

SILICONES IN BIOMEDICAL APPLICATIONS

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Chapter 32

Silicones in Biomedical Applications

Barry Arkles and Peter Redinger

ABSTRACT

The chemistry and mechanical properties of silicones including surface treatments, filler-free silicones, low and high consistency silicones and new-generation silicone-thermoplastics is reviewed in the context of medical device applications. Specific medical devices considered are low thrombogenic potential coatings, membranes for oxygenation and dialysis, pressure sensitive adhesives, mammary (soft tissue) prosthesis, cardiovascular devices, microelectrodes and contact lenses. Silicone IPN's silicone-urethanes, silicone-polycarbonates and silicone-methacrylates are emphasized.

Encouraging toxicological data led to the fabrication of the first medical device, a catheter, by 1950 [10]. Silicone polymers now have over thirty years of *in vivo* testing, and are found in an increasingly wide range of medical applications. The silicones that have evolved to meet these requirements have a broad range of properties which are difficult to discuss systematically. This review of silicones will be an overview from the materials perspective illustrated with specific medical device applications.

While applications for biomedical materials may be divided into short-term body contact and long-term implantation, in the case of silicones differences in materials recommendations are associated with manufacturing practice and control. Modifications of mechanical or chemical characteristics of silicones generally play no *a priori* role in determining implantability.

Several basic fabrication techniques are associated with silicones incorporated into medical devices.

| | |
|---------------|---------|
| Coating | Sealing |
| Bonding | Casting |
| Encapsulation | Molding |

The selection of a particular silicone is dictated both by its fabrication and performance characteristics. A single silicone material frequently is not limited to a single fabrication technique. Widely different silicones fabricated by different techniques can satisfy the same application requirements.

Silicone materials used in biomedical applications can be classified as follows:

1 Monomers and Low Molecular Weight Resins

1.1 Surface treatments and molecular coatings

1.2 Monomers for cureable systems and crosslinking

2 Prepolymer Fluids and Unfilled Polymers

2.1 Filler-Free Silicone Rubber

2.2 Pressure Sensitive Adhesives

2.3 Gels

3 Elastomers (including RTV, LTV and HTV systems)

The most general purpose and widely recognized silicones including those used in many soft tissue replacement, catheter, flexing joint, drain, shunt, adhesive and impression applications.

3.1 Adhesives and Sealants

3.2 Low-Consistency Elastomers

3.3 High-Consistency Elastomers

4 Thermoplastics

"New-generation" silicones with greater mechanical properties including copolymers, graft polymers and IPN's used in cardiovascular devices, contact lenses and membranes.

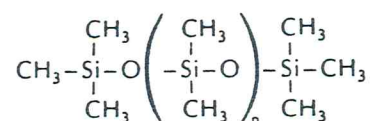
4.1 Silicone-Urethanes

4.2 Silicone-Polycarbonates

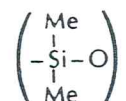
4.3 Silicone-Methacrylates

REVIEW OF SILICONE CHEMISTRY

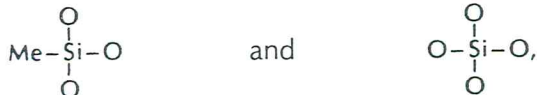
The simplest silicones are the polydimethylsiloxanes. The polymers, when not cross-linked, have found widespread use in the manufacture of cosmetics, food-processing materials, and deflatulant preparations. They are valued for their lubricity, ability to lower surface tension and act as antifoam agents. Polydimethylsiloxanes exhibit the basic structure of the silicones.



The repeating unit of the polymer



the dimethylsiloxane unit, is referred to as a D unit, while the termination unit ($\text{Me}_3\text{SiO}-$), the trimethylsiloxy group, is referred to as an M unit. The structure shown above can thus be described as an M unit. The silicones do not have to remain linear; they can obtain branch points,



referred to as T groups and Q groups, which introduce rigidity into the structure. This relatively simple chemistry results in fluids at low molecular weights and leads to cross-linkable gums at high molecular weights.

