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Tensile properties of orthodontic elastomeric chains

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SUMMARY The purpose of this investigation was to assess the permanent elongation, tensile strength and toughness of elastomeric chains. Two types (open and closed chains, i.e. with and without an intermodular link) of three brands of elastomeric module yielding six groups were included in the study. Specimens with equal numbers of loops were measured and classified into four groups based on their ageing state: (a) as-received; (b) subjected to a 24 hour steady strain in air determined as 50 per cent of original length; (c) exposed intraorally for 24 hours; and (d) retrieved following 3 weeks of intraoral exposure. All specimens were subjected to tensile stress and their behaviour was analysed with three-way ANOVA and Tukey's multiple comparison test at $\alpha = 0.05$ level of significance, with brand, type (open or closed chains), and treatment variables serving as discriminating variables.

Modular geometry or design were found not to have significant effects on elongation, probably due to the substantial variation in chain shape, modular size, and link length among products of the same category. Similarly, no correlation was identified between specimen treatment and the tensile strength of elastomers. The toughness results were not consistent with the group rankings for tensile strength, probably because of variation in the elastic and plastic deformation of specimens upon loading.

Introduction

Orthodontic elastomeric module chains are polyurethanes, thermosetting polymer products of a step-reaction polymerization process, possessing a –(NH)-(C=O)-O- unit (Billmeyer, 1984). The elastomeric chains currently available are fabricated either by die-cut stamping or injection moulding and can be either poly(ether)urethane or poly(ester)urethane (Renick *et al.*, 1999). Apart from the manufacturing processes and composition, the variation between nominally identical products with respect to time-dependent force decay (Ferriter *et al.*, 1990) may further be attributed to: variations in module manufacturing techniques such as die-cut stamping or injection moulding; effects due to additives; and different morphological (ellipsoid or circular modules) or dimensional characteristics (presence or absence of an intermodular link) of the chains (Eliades *et al.*, 2001).

The mechanisms for permanent deformation of polymeric materials include molecular chain stretching, slippage between adjacent molecular chains, and molecular chain scission (Billmeyer, 1984). During stretching the material is subject to an instantaneous elastic deformation. As the load is maintained, there is a retarded elastic deformation, as well as an irreversible viscous deformation, producing permanent elongation (depending on the load). During unloading, the instantaneous elastic strain is immediately recovered, followed by a decay of the retarded elastic strain. The term 'creep compliance' encompasses all three aspects: the instantaneous elastic deformation, the retarded elastic deformation, and the permanent deformation (Brantley *et al.*, 2001). The mechanical properties of elastomers are influenced by the rate and duration of loading as well as by environmental conditions (Rock *et al.*, 1985; Stevenson and Kusy, 1995).

The performance of orthodontic elastomeric modules in tension is of specific interest because of the relevance to the clinical application of these materials. In the stretched state, a portion of the work of elongation is dissipated as heat, while another part produces molecular reorientation and permanent deformation. If the changes in the dimensions of the material are induced early, i.e. within 24 hours of the initiation of the retraction, the dimensions of the module would be increased for the full term of the remaining intraoral use, adversely affecting the performance of the device as a tooth-loading mechanism. In addition, tensile strength testing provides an assessment of the resistance to fracture of the chain across the direction of loading. This particular property may facilitate assessment of the probability of rupture of the chain as well as the impact of intraoral ageing (Eliades *et al.*, 1999) on the strength of the material.

The hypothesis tested was that intraoral exposure of elastomeric chains alters their tensile strength. Therefore, the purpose of this study was to assess the tensile strength and permanent elongation of elastomeric chains *in vitro* and *in vivo*.

Materials and methods

Two types (open and closed chains, i.e. with and without an intermodular link) of three brands of elastomeric module producing six groups were included in the study. Table 1 lists the brand names and types of material tested.

Table 1 The elastomeric chains included in the study.

Product code	Brand (manufacturer)	Design
I	Elasto-Force (Dentaurum, Pforzheim, Germany)	Closed (without an intermodular link)
II	Generation II (Ormco, Glendora, California, USA)	Closed
III	Alastik (3M/Unitek, Monrovia, California, USA)	Closed
IV	Elasto-Force	Open (with an intermodular link)
V	Generation II	Open
VI	Alastik	Open

The specimens were prepared by cutting multiple series of specimens of each type and brand containing 10 samples each (total 60), having equal numbers of loops (six) from the spools, with the use of a sharp ligature cutter. The choice of the number of loops as opposed to a standardized length of chain was based on the variability in shape and design noted among the brands selected. This precluded using the parameter of initial length as a reliable normalization factor for the fabrication of specimens. Care was taken to avoid extended handling during cutting as this might have incorporated stresses in the material prior to testing.

