CHAPTER II

LITERATURE REVIEW

Orthodontic tooth movement requires the application of a force-delivering system, capable of eliciting the desired response of the individual dental units. The ideal force-delivering system should have attempted to meet the following criteria: 1) provide optimal tooth moving forces to elicit the desired effects, 2) be comfortable and hygienic to the patient, 3) require minimal operator manipulation and chair time, 4) require minimal patient cooperation, and 5) be economical. Numerous force-delivering systems have been proposed to fulfill these criteria. These included intra-arch and interarch force systems, coil springs, latex elastics, elastic threads, sectional arch wire auxiliaries and elastomeric auxiliaries. Although all of these systems had their inherent shortcomings, many researches suggested that the elastomeric auxiliaries had the greatest potential to fulfill these requirements (Sonis et al., 1986).

ORTHODONTIC ELASTOMERIC CHAINS

The most important property of an elastomeric material used in orthodontics was its ability to exert a useful force over a period of clinical use. This avoided the need for too frequent recalls and obviated the need for the patient to replace elastics. The decreasing in force after a period of time in the mouth might be the result from a reduction in extension due to tooth movement, and also from the degradation of elastic material due to mechanical stress and exposure to the oral environment (Rock, 1986).

The synthetic orthodontic elastomeric chains are amorphous polymers made from polyurethane materials, but their exact composition is proprietary information. They have been used by orthodontists since the 1960s. These polyurethane materials have largely replaced latex elastics for intra-arch tooth movement. An improvement in the material itself or in the manufacturing process was needed to provide the orthodontist with a more consistent means of delivering force over an extended period of time. In 1981, Ormco Corporation changed the polyurethane resin used in the manufacture of their power products. Rocky Mountain Orthodontics entered the elastomeric chain market in 1982 with what they claimed to be a " tough and tasteless elastic material...that delivers a uniform continuous force " (Rocky Mountain® Orthodontics Catalog, 1992). TP Laboratories markets claimed that Elast-O Chain, which was eleminated the solid bar between the rings, provided light continuous traction force (TP® Orthodontics Product Catalog, 1995). This supposedly results in an increase in resiliency and provides lighter forces over a longer period of time. American Orthodontic Corporation produced plastic chains, which were latex free and hypo-allergenic FDA approved polyurethane with consistent force available (American Orthodontic Product Catalog, 1998).

The manufacturing processes of orthodontic elastomeric chains were extrusion process (Leone®, TP® laboratories and ORMCO®), and injection-molded process (Unitek®). Hershey and Reynolds (1975) noted that modules manufactured by a die-cut stamping process were more consistent in the amount of force produced as compared to injection-molded materials.

Recently, the variety of colored elastomeric chains have been introduced and have achieved notable popularity. In general, the colored chains behaved similarly to the grey chain in the same manufacture, but in some cases the coloring materials affected to the force delivery properties (Baty et al., 1994). Similary to fluoride-releasing elastomers, a commercial company claimed about its capable of releasing low concentrations of fluoride that could inhibit demineralization and promote remineralization. The force degradation from fluoride-releasing elastomeric chains exhibited a very rapid loss over the first week (Storie et al, 1994).

Furthermore, numerous investigators had studied the remaining force and force degradation of elastomeric chains, and also several factors that influenced on them.

1. THE ELASTIC MATERIALS

Andreasen and Bishara (1970) compared the behavior of elastic and Alastik (plastic elastic) materials and advocated that under all environmental conditions most of the force loss occured within 24 hours. Moreover, the Alastiks demonstrated significant decay rates of forces, losing 75 percent of their initial forces over the first day and only 40 percent for the elastics. The greatest percent force decay rate occurred during the first hour. After the initial 24 hours, the remaining force of Alastiks and elastics were reasonably constant throughout the three-week period. Furthermore, the longer the material was left in the mouth, the rate of force decay was less. They also suggested stretching Alastik chains three to four times intended load to compensate for subsequent force decay.

Bishara and Andreasen (1970) evaluated the changing of force of rubber elastics and plastic Alastiks during a three-week period. They concluded that most of the force decay occured within the first day and continued a slow rate during the rest of the three-week period. The rate of force decay was greater in the first few hours. The only difference between elastics and Alastiks was that initial drop of force was much greater in plastic Alastiks than in rubber elastics. Thus, a relatively constant force was observed in both materials after the first day, but there was relatively more force remaining in the rubber elastics than in the plastic Alastiks after three weeks. Both materials underwent permanent deformation in shape, but there was more deformation with plastic Alastiks. The greater the distance the plastic Alastiks were stretched, the more deformation.

2. FORCE DECAY RATE AND PATTERN

Hershey and Reynolds (1975) studied the behavior of plastic modules when used as orthodontic space-closing appliances and compared a variety of elastomeric auxiliaries from several manufacturers. By means of a framework that simulated tooth movement of 0.25 and 0.50 millimeters per week or none. They demonstrated that all tested modules produced similar decay curves. The rapid loss of force was slightly more than 50 percent of initial force in the first 24 hours. The modules retained an average of 40 percent of their initial force after four weeks at the zero rate of closure and almost exactly the same after six weeks. The rate of force loss increased with simulated tooth movement. The remaining forces after one week were 1/3 and 1/4 of the initial forces with simulated tooth movement 0.25 and 0.50 millimeters per week respectively. They recommended that the plastic modules produced effective tooth movement force throughout a four-week to six-week period.