Table 1. Silicone Substitutions.

| Structure | Name | Application |
|--------------------------------------|-----------------|------------------------------------------------------------------------------------------------------------|
| CH_3- | methyl | Basic substitution found in virtually all silicones; a cross-linking point with peroxide cures |
| C_6H_5- | phenyl | Increases modulus, thermal, and UV stability; raises refractive index; decreases reactivity with peroxides |
| $\text{CF}_3\text{CH}_2\text{CH}_2-$ | trifluoropropyl | Increases stability, increases solvent resistance |
| $\text{H}-$ | hydride | Introduces metal-catalyzed and vinyl-addition cross-link sites |
| $\text{HO}-$ | silanol | Cross-linking point for condensation and metal-catalyzed cross-linking |
| $\text{CH}_2=\text{CH}-$ | vinyl | Increases peroxide reactivity; introduces cross-link points for vinyl addition |

A variety of groups, including phenyl, vinyl, and hydrogen can substitute for the methyl group in a silicone. This is significant inasmuch as the substitution, branching and molecular weight of a silicone polymer will dictate the method by which curing or cross-linking can be accomplished. Fundamentally, there are four processes employed to cure silicones. In high temperature vulcanizing (HTV) systems, polymers containing methyl or vinyl groups are cross-linked with peroxides. In room-temperature vulcanizing (RTV) systems, two cure methods are used. In the older methods silanols are condensed with a moisture

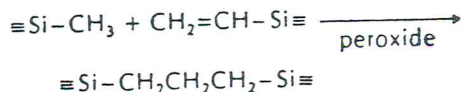
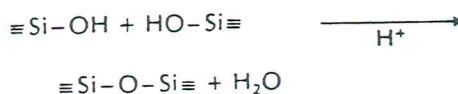
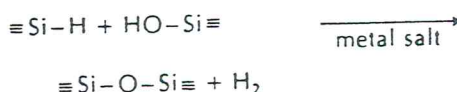
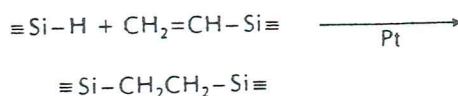
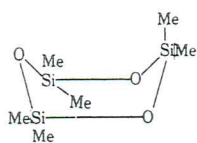
Peroxide**Condensation****Metal salt****Vinyl addition**

Figure 1. Silicone-curing systems.

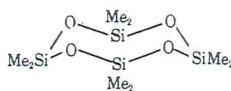
sensitive silane cross-linker or a metal salt catalyzed reaction between silicon hydrides and silanols takes place. The latter reaction liberates hydrogen which can be used to produce foamed products. Vinyl addition systems in which a platinum complex catalyzes the addition of silicone hydrides to vinyl substituted silanes was initially used in low temperature vulcanizing (LTV) systems but has been gradually extended to RTV's and HTV's.

The primary monomeric raw materials for silicone polymers are chlorosilanes. These materials are corrosive, flammable and generally difficult to handle. They are used directly or as their alkoxy derivatives only in surface treatment applications. The direct intermediates for polymers are cyclic siloxanes and low molecular weight silanols. It is important to note that biological inertness of polymers derived from these monomers does not imply biological inertness of the monomers themselves. The biological activity of many low molecular weight silanes has been established. The commonly used monomers shown in Figure 2 have been shown to have extremely low levels of toxicity.

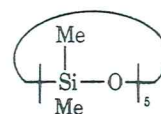
Silicones in Biomedical Applications



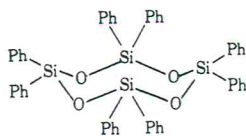
hexamethyl-
cyclotrisiloxane



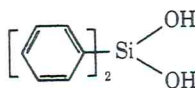
octamethyl-
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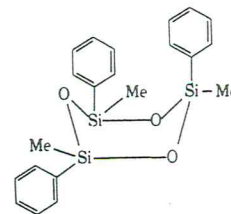
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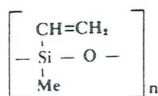
octaphenylcylco-
tetrasiloxane



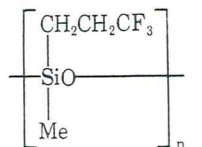
diphenylsilane-
diol



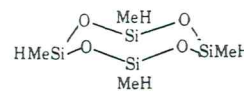
trimethyltriphenyl-
cyclotrisiloxane



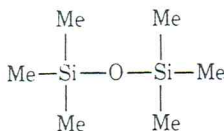
vinylmethyl-
cyclosiloxanes



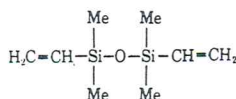
trifluoropropylmethyl-
cyclosiloxanes



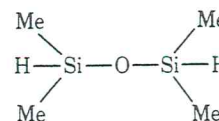
methylhydro-
cyclosiloxane



hexamethyl-
disiloxane



divinyltetra-
methyldisiloxane



tetramethyl-
disiloxane

Figure 2. Silicone monomers and terminators.