The initial length (L_0) of the prepared specimens was measured with a digital calliper (Mitutoyo, Tokyo, Japan). The specimens of each group were classified into four groups:

1. elastomeric chains in the as-received state;
2. chains subjected to a 24 hour steady strain in air, determined as 50 per cent of its original length. A percentage of elongation was chosen because of the variability in the design and dimensions of the modules of the chains, which made standardization of absolute values of extension unreliable;
3. retrieved modules elongated intraorally at approximately 50 per cent extension relative to the as-received state. Specimens comprising the latter group were retrieved following 24 hours of exposure in the oral cavity of the same patient of good oral health under orthodontic treatment with edgewise, 0.018 inch, Hilgers prescription, Ormco Diamond brackets (Ormco) and receiving no medication *per os*. This facilitated comparison between material alterations induced by *in vitro* stretching and intraoral extension during the same time period;
4. chains retrieved after a 3 week exposure to the oral environment of 10 patients receiving orthodontic therapy by the same orthodontist with brackets identical to the 24 hour retrieval experiment. The modules were again elongated approximately 50 per cent relative to the as-received state.

All intraorally exposed specimens were rinsed with copious amounts of distilled water to remove the loosely bound intraorally formed integuments. The final length (L) of the specimens was measured at the end of the testing period. Because of the absorption of oral fluids, the resultant swelling of each modular unit and the different engagement pattern of elastomerics with brackets intraorally, the dimensions of retrieved modules were altered by a set of parameters additional to the stretching and, therefore, elongation measurements were only reliably performed for the specimens of the first group, i.e. stretched in air. The percentage elongation was calculated as: $\varepsilon = [(L - L_0)/L_0].100$.

For the tensile strength test, new specimens were fabricated by cutting 12 loops of each chain spool of the materials included in Table 1; in this case the six central loops were subjected to stretching and the three loops on each side were used to alleviate the excessive stress concentration on the terminal loops. Ten specimens of each of the six material groups exposed to the four conditions of the experiment were mounted on a calibrated universal testing machine (Model 6022, Instron Corp., Canton, Massachusetts, USA) (Figure 1) and were subjected to tensile extension at a rate of 5 mm/minute until failure. The testing configuration consisted of fabricating two hooks from 0.050 inch diameter stainless steel wire and attaching the elastic chain to the hooks. The choice of a large-diameter round wire was to avoid the edges of the rectangular wire, which may have applied increased stress on the chain. A 0.050 inch wire is also sufficiently stiff to exclude any absorption of stress during testing.

The modules were subjected to tensile stress and the breaking force (N) was recorded for each specimen. The toughness (J) index of each specimen, representing the energy absorbed to fracture, was estimated by calculating the area under the curve (Takahara *et al.*, 1992). It should be noted that the raw data were in terms of force versus extension curves, as it was not possible to convert the force precisely into stress units (Nm^{-2}) as the cross-sectional area of the material varied. Similarly, toughness values were not obtained in the normal units of Jm^{-3}, but in (milli)Joules.

The results of the elongation experiment were analysed with a two-way ANOVA with brand and type (open or closed chains) serving as discriminating variables, while the results of the tensile strength and toughness measurements were statistically analysed with a