Wong (1976) suggested that the force decay of synthetic elastomers, which were stretched over a specific length and time period, exhibited a great loss in force. This loss could be as great as 73 percent during the first day. The decay of force continued at a slower rate during the rest of the 21-day period. The force decay under constant force application to latex, elastic, polymer chains showed that the greatest amount of force decay occured during the first three hours in the 37°C water bath. The force remained relatively the same throughout the rest of the test period.

3. INITIAL FORCE MAGNITUDE

Hershey and Reynold (1975) found no significant difference in the decay curves for the sample groups exhibiting different initial forces. Furthermore, they stated that the percentage of force loss was similar for modules stretched to high and low initial forces. The decay characteristics were not related to magnitude of the initial force. A difference in initial load had no effect on the degree of stress relaxation.

Young and Sandrik (1979) investigated the exerted force from Unitek CK and C2 grey Alastiks. They found that the remaining force after 24 hours of CK and C2 Alastiks was 43.6 percent and 56.4 percent respectively. The C2 Alastiks with 180 grams initial load had significantly greater force decay than with 90 grams initial load. Furthermore, at 99 percent confidence level C2 Alastiks showed a significantly higher force remaining compared with the CK group in the same 90-gram initial load.

De Govana et al. (1985) compared the percentage of remaining force from short and long modules of elastomeric auxiliaries from several manufactures with constant and decreasing extension length, 0.50 millimeters per week. They found that decreasing extension which was simulated tooth movement, significantly decreased the mean percentage of remaining force in both short and long modules. Both the constant length and the length with simulated tooth movement, short modules exerted higher initial force values and generally retained a higher mean percentage of remaining force for each material than the long modules.

Lu et al. (1993) determined the initial force and the force decay for a six-week stretching period, at three variously stretched dimension with simulated tooth movement, 0.5 millimeters per week, in 37°C distilled water bath and pH 7.0 condition. They concluded that the greater the initial force, the greater the force decay, and the greater the submerged time, the more the force decay of all three stretched lengths.

4. OTHER ELASTOMERIC PROPERTY RELATED FACTORS

Other factors such as rate of extension, prestretching mechanic, and load-extension behavior influenced on elastomeric properties of elastomeric chains.

4.1 Rate of Extension

Kovatch et al. (1976) assessed the response to the loading and the ultimate strength of the modules. He found that load-extension curve of elastomeric behavior depended on the rate of extension as well as the amount of extension. The modules tested at the faster speeds had higher moduli and were stronger, but as the extension continued, the trend was reversed. At breakage the more slowly stretched moduled were the strongest force. The initial load decay was more rapidly for both faster extension and higher load.

4.2 Prestretching Mechanic

Many investigators studied the effect of prestretching on force degradation of plastic module. Brooks and Hershey (1976) concluded that the prestretched module had much greater remaining force and could partially have decreased the unwanted reduction in force.

Wong (1976) suggested that the synthetic elastomers should be prestretched before being placed in the mouth. The elastomeric materials needed to be prestretched one third of their length to prestress the molecular polymer chain. This procedure increased the strength of the material.

Brantley et al. (1979) studied the effect of prestretching on force degradation of plastic modules which were maintained at the same fixed stretched distance, approximately 100 percent extension, during prestretching and force degradation testing. The results showed that prestretching modules had nearly constant forces when tested immediately after prestretching.

Young and Sandrik (1979) investigated the influence of prestretching on stress relaxation of two different orthodontic elastic polymers in different prestretching distances, in de-ionized water at 37°C condition with constant initial load and constant dimension over the 24 hours force testing period. The results showed that prestretching decreased the rapid force loss of the elastic polymer in CK Alastiks, but not in C2 Alastiks. They stated that the difference between the behavior of these two materials might be due to the difference in shape and/or polymer composition.

4.3 Load-Extension Behaviors of Elastomeric Chains

Wong (1976) indicated that if the elastomeric materials were overstretched, a slow set occured and they went back to the original state in time. If the material was overstretched to near breaking point over and over again and remained fixed in its extension, permanent plastic deformation occured. The extent of resiliency and plastic deformation also depended on how fast and how long the material was stretched. The modulus of elasticity of both latex and synthetic elastomers were much lower after immersion in the 37°C water bath than in dry condition.

Rock et al. (1985) reported the dimensions and force-extension characteristics of 13 commercially orthodontic elastomeric chains. The relationship between force and extension was nonlinear over the range of investigation. The initial part of this curve was linear, until a transitional point at which the curve flattened so that continued extension produced a lower increase in force. Beyond this part of the trace the force generated per unit of extension increased again although the final part of the curve was less steep than the initial part. The shape of the force-extension curve varied for each material. At the transitional point the force exerted by the great majority of specimens was between 2 newtons to 3 newtons and 100 percent extension of original length of elastomeric chain generally exerted 4 newtons to 5 newtons. For small extension, the stiffness showed an inverse relationship to the number of loops. Furthermore, he suggested that for practical proposes an extension of between 50 percent and 75 percent of the initial length would produce an appropriate orthodontic force.