Two general methods are employed to produce silicone polymers. Gums and non-silanol terminated fluids are produced by base catalyzed (anionic) polymerization. Under typical conditions, potassium, sodium, or tetramethylammonium silanolate catalysts are introduced into a mixture of monomers and end-cappers. The mixtures are equilibrated by heating under moderate conditions (60-150°C) [11,12]. Silanol terminated polymers are produced by a variety of techniques including hydrolysis of chlorosilanes with water or indirectly with methanol, metal salt catalyzed telomerization of cyclics in the presence of dimethyl-

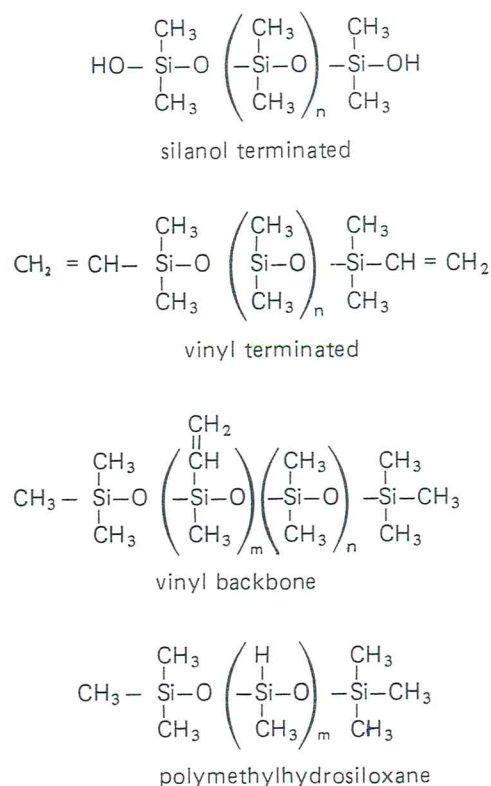


Figure 3. Silicone prepolymers.

dichlorosilane followed by hydrolysis, and equilibration of cyclics with low molecular weight silanol fluids which act as chain terminators [13,14,15]. Typical reactive silicone polymers are shown in Figure 3. They include silanol, vinyl and hydride functional materials of various degrees of polymerization.

Most silicones are filled with particulate silica. Particulate silica has a reinforcing effect on silicone elastomers that results in increased modulus, tensile strength, tear strength and abrasion resistance. The reinforcing effect depends on three properties of the filler: particle size, structure, and surface interaction. The particle size together with the filler loading determines the area available for surface interaction. The structure determines the extent polymer movement is restricted under deformation. The surface interaction determines the effectiveness with which particle size and structure control elastomer properties. Parameters associated with surface interaction include population and type of hydroxyl groups, adsorbed moisture and chemical treatment. It is postulated that the redistribution of stress among polymer chains in reinforced systems is accomplished by two-dimensional mobility of polymer chains that are otherwise bound tightly to the silica surface.

Chemical treatment of silica may greatly change its interaction with silicone polymers. Fumed silica behavior can be altered by inactivating the isolated silanol groups with various silanes. Materials used to modify surface characteristics include hexamethyldisilazane, divinyltetramethyldisilazane and cyclic and short linear siloxanes. Under-treatment of silica

results in "crepe hardening," a characteristic condition in which uncured elastomer exhibits crumbling rather than plasticity under applied stress. Over-treatment of filler results in non-responsive or "dead" elastomers. High tear strength elastomers result from filler treatments which introduce vinyl groups.

1 Monomers and Low Molecular Weight Resins

1.1 Surface treatments and molecular coatings

Reactive monomeric and oligomeric silanes are employed to impart specific surface properties to inorganic and silicone substrates. The materials have the general structure: $R_{4-n}SiX_n$ where $n = 1$ to 3 . R is an organic radical in which a carbon is bound to silicon. X is an electronegative moiety such as alkoxy or chlorine. The organic radical is hydrolytically stable and the electronegative moiety is not. The electronegative group is removed by hydrolysis to form an unstable silanol which condenses with the $-OH$ group of substrates. If there are three hydrolyzeable moieties telomerization occurs during the deposition process and coatings 6-8 molecular layers thick usually result. Details of this chemistry have been reviewed [16].

Monomeric and oligomeric silanes are not sold as medical grade materials. They are widely used in industrial products including composites, coatings and adhesives. They are used to form water-repellent surfaces. It is imperative that the user of these materials for medical devices insures that excessive monomer and hydrolysis byproducts are removed after surface treatments.

