such as wire and bracket material, arch wire bending, ligation and preadjusted bracket prescription. Variables including bracket width, saliva, and vibration were investigated to determine their roles in the reduction of the "clinical" coefficient of friction *in vivo*, as compared to *in vitro* tests. All sliding surfaces were made of similar stainless steel to eliminate materials that could additionally influence friction. All friction testing assemblies were fabricated as accurately as possible on a stone model poured from a recent alginate impression and made passive before insertion. The use of different length tubes bilaterally facilitated the comparison of the effect of bracket width on friction. The normal force of ligation, and the effect of applied moment from brackets was eliminated by the use of tubes instead of preadjusted orthodontic brackets. In addition, the friction testing assemblies were designed as independent units which could undergo identical testing procedures *in vitro* and *in vivo.*

A series of preliminary studies were completed to design and test retraction mechanisms that would provide reproducible forces useful to study the phenomenon of friction. Initial *in vitro* experiments found that closed coil Nitinol springs were suitable for the purpose of retraction of a tube over an orthodontic arch wire. (see Figure 3.2.1) By varying the length of these springs, their force deflection characteristics were made compatible with the dimensions and forces required for the friction testing assembly. Multiple *in vitro* force-deflection calibration experiments produced similar results for similar lengths of closed coil Nitinol springs. After the retraction mechanism was developed, a method of creating a moment on the sliding tube was devised. This moment was useful in creating the desired frictional forces between tube and arch wire. A moment generating spring was used on the basis that it produced a consistent moment, the magnitude of which was easily adjusted.

3.2 APPLIANCE DESIGN

Instead of using brackets in the friction testing apparatus, stainless steel tubes were cut from longer length tubes of dimension 0.018 inch height and 0.025 inch width (American Orthodontics®). The two tubes cut for the experimental apparatus were in two lengths representative of narrow and wide orthodontic brackets (4.0mm, and 8.0mm). A diamond disk running at slow speeds was used to avoid heating the surfaces, and thus altering the properties of the stainless steel. The tubes then had their ends polished with a standard rubber wheel to ensure smoothness and to eliminate metal fragments which would otherwise gouge or scrape the arch wire. Each tube was then tested for ease of sliding on a clean piece of 0.017 x 0.025 inch stainless steel arch wire before fabricating the friction testing apparatus. The stainless steel tubes had their outer surfaces roughened to facilitate the attachment of the moment generating springs. The prepared tubes were placed on a piece of 0.017 x 0.025 inch stainless steel arch wire which was waxed into a jig which facilitated the attachment of the moment generating springs to the tubes. Light cured orthodontic composite resin (filled and unfilled, Spectrum-American Orthodontics®) was used to bond the moment generating springs to the stainless steel tubes. This resin was also used to bond a 10mm Nitinol closed coil spring from the side of the stainless steel tubes to the crimped surgical hook on the friction testing apparatus. (See Figure 3.2.1) This Nitinol spring had a lumen diameter of 0.030 inches, and was constructed of 0.0075 inch wire. This spring was used to generate the retraction force on the tubes to initiate distal movement of the tube.

Figure 3.2.1 Diagram of Apparatus



A piece of orthodontic band material (3M-Unitek®) approximately 3mm by 2mm was cut and spot welded to a straight length of stainless steel arch wire of 0.018 inch dimension. (American Orthodontics®) The 0.018 inch stainless steel arch wire was bent to form the shape of a moment generating spring. The band material was used to provide a surface for the attachment of the moment generating spring to the stainless steel tube.

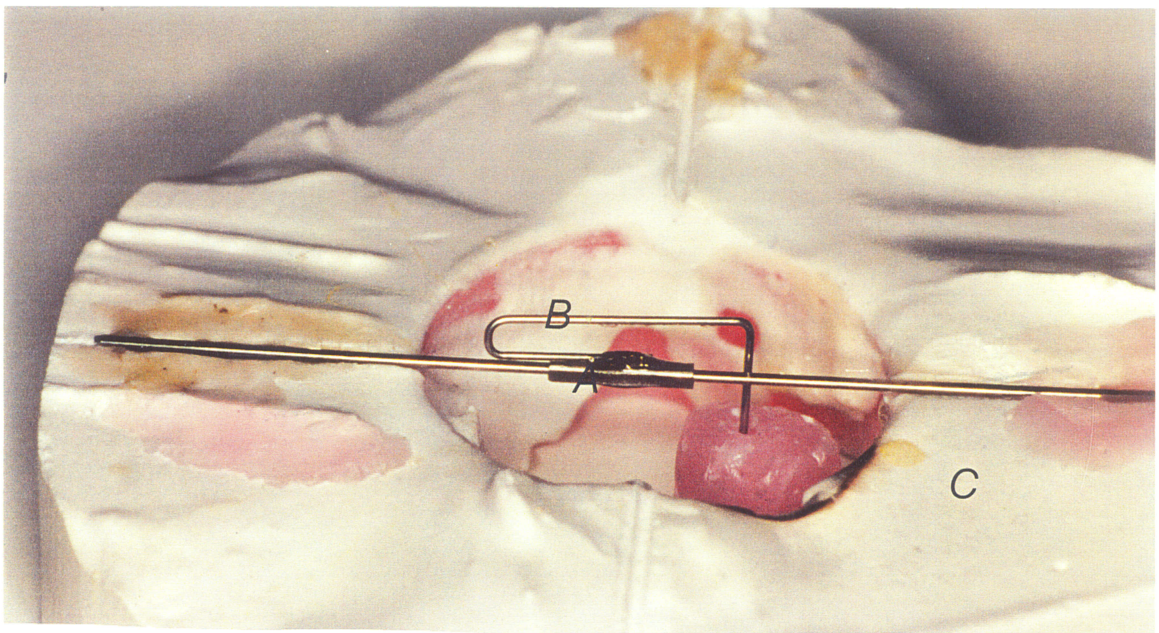


Figure 3.2.2 Bonding of Sliding Tube to Archwire
A: Sliding tube B: Moment generating spring C: Jig for bonding

The moment generating spring was bent so that the hook used to apply force to the utility arch wire rested passively 1-2 mm below the arch wire. Once each tube was bonded to each moment generating spring, the moment generating springs were calibrated to measure how much force was applied when the springs were in the activated position of resting on top of the utility arch wire.

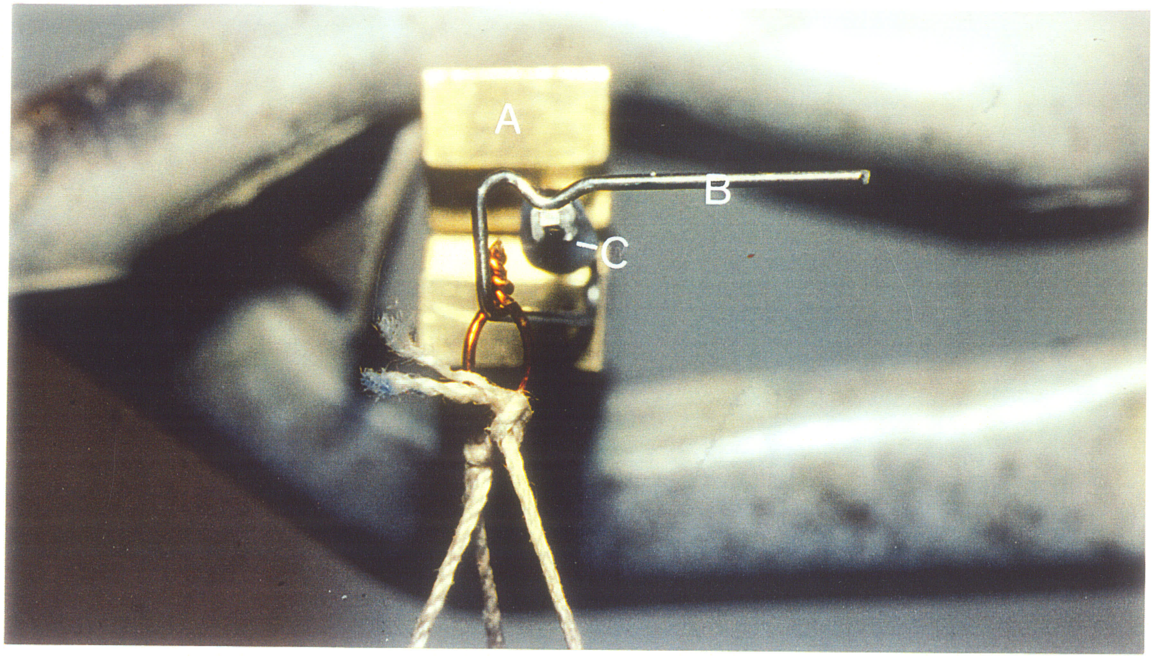


Figure 3.2.3 Moment Generating Spring Calibration **(A: Brass jig to support arch wire; B: moment generating spring being pulled down to "end on" view of arch wire; C: Sliding tube on arch wire)**

Each moment generating spring, attached to its sliding tube, was inserted upside down on a horizontally mounted piece of 0.017 x 0.025 inch arch wire, which was fixed by a brass jig to a retort stand. A small plastic specimen dish suspended from the moment generating spring was used to support weights to evaluate the force produced by each moment generating spring when in the activated position. The moment generating spring and tube resting on the horizontal arch wire were viewed end-on with a vernier microscope to determine of what weight brought the moment generating spring to an activated position relative to the horizontal arch wire. The specimen dish containing this mass was then weighed on a Mettler PJ6000 balance to measure the normal force of the moment generating spring in the activated position.

The utility arch wire used to support the sliding tube was fabricated of 0.017 x 0.025 inch stainless steel arch wire. The utility arch was fabricated on a plaster model of each subject's maxillary arch to ensure an accurate and passive fit of the arch in all dimensions. The utility arch sections used for testing of the wide and narrow sliding tubes were stepped gingivally from the canine to the molar attachment to avoid interference with sliding. However, the utility arch was placed close to the attached gingiva of the alveolar process of the maxilla to avoid discomfort when chewing. Assignment to the right and left sides for wide and narrow tubes was done randomly, with a coin toss. The tubes and moment generating springs were slipped on to the utility arch wire as it was being bent to fit the plaster model. Once the utility arch was formed to the plaster model, a crimpable surgical hook (American Orthodontics®) was placed on a vertical section of the utility arch wire just mesial to the molar band attachment. The Nitinol closed coil springs used for tube retraction were then bonded from the sliding tubes to the surgical hooks.

When completed, the friction testing assembly was an independent unit which inserted into the 0.018 x 0.025 inch auxiliary tube on a first molar band. The friction testing apparatus contained the sliding stainless steel tubes with moment generating springs which slid along straight sections of the utility arch as well as the Nitinol retraction springs which were used to provide the retraction force for the sliding tubes.

3.3 PATIENT SELECTION

Ten subjects were selected from patients undergoing orthodontic treatment in the Graduate Orthodontic Clinic of the University of Manitoba based on the following criteria:

1) Patients of any age who were undergoing orthodontic treatment with fixed appliances utilizing continuous straight wire therapy in the upper arch. First molar bands must have 0.018x0.025 inch auxiliary tubes.

2) Adequate vestibular height, preventing the buccal frena from interfering with operation of the friction testing assembly.

3) Minimal gingival inflammation evaluated by lack of bleeding on probing.

4) Willingness of the patient to tolerate the making of impressions, and the bulkiness of the appliance, and capable of maintaining the discipline required during the testing procedures.

5) Absence of signs and/or symptoms of temporomandibular disorder.

All subjects read an information sheet detailing the purpose, procedures, and risks and benefits of participation in the study. All subjects signed a consent form, approved by the Committee on Research Involving Human Subjects of the Faculty of Dentistry, University of Manitoba, which stipulated that the information gathered from subsequent records was to be used for research purposes. (See Appendix A-consent sheet, ethics committee sheet)

3.4 *IN VITRO* EXPERIMENTAL PROTOCOL

The spring constants of the Nitinol closed coil springs were determined, *in vitro*, using a vernier microscope to measure displacement under load. Five 0.030 x 0.0075 inch diameter closed coil springs (American Orthodontics®) were suspended from a fixed point. The inferior eyelet of each spring was tied to a plastic specimen dish. Weights were added to the specimen dish and the deflection of the spring was measured. The data describing the force and deflection of the spring was used to calculate five force-deflection curves for each of the closed coil springs. (See Appendix B) This enabled the calculation of the retraction force placed upon the sliding tubes by the Nitinol retraction springs at known linear displacements of the springs. Hysteresis of the Nitinol closed coil springs between loading and unloading was less than 11% of total force , with displacements less than 5mm.