5. THE ENVIRONMENTAL FACTORS

There are various studies on the effects of several environmental conditions, such as temperature and testing media, on stress relaxation of elastomeric chains. Some researchers reported that the stress relaxation behavior was different when the elastomeric chains exposed to different conditions.

5.1 Temperature

Brooks and Hershey (1976) determined the effect of heat on the force-decay curves of plastic orthodontic modules. The results showed that when the plastic modules were subjected to heat, the force that exerted by stretched plastic modules were substantially reduced.

De Genova et al. (1985) evaluated how thermal cycling affected on the force decay patterns of three elastomeric products. The modules were tested either in conditions of constant temperature (37°C) or in a thermal-cycled environment (15°C to 45°C). They found that the samples subjected to thermal cycling retained a significant higher percentage of remaining force than the samples held at a constant temperature of 37°C. Thermal cycling, which was simulated oral conditions, favored the production of higher force values.

5.2 Testing Media

Young and Sandrik (1979) investigated the Alastik C2 chains in 100 percent humidity and in de-ionized water at 37°C to determine their environmetal effects on the remaining force of specimens. The t-test showed no statistically significant difference between both groups.

von Fraunhofer et al. (1992) examined the effects of artificial saliva and topical fluoride treatment on the elastic properties and load relaxation of three brands of orthodontic elastomeric chains. This study showed that exposure to liquid media affected the elastic properties of the elastomeric chains and increased distractions required to deliver the same amount of force in comparison with the control, in the air. In the load relaxation test, all of elastomeric chains showed a significant loss of force delivery over the test and greater load relaxation occured in distilled water than in air.

Baty et al. (1994) studied a force-delivery to determine the distraction required to generate a force of elastomeric chains, which were immersed in distill water and artificial saliva. The results showed that the distraction necessary to deliver a force increased for all chains after the first 4 hours to 24 hours exposure to distilled water and artificial saliva.

Taloumis (1997) evaluated force decay, permanent deformation and the relationship between ligature dimension and force of elastomeric ligatures. He concluded that dimensional changes due solely to moisture and heat had a pronounced effect on force decay and permanent deformation. The unstretched ligatures absorbed moisture, and they had dimensional changes of wall thickness, inside diameter and outside diameter.

Most of the studies concerning the force degradation of elastomeric elements have been performed *in vitro*. During clinical used of such elements; however, factors that were not simulated in the *in vitro* studies add to the influence on the elastic element. Kuster (1986) noticed that factors might be chemical from saliva, food or oral hygiene products, thermal due to the ingestion of hot and cold foods, and mechanical due to mastication and oral hygiene procedures. These factors probably changed the properties of the elastomeric components in a way that could not be simulated *in vitro* experiments.

Ash and Nikolai (1978) compared the relaxation patterns in dry air 37°C, in water 37°C and in oral environment of Unitek CK and K1, injection-molded polyurethane based. The results showed that the relaxation pattern in the three environments were similar in form. Force degradation was proved to be more rapid in water and oral environments than in air.

Howard and Nikolai (1979) compared Nylon-covered latex and extruded polymeric threads in air at room temperature (at or near 24°C) and *in vivo*. They concluded that simulations of oral cavity conditions in laboratory studies had not accurately predicted relaxation characteristics of elastic threads *in vivo*. Besides environmental differences, mastication and tooth brushing superimposed incremental loading patterns upon the already stretched threads. Relaxation in air was apparently greater than *in vivo* over the first several days following initial loading.

Kuster et al. (1986) tested Unitek and Ormco manufactures of elastic chain for force development in laboratory, in air at room temperature, and intraoral experiments. They found that the two brands behaved similarly. The decline of the force value during intra-oral use was greater than in the laboratory test.

Rock et al. (1986) considered the effect of one month of clinical use upon the mechanical properties of the elastomeric chains. After four weeks in the mouth the stiffness of each elastomer had decreased and the force exerted by each chain at the final extension was also considerably reduced from the initial value and the amount of tooth movement was variable in distance.

In brief, several factors influenced on elastic properties of elastomeric chains. Since some factors can not be control, so the clinician should carefully select and use proper type of elastomeric chain to produce the optimal force for the best result and the most effectiveness for tooth movement.

FORCE AND TOOTH MOVEMENT

Throughout the development of the optimal force theory, various ideal force magnitudes have been suggested, in some cases for apparently identical movements of similar teeth. Several investigators concluded that perhaps no optimum force values exist, because of the differences in metabolic responses among individuals, and that further development of the theory may be unwarranted.

Storey and Smith (1952) concluded that there was an optimal range of force values that produced a maximum rate of distal movement of the canines, but it did not produce any discernible movement of the molar anchor. They used spring to supply the tooth-moving forces and found that the optimal force range for moving the canine distally extended from 150 grams to 200 grams. By increasing the force level above this optimal range, the rate of distal movement of the canine was decreased and finally approached zero. Moreover, with increase of force from the springs, appreciable movement of the molar anchor unit occurred with the high range of force values; 300 grams to 500 grams.