Silane surface treatments are employed in clinical analytical techniques to reduce interaction of blood with hydrophilic surfaces. Water-soluble derivatives of octadecylsilane are employed in the treatment of glassware to reduce adsorption of erythrocytes. Oligomeric silanes with active chlorine or alkoxy termination not only exhibit the hydrophobicity of octadecyl surface treatments, but have a high degree of oleophobicity as well. They reduce protein adsorption on glass surfaces. Other silanes perform more active roles, enhancing adsorption or bonding of specific proteins such as heparin, bioactive groups such as fungistats, or whole cells and cell organelles. Silanes have been employed to covalently bond enzymes and cofactors to inorganic substrates. A summary of surface treatments is found in Table 2.

Table 2. Surface Treatments and Molecular Coatings

| Silane Type | Application | Tradename | ref. |
|-----------------------------------|----------------------------------------------------------------------------------|----------------------------|-------|
| Octadecyl | reduces protein adsorption reduces thrombogenicity | Glassclad® 18 Siliclad | 18,19 |
| Chlorinated siloxane telomer | reduces protein adsorption reduces thrombogenicity increases oleophobicity | Glassclad® 6C Dri-Film® | 17 |
| Aminopropyl | adsorbs heparin | | 17 |
| Polyethylenimine | adsorbs proteins | Glassclad® IM | 19 |
| Ethyl | adsorbs erythrocytes | | 19 |
| Octadecyl Quat | fungistatic treatment | D.C.® 5700 | 20,21 |
| Amine, p-nitrobenzamide, mercapto | immobilization of enzymes | | 22,23 |
| Heparin | anticoagulant | Glassclad® HP | 24 |

Glassclad® J.M. of Petrarch Systems, Dri-Film J. M. at General Electric D.C. T.M. at Dow Corning

1.2 Monomers for Cureable Systems

The hydrolysis and condensation of alkoxysilanes leads to resinous products which have traditionally been used as coatings and encapsulants. In a manner consistent with this chemistry, slow release anti-tumor devices have been produced by hydrolyzing the silanes in the presence of various drugs [25].

The direct polymerization of monomeric siloxanes such as hexamethylcyclotrisiloxane (D_3) and octamethyl-cyclotetrasiloxane (D_4) is of current interest in the preparation of filler-free silicone rubber (FFSR). On many occasions undesirable biological responses to filled standard silicone rubber (SSR) has been attributed to the filler rather than the silicone itself. Physiological reaction to fillers is especially notable in devices with large surface areas such as oxygenator and dialysis membranes. FFSR is more often achieved by reaction of prepolymers than polymerization of monomers. The monomeric approach is discussed in this section. The prepolymer approach is discussed in the next section.

Plasma and radiation polymerization of D_3 has been reported by Chawla for the production of filler-free silicone coatings and bulk moldings [26-29]. In this method D_3 is melted and cast or applied to a polymer substrate then subjected to high energy polymerization. The materials have tensile strengths of 2.5 MPa and an elongation of 500%. In work with whole canine blood adsorption of platelets and leukocytes is reduced 30-35% when compared to SSR.

2 Prepolymer Fluids and Unfilled Polymers

2.1 Filler-Free Silicone Rubber

Filler-free silicone rubber can be prepared by cross-linking unfilled silicone gum stocks. Membranes for artificial lungs, are fabricated by a method in which 1-1.5% dichlorobenzoyl peroxide is used to cross-link a polydimethylsiloxane gum. The gum is cured at 100°C, for five minutes, followed by a four hour postcure at 150°C. Another method that lends itself to the preparation of thin films requires that a vinyl-modified dimethylsiloxane gum and peroxide be dissolved in toluene and cast into a film. After the film is dried at 75°C, it is cured in air at 180°C and subjected to ultraviolet radiation from germicidal lamps [30,31].

The importance of silicone membranes in oxygenation and dialysis cannot be overemphasized. The permeability of oxygen through silicones is more than ten times greater than permeable materials like natural rubber and low density polyethylene. It is over 100 times greater than butyl rubber and nylon. The permeability (P) of selected gases through polydimethylsiloxane elastomer has been reported [32]. An incomplete summary follows.