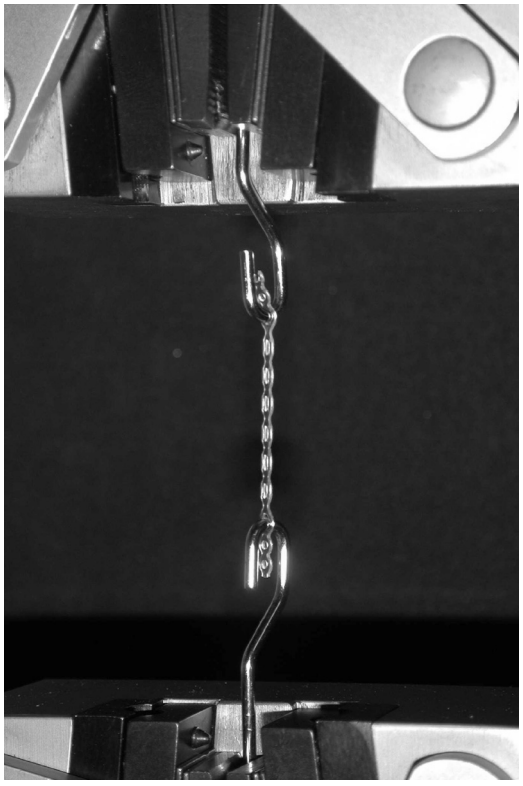

Figure 1 A stretched elastomer mounted on the experimental configuration.

three-way ANOVA with brand, design (closed versus open) and state (as-received, 24 hour stretched in air, 24 hour retrieved, and 3 week retrieved) variables. Further differences among groups were examined with Tukey's multiple comparisons test at $\alpha = 0.05$ level of significance.

Results

Table 2 shows the results of permanent elongation measurements following extension in air and the corresponding statistically significant differences. The statistical analysis did not show a variation between closed and open chains with respect to elongation despite the fact that the highest elongation was presented by two closed-type products (groups VI and IV).

Table 2 Permanent elongation (percentage) of chain groups after 50 per cent extension in air for 24 hours.

Code	Percentage elongation (mean ± SD)	Tukey's grouping
VI	9.2 ± 0.7	a
IV	6.6 ± 0.7	b
II	3.9 ± 0.2	c
III	3.1 ± 0.2	c
V	2.1 ± 0.2	d
I	1.6 ± 0.1	d

Means with the same letter are not statistically different at $\alpha = 0.05$ level of significance.
SD, standard deviation.

Figure 2 illustrates the force/extension curve of as-received (a) and retrieved (b) specimens following 3 months *in vivo*, of the same brand and type. The curve for the retrieved material denotes a stiffer body.

Tables 3–6 summarize the results of the tensile strength and toughness measurements of the chains exposed to the four experimental conditions. Some statistically significant differences were found in tensile strength and toughness among the various chain types subjected to the same treatment. ANOVA indicated that all interaction terms (brand, design, treatment) demonstrated statistically significant values precluding the assignment of effects to individual variables (Sokal and Rohlf, 1995). Thus, no direct correlation was found between the tensile strength and toughness of the chains submitted to the same treatment and no statistically significant differences were detected in tensile strength among the four testing conditions for each chain type. Nevertheless, statistically significant lower fracture values were noticed in the following treatments: 3 weeks *in vivo* (I), 3 weeks *in vivo* and 24 hours *in vivo* (II and IV), and 24 hours in air (III and VI). These results do not confirm any direct association between the design of the chain (closed, open) and the tensile properties evaluated.

Discussion

The results of this study suggest that as early as 24 hours following a 50 per cent fixed elongation in air, most elastomers present an unfavourable permanent deformation, which may extend up to 10 per cent of the original length. The alteration in elastomeric chain length associated with their intraoral exposure did not allow for the determination of permanent elongation in retrieved specimens, although anecdotal clinical experience indicates that this elongation is higher in aged samples. Elongation of modules leads to a reduced

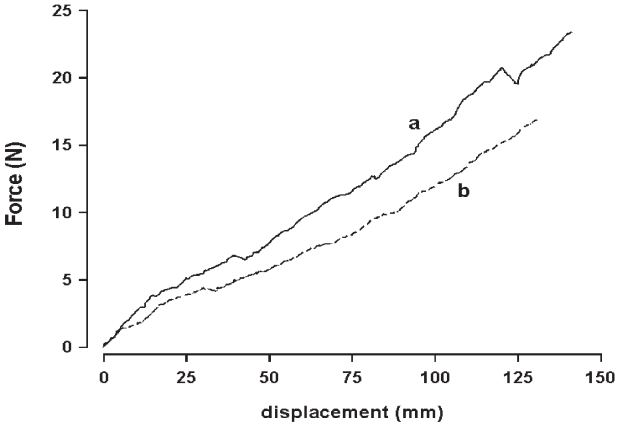

Figure 2 Force/extension plots for an (a) as-received and (b) retrieved Alastik elastomer specimen.

Table 3 Tensile strength (N) and toughness (mJ) of the reference specimens.