Reitan (1957) suggested that in a continuous tooth movement of individual teeth, it was practical to apply an initial force as light as around 25 grams in adults and around 40 grams in younger patients. After this first stage, the force applied for further tooth movement might be increased. The appropriate amount of force to be applied in the final stage of a continuous bodily tooth movement of a single tooth might be varied according to the root length. For upper canines, it might be necessary to apply 150 grams to 250 grams, and occasionally slightly more during the final stage of closure; but for lower canines were range 100 grams to 200 grams, and less force for premolars were required.

Burstone and Groves (1961) investigated the relationship between force magnitude and rate of tooth movement, and found that the optimal force produced the optimal rates of tooth movement and increasing the magnitude of force did not produce any increase in the rate of tooth movement. A positive correlative was observed between force magnitude and pain and tooth mobility.

Hixon et al (1969) found no data to support the theory of the optimal force applied to canine retraction for a clinically useful period of time. Their latter study in 1970 about the relationship between force and tooth movement showed large differences among patients with regard to root area, time of beginning tooth movement and rate of tooth movement. The magnitude of the variation in each of this factors appeared to be far more important than differences in magnitude of force (above 100 grams).

Gianelly and Goldman (1971) concluded that the optimal force for moving a tooth was one which produced the desired tooth movement rapidly with minimum tissue damage and clinical discomfort. Ideally, the force should not disturb the vitality of the periodontal ligament and should elicit efficient bone remodeling. Direct bone resorption should be occured and the tooth should be more relatively smoothly in an uninteruppted manner.

Quantitatively, several experiments determined " optimum " forces for small tooth e.g. incisor and large tooth e.g. canine were shown in Table 2.1.

Table 2.1 Optimum force for tooth movement (Modified from Gianelly and Goldman, 1971)

Type of tooth movement	Optimum force
a. tipping	small tooth 20-30 gm.
	large tooth 50-75 gm.
b. controlled root movement	small tooth 50 gm.
(root torque)	large tooth 120-150 gm.
c. bodily movement	small tooth 40-50 gm.
	large tooth 150 gm.
d. extrusion	25-30 gm.
e. intrusion	15-50 gm.

After the initial movement, the force probably could be increased because the resorptive process was active and the ligament space might be wider. For example, the force for bodily movement of large teeth such as canines could probably be increased to over 150 grams. The force determinations were usually done on animal-experiment. However, the tooth size in most instances was comparable to humans and the data were therefore applicable in a general sense.

Boester and Johnston (1974) studied canine retraction by sectional arch and found that a light force, 2 ounces (55 grams), produced significantly less movement than 5, 8 and 11 ounces (140, 225 and 310 grams respectively) of applied force. In addition, there was no significant difference among the 5, 8 and 11 ounces, which produced about the same amount of tooth movement.

Nikolai (1975) studied amount of the optimum force applied to canine retraction by canine-retraction appliance. For a particular patient, 60 grams was a suitable distal driving force for maxillary canine tipping, then 210 grams should be proper for bodily movement. The optimum force range extended into that of heavy forces was not clear, but indications were that more rapid and/or greater displacements of teeth may occurred in the third stage of tooth movement with heavier force, which counterbalanced the longer second stage.

Andresen and Zwanziger (1980) evaluated the clinical differential force concept as applied to the edgewise bracket. They suggested that the factor of friction should be mentioned relative in sliding an edgewise bracket over an archwire and varying degrees of friction impeded tooth movement. When using the Siamese edgewise bracket with continuous archwires, it did not always follow that only the canine moved backward, but the buccal segment or anchor unit did not move after 100 to 150 grams of force application. Likewise, with 400 to 500 grams of applied force, it did not always that only the buccal segment or anchor unit moved forward, but the canine did not move.

Quinn and Yoshikawa (1985) estimated that a maxillary efficient canine retraction force from the clinical data was between 100 and 200 grams. This was yield mean compressive stresses for the average cuspid root (assuming one half of the root surface to be under compressive stress) of approximately 70 to 140 grams per square centimeter.

Lu et al (1993) reported that the initial force for moving canines were neither larger than 400 grams nor less than 150 grams. The initial force which were greater than 400 grams made the patient more discomfort at the initial period and the heavy force might cause severe root resorption for retracting a canine into the space of an extracted first premolar.

Frank and Nikolai (1980) suggested that the presence of friction in fixed orthodontic appliances was undeniable and was recognized by the clinicians. Friction impeded displacement in the mechanics of canine retraction along a continuous arch wire into the site of a premolar extraction. All practitioners should be awared about friction, so that the activated forces from the appliances should be greater than those required to move teeth.

In conclusion, the optimal force for canine retraction is still in controversy. It was effected by several factors such as individual biology, treatment techniques (appliances), treatment mechanics and also friction resistances. Therefore, clinician should adjust the applied force for the best results and the most effectiveness for tooth movement.

BASIS OF SYNTHETIC POLYMERS

Polymer is a large molecule of carbon-hydrogen chain elements that tend to gather and behave in basic groups. The hydrogen in a basic group could be substituted by some other element or group to form a new group. The basic small molecules are called " monomers ", and when they polymerize into very large molecules, they form " polymers " (Harper, 1975; Richards et al, 1967).

The creation of chemical linkages between monomers to form polymers are different patterns. In some cases the repeated attachment from its links to be a long chain is linear. Most linear polymers can be made to soften and take on new shapes by heat and pressure. In other cases the chains are branched or cross-linked. The individual molecules in branched chain are still discrete. The cross-linked occurs by the linking of atoms between or across two linear polymers and produce a three-dimensional network polymer that is stable to heat and can not be made to flow or melt (Harper, 1975; Billmeyer, 1984).