Permeabilities of Gases through Polydimethylsiloxane Elastomer

| | $\frac{\text{cc of gas} \cdot (\text{cm})}{\text{sec} \cdot \text{cm}^2 \cdot \text{cmHg}} \times 10^9$ | | |
|----------------|---------------------------------------------------------------------------------------------------------|----------|------|
| water | 3600 | nitrogen | 28 |
| carbon dioxide | 325 | methane | 95 |
| oxygen | 60 | methanol | 1390 |

In order to grasp the significance of the permeation rates, it is useful to make some conver-

sions and approximations of the reported units, simplifying it within 10% to the following:

$$P = \frac{\text{liter, mil thick}}{\text{hour, meter}^2, \text{atmosphere}}$$

This indicates, for example, that oxygen would diffuse at approximately 60 liters per hour through a square meter of silicone one mil thick under a pressure of one atmosphere. Man at rest consumes 18 liters of oxygen per hour.

2.2 Pressure Sensitive Adhesives

Most of the unfilled cured silicone polymers employed in medical devices can be described as pseudointerpenetrating polymer networks. The materials contain two polymer components. One silicone is a matrix material with a linear structure. The second silicone component has a branched or cross-linked structure that extends homogeneously throughout the matrix. The systems behave in a uniform cohesive manner, although unlike true interpenetrating polymer networks only one component is cross-linked.

Pressure sensitive adhesives are generally formulated from a high viscosity silanol terminated fluid and a low molecular weight resinous silicone. The two component system is supplied in a solvent. The materials are used externally for adhering ileostomy and colostomy appliances, attaching electrodes and adhering maxillofacial medical devices. The properties of silicone PSA's are exemplified by Dow Corning® 355.

Silicone Pressure Sensitive Adhesive

| | |
|---------------------------------|--------------------------|
| solids content, % | 18.5 |
| solvent | trichlorotrifluoroethane |
| adhesive strength Al/Al, oz/in. | 100 |

2.3 Gels

Silicone based pseudointerpenetrating polymer networks which result in clear resilient gels with cohesive properties are useful in orthopedic pads [33]. The gels used for orthopedic pads in breast implants are generally produced by vinyl addition chemistry. The first gels used for these purposes were linear siloxanes with low levels of cross-linking; for example, a 500-700 centistoke dimethylsiloxane fluid with 1.0% vinylmethyl backbone was cross-linked by platinum-catalyzed addition to 4 to 5 ctsk polymethylhydrosiloxane [34]. In most updated formulations, vinyl terminated siloxanes of 1000 ctsk or greater are cross-linked with a methylhydro-dimethylsiloxane copolymer using a complex of neutralized chloroplatinic acid in divinyltetramethyldisiloxane or vinylmethylcyclsiloxanes. However, this cross-linked system constitutes only 10-20% of the gel weight; the matrix material is 350 to 1000 ctsk polydimethylsiloxane. The cross-linked silicone produces an interpenetrating polymer network within the matrix silicone, which creates a responsive gel with good cohesion [35-38]. Another approach to gel formation is to produce a hydrophilic copolymer between a vinylsiloxane with hydrolyzeable groups and vinylpyrrolidone [39,40].

An undesirable property of pseudointerpenetrating silicone networks is for a small

portion of the matrix polymer to bleed out. The matrix material can bleed through fully cross-linked polydimethylsiloxane mammary shells. Methods of reducing bleed include coating or preparing shells from phenyl or trifluoropropyl containing siloxanes [41]. The different solubility parameters for these materials allow them to behave as barriers for polydimethylsiloxanes.

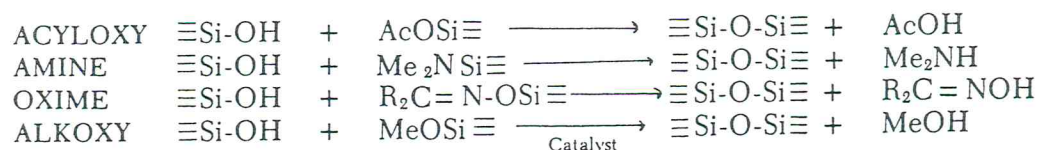
3 Elastomers

The most widely recognized silicones are filled or reinforced elastomeric products. The two main classifications of materials are: low-consistency – not requiring simultaneous application of heat and pressure for processing; high-consistency – requiring simultaneous application of heat and pressure. All of the cure systems mentioned earlier are employed in the elastomeric products. The multitude of products allows the selection of only the most representative examples in this section.