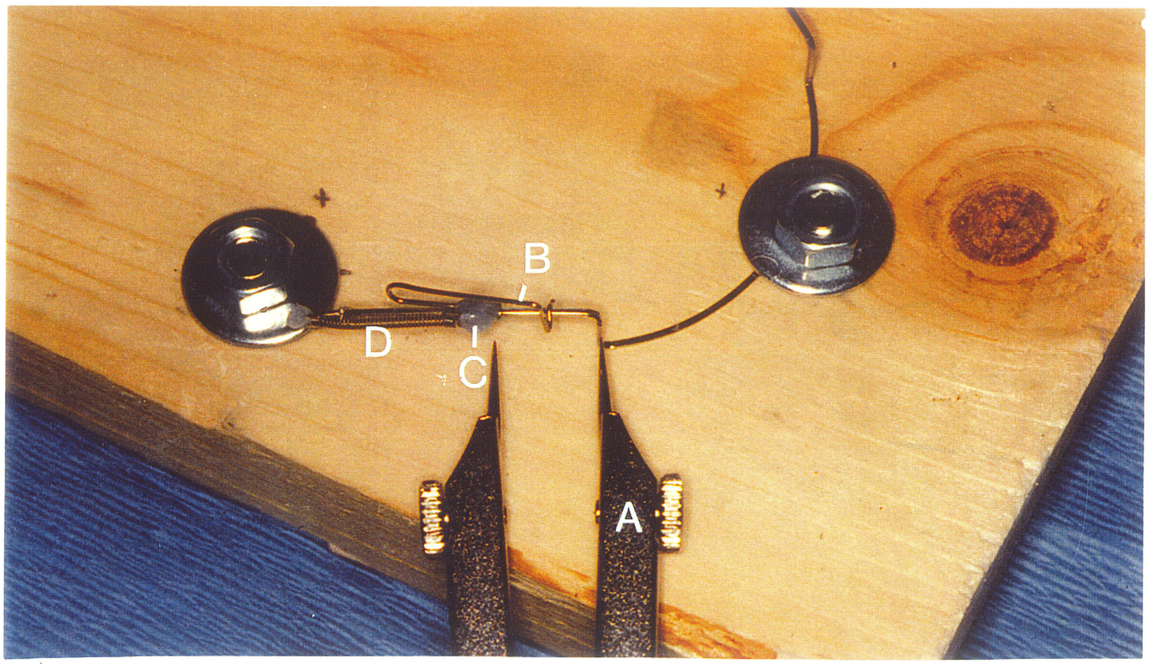


Figure 3.4.1 ***In Vitro*** **Testing Jig (A: Calipers used to measure tube displacement; B: Moment generating spring; C: Sliding tube ; D: Nitinol retraction spring)**

After each friction testing assembly was fabricated, it was disinfected for intraoral use by a 10 hour immersion in Coldspor®, a gluteraldehyde/compatible dual phenolic compound used for cold sterilization of non-autoclavable materials. This type of disinfection was necessary because normal heat or steam sterilization procedures could possibly heat-treat the Nitinol closed coil springs which would affect their force characteristics as well as affecting the strength of the composite resin used for attachment of the moment generating springs.

After sterilization, each friction testing assembly was washed with water, dried with a pressurized air spray, and secured to a stabilizing jig. An initial measurement was made from a fixed point on the utility arch to the mesial edge of the sliding tube in its rest position with the moment generating spring inactivated. The measurement was made with a pair of pointed dividing heads to the nearest 0.5 mm and recorded, as the rest length, for each Nitinol retraction spring for each narrow and wide tube, repectively. The error of measurement of spring elongation of +/- 0.25 mm resulted in a maximal error of $\mu_s = 0.01$. Both moment generating springs were then activated by placing the hook on top of the straight section of utility arch wire causing a moment to be placed on the sliding tubes. The moment developed between the sliding tube and the utility arch wire created friction which would inhibit the retraction of the sliding tube by the Nitinol closed coil spring. Each sliding tube was then pushed to the most mesial position possible on the straight section of the utility arch wire and allowed to rebound distally to rest. When each sliding tube came to rest, a measurement was made from the most mesial edge of the sliding tubes to the same fixed point on the utility arch wire as done previously. This procedure was completed three times on both sides of the utility arch containing the narrow and wide sliding tubes. The distance so measured was subtracted from the initial rest length to produce the amount of elongation of each Nitinol closed coil spring. The friction generated as a result of the applied moment prevented the retraction spring from retracting the sliding tube back to the rest position. The design of the friction testing assembly ensured that the force of retraction applied by the Nitinol closed coil spring when the sliding tube came to rest was equivalent to the force of friction resisting sliding tube movement. This assumption permitted the calculation of an *in vitro* static coefficient of friction for each width of sliding tube for each trial.

3.5 *IN VIVO* EXPERIMENTAL PROTOCOL

Each patient attended the University of Manitoba, Faculty of Dentistry, Graduate Orthodontic clinic for three research appointments. The initial appointment was used to obtain informed consent, complete a medical and dental history form, and ascertain if food allergies or medical conditions would affect participation in the project. Alginate impressions of the subject's upper arch were made to facilitate the fabrication of the utility arch prior to the experiment. After the initial *in vitro* trial, the second appointment was completed. At this appointment, the utility arch was inserted into the 0.018 x 0.025 inch auxiliary molar tube of the first molar bands. Adjustments were made to the utility arch to ensure its passive fit to avoid arch wire bending. Once this was completed, the anterior of the utility arch was tied to the patients continuous upper arch wire with two stainless steel ligatures of 0.012 inch dimension. The utility arch was then given 5 minutes to acclimatize to the patients intraoral environment. The subject then randomly decided, by coin toss, which side was to be tested first in the *in vivo* trial. The moment generating spring was placed in the activated position on one side of the utility arch and the sliding tubes were moved mesially as in the *in vitro* trial. The subjects were then given a standardized size of flavourless chewing gum base (Wrigley's Chewing Gum®) and instructed to masticate on a given side of their mouths for exactly 60 seconds. Subjects chewed to the beat of a metronome set at 60 beats/ second to standardize the number of chews per trial. Each subject was also timed independently with a stopwatch to ensure each trial lasted exactly 60 seconds. The Wrigley's® gum base was chosen because it was a consistently chewy material which created maximum vibration of the teeth while chewing. In addition, it did not inhibit or interfere with the mechanism of the friction testing assembly. The goal of mastication of the chewing gum base was to impart vibration to the teeth on to which the friction testing assembly was anchored. The vibration generated was required to test the hypothesis that intraoral vibration during mastication does not reduce the *in vivo* coefficient of friction compared to *in vitro* measurements. Six *in vivo* measurements were made of each of the narrow and wide tubes creating twelve measurements for each subject per *in vivo* trial. In order to identify if any changes occurred in the friction testing assembly as a result of the first series of *in vivo* trials, a second *in vitro* trial was performed before the second *in vivo* trial . All sliding tube displacement data was recorded manually to the nearest 0.5mm.

Figure 3.5.1 Friction Testing Apparatus - *In vivo* inactivated

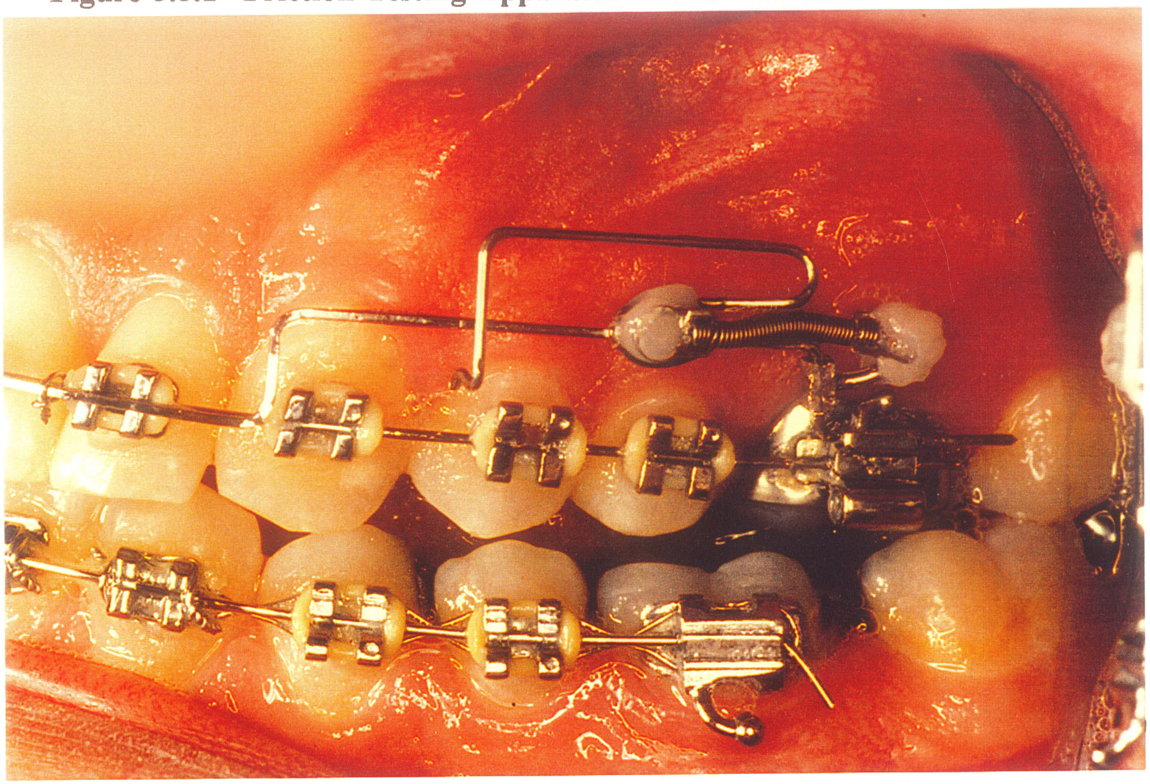
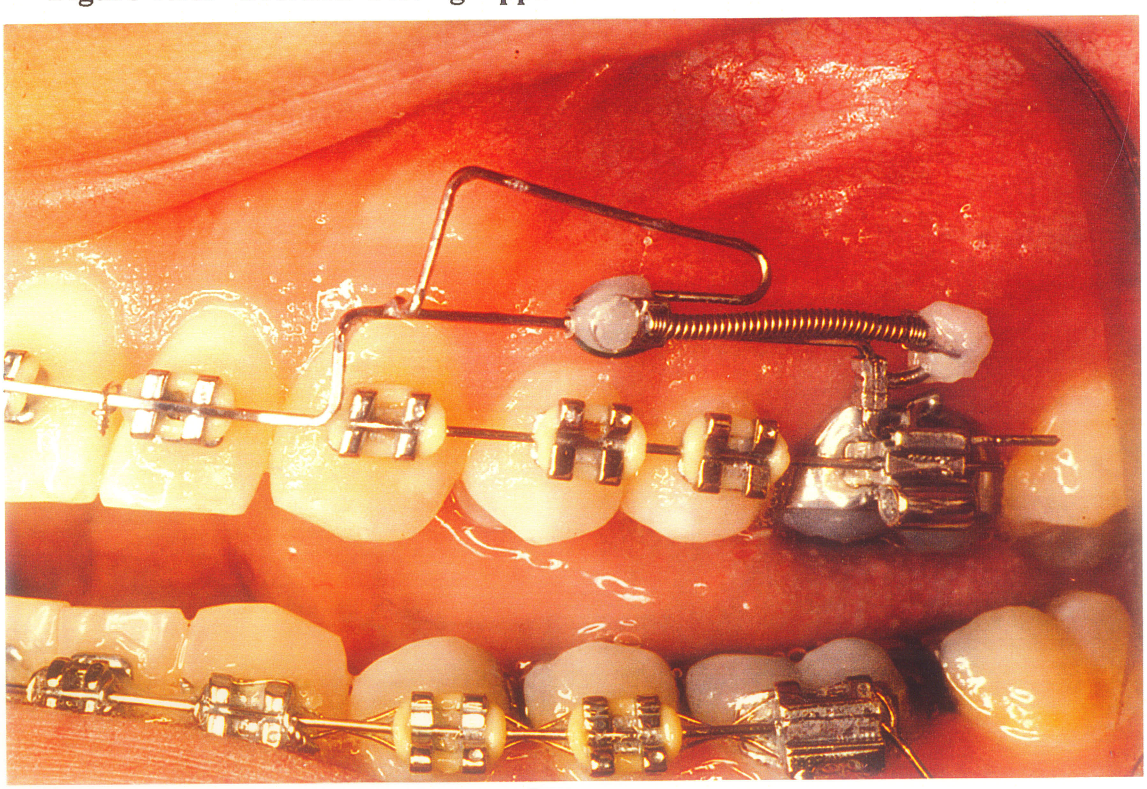