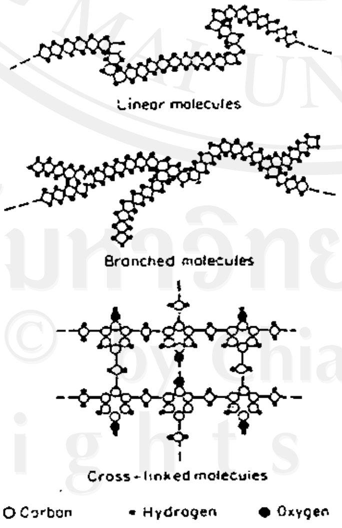

Figure 2.1 Some possible molecular structures in polymers (Harper, 1975).

Molecular Forces and Chemical Bonding in Polymers

The predominant bond in polymers is covalent bond, which is formed when one or more pairs of valence electrons are shared between two atoms, resulting in stable electronic shells. Even when all the primary valences within covalent molecules are saturated, there are still forces acting between the molecules. These generally known as " secondary valence " or " intermolecular forces ", or " van der Waals forces " (Billmeyer, 1984).

The interchain forces of non cross-linked polymers such as hydrogen bonding can often be strong and specific, but the interactions between chains in many polymers are only van der Waals attraction. Though polymers do not generally crystallize, they often have regions in which polymer molecules are oriented to one another much as they would be in a crystal. Amorphous regions contain between polymer molecules with more random orientation. The balance between crystallites and amorphous regions has a significant influence on polymer properties. When the forces between chains in an amorphous polymer are weak, the chains are free to slip by one another. The polymer will have low tensile strength and be subject to plastic flow (Richards et al., 1967).

Unoriented crystalline polymers are characterized by many randomly oriented crystallite regions. Since these regions melt rather sharply, such polymers have a well-defined melting temperature above which they are amorphous and undergo plastic flow that permits them to be easily molded.

Oriented crystalline polymers are characterized by oriented of crystalline regions with respect to one another with very useful properties, in particular, a much higher tensile strength than the unoriented polymer. When plastic flow and elongation occur, the crystalline regions are drawn together and oriented along the direction of stress.

There are numerous minor classifications for polymers, depending upon how one wishes to categorize them. However, nearly all can be placed in one of two major classifications, which were categorized by organized pattern.

These two major plastic-material classes are thermosetting materials and thermoplastic materials, as shown in Figure 2.2.

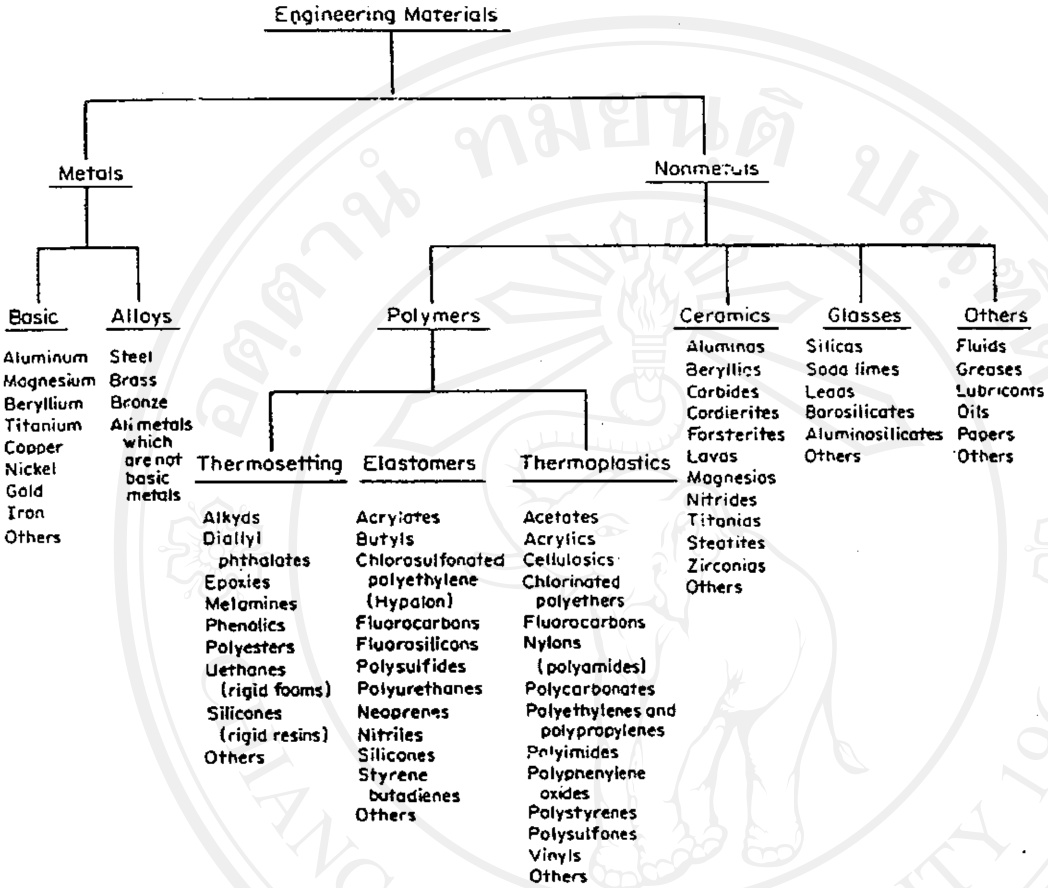

Figure 2.2 Classification of engineering materials, showing polymer classes (Harper, 1975)