3.1 Adhesives and Sealants

Silicone adhesives and sealants are low consistency room temperature vulcanizing liquid rubbers which offer considerable latitude in formulation and properties. The main components of condensation cure RTV's are silanol terminated polydimethylsiloxanes. The viscosity of the polymer can range from 500 to 1,000,000 ctsk. Often a blend of polymers will be used in compounding. This allows variations in cross-link density, cure times and working consistencies.

Most RTV's of this class have moisture sensitive cure systems. They contain a silane cross-linker with moisture sensitive hydrolyzeable groups. The most common systems are listed below:



The reactions of these systems are often catalyzed with carboxylic acid salts of tin, zinc or iron. A fast catalyst is tin octoate. A slow catalyst is dibutyltin dilaurate. Titanates are also used to catalyze these reactions. In general, amine and oxime materials react the fastest, followed by acyloxy then alkoxy. Excess silane gives slower reacting materials with longer pot-life. Adjusting cure times is best accomplished during compounding.

Choosing a material for a particular application requires consideration of physical properties and processing. One part RTV's tend to be used as adhesives. Two part RTV's tend to be used as sealants and encapsulants. Either type can usually be dispersed in solvents as required for processing.

3.2 Low-Consistency Elastomer – Addition Cure

Addition cure materials tend to have greater strength than condensation cure products. They are thought of as clean systems since there are no by-products or hydrolyzeable leaving group during cure. Cure rate may be adjusted from minutes to hours depending upon the use of inhibitors or level of catalyst.

The viscosity of material may vary from 1000 to 1,000,000 ctsk. Lower molecular weight polymers are used to formulate dental impression materials [42,43], while higher molecular weights are used in liquid injection molding compounds. Higher molecular weight, high consistency products with properties similar to traditional HTV's are available but will not be discussed here.

Addition cure materials are supplied as two part systems. Generally the catalyst, a platinum complex, is in the "A" part and the cross-linker, a hydride, is in the "B" part. Cross-link density may be adjusted by the distribution of hydride along the siloxane chain and length of the chain. Short chains with high hydride give material with a "snappy" character and are used in formulating products with high tensile and tear strengths. Longer chains with lower hydride content give greater elongations. Typical levels of platinum in these formulations range from 5 to 15 ppm. Inhibitors may be used to moderate the rate of cure. They differ from "poisons" in that their interaction with the platinum complex is reversible. Poisons include sulfides and sulfur containing rubbers and certain amines.

Very few of the additional cure systems are sold specifically as medical grade materials. One such material which is used for biomedical encapsulating and moldmaking is Dow Corning® MDX-4-4210. Typical properties are given below.

| | |
|-------------------------------|--------|
| viscosity, as supplied, ctsk. | 80,000 |
| Cured Properties | |
| Durometer, Shore A | >25 |
| Tensile Strength, MPa | >3.8 |
| Elongation, % | >350 |
| Specific Gravity | 1.12 |

3.3 High Consistency Elastomers

The oldest and most widely used silicones are the peroxide cure high temperature vulcanizing (HTV) systems. They are used for extrusion of catheter tubing, compression molding of shunts and flexing joints and calendaring sheeting, both Dacron reinforced and non-reinforced. Uncured forms can be fabricated by manual or "lay-up" techniques. The reader is referred to Lynch [2] for an excellent review of processing and properties of HTV's.

Table 3. Typical Properties of High Consistency Silicone

| | |
|------------------------------------|--------------------------|
| specific gravity | 1.12-1.25 |
| durometer Shore A | 25-75 |
| tensile strength, MPa | 5.9-8.3 |
| elongation, % | 350-700 |
| tear strength, lb-in | 70-160 |
| compression set, 24 hours at 175°C | 10-40% |
| dielectric constant | 2.8-2.9 |
| dissipation factor | $6.0-9.0 \times 10^{-4}$ |

4 Thermoplastic Silicones

Thermoplastic silicones include silicone copolymers, block copolymers and some alloys. A substantial portion of the polymer composition does not contain a siloxane backbone. Generally speaking the materials have radically different properties than conventional silicones. Mechanical strengths are higher, but a variety of other striking property differences are introduced.