Figure 3.5.2 Friction Testing Apparatus - *In vivo* activated



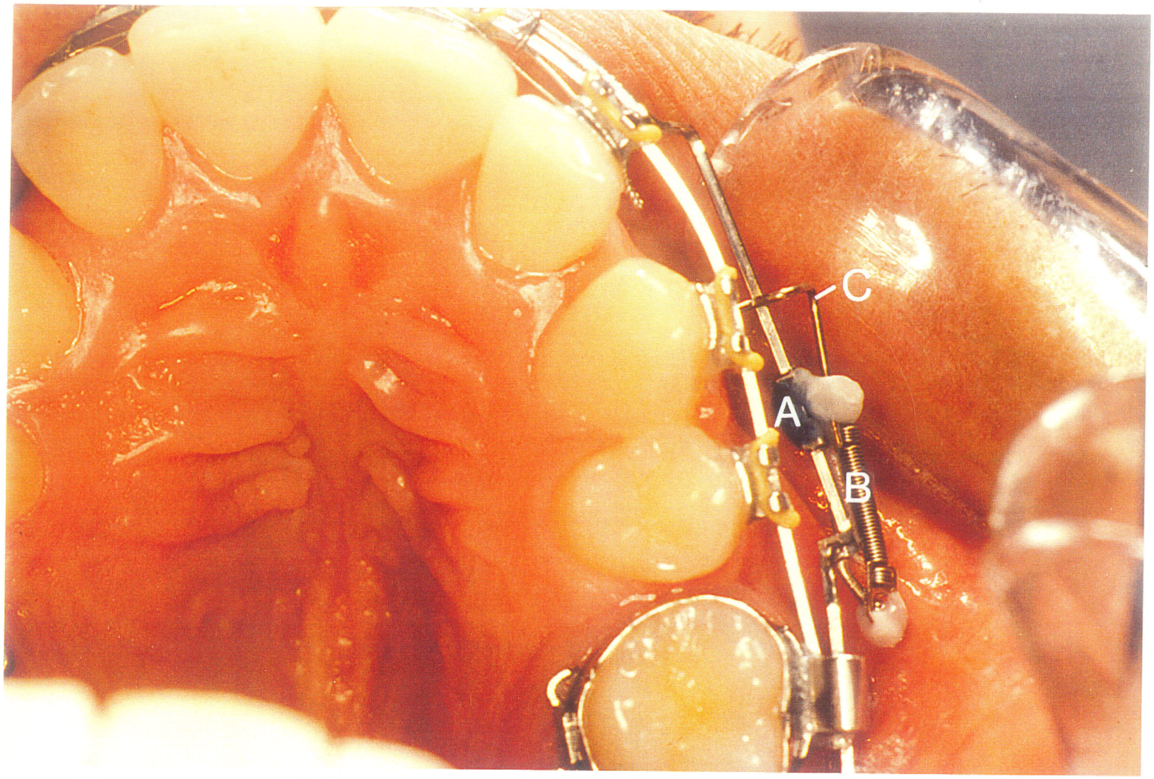


Figure 3.5.3 Occlusal view of Friction Testing Apparatus
A: Sliding tube B: Retraction spring C: Moment generating spring

3.6 DATA ANALYSIS

The force / displacement or spring constant of the Nitinol coil springs facilitated the calculation of the retraction force on the tube at a given displacement of each of the Nitinol coil springs. Since each sliding tube was in a state of equilibrium at the time of measurements, it can be assumed that the force of retraction of the Nitinol closed coil spring at that moment was equivalent to the force of friction preventing the sliding tubes movement. With the normal force of the moment generating spring on the utility arch wire previously calculated, it was possible to calculate the static coefficient of friction (μ_s) for the sliding tubes from the formula:

$$\mu_s = \text{Force}_{\text{Retraction}} \;/\; \text{Force}_{\text{Normal}}$$

The data collected from 20 trials on 10 subjects was entered manually into an IBM PC computer. The NCSS (Number Cruncher Statistical System) program was used to calculate a multiway analysis of variance.

3.7 COMPARISON OF *IN VITRO* AND *IN VIVO* COEFFICIENTS OF FRICTION

Results from several pilot studies using the friction testing apparatus showed that when normal forces between sliding tube and arch wire were above 200 grams, little observable change was exhibited in the static coefficient of friction between the *in vitro* and *in vivo testing* environments. Subsequent friction testing utilized sliding tube normal forces which ranged from 104 grams to 154 grams for the short tubes with a mean equal to 119 grams with a standard deviation equal to 15 grams. Normal forces ranged from 56 grams to 108 grams for the long tubes with a mean equal to 78 grams with a standard deviation equal to 16 grams. The normal forces used for the long and short tubes were chosen on the basis that they were within the range of force values that would permit a significant change in the static coefficient of friction, when exposed to intraoral vibration. It is important to note that comparison between tube lengths in similar or different testing environments is not as applicable due to the differences in normal forces for the different length tubes employed. Normal forces differed by 35% on average between the short and long tubes, with the long tubes having a mean normal force 41 grams less than the short tubes. This was intentionally done to maximize the observed tube displacement between the *in vitro* and *in vivo* testing environments, and to minimize the percentage of measurement error.

Figure 3.8.1 Freebody Diagram of Friction Testing Assembly

Legend

d = width of sliding tube
L = length of moment spring
F_1 = force generated by the moment spring
$F_2 = F_3$ = forces of the couple generated by the moment spring, upon the tube by the utility arch wire

The normal force of sliding tube placed upon arch wire is equal to;

$$F_{N1} = F_2 + F_{1/2}$$
$$F_{N2} = F_{1/2} - F_3$$

Since $F_{N1} > F_{N2}$, it represents the largest normal force of the sliding tube upon the arch wire

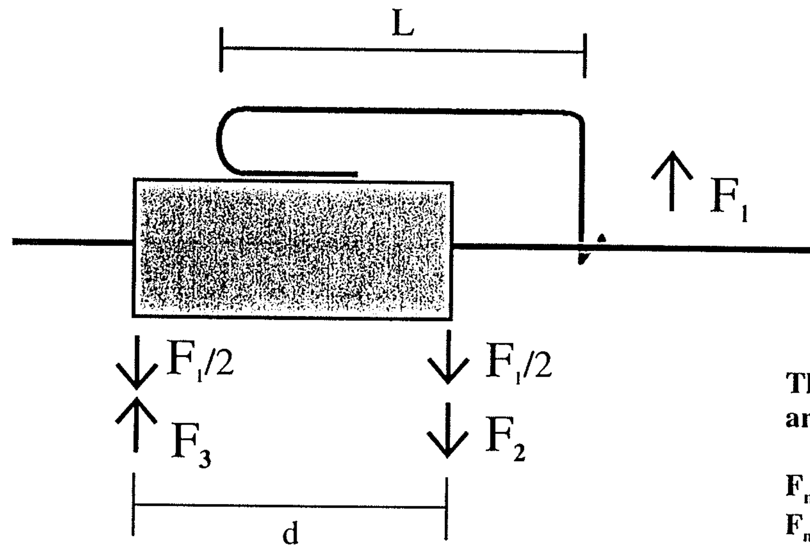
At equilibrium,

Moment generated by the moment Spring = Moment of the Tube upon arch wire

$$Ms = Mt$$
$$(F_1)(L) = (d)(F_2)$$

$$F_2 = \frac{(L)(F_1)}{(d)}$$

Free Body Diagram of Friction Testing Assembly



The normal forces of the sliding tube upon the arch wire are equal to;

$F_{n1} = F_2 + F_1/2$ where $Fn1 > Fn2$
$F_{n2} = F_1/2 - F_3$

At equilibrium;
Moment of Spring = Moment of Tube

$$Ms = Mt$$
$$(F_1)(L) = (d)(F_2)$$
$$F_2 = \frac{(L)(F_1)}{d}$$

Legend
d = width of sliding tube
L = length of moment spring
F_1 = force generated by the moment spring
$F_2 = F_3$ = forces of the couple generated by the moment spring upon the tube by the utility arch wire

3.9 ERROR OF THE METHOD

Error in fabrication of the friction testing assembly was minimized by using the same apparatus for testing in *in vitro* and *in vivo* environments. Alternating the *in vitro* and the *in vivo* trials increased the likelihood of detecting if the testing procedure altered the function of the friction testing apparatus between sessions.

Three different error studies were undertaken to help identify sources of variance in the results. Before clinical trials were begun, a consistency test was performed to identify how consistently the friction testing assemblies could be manufactured. Four friction testing assemblies were fabricated containing eight sliding tubes of two sizes, 8.0 mm and 4.0 mm respectively. Each moment generating spring of the similar length tubes was fabricated with as comparable activation as possible to create identical testing situations. These four friction testing assemblies were then subjected to two *in vitro* testing procedures to calculate the variance of the coefficient of friction. Within these similar testing situations, the theoretical static coefficient of friction should only be dependant upon the materials in contact. Results from the consistency study for short tubes gave static coefficients of friction ranging from 0.12 to 0.11, or +/- 7% variation, whereas the results from the long tube consistency study gave static coefficients of friction ranging from 0.14 to 0.12, or +/- 6% variation. This amount of variance in the coefficient of friction among different tube widths was judged as acceptable variation in experimental design for similar materials.

A second error study was completed after the clinical trials had finished to evaluate the effect of saliva on the *in vitro* static coefficient of friction. Natural saliva was the only option for this type of experiment, as only whole unstimulated natural saliva could duplicate the *in vivo* effect of saliva containing glycoproteins, necessary for duplicating the rheology of saliva (Dawes, 1996). Whole unstimulated saliva was collected in a beaker from one individual over a period of 1.5 hours. The 10 friction testing assemblies were tested dry using the *in vitro* testing procedure as outlined previously. The 10 friction testing assemblies were then covered by the collected saliva and retested in an identical manner. Results were recorded and comparison of the wet and dry states was completed using a paired t-test. (Sigmaplot, (Jancen Corp.®))

A third error study was completed to evaluate the effect of arch wire bending on the coefficient of friction of the different length of sliding tubes. A straight length of 0.017x 0.025 inch stainless steel wire held in tension was manipulated to gradually increase the curvature of the arch wire. (Refer to Figure 3.9.1) A vernier microscope was mounted perpendicular to this arch wire to accurately measure the vertical distance of displacement of the middle of the arch wire span. Identical *in vitro* testing procedures were then repeated at successively greater arch wire deflections to evaluate if arch wire deflection and sliding tube length significantly affected the static coefficient of friction. Testing was also completed with zero arch wire deflection which served as a base line data for comparison.