Code	Tensile strength (mean ± SD)	Tukey's grouping for strength	Code	Toughness (mean ± SD)	Tukey's grouping for toughness
VI	25.0 ± 1.9	a	VI	720 ± 21	a
I	23.1 ± 1.6	a b	V	577 ± 24	b
III	21.5 ± 3.0	b c	IV	548 ± 24	b
II	21.3 ± 4.0	c	III	545 ± 13	b
IV	21.1 ± 4.0	c	II	490 ± 1	c
V	19.5 ± 9.0	c	I	485 ± 24	c

Means with the same letter are not statistically different at $\alpha = 0.05$ level of significance.
SD, standard deviation.

Table 4 Tensile strength (N) and toughness (mJ) of the chains subjected to 50 per cent extension in air.

Code	Tensile strength (mean ± SD)	Tukey's grouping for strength	Code	Toughness (mean ± SD)	Tukey's grouping for toughness
VI	24.0 ± 1.2	a	VI	665 ± 34	a
III	23.8 ± 1.5	a	III	529 ± 31	b
II	23.5 ± 1.7	a	IV	518 ± 23	b
I	23.3 ± 2.4	a	I	495 ± 16	b
IV	19.5 ± 1.2	b	V	481 ± 23	b
V	17.5 ± 1.4	b	II	460 ± 42	c

Means with the same letter are not statistically different at $\alpha = 0.05$ level of significance.
SD, standard deviation.

Table 5 Tensile strength (N) and toughness (mJ) of the chains following exposure to the intraoral environment for 24 hours at 50 per cent extension.

Code	Tensile strength (mean ± SD)	Tukey's grouping for strength	Code	Toughness (mean ± SD)	Tukey's grouping for toughness
I	28.0 ± 4.2	a	VI	515 ± 60	a
IV	21.4 ± 6.3	a	I	485 ± 23	a
II	20.1 ± 3.4	a	V	482 ± 32	a
VI	19.5 ± 9.1	a b	IV	474 ± 29	a
V	18.7 ± 8.2	b	II	395 ± 13	b
III	17.5 ± 4.1	b	III	375 ± 24	b

Means with the same letter are not statistically different at $\alpha = 0.05$ level of significance.
SD, standard deviation.

Table 6 Tensile strength (N) and toughness (mJ) of the chains following exposure to the intraoral environment for 3 weeks at 50 per cent extension.

Code	Tensile strength (mean ± SD)	Tukey's grouping for strength	Code	Toughness (mean ± SD)	Tukey's grouping for toughness
VI	25.7 ± 2.5	a	VI	685 ± 60	a
IV	23.0 ± 1.3	a b	III	490 ± 32	b
III	22.5 ± 2.3	a b	IV	487 ± 21	b
I	20.1 ± 2.0	b c	V	463 ± 13	b
II	19.2 ± 3.4	b c	II	432 ± 19	b
V	17.5 ± 2.3	c	I	375 ± 23	c

Means with the same letter are not statistically different at $\alpha = 0.05$ level of significance.
SD, standard deviation.

tensile force, which reduces the efficacy of this system for the retraction of teeth.

The trend seen for the open elastomeric products (with an intermodular link) to present higher elongations may be partly explained by the increased concentration of the load and thereby elongation at the intermodular link region. This is in contrast to closed elastomeric modules where the strain developed at the modular rings is much higher (Eliades *et al.*, 2001). Nevertheless, no statistical significance was identified among closed or open chains with respect to elongation variants. This may be partially attributed to the increased variability in the link length noted between open elastomers of various brands, with the open presenting intermodular link lengths extending from 1 to 3 mm, while most modules had different shapes ranging from elliptical to circular.

The issue of force relaxation of elastomeric chains has attracted the interest of many investigators because of the clinical significance of the performance of materials (Brantley *et al.*, 1979; Rock *et al.*, 1985; von Fraunhofer *et al.*, 1992; Chau *et al.*, 1993). In spite of the extensive evidence presented on this subject, there is a lack of information on the structural changes occurring during stretching and unloading, including molecular conformation of the polymeric material. A typical stress–strain curve for an elastomeric module is characterized by a deviation from the Hookean elasticity, seen in the curved initial region of the plot as shown in Figure 2. Permanent deformation is the result of irreversible displacements between the polymeric chains, possibly including chain rupturing of crosslinks. The limit for entropy-driven elasticity in these materials occurs at the point of inflection where the curvature of the plot changes from concave-down to concave-up, whereas the region of elastic deformation corresponds to reversible changes in the separation between molecules in the polymeric structure.