Thermosetting Plastics

Thermosetting plastics or thermosets are cured, set, or hardened into a permanent shape. This curing is an irreversible cross-linking reaction under heat, so a three-dimensional rigid cross-linked molecular structure is made. However, some thermosetting materials initiated or completed curing at room temperature. Although the cured part can be softened by heat, it can not be remelted or restored to the flowable state that existed before the plastic resin was cured.

Thermoplastics

Thermoplastics do not cured or set under heat as thermosets. Thermoplastics are formed by a flexible molecular structure, either linear or branched. They merely soften to a flowable state when heated, and they can be forced or transferred from a heated cavity into a cool mold by pressure. They can be remelted and rehardened by cooling many times. Thermal aging, which brought about by repeated exposure to the high temperatures for melting, causes eventual degradation of the material. Thus, the number of reheat cycles is limited.

Although Figure 2.2 shows elastomers separately as a separate application group. They are either thermoplastic or thermosetting, depending on their chemical nature (Harper, 1975).

Elastomers

Elastomers includes the complete spectrum of elastic or rubberlike polymers which are sometimes randomly refered to as rubbers, synthetic rubbers, or elastomers. More properly, however, rubber is a natural material and synthetic rubbers are polymers which have been synthesized to reproduce consistently the best properties of natural rubber (Harper, 1975).

Elastomers are intermediate in character between amorphous and crystalline polymers. When a polymer is largely amorphous with only weak forces between chains, elastic properties result. However, the presence of some crystallite regions is necessary to prevent plastic flow. A judicious balancing of these two regions produces an elastomer. On elongation, the amorphous regions are stretched, so the chains become nearly parallel and producing a semicrystalline state. However, the forces between polymer chains of an elastomer are too weak to maintain the semicrystalline state. When tension is released, the polymer snap back to its original form, largely amorphous. Cross-linking is also necessary to give elastomers structural stability, and non-crosslinked elastomers usually show very pronounced cold flow and easily lose their shape (Richard et al., 1967).

Elastomers are characterized by large deformability, rigidity, large energy-storage capacity, non-linear stress-strain curves, high hysteresis (the lagging of strain behigh stress during deformation, ASTM D 1566), large variation with stiffness, temperature and rate of unloading, and compressibility of the same order of magnitude as most liquids. Certain of the elastomeric materials possess additional useful characteristics to a relative degree, such as corrosive chemical resistance, oil resistance, ozone resistance, temperature resistance, and other environmental conditions (Harper, 1975).

Processing Methods of Polymers

There are many plastic-fabrication processes, which produce a wide variety of plastics. The fabrication process and the tooling determine the forms or shapes which are produced (Harper, 1975; Billmeyer, 1984).

A. Molding process are those in which a finely divided plastic is forced by the application of heat and pressure to flow into, fill and conform to the shape of a cavity (mold).

Most thermoplastic materials are molded by injection molding process. The polymer is preheated in a cylindrical chamber to a temperature at which it will flow and then is forced into a relatively cold, closed mold cavity with high pressures by a reciprocating screw that serves the dual purposes of providing the molten polymer mass and forcing into the mold.

B. Extrusion process consists, basically, of forcing heated, melted plastic continuously through a die which has an opening shaped to produce a desired finished cross section. Normally, it is used for processing thermoplastic materials, but it can be used for processing thermosetting materials. The main application of extrusion is the production of continuous lengths of useful forms and cross sections. After the melted plastic is extruded through the die, the extruded material is cool hardened by air or water.

POLYURETHANE ELASTOMER

The polyurethanes are not direct polymers of urethane, but are derived through a process of reactions of either polyethers or polyesters with di- or polyisocyanites to produce a complex structure of urethane linkage. Thus, the major component of a complex polymer is a prepolymer, composed of a linear polyester or polyether that has been extended several times in structural chain length by coupling through urethane linkage (Roff and Scott, 1971).

Many chemical combinations lead to many physical variants. The whole macromolecule consists essentially of hard segments (polyurethane) and soft, flexible segments (polyester or polyether), Figure 2.3. The flexible segments can be readily extended by stretching, giving extensibility and flexibility. The hard segments are hydrogen bonded to one another, which lead to the stability and strength of polyurethane structures (Morton-Jones and Ellis, 1986).