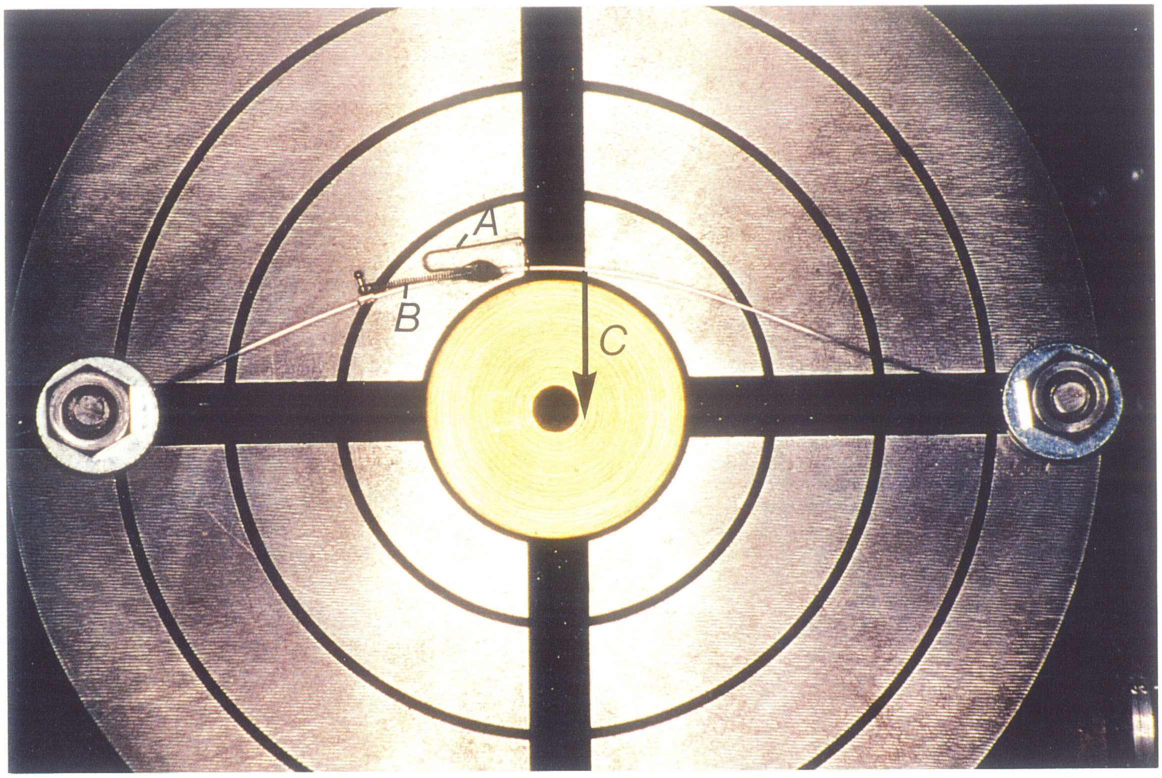


Figure 3.9.1 Archwire Deflection Experiment
A: Moment generating spring B: Retraction spring
C: Vertical deflection of arch wire

CHAPTER 4

RESULTS

RESULTS

4.1 INTRODUCTION

Data was collected and analyzed for 20 *in vivo* and 20 *in vitro* trials on 10 subjects using each subject's experimental apparatus for both environments. For each subject, a comparison of the static coefficient of friction between the *in vitro* and *in vivo* testing environment is presented. The results are displayed in a graphical and tabular form representing the mean static coefficient of friction for each testing situation. Six *in vivo* and six *in vitro* measurements were performed for the short and long tubes respectively, at two different testing sessions.

4.2 MULITWAY ANALYSIS OF VARIANCE

The percentage change of the static coefficient of friction was calculated between testing environments for each testing session. Means of the static coefficients of friction were calculated by averaging coefficients within similar testing enviroments. A mutliway analysis of variance (ANOVA) was used to identify significant differences between subjects and within subjects with respect to: testing environment, tube length, testing environment with tube length, and tube length with subject tested.

Multiway ANOVA	df	S.S.	M.S.	F ratio	Probability
Between subjects	9	0.0331	0.0037	4.842	S, ($p<0.001$)
Within subjects					
short tube / long tube	1	0.0499	0.0499	65.592	S, ($p<0.001$)
in vivo / in vitro	1	0.1846	0.1846	242.895	S, ($p<0.001$)
tube length / environment	1	0.0001	0.0001	0.013	NS, ($p>0.05$)
subject / environment	9	0.0170	0.0019	2.487	S, ($p<0.05$)
tube length / subject	9	0.0449	0.0050	6.566	S, ($p<0.001$)
error	49	0.0372	0.0008		

Table 4.2.1 Multiway Analysis of Variance

A multiway analysis of variance of the data from 10 subjects having two trials in either testing environment revealed that intraoral vibration decreased the *in vivo* static coefficient of friction as compared to *in vitro* results ($p<0.001$). No significant difference was identified between testing environment and tube length, indicating that the differences between the *in vivo* and *in vitro* testing environments were similar for both long and short tubes for all subjects tested. This demonstrates that saliva and intraoral vibration had the same influence on the frictional resistance of the long and short sliding tubes. Long and short tubes exhibited different coefficients of friction for all subjects tested ($p<0.001$); the long tubes having the higher value. The effect of tube length was different among the subjects tested ($p<0.001$). This could be a result of the differences in the quality and quantity saliva and intraoral vibration among subjects. The *in vivo* and *in vitro* differences as well as the differences between long and short tubes varied significantly among the ten subjects tested. ($p<0.05$) However these differences, because of their small magnitude, were not felt to be clinically significant.

4.3 SUMMARY OF *IN VITRO* AND *IN VIVO* RESULTS

	Short - *in vitro*	**Short - *in vivo***	**Long - *in vitro***	**Long - *in vivo***
MEAN	0.21	0.12	0.26	0.17
S.D.	0.05	0.04	0.03	0.05
S.E.	0.01	0.01	0.01	0.01
95% CONFID.	0.02	0.02	0.02	0.02
MAX VALUE	0.30	0.20	0.31	0.24
MIN VALUE	0.15	0.06	0.19	0.09

Table 4.3.1 Summary of *In Vitro* and *In Vivo* Results

The mean coefficient of friction for the short tubes decreased by an average of 0.09 between *in vitro* and *in vivo* testing. The mean coefficient of friction for the long tubes decreased by 0.09 between *in vitro* and *in vivo* testing. The change in the mean coefficient of friction for the long and short tubes was found to be very similar. (See figure 4.3.1.)

Figure 4.3.1 Change in the Mean Coefficient of Friction for Long and Short Tubes

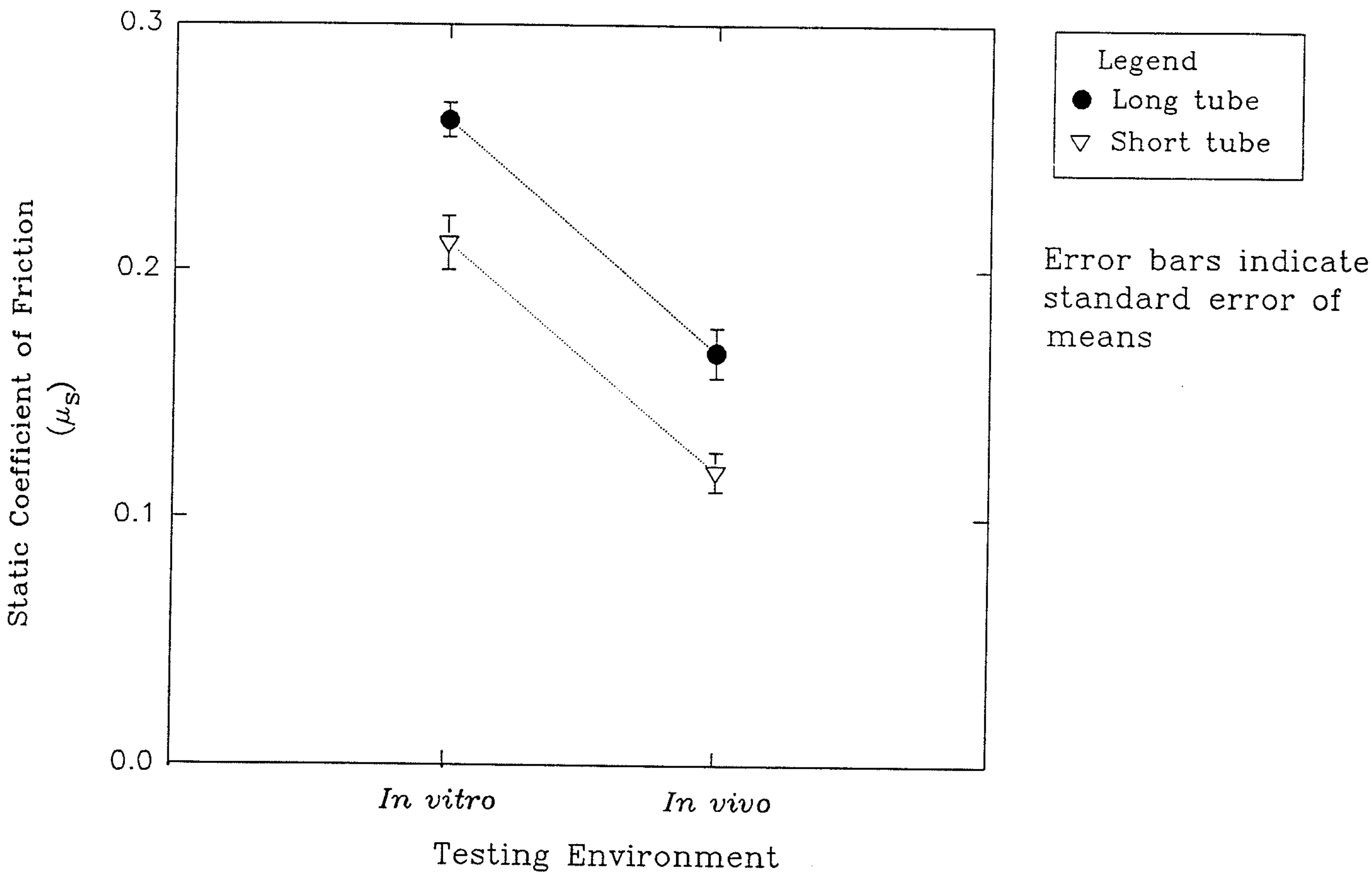
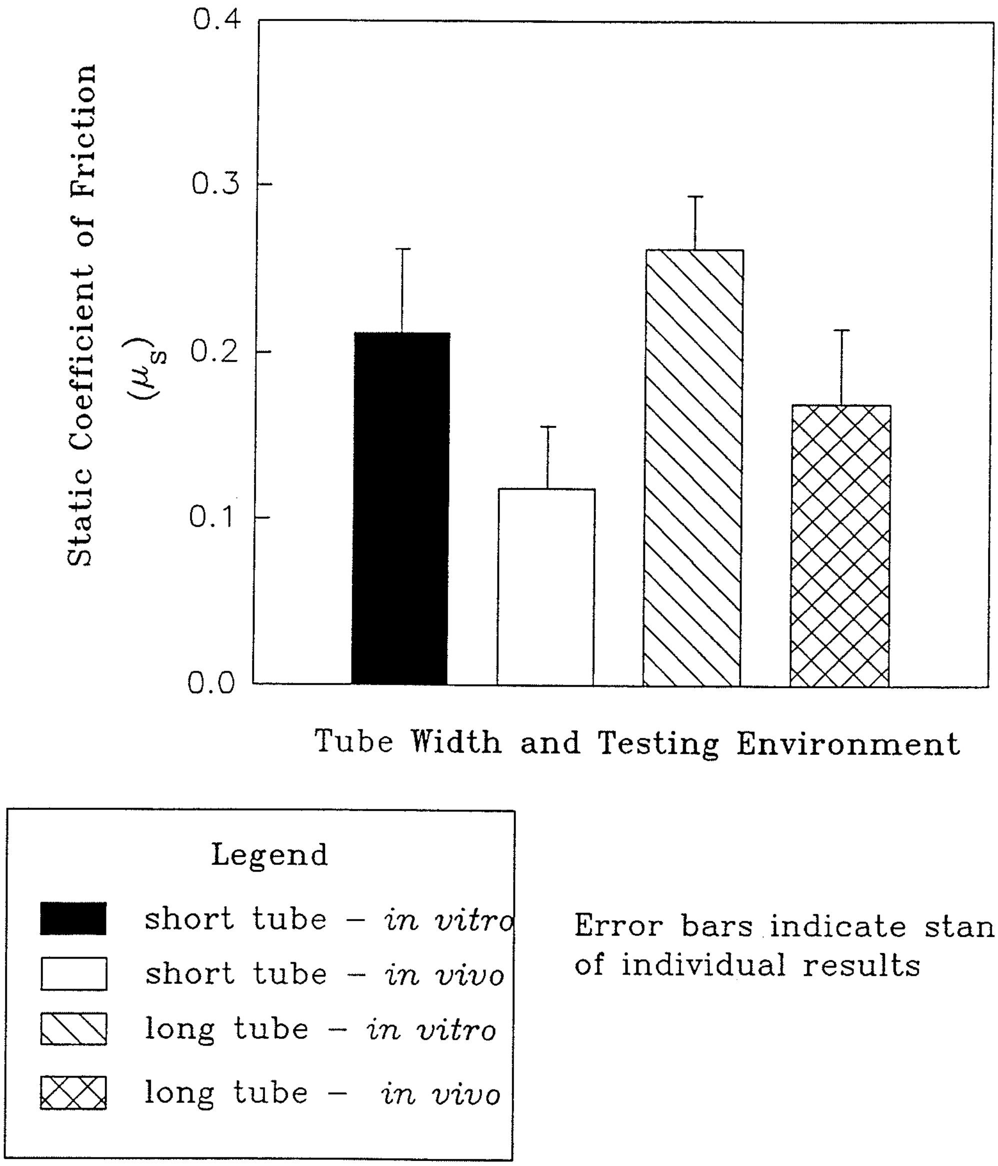