4.1 Silicone-Urethanes

The mechanical, fatigue and physiological response characteristics of silicone homopolymers are not comparable to those of silicone-urethanes used in blood pumps, intra-aortic balloons, total heart replacements and related devices [44]. Fatigue strength, toughness, flexibility and low interaction with plasma proteins are bio-engineering criteria applicable to silicone-urethanes. The most widely recognized material in this category is Cardiothane 51 (formerly Avcothane 51) produced by Kontron Cardiovascular, Inc. The material is described as a copolymer of aromatic polyether urethane and polydimethyl-siloxane [45]. The copolymer is proposed to crosslink by the interaction of acetoxy-terminated siloxane blocks with substituted urethane nitrogens [46]. The polymer is not homogeneous and may be described as a mixture of 10% moisture cured polydimethylsiloxane in a urethane matrix, with bulk phase separation prevented by small amounts of silicone-urethane surfactant formed during cure [47]. Cardiothane is primarily a casting resin. Molding and extrusion grades of silicone-polyester urethane IPN's are available from Petrarch Systems.

While not yet available on a commercial basis, several true urethane-siloxane copolymers with promising properties are in development. One approach is to use a carbinol (C-OH) transition between the siloxane and urethane [48,49]. A carbinol terminated silicone can be created by reacting silanol-terminated polymers with ethylene oxide to yield a hydroxy-ethyl ether ($\text{SiOCH}_2\text{CH}_2\text{OH}$), and by adding allylic alcohols to hydride-terminated polymers to give materials with greater hydrolytic stability ($\text{SiCH}_2\text{CH}_2\text{CH}_2\text{OH}$). A second approach is to use a urea transition between the siloxane and urethane which results from an aminoalkyl terminated siloxane [50]. (Table 4).

Table 4. Silicone-Urethane Mechanical Properties

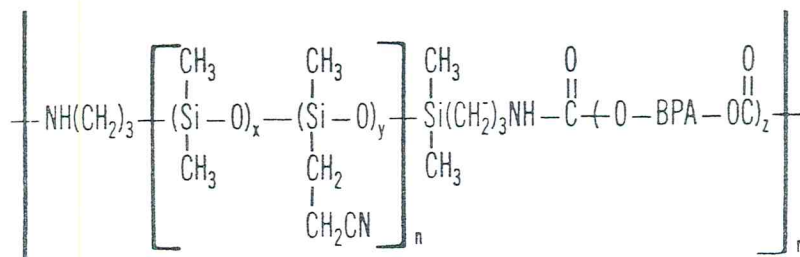
| | | |
|-------------------------|------------------|--------------------|
| name | Cardiothane 51 | PTUE 102 |
| manufacturer | Kontron | Petrarch |
| processing | dipping, casting | molding, extrusion |
| silicone % | 10 | 10 |
| specific gravity | .95 | 1.00 |
| tensile strength, MPa | 43.8 | 41.4 |
| elongation, % | 525 | 450 |
| tear strength lb/in | 490 | 750 |
| water absorption, 24 hr | 1.6% | .03% |
| hardness, Shore | A72 | D50 |

4.2 Silicone-Polycarbonates

The earliest silicone-polycarbonates were produced by condensing preformed silicone blocks with Bisphenol A by phosgenation in the presence of amine [51-54]. Later syntheses of these materials improved reproducibility and properties of these materials by using polycarbonate prepolymers and amine terminated siloxane blocks [55]. The physical characteristics of these materials have been extremely well defined [56,57]. Silicone-polycarbonates have high mechanical and tear strengths, while maintaining a high degree of oxygen and water permeability. Films may be extruded or cast at thicknesses below 1 mil.

Blood oxygenation, dialysis and microelectrode membranes are the usual biomedical applications for silicone-polycarbonates. The simple unmodified materials are used in blood oxygenation [58]. Techniques for preparing thin high temperature membranes for oxygen transport have been described [59-61]. Silicone-polycarbonates have been cast as defect free membranes 0.015 microns thick.

In microelectrode applications simple block copolymers of Bisphenol A polycarbonate have inadequate hydrolytic stability and their dielectric constants are too low to permit ion transport. For this reason, many current microelectrode materials incorporate ion-carrier groups in the siloxane block to allow molecular transport as well as a urethane transition to the polycarbonate to increase hydrolytic stability. The combination of mechanical, dielectric, permeation and film formation properties has allowed the preparation of membranes useful for monitoring physiological changes in concentrations of hydrogen, potassium, sodium and ammonium ions [62-65]. A typical structure is:



Silicone-polycarbonates can be injection molded and extruded. Perhaps their most unique process characteristic is the ability to cast films from solutions in methylene chloride or trichloropropane by dropping the solutions onto warm water and allowing the solvent to evaporate. The mechanical properties of these materials are far greater than unfilled silicones.