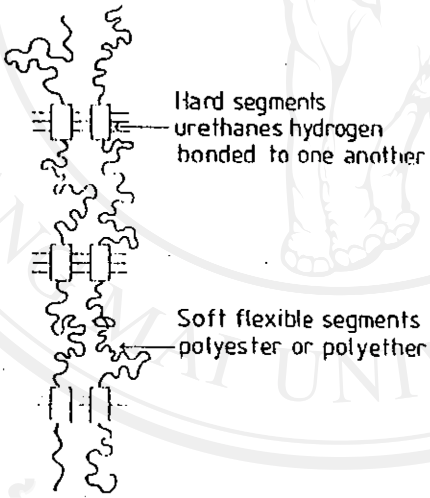

Figure 2.3 Polyurethane block structure (Morton-Jones and Ellis, 1986)

Load-Deformation Behavior of Orthodontic Elastomeric Module

An elastomer is a rubber or rubberlike, often synthetic, polymeric material. The nature of the bonding of atoms and molecules, and the imperfections, make the material inherently more flexible than a same size and shape crystalline solid. The load-deformation curve in tension of polymeric material may present by two-part or three-part diagram. A typical two-part diagram is presented in Figure 2.4, the solid curve represents loading until rupture. The linear portion of the plot reflects principally in the uncoiling of long molecular chains. A typical three-part diagram shows non-linear initial portion because the secondary bonds must be broken before the major uncoiling of chain molecules can begin. This part of the curve exhibits decreasing slope with increasing applied force. After the uncoiling is completed, material stiffness increases as primary bonds begin to be stretched. The slope of the curve grows and rupture eventually occurs with continued increasing of the load. Accurate determination of the coordinates of the elasic limit of the module is difficult because of the time-dependent nature of the mechanical behavior of the material. Most elastomers are relatively brittle, however, their elastic limits are " high " on the load-deformation curves, beyond the linear initial or middle portions of these plots. Therefore, inelastic action is unexpected in clinical use, as the name " elastomer " suggests.

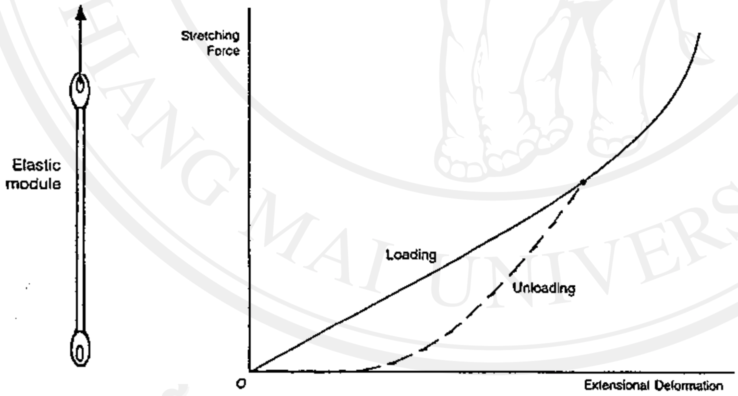

Figure 2.4 The force-deformation diagram for the stretched elastomeric module in direct tension (Modified from Nikolai, 1985)

The dash curve in Figure 2.4 is representative of rather rapid unloading of the module. Unloaded from a point below its elastic limit, some deformation remains immediately after removal of the activating force, but over time, full recovery of the preload, passive configuration takes place as secondary bonds are re-established (dashed line). Besides the amount of elongation, the force from orthodontic module depends upon the time passed while under load. Hence, stiffness, as the slope of the load-deformation curve, can only be approximated because it is also changing with time (Nikolai, 1985).

When a force is applied to the polymer chains, they undergo both chain slippage and chain stretching and uncoiling. However, the chain slippage is eventually predominating and leads to viscous behavior that is slow and irreversible. While chain stretching and uncoiling leads to elastic behavior that is quick and reversible (De Gonova et al, 1985).

Liu et al. (1993) reported that both natural latex rubber and polyurethane elastics exhibit viscoelastic rather than perfectly elastic behavior.

Mechanical and Structural Properties

Noncrystalline materials, such as the polymers seldom exhibit any useful level of ductility and often do not obey Hooke's Law. As a result, there is little distinction between elastic strength and ultimate strength. Although not exactly correct, the elastic limit may be taken as the point where the slope of the stress-strain curve begins to increase substantially toward rupture.

When the molecular materials, particularly certain polymers, are subjected to a moderate tensile load, and holded that constant load while monitoring deformation. A continuing elongation, creep, will be observed over a period of hours even in room or oral temperature, and a strain-time curve may be generated similar to the plot of Figure 2.5. If the temperature or time period is sufficiently high or long, the linear portion of the curve will give way to a nonlinear increase in strain followed by rupture, even though the stress level may be substantially below the ultimate strength of the material as gained from a static, tension test at room temperature. A material that creeps will also " relax ". Relaxation is the decrease with time of load carried under conditions of constant strain (Nikolai, 1985).

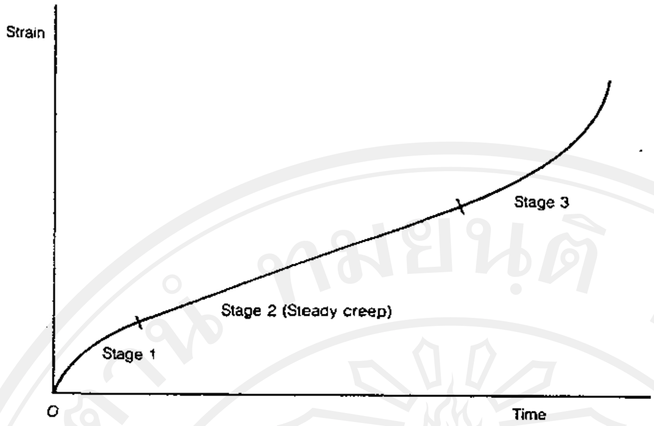

Figure 2.5 A strain-time plot illustrating the three stages of creep (Nikolai, 1985)

The exact nature of the tensile response of a polymeric material depends upon the chemical structure of the polymer, conditions of sample preparation, molecular-weight distribution, crystallinity, and the extent of any crosslinking or branching. The mechanical response also depends in a very significant way on temperature and the rate of deformation. Any amorphous polymer can exhibit the entire range of tensile behavior, from brittle to rubbery response, by increasing the testing temperature or by decreasing the rate of deformation (Fried, 1995).