Figure 4.3.2

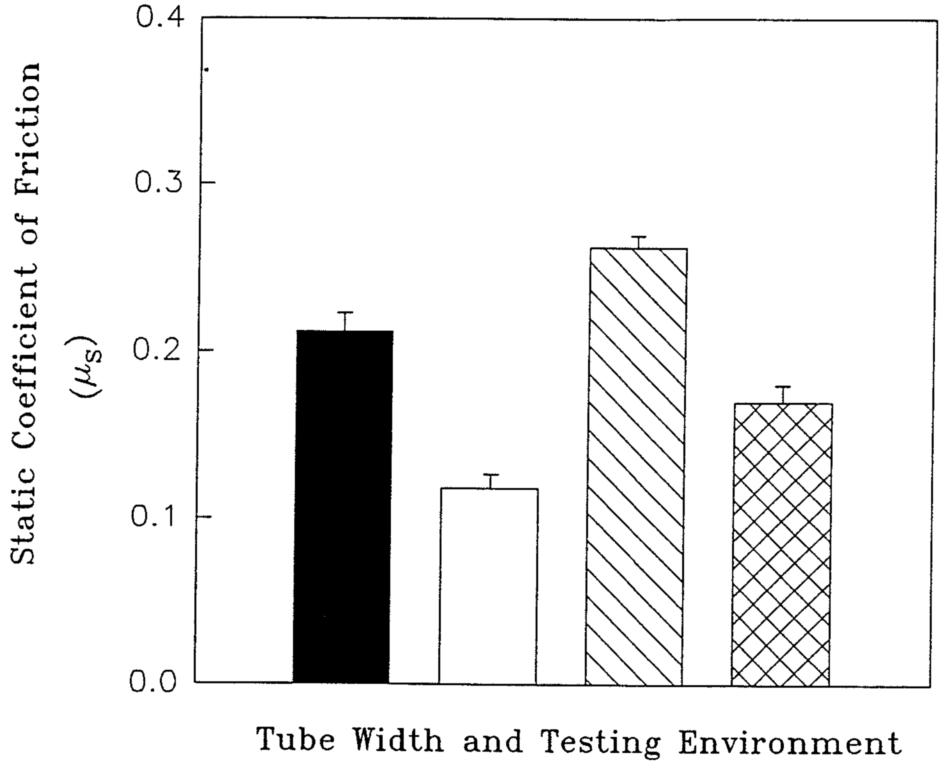
Mean Coefficients of Friction between Testing Environments



Error bars indicate standard deviation of individual results

Figure 4.3.3

Mean Coefficients of Friction between Testing Environments



Legend

- short tube – *in vitro*
- short tube – *in vivo*
- long tube – *in vitro*
- long tube – *in vivo*

Error bars indicate standard error of mean

4.4 INFLUENCE OF SALIVA ON *IN VITRO* COEFFICIENTS OF FRICTION

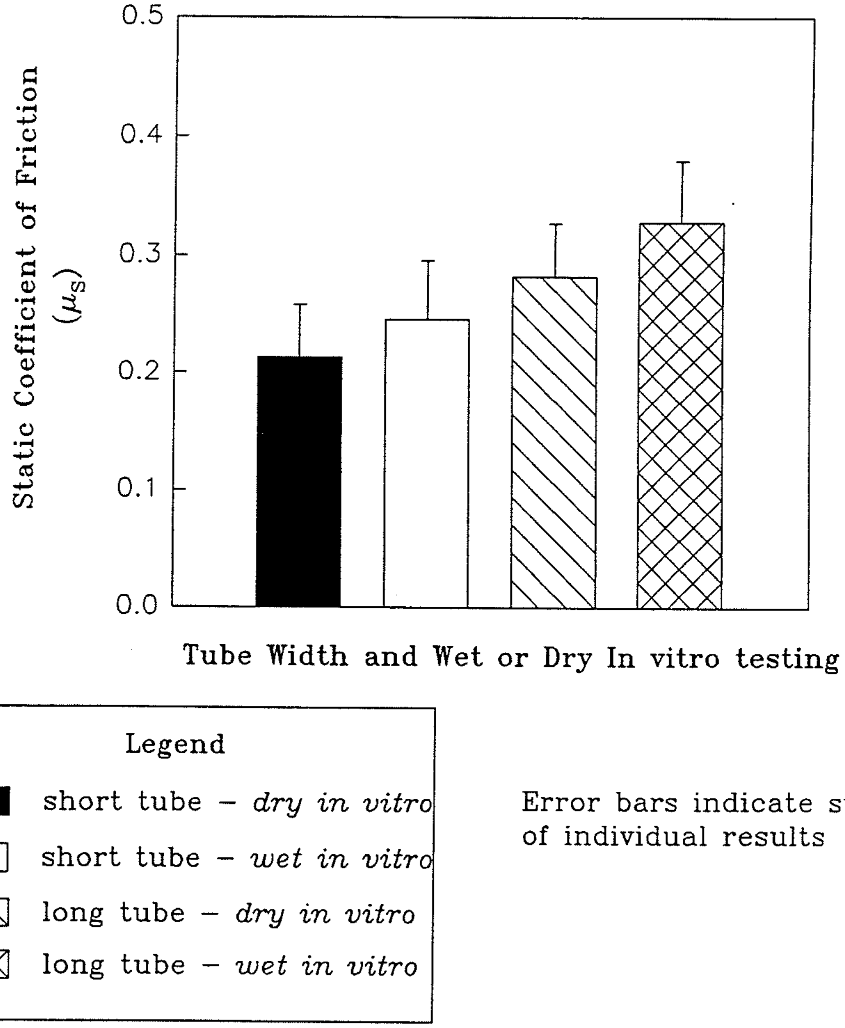
To accurately compare the effect of the presence of saliva on *in vitro* static coefficients of friction, the friction testing assemblies were retested with the *in vitro* testing protocol in the presence of saliva. Evaluation of the friction testing assemblies in the presence of saliva helps isolate the influence of intraoral vibration upon frictional resistance.

Appliance	Long μ_s - wet	Long μ_s - dry	Short μ_s - wet	Short μ_s - dry
AB	0.33	0.23	0.27	0.18
SG	0.27	0.32	0.35	0.23
JW	0.39	0.32	0.18	0.18
NG	0.35	0.35	0.30	0.27
TD	0.23	0.21	0.21	0.21
KC	0.35	0.28	0.26	0.29
NH	0.36	0.25	0.22	0.19
CD	0.38	0.28	0.21	0.15
VB	0.34	0.29	0.25	0.25
KW	0.29	0.23	0.22	0.19
SUMMARY				
MEAN	0.33	0.28	0.25	0.21
S.D.	0.05	0.05	0.05	0.05
S.E.	0.02	0.01	0.02	0.01
Paired T-tests	means different $t=2.987$, $p<0.02$		means different $t=2.354$, $p<0.05$	

Table 4.4.1 Influence of Saliva on the *In Vitro* Coefficient of Friction

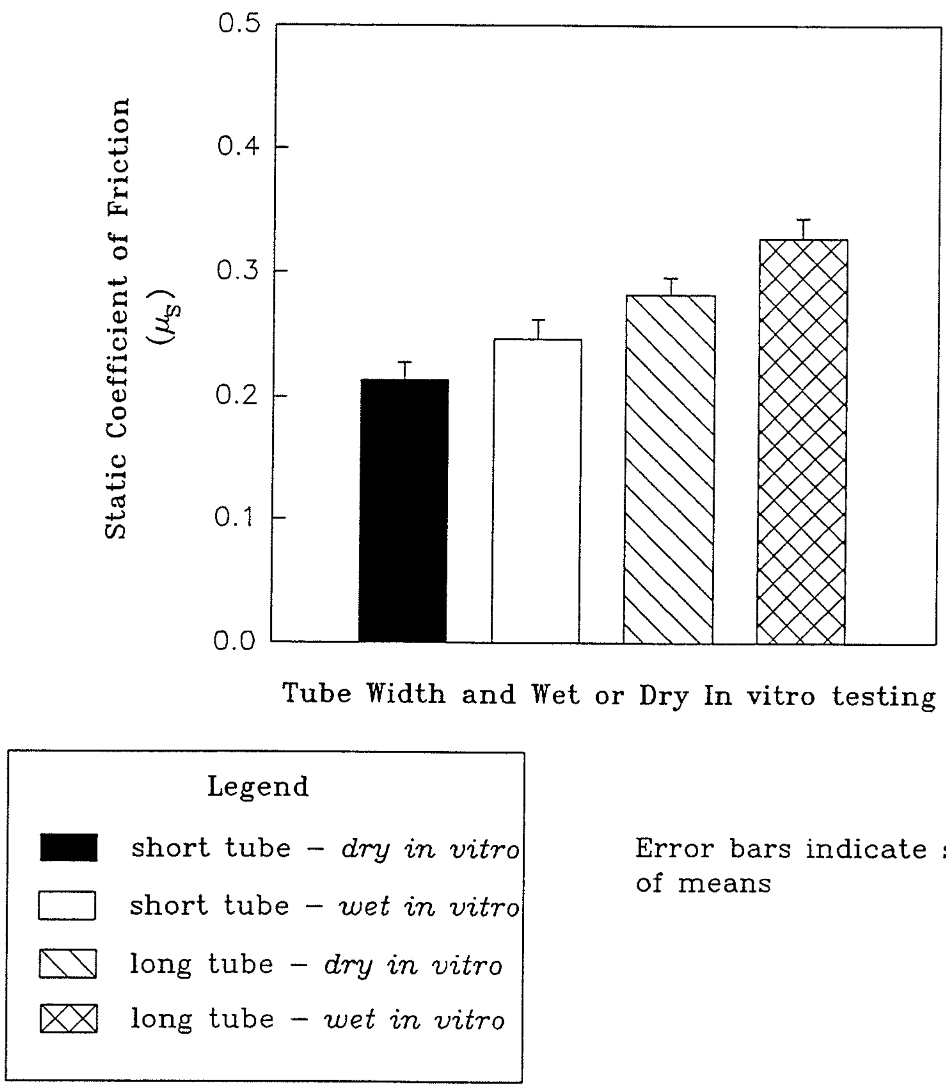
The presence of saliva increased the mean *in vitro* coefficient of friction for both long and short tubes. Paired t-tests of the long and short tube data revealed that the mean wet and dry *in vitro* coefficients of friction for long and short tubes were different ($p<0.02$ long tubes, $p<0.05$ short tubes). The standard deviation and standard error calculated for the *in vitro* saliva experiments were similar for both testing conditions and tube lengths.

Figure 4.4.1
Influence of Saliva on Friction between Testing Environments



Error bars indicate standard deviation of individual results

Figure 4.4.2
Influence of Saliva on Friction between Testing Environments



Error bars indicate standard error of means

4.5 EFFECT OF VERTICAL ARCHWIRE DEFLECTION ON *IN VITRO* COEFFICIENT OF FRICTION

Comparision of the mean *in vitro* and *in vivo* static coefficients of friction revealed that the short tubes experienced less frictional resistance than long tubes. The mechanics of frictional resistance suggest that long tubes should have experienced less frictional resistance than short tubes because the extra tube length produces lower normal forces on the arch wire. One hypothesis for this peculiar result was that vertical arch wire deflection during frictional resistance testing was inhibiting the movement of the longer tubes.

Therefore, to investigate the effect of vertical arch wire deflection on the *in vitro* static coefficients of friction, the friction testing assemblies were retested with the *in vitro* testing protocol with incrementally greater arch wire deflection.