The differences in tensile strength found between the reference state and the 50 per cent extension in air may reflect variations in the molecular arrangement of materials. The different rankings in tensile strength and toughness imply that some elastomers undergo different degrees of elastic and plastic deformation after preloading. The modes of deformation are further modified by the presence of plasticizing agents, i.e. water, oral fluids or stiffening agents, such as intraoral calcification, and depend on the intraoral conditions and intraoral exposure period (Eliades *et al.*, 1999).

The deviation of the stress–strain plot from an ideal straight line is probably related to the low loading rate chosen, as high loading rates may modify the behaviour of the material. In general, at low temperatures or high strain rates the material behaves like a stiff body because there is no time allowed for the chain segments to be mobilized and slippage is absent, thus the primary stiff valence bonds govern the profile of the material. At high temperatures or low strain rate, chain slippage dominates and the material becomes more compliant (Darvell, 1997). The 5 mm/minute rate employed in this study was considered substantially low compared with the instantaneous chain stretching which occurs prior to chain placement. However, this rate is routinely used in relevant studies and also the emphasis is mostly on the long-term survival of the material, and not on the strength of the material before its engagement into the bracket, which is subject to operator control.

For elastomerics subjected to strains with a λ ratio (L/L_0) of less than 1.1, the stress at a fixed strain decreases with increasing temperature, while at larger λ the stress increases with increasing temperature. The process is termed thermoelastic inversion (Darvell, 1997). Three major models have been proposed to explain the behaviour of elastomers on stretching and unstretching from a statistical mechanical perspective (Gedde, 1995). These theories assume that crosslinks are fixed in space; the entropy of the network is the sum of the entropies of individual chains; all different conformational states have the same energy; the deformation at the molecular level resembles that of the macroscopic level; and an unstressed network is isotropic. The introduction of many theories derives from the inability of a rigid hypothesis to explain the behaviour of elastomers, especially when deformation ratios exceed four. The reason is that all these theories integrate a number of assumptions which do not exist in reality. These include the absence of loose ends in the structure of the material, which holds true only for an infinite network, and the lack of fillers in the synthesis of the elastomers (Darvell, 1997). Loose ends, or uncrosslinked polymers, do not take part in the formation of the retracting force, thus reducing the number of load-carrying chain segments. Similarly, fillers in the elastomers in the form of colour pigments or other substances intended to increase the strength of the materials, may have a pronounced effect on the behaviour of elastomers in stretching. If the filler particle included in the polymer structure has a larger modulus than the surrounding structure, it will not extend to the same amount as the remaining material. This means that the ends of the material in contact with fillers across the direction of tension must be stretched more than the adjacent polymer fibrils to counteract the inability of fillers to stretch (Darvell, 1997). Filler content may thus be critical for the microscopic strain, as more fillers packed closely will induce larger stretching of the intervening polymer cylinders. Evidence supporting this effect has been presented (Baty *et al.*, 1994) illustrating steeper force decay for coloured specimens.

It is desirable that a stretched elastomeric chain possesses high tensile strength to avoid premature rupture. Higher values of these mechanical properties are associated with crystalline polymers where stress is employed to alter the orientation of the molecular arrangements

(Renick *et al.*, 1999). This effect should also provide some increase in the polymer crystallinity and mechanical properties when the polyurethane chains are stretched, as in the unstretched state the elastomeric modules have a more amorphous structure.

The composition of the environment plays a major role in the total life of the polymer. This effect is particularly important for biomedical applications, such as in orthodontics. In this case, synergistic effects may arise from several factors such as loading, variations in temperature, solvents or pH fluctuations (Huget *et al.*, 1990) leading to a reduced fatigue limit for a polymer specimen (Stokes *et al.*, 1987).

In contrast to the foregoing considerations, the results of this study did not demonstrate any correlation between ageing and variation in tensile strength. There are a number of reasons explaining this discrepancy including: (a) the unique environment and application of these materials in orthodontics as opposed to implantable devices where most of the information available for the maturation and properties of these materials derives (Stokes *et al.*, 1987); (b) the relatively reduced force magnitude developed during stretching in orthodontics.