Activation and Deactivation Processes

Generally, the initial portion of the force-deformation plot of the amorphous nature of elastic materials is often nonlinear. During between-appointments periods, the activated elastic module often maintains much of its initial stretch for a substantial length of time. Because of the secondary bonding in certain amorphous materials are relatively weak when maintained in a stretched configuration, many of these materials lose tension over time. The force-time plot of polymeric element under constant strain conditions is illustrated in Figure 2.6. The exhibited phenomenon is relaxation, loss of force with time and static extension (Nikolai, 1985).

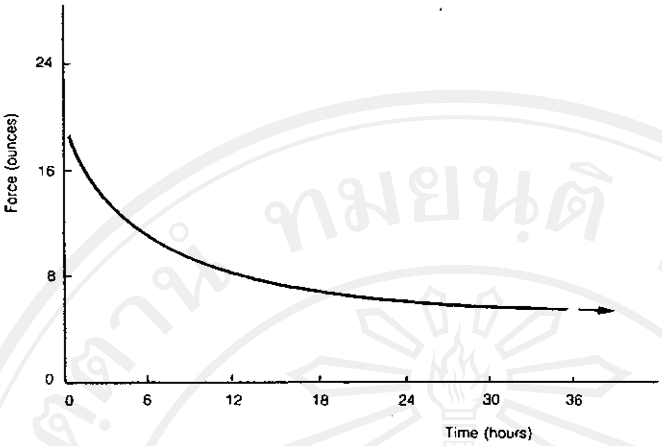

Figure 2.6 A force-time plot of polymeric materials under static extension (Nikolai, 1985)

The activation of the elastic module provides it with available energy to do work on another system; however although the activating process should not take the material of the module beyond its elastic limit, the amount of available energy may decrease with time due to relaxation. In the relatively long-term deactivation process, it is generally impossible to quantitatively separate the energy transferred from the elastic module into external-work and loss portions.

To activate a pulling force in a specific direction within the orthodontic appliance, or to transmit a tensile force from one site to another, an elastics may be used in placed of the tension spring. The popularity of the elastic, generally rubber or rubberlike polymeric materials, is associated with its favorable characteristics of high flexibility, small cross-sectional dimensions, and low cost. Free-body diagrams of the elastic module with direct tension loaded is shown in Figure 2.7. A typical load-deformation is also presented. An undesirable feature of the elastic, associated with many polymeric materials, is the tendency toward relaxation.

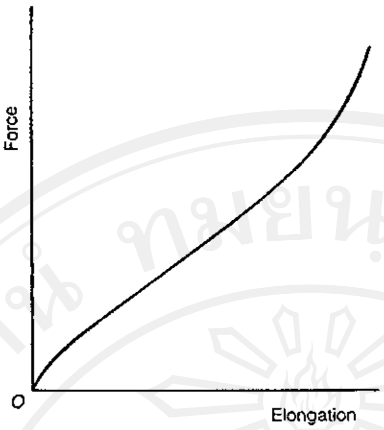

Figure 2.7 An example load-deformation diagram of the elastic module with direct tension loaded (Nikolai, 1985)

Figure 2.8 is a four-part plot for an elastic showing 1) initial activation, loading 2) maintenance of constant strain for a finite time period, relaxation 3) rather rapid unloading to a stress-free state, and 4) recovery in the absence of reactivation. The fourth segment of the plot for some polymers with sufficient time, the original passive configuration may be nearly regained if the elastic limit of the material was not exceeded during activation.

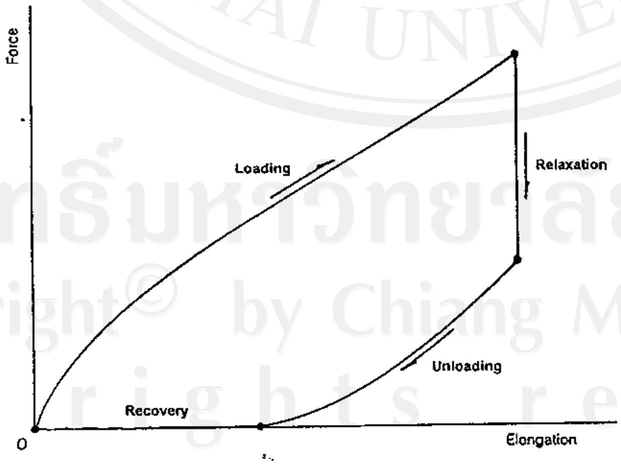

Figure 2.8 A four-part hysteresis diagram for an elastic appliance element (Modified from Nikolai, 1985)