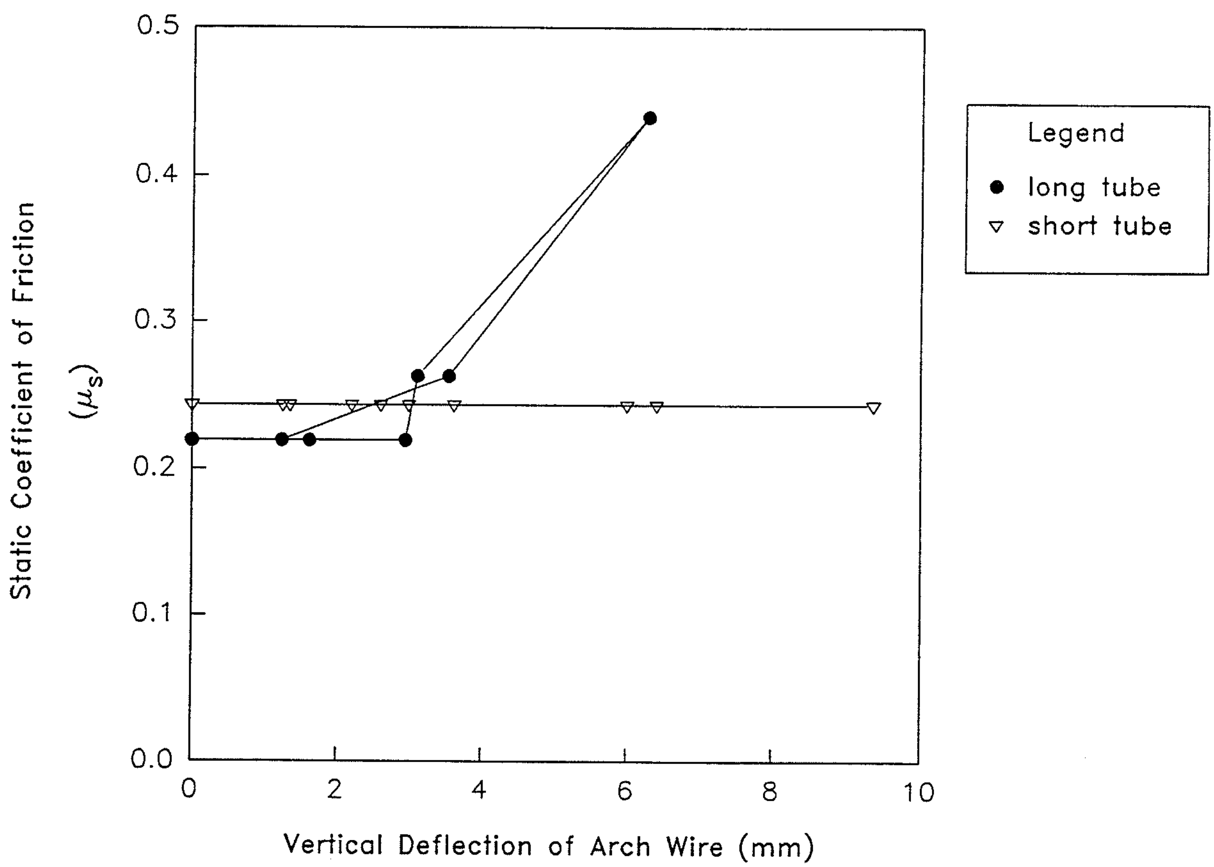
Arch Wire Deflection (mm)	Long Tube μ_s - dry
0	0.22
1.2	0.22
3.6	0.26
6.3	0.44
3.1	0.26
3.0	0.22
1.6	0.22
0	0.22
Arch Wire Deflection (mm)	**Short Tube μ_s - dry**
0	0.24
1.2 - 9.7	0.24

Table 4.5.1 Effect of Vertical Archwire Deflection on *In Vitro* Coefficient of Friction

The static coefficient of friction for the long sliding tubes was much more sensitive to vertical deflection of the arch wire than the short tubes. The long tube's static coefficient of friction increased 18% with 3.6 mm deflection, and 100% with 6.28 mm deflection. The static coeffficient of friction of the short sliding tubes was unaffected by vertical arch wire deflection.

Figure 4.5.1

Affect of Vertical Arch Wire Deflection on *In Vitro* Coefficients of Friction



4.6 COMPARISON OF INITIAL AND FINAL DRY *IN VITRO* COEFFICIENTS OF FRICTION

A second hypothesis for the cause of the increased frictional resistance of the long sliding tubes could be the inability to properly clean and remove dried organic contaminants from the tubes after each *in vivo* trial. Dried organic contaminants that could not be removed from the long tubes might increase the resistance at the arch wire-sliding tube interface, thereby increasing the coefficient of friction of the long tubes . Comparison of the initial clean dry long tube coefficient of friction with the same tube final dry coefficient of friction, after completing all *in vivo* tests, would help determine if this effect was significant.

To investigate the effect of dried organic contaminants on the static coefficient of friction of the long sliding tubes, the initial *in vitro* static coefficient of friction of each long sliding tube was compared to a final *in vitro* coefficeint of friction taken at the commencement of the *in vitro* saliva experiment.

Appliance	First Dry *In vitro* μ_s	Last Dry *In vitro* μ_s
AB	0.30	0.30
SG	0.30	0.32
JW	0.25	0.32
NG	0.26	0.35
TD	0.23	0.22
KC	0.25	0.28
NH	0.22	0.25
CD	0.28	0.28
VB	0.29	0.29
KW	0.23	0.23
MEAN	0.26	0.28
S.D.	0.03	0.04
S.E.	0.01	0.01
Paired t test	means not different, t=2.158, NS	

Table 4.6.1 Comparison of Initial and Final Dry *in Vitro* Coefficients of Friction

The mean final long tube *in vitro* coefficient of friction was not found to be significantly higher than the mean initial long tube coefficient of friction, by a paired t test. Similarly, a variance ratio test identified that the means were not significantly different. However, the standard deviation for the final dry *in vitro* coefficient of friction is notably larger, indicating increased variation about the calculated mean, but not to the point of statistical significance. Most likely, organic contaminants within the longer sliding tubes did not increase the static coefficient of friction.

CHAPTER 5

DISCUSSION

DISCUSSION

5.1 INTRODUCTION

A specially designed device was used in this study to test the hypothesis that intraoral vibration during mastication does not reduce the *in vivo* coefficient of friction of orthodontic appliances using sliding mechanics. The friction testing assembly in this experiment provided a consistent and relevant method for comparing the static coefficients of friction for a given tube length between *in vitro* and *in vivo* environments. Data was collected and analyzed from 240 trials on 10 subjects, providing 24 individual coefficient of friction measurements.

The findings from the experiments permitted rejection of the null hypothesis. That is, the *in vivo* coefficient of friction of a device mimicking an orthodontic appliance was reduced with mastication and this effect was attributed to the effect of intraoral vibration. It was also found that frictional resistance was not totally eliminated by intraoral vibration.

5.2 DIFFERENCES IN FRICTION MEASUREMENTS RECORDED IN THE *IN VITRO* AND *IN VIVO* ENVIRONMENTS

A multiway analysis of variance (ANOVA) identified that intraoral vibration decreased the *in vivo* static coefficient of friction as compared to *in vitro* results using the same apparatus. In addition, the ANOVA identified that the differences between testing environments were significant for all subjects tested. This confirms that unless the experimental technique includes vibration during testing, *in vitro* frictional resistance testing of orthodontic products, by itself, does not give significant insight into the frictional resistance observed clinically.

The ANOVA also identified that long tubes had significantly different coefficients of friction than short tubes, and these differences were observed in all subjects.

No significant differences were identified between testing environment and tube length, indicating that differences in the mean static coefficient of friction due to influences of testing environment were similar for both tube lengths, in all subjects tested. The decrease in the calculated coefficient of friction from *in vitro* to *in vivo* environment was attributed to vibration resulting from mastication in the presence of saliva. These effects were similar for both wide and narrow tubes. Thus, the imprecision of *in vitro* testing in previous studies and by orthodontic manufactures would be similar whether undertaken with wide and narrow orthodontic brackets.

5.3 DIFFERENCES DUE TO TUBE LENGTH

Long tubes were found to have higher static coefficients of friction than short tubes for both *in vitro* and *in vivo* testing. In addition, the mean coefficient of friction for *in vivo* long tube testing was 30% higher than for short tube testing under similar conditions. This finding is contrary to what is expected, considering the mechanism of frictional resistance when using sliding mechanics. Several possibilities could explain this aberrant finding. Firstly, organic contaminants may have entered the lumen of the long tubes that could not be removed with cleaning as effectively as with short tubes. To investigate this possibility, a comparison of initial and final dry *in vitro* static coefficients of friction for the long tubes was completed. No significant difference between mean static coefficients of friction for initial and final dry *in vitro* testing was found. However, the standard deviation for the mean final dry *in vitro* coefficient for the long tubes was

25 % larger than the standard deviation for the mean initial dry *in vitro* coefficient. This difference in standard deviation indicates greater variation about the mean, perhaps caused by dried organic contaminants entering a few of the longer tubes.

A second possible explanation for the behaviour of the long tubes is the effect of arch wire deflection causing binding of the long tubes. Arch wire deflection experiments indicated that the long tubes were very sensitive to vertical arch wire deflection. Short tube frictional resistance was unaffected by vertical arch wire deflection, yet deflections of 3.5mm, caused the long tube frictional resistance to increase 18%. Thus, small deflections of the utility arch wire could produce an increase that would account for at least one half of the difference observed between the *in vitro* and *in vivo* testing environments. Arch wire deflections approximating 3.5mm, could have significantly increased the frictional resistance of the long sliding tubes. Another possible reason for long tubes having a greater mean coefficient of friction than short tubes is the variation in normal forces compared to tractional forces for the given tube lengths.

	Long tube	Short tube	Long to Short Difference
Mean Tractional Force (F_T)	20.9	25.0	Long < Short by 16 %
Mean Normal Force (F_N)	77.9	118.9	Long < Short by 35 %
Calculated $F_T/F_N = \mu_s$	0.27	0.21	Long > Short by 22 %

Table 5.3.1 Comparison of the *In vitro* Tractional and Normal Forces between Tube Lengths

It was found that the mean tractional force on the long tubes was 16% less than on the short tubes, while the mean normal force on the long tubes was 35% less than the normal force on the short tubes. Since the calculation of the static coefficient of friction involves a ratio of tractional force to normal force, a larger decrease in the denominator than in the numerator would create an increased value for the static coefficient of friction. Perhaps the variation of the normal forces to tractional forces in the experiment artificially increased the static coefficient of friction for the long tubes. Paradoxically, the mechanics of frictional resistance dictate that the lower normal forces experienced by the long tubes should make the longer tubes more sensitive to vibration.

The effect of tube length was found to be significantly different among all the subjects tested. This indicates that the influence of tube length on frictional resistance differed between subjects or this could be due to the variation in arch wire bending across the 0.017 x 0.025 inch dimension. These differences are most likely due to differences in the magnitude of masticatory force as well as the quantity and quality of saliva and the influence these variables have on intraoral vibration.

Static coefficients of friction calculated from identical trials on identical subjects on different days did exhibit a mean variation of approximately 10% for both tube lengths. This was most likely a result of the variable nature of the testing apparatus as well as the variable nature of masticatory forces and saliva.

5.4 DIFFERENCES DUE TO PRESENCE OF SALIVA

Paired t test comparison of the dry and wet *in vitro* coefficients of friction revealed that the presence of saliva increased the *in vitro* static coefficient friction for both long ($p<0.02$) and short tubes ($p<0.05$). The increase in the coefficient of friction from dry to wet *in vitro* testing was greater for the long tubes than for the short tubes. It is possible that the differences in the frictional resistance between the *in vivo* and *in vitro* environments are larger than were observed in our experimental design, because the *in vivo* static coefficient of friction was significantly less than the dry *in vitro* coefficient of friction in all subjects tested. Several theories may be used to explain why the presence of saliva increased the *in vitro* coefficient of friction. Kusy *et al.* (1991) believed that stainless steel brackets sliding on stainless steel arch wire may experience some adhesive behaviour in the wet versus dry state. In addition, Kusy hypothesized that saliva may chemically break down chromium oxide surface characteristics which render certain arch wire surfaces chemically inactive, or alternatively, saliva could be acting as an adhesive because of surface tension effects. Stannard *et al.* (1986) stated that water and other polar liquids are known to increase adhesion and attraction among polar materials and thus increase friction. This behaviour has been observed for several different dental materials in the presence of saliva, and has been explained by the adhesion theory of friction. Clinically, high pressure contacts between arch wires and bracket slots create a situation of classical dry friction. Pressure from normal forces is thought to expel saliva at areas of metal to metal contact. Dry surface friction increases with increasing normal forces, thereby creating more stick-slip phenomenon *in vivo*. Saliva may only act as a lubricant at low loads which are easily exceeded with the normal forces generated with the use of sliding mechanics. It should be noted that this study utilized normal forces which exceeded the capacity for boundary lubrication to occur, and as a result, strictly evaluated only the influence of intraoral vibration upon classical dry friction.