Thus, while it is known that an alteration in environmental variables substantially modifies the force decay rates of elastomers (Rock *et al.*, 1985; de Genova *et al.*, 1985; Chau *et al.*, 1993), 'terminal' properties of the material, such as the tensile strength, seem not to be sensitive to environmental alterations.

The applicability of the results of this study to various degrees of stretching is uncertain and, therefore, changes or similarities among the materials tested may be different when stretching is substantially increased.

Conclusions

In vitro stretching of elastomeric chains induced permanent deformation in the form of elongation. Chain geometry or design did not affect the permanent elongation of chains, probably because of the substantial variation in chain shape, size and link length among products of the same category (open or closed).

No correlation was identified with as-received, stretched in air, and retrieved specimens of elastic chains with respect to tensile strength and toughness.

The toughness of specimens was not consistent with the group rankings of tensile strength. This may be due to variations in elastic and plastic deformation of specimens on loading.

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References

Baty D L, Volz J E, von Fraunhofer J A 1994 Force delivery properties of colored elastomeric modules. American Journal of Orthodontics and Dentofacial Orthopedics 106: 40–46

Billmeyer Jr F W 1984 Thermosetting resins. In: Textbook of polymer science. Wiley, New York, pp. 36–457

Brantley W A, Salander S, Myers L C, Winders R V 1979 Effect of prestretching on force degradation characteristics of plastic modules. Angle Orthodontist 49: 37–43

Brantley W A, Eliades T, Litksy A S 2001 Mechanics and mechanical testing of orthodontic materials. In: Brantley W A, Eliades T (eds) Orthodontic materials: scientific and clinical aspects. Thieme, Stuttgart, pp. 27–49

Chau L T, Wang W M, Tarng T H, Chen J W 1993 Force decay of elastomeric chains. American Journal of Orthodontics and Dentofacial Orthopedics 104: 373–377

Darvell B W (ed.) 1997 Materials science for dentistry, 4th edn. University of Hong Kong, pp. 43–56

de Genova D C, McIness-Ledoux P, Weinberg R, Shaye R 1985 Force degradation of orthodontic elastomeric chains. A product comparison study. American Journal of Orthodontics 87: 377–384

Eliades T, Eliades G, Watts D C 1999 Structural conformation of *in vitro* and *in vivo* aged orthodontic elastomeric modules. European Journal of Orthodontics 21: 649–658

Eliades T, Eliades G, Brantley W A, Watts D C 2001 Elastomeric ligatures and chains. In: Brantley W A, Eliades T (eds) Orthodontic materials: scientific and clinical aspects. Thieme, Stuttgart, pp. 173–189

Ferriter J P, Meyers C E, Lorton L 1990 The effects of hydrogen ion concentration on the force-degradation rate of orthodontic polyurethane chain elastics. American Journal of Orthodontics and Dentofacial Orthopedics 98: 404–410

Gedde U M 1995 Polymer physics. Chapman & Hall, London, pp. 39–75

Huget E F, Patrick K S, Nunez L J 1990 Observations on the elastic behavior of a synthetic orthodontic elastomer. Journal of Dental Research 69: 496–501

Renick M R, Brantley W A, Webb C S, Beck F M, Vig K, Culbertson B M 1999 DSC studies of orthodontic plastic modules I. As-received products. Journal of Dental Research 78: 320

Rock W P, Wilson H J, Fisher S E 1985 A laboratory investigation of orthodontic elastomeric chains. British Journal of Orthodontics 12: 202–207

Sokal R R, Rohlf F J 1995 Biometry: the principles and practice of statistics in biological research, 3rd edn. Freeman, New York

Stevenson J S, Kusy R P 1995 Structural degradation of polyurethane based elastomeric modules. Journal of Materials Science–Materials in Medicine 6: 377–384

Stokes K, Coury A J, Urbanski P 1987 Autooxidative degradation of implanted polyether polyurethane devices. Journal of Biomaterials Applications 1: 411–448

Takahara A, Hergenrother R W, Coury A J, Cooper S L 1992 Effect of soft segment chemistry on the biostability of segmented polyurethanes. II. *In vitro* hydrolytic degradation and lipid sorption. Journal of Biomedical Materials Research 26: 801–818

von Fraunhofer J A, Coffelt M T P, Orbell G M 1992 The effect of artificial saliva and topical fluoride treatments on the degradation of the elastic properties of orthodontic chains. Angle Orthodontist 62: 265–274