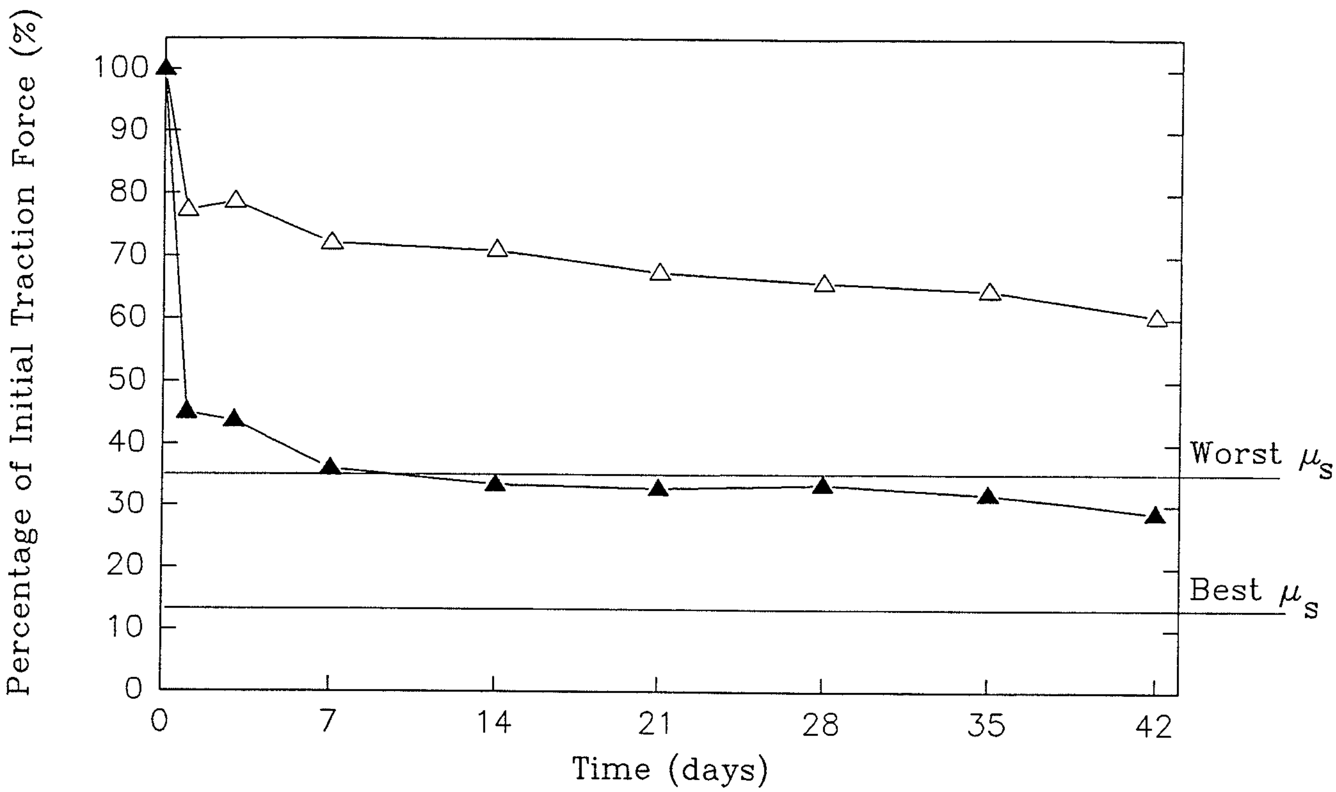
5.5 CLINICAL SIGNIFICANCE OF RESULTS

Results obtained from this project have relevance to the current clinical practice of orthodontics. The *in vivo* results from this study can offer some measure of the lowest coefficients of friction that are likely to be found in the intraoral environment, given that ligation effects were not taken into account.

Elastics are commonly utilized to produce the force required to slide teeth along orthodontic arch wire. Unfortunately, there is rapid decay of the forces produced by the various elastics in use today. (Killiany *et al.*, 1985) It is not unusual for force levels to decay to 50% of initial values within 24 hours of placement intraorally. This phenomenon raises an important issue concerning the ability of elastics to produce tractional forces of sufficient magnitude over a period of time necessary to produce effective tooth movement. For example, the variation in force decay of two elastic chains is shown in figure 5.5.1.(data taken from Killany *et al.*,1985) Note that on the graph, a "worst case" and "best case" of *in vivo* coefficients of friction are identified from results found in this study. In the best case (lowest *in vivo* coefficient of friction; $\mu_S = 0.09$), 12% (19gm) of the assumed total tractional force is required to overcome the frictional resistance. In the worst case (highest *in vivo* coefficient of friction; $\mu_S = 0.24$), 35% (52gm) of the assumed total tractional force is required to overcome the frictional resistance. These results indicate that it is possible to lose from 12% to 35% of applied tractional force to overcome frictional resistance, when a 3.5 mm wide bracket is used to translate a tooth with a center of resistance 10 mm apical to the bracket slot. It is apparent that force dissipation through frictional resistance can result in the teeth experiencing less net tractional forces than expected.

Figure 5.5.1

Percentage of Tractional Force Lost to Friction



Legend
△ RMO Energy Chain
▲ AO Plastic Chain

Assuming a traction force of 150 grams, using a 3.5mm bracket, 10mm distance from the center of resistance of a tooth. This does not include the effects of ligation.

Data source: (Killiany et al., 1985)

It is essential that the frictional characteristics of all the materials we used in clinical practice be well characterized so that the quantity of tractional force that is lost to friction can be better estimated. If it is possible to calculate the amount of frictional resistance found in any tooth moving system, and what percentage of tractional force is lost to friction, then the character and duration of tractional force could be better designed to maximize the efficiency of the biological response. Use of continuous forces that remain high enough to compensate for forces lost to friction will result in more efficient tooth movement, that is less deleterious to the supporting tissues.

5.6 MANAGEMENT OF FRICTION IN CLINICAL PRACTICE

The results of this project demonstrate that intraoral vibration dramatically reduces frictional resistance as compared to *in vitro* tests. Consequently, asking orthodontic patients to chew gum during treatment involving sliding mechanics is one method by which an orthodontist may increase intraoral vibration to decrease frictional resistance. Hwang *et al.*(1994) demonstrated that having orthodontic patients chew on an arch form fabricated from a plastic polymer (KM Thera-Bite®) reduced pain during orthodontic treatment. Hwang *et al.*(1994), believe that the major source of pain during orthodontic treatment is probably due to the creation of ischemic areas that undergo sterile necrosis within the periodontal ligament. Chewing something that causes individual teeth to jiggle within their alveolar sockets was recommended as a means of loosening the tightly grouped fibers around the nerves and blood vessels, restoring normal vascular and lymphatic circulation, and preventing inflammation and edema. Theoretically, chewing Thera-Bite® wafers could break up binding friction that occurs with sliding mechanics, without causing the bond and attachment failure, seen commonly with regular gum chewing.

Currently, most orthodontists who utilize sliding mechanics attempt to minimize friction by using round wire of smaller dimension than the bracket slots. This is based on the belief that the use of round wire will eliminate the friction experienced with rectangular arch wire in preadjusted brackets. The use of smaller dimension round wire creates increased intra-bracket space which allows greater bracket slot angulation change before the edges of the bracket bind on the arch wire as a result of tooth tipping. However, greater angulation change increases the normal force of the bracket upon the arch wire, thereby increasing frictional resistance. Also, round wires of small dimension do not possess the vertical stiffness needed to control the side effects of sliding mechanics, often leading to anterior extrusion or tipping of the teeth adjacent to the teeth being moved. In addition, the vertical stiffness of arch wires prevents tooth tipping during sliding mechanics. Orthodontists who appreciate the pitfalls of undersized and flexible arch wires have elected to place stiff rectangular steel arch wires for sliding mechanics. Initially, enough time is allowed for all brackets to become passive with respect to second and third order bracket prescription, before any traction force is applied. This technique, allows reduction of friction due to second and third order discrepancies, without tooth tipping and loss of vertical control. With the use of larger rectangular arch wires, care must be taken to eliminate an arch wires with nicks or scratches. Any physical impediment to bracket movement on an arch wire will be magnified if it occupies the entire bracket slot. The current use of rectangular arch wires with increased flexibility permits earlier insertion of arch wires with greater dimension which may decrease the treatment time necessary to achieve passive fit of larger dimension rectangular stainless steel wire, which has the inherent stiffness required for sliding mechanics.

Many clinicians have resorted to using loosely tied stainless steel ligatures to reduce the normal force of ligation to facilitate tooth movement with sliding mechanics. However, decreasing the normal force of ligation decreases the rotational constraint present when using sliding mechanics. New bracket designs have been developed which selectively lift the normal force of ligation off the arch wire during sliding mechanics to facilitate tooth movement. These brackets do facilitate tooth movement, but with the sacrifice of rotational control (Ogata *et al.*,1996). Some clinicians believe that placing stainless steel ligation only upon an individual bracket wing will prevent tooth rotation and may decrease frictional resistance. Ligation of only the distal wing when using twin brackets for canine retraction is an example of this technique. This belief is theoretically incorrect as the ligature on the wing closest to the anchor segment is the part of the ligature which has the largest normal force applied to the arch wire, acting as the specific location for generation of frictional resistance (Bednar, *et al.*, 1993). Pollit (1996) identified that the direction of twist of stainless steel ligation may influence frictional resistance enough to create differential anchorage. Pollit explains that twisting stainless steel ligation from gingival to incisal decreases the normal force upon the arch wire leading to decreased frictional resistance.

Other orthodontists have devised creative uses for frictional resistance to control anchorage. The use of uprighting springs to decrease or increase friction by restricting or increasing bracket angulation change have also been used in an attempt to make friction work for the orthodontist. Alternative methods of ligation, such as twisting elastomeric ligation in a figure of eight, or using elastomers under self ligating brackets to form a brake, dramatically increase the normal force of ligation, thereby increasing frictional resistance of the anchor segment. Other clinicians have attempted to use ion implantation to reduce the frictional resistance of individual brackets or sections of arch wire to facilitate sliding mechanics. However, no *in vivo* evaluation of any of the above mentioned techniques has been completed to substantiate these claims.

It is evident that friction is a major component affecting the duration of orthodontic treatment. Identification of the factors influencing friction will facilitate the design of mechanics that will maximize the biological response to effective tractional forces, making orthodontic treatment more efficient. The friction testing apparatus used in this study may be adapted to test many of these asumptions in the *in vivo* environment, providing important needed information about factors that influence friction.

5.7 LIMITATIONS OF THE PROJECT

The design of this study permitted the evaluation of the influence of saliva, bracket width and chewing vibration on frictional resistance as discrete variables. No attempt was made to simulate vibration *in vitro.* The combination of saliva and vibration were not utilized to evaluate the influence of bracket width in the *in vitro* testing environment.

The design of the friction testing apparatus as well as the method used to solve for frictional resistance ignored any frictional resistance generated between the moment generating spring and the arch wire. This is a reasonable assumption because higher normal forces exist at the corners of the sliding tubes than at the contact point of the moment generating spring and utility arch wire. However, the contact of the moment generating spring to the utility arch wire creates an additional surface which could catch arch wire scratches or metal tags during frictional resistance testing. This omission was consistent for all testing environments and trials, and was undertaken to simplify calculations. Use of the exact same apparatus in both testing environments ensured that the effect of this omission uniformly affected all results. A single sample of saliva was used to investigate the effects of saliva *in vitro*, and variations in salivary composition were not investigated. In addition, no effort was made to control the amount of the force generated during mastication or eliminate variation in masticatory forces.

The same investigator designed, manufactured and completed all frictional resistance testing eliminating the possibility of a blind investigation, and allowing the possibility of subjective measurement bias.

CHAPTER 6

CONCLUSIONS

AND

RECOMMENDATIONS FOR FUTURE RESEARCH

CONCLUSIONS

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

1) Intraoral vibration decreased the *in vivo* static coefficient of friction as compared to *in vitro* results. The decrease of the static coefficient of friction from the *in vitro* to *in vivo* testing environment was similar for both sliding tube lengths.

2) Long and short tubes exhibited different coefficients of friction for all subjects tested with the effect of tube length being significantly different among all subjects. This could be a result of the influence of the differing quality and quantity of intraoral vibration and saliva in each subject.

3) Long tubes were found to have higher static coefficients of friction than short tubes for both *in vitro* and *in vivo* testing. This could be due to the different normal forces utilized for each length of tube which were selected to maximize the effect of intraoral vibration, as well as the different sensitivity to vertical arch wire deflection.

4) Saliva and intraoral vibration had the same influence on the frictional resistance of both long and short sliding tubes.

5) The static coefficient of friction for the long sliding tubes was sensitive to vertical arch wire deflection while, the static coefficient of friction for the short tubes was not.

6) Organic contaminants that may have entered the lumen of the long tubes did not influence the frictional resistance of the long tubes.

7) The presence of saliva increased the *in vitro* static coefficient friction for both long and short tubes. The increase in the coefficient of friction from dry to wet *in vitro* testing was greater for the long tubes than for the short tubes.

8) High normal forces generated from angulation change of the sliding tubes eliminated the reduction of the static coefficient of friction by intraoral vibration. In addition, arch wire surface characteristics such as nicks and scratches that act as impediments to sliding tube movement also eliminated the reduction of the static coefficient of friction by intraoral vibration.

9) Frictional resistance was not completely eliminated with intraoral vibration.

RECOMMENDATIONS FOR FUTURE RESEARCH

Based upon the results of this investigation, recommendations for future studies in frictional resistance in orthodontic mechanics should include;

1) Development of a technique to vary the normal force of ligation and measure its influence upon frictional resistance *in vivo*. The use of sections of spring steel tied to the vertical slots of orthodontic brackets is a possibility. In addition, investigate the significance of ligation material on frictional resistance *in vivo.*

2) Identification of a range of normal forces that will minimize frictional resistance *in vivo* with commonly used orthodontic appliances. This will enable us to gain a better understanding of what quantity and quality of tractional forces are required to permit efficient tooth movement.

3) Completion of similar experiments with sliding tubes of shorter length, more similar to commonly used orthodontic bracket widths. The 8 mm long sliding tubes used in this study were too sensitive to vertical deflections in the utility arch wire. These studies should include evaluation of the effect of *in vitro* vibration testing on the static coefficient of friction.

4) Completion of *in vivo* investigations comparing and contrasting frictional resistance with orthodontic brackets utilizing a standardized method of ligation with different materials.

5) Using analogous experimental methods to investigate the influence of auxiliaries that control second order changes upon frictional resistance *in vivo.* Utilizing contemporary power arms and uprighting springs may be a suggestion.

6) The development of a more sophisticated approach to measure the quantity, character and duration of forces transmitted to teeth with contemporary orthodontic appliances. This information, as well as the resultant tooth movement, will advance the understanding of the phenomenon of friction in orthodontic mechanics.

7) Using *in vivo* frictional resistance evaluation as testing procedure for new arch wires, brackets, and methods of ligation, to give more accurate insight into the clinical performance of these new orthodontic products.

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APPENDIX A

INFORMATION FOR SUBJECTS OF FRICTIONAL FORCE CLINICAL STUDY

Friction between braces and wires used to straighten teeth is thought to slow the movement of teeth and thus lengthen the time to complete treatment. The purpose of this study is to compare the friction between a wire and sliding tube in the mouth and in a laboratory to see if any differences exist. The principal investigator is qualified as a general dentist and is a graduate student in the graduate orthodontic program at the University of Manitoba.

The basic techniques and materials used for this study are common to orthodontic treatment. These techniques are non-invasive, and are not associated with any known harm to human tissues. To be accepted for this project, you must have braces on your upper teeth and do an excellent job at keeping your teeth clean.

Patients for the frictional force study will be required to have an impression made of their upper teeth to help the investigator make a special wire that will fit over top of regular braces and wire. The special wire will have a sliding tube with a spring attached to it. The special wire will be placed into a slot on the back braces and will be tied to front braces to keep it in place. This special wire will not effect the braces on the teeth or interfere with normal orthodontic treatment. Patients will be required to chew gum as the investigator takes measurements of how far the sliding tube moves. Photographs of the special wire in the patients mouth will be taken for records.

You will have at least one to two weeks to consider if you wish to participate in the study. If you participate in the study you will have, in addition to your regular orthodontic appointments: an initial appointment to explain the study and make an impression of your upper teeth, and two identical testing sessions approximately two weeks apart. These three appointments will require approximately 6 hours of your time.

The general health and well being of the patients participating should not be affected by any techniques applied in this study. Furthermore, immediate and obvious benefits to the participants as a result of taking part in this study are not expected. The risks associated with this study are identical to those in clinical practice and include: allergic reactions to the materials used, or to carrots and chewing gum; accidental swallowing or inhaling of materials; irritation of the gum tissue beside the special wire and/or sliding tubes, and slight jaw tiredness at the completion of the experiment. The special wire and tube design has been thoroughly tested and will be made so that all possible risks are minimized.

Each patient will be compensated $75 for participation in the study ($25/ 2 hour session) as remuneration for time, transportation, and parking. There is no obligation to participate and you may refuse to take part. You may also participate and then withdraw from the study at any time, without penalty, or compromise to ongoing orthodontic treatment. The information gathered from this study will be used solely for research purposes and the names of any participants will not be revealed.

APPENDIX A

INFORMED CONSENT FOR FRICTIONAL FORCE CLINICAL STUDY

I, ______________________________ (please print name), have agreed to participate in a study concerned with frictional forces, to be conducted by Dr. Mark Ziedenberg, Graduate Orthodontic Student. I have read the information sheet about the study and it has also been explained to me, by Dr. Ziedenberg. All questions regarding the frictional force study have been answered to my satisfaction. I understand that this study will involve the wearing of a special wire with a sliding tube while I perform ordinary chewing tasks.

I understand that the study requires 3 separate sessions, requiring approximately 6 hours of my time as follows: 1) an initial 1-2 hour session where information about the study will be given and an impression of my upper teeth will be taken, 2) a two hour second session 2 weeks later where a wire with a tube and spring will be place in my mouth and I will be asked to chew gum and chew and swallow pieces of raw carrot and 3) a third session approximately 2 weeks later which will have the same protocol as the second.

I understand that measurements of the tube movement as well as photographs of my mouth will be taken. I understand that I am under no obligation to participate, and that I can refuse to participate without compromise to my ongoing orthodontic treatment at the University of Manitoba.

I understand that all the materials used in this study are common to modern orthodontic treatment, and I understand the risks and benefits associated with their use. I understand that taking part in this study will in no way effect the progress of my orthodontic treatment.

I understand that there are no specific, personal benefits to be realized as a result of my participation in this study, but that the results of the research are expected to contribute to a better understanding of frictional forces and their role in orthodontic treatment. I understand that I will receive monetary compensation for taking part in the study. ($25 / session) The information from this study will become the property of the University of Manitoba, and may appear in scientific publications and presentations, but the names of the participants will be protected and will remain anonymous.

I have volunteered to take part in this study on my own, and I realize that I am able to withdraw from the study at any time, without penalty or compromise to my orthodontic treatment.

Signature of Participant: ______________________________

Signature of Parent/Guardian: ______________________________

Date: ______________________________

Signature of Witness: ______________________________

APPENDIX A

ETHICS COMMITTEE APPROVAL FORM

The University of Manitoba
Faculty of Dentistry

COMMITTEE ON RESEARCH INVOLVING HUMAN SUBJECTS

Date:	June 27, 1995
Committee Reference	EC33/95P
Names of investigators:	Drs. Mark Ziedenberg, J.C. Nickel, K.R. McLachlan, D. Singer
Your project entitled:	In Vivo Testing of Friction on Orthodontic Archwire,

has been approved by the Committee.

PLEASE NOTE

Any significant changes in the approved protocol must be reported to the Chair of the committee for the Committee's consideration and decision, prior to the implementation of the changes in the protocol.

Yours sincerely,

Colin Dawes

Colin Dawes B.Sc., B.D.S., Ph.D.
Chair, Committee on Research
Involving Human Subjects

APPENDIX B

A series of preliminary studies were completed to design and test different retraction mechanisms that would provide reproducible forces useful to study the phenomenon of friction. Initial *in vitro* experiments found that closed coil Nitinol springs were a suitable tool for the purpose of retraction of a tube over an orthodontic arch wire. By varying the length of these springs, their force deflection characteristics were made compatible with the dimensions and forces required for the friction testing assembly. Multiple *in vitro* force-deflection calibration experiments produced consistent results for similar lengths of closed coil Nitinol springs.

The Nitinol springs that were chosen for this investigation had a lumen diameter of 0.030 inches, and were constructed of 0.0075 inch Nitinol wire. The Nitinol springs were 10mm long at rest, measured from eyelet to eyelet.

Five 10mm Nitinol closed coil springs were tested for consistency of their force-deflection characteristics. Each Nitinol spring was fixed to a retort stand through a superior eyelet. A specimen dish was attached to each spring's inferior eyelet so that incrementally larger weights could be added. As each weight was added, a Vernier microscope was used to measure the change in length of each Nitinol spring under defined load. Springs were loaded and unloaded in order to examine if any hysteresis occurred in the force-deflection relationship. A linear regression of five force deflection curves was completed with Lotus 1-2-3, Release 5®. This linear regression yielded the following data;

Linear Regression Data	
Slope	7
Intercept	3
r	0.99
R^2	0.98
N	72
T"	54.5
df	70

Table B.1 Results of Linear Regression of Force Deflection Curves

The results of the linear regression produced an expression for the force-deflection characteristics of the Nitinol closed coil springs;

Force in grams, at deflection = 7 (deflection in mm) + 3

This expression permitted the calculation of force created by the Nitinol closed coil springs at specific elongations. The force produced by the Nitinol closed coil springs acted as the traction force in the friction testing assembly.

APPENDIX C

Subject	Trial # mode	Normal Force (gm)	Short tube μ_s	% Change	Normal Force (gm)	Long tube μ_s	% Change
AB1	1 / A	109	0.15		108	0.30	
	1 / B	109	0.09	-41	108	0.24	-19
AB2	2 / A	109	0.21		108	0.30	
	2 / B	109	0.09	-59	108	0.21	-29
SG1	1 / A	113	0.21		95	0.31	
	1 / B	113	0.15	-27	95	0.23	-27
SG2	2 / A	113	0.29		95	0.24	
	2 / B	113	0.20	-32	95	0.20	-16
JW1	1 / A	124	0.18		90	0.25	
	1 / B	124	0.10	-48	90	0.14	-43
JW2	2 / A	124	0.18		90	0.29	
	2 / B	124	0.10	-43	90	0.14	-51
NG1	1 / A	106	0.26		73	0.26	
	1 / B	106	0.15	-44	73	0.17	-34
NG2	2 / A	106	0.30		73	0.26	
	2 / B	106	0.12	-61	73	0.21	-19

Table C.1 **Data from *In Vitro* and *In Vivo* Trials**

Legend **A : *In Vitro* trials**
B : *In Vivo* trials

Subject Name	Trial #/ mode	Normal Force (gm)	Short tube μ_s	% Change	Normal Force (gm)	Long tube μ_s	% Change
TD1	1 / A	153	0.19		85	0.23	
	2 / B	153	0.17	-11	85	0.16	-31
TD2	2 / A	153	0.23		85	0.27	
	2 / B	153	0.17	-26	85	0.15	-43
KC1	1 / A	124	0.16		64	0.25	
	1 / B	124	0.10	-37	64	0.17	-34
KC2	2 / A	124	0.29		64	0.30	
	2 / B	124	0.10	-64	64	0.17	-45
NH1	1 / A	117	0.19		72	0.22	
	1 / B	117	0.11	-41	72	0.09	-61
NH2	2 / A	117	0.27		72	0.19	
	2 / B	117	0.11	-57	72	0.09	-61
CD1	1 / A	109	0.15		69	0.28	
	1 / B	109	0.09	-41	69	0.23	-19
CD2	2 / A	109	0.15		69	0.28	
	2 / B	109	0.10	-34	69	0.20	-28
VB1	1 / A	130	0.22		67	0.29	
	1 / B	130	0.06	-73	67	0.11	-62
VB2	2 / A	130	0.22		67	0.29	
	2 / B	130	0.07	-74	67	0.17	-40
KW1	1 / A	104	0.19		56	0.23	
	1 / B	104	0.14	-23	56	0.14	-38
KW2	2 / A	104	0.19		56	0.23	
	2 / B	104	0.12	-37	56	0.11	-51

Table C.1b Data from *In Vitro* and *In Vivo* Trials