SILICONE-POLYCARBONATE 50:50 BLOCK COPOLYMERS

| | |
|--------------------------|-------------------------------------|
| specific gravity | 1.07–1.08 |
| tensile strength | 19.0–22.0 MPa |
| elongation | 270–350% |
| water vapor transmission | 16–20 g/24hr/100m ² /mil |

4.3 Silicone-Methacrylates – Ophthalmic Materials

Once it was recognized that maximum contact lens comfort was achieved when the eye could “breathe,” silicones became likely candidates for lens materials. The primary design

parameters which lead to selection of silicone materials are permeability (Dk) and Equivalent Oxygen Percentage (EOP), a finished lens measurement that considers oxygen demand of ocular tissue. Other parameters which require modification of basic silicone properties are wettability (contact angle), dimensional stability and refractive index.

A minimum EOP of 5-7% has been proposed for finished lenses [66,67]. This is equivalent to the oxygen available to the eye during sleep (closed eyelid). An EOP below 2 is known to cause corneal edema. Figures 4 and 5 show how the EOP relates to permeability for various lens thicknesses and various lens materials.

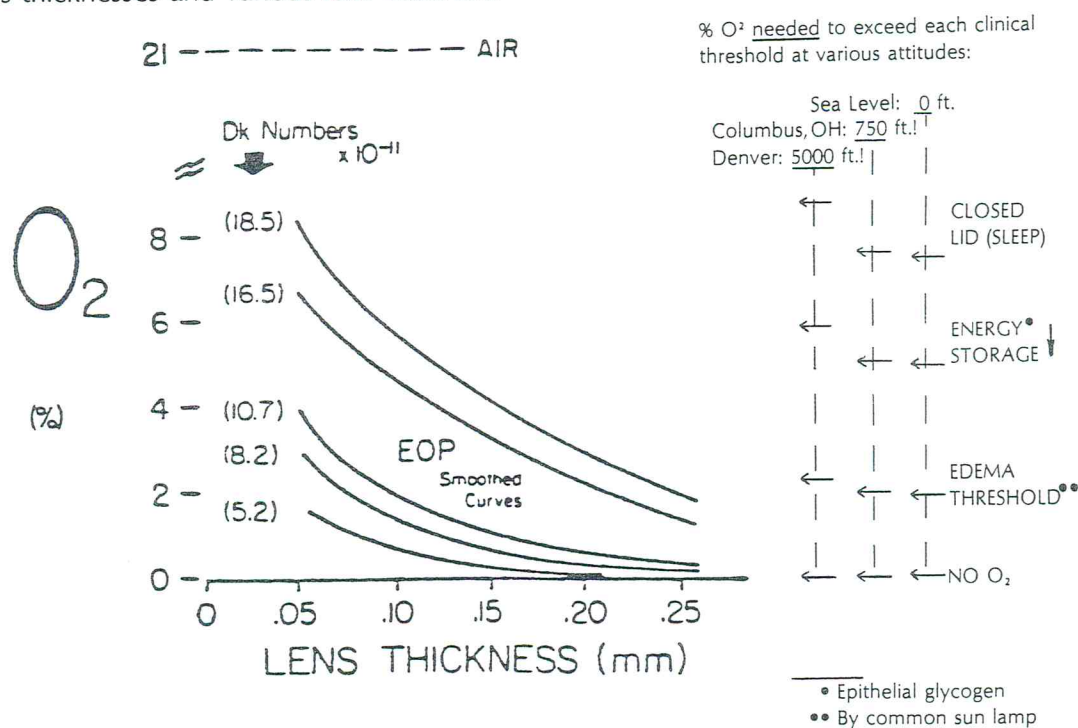


Figure 4.

Trends in the early '70s toward soft lenses bolstered interest in pure siloxane backbone materials which have the highest Dks. A variety of mechanical, optical and wetting problems posed a considerable challenge. No siloxane backbone lenses have achieved commercial acceptance. A recently introduced lens which incorporates siloxane backbone technology is offered by Dow Corning as Silsoft®. A newer approach is to employ siloxane modified methacrylate copolymers. This polymer technology leads primarily to hard lenses and is now an established commercial technology. The most conspicuous example is the Polycon Silofocon® lens offered by Syntex.

The enhancement of oxygen permeability in siloxane lens systems is associated with high relative proportions of silicon – oxygen and silicon-carbon bonds. These long bonds lead to a free volume element which is in the case of polydimethylsiloxane 5-6 times greater than that in polymethacrylates [68]. Both polydimethylsiloxane and polyvinyltrimethylsilane exhibit high oxygen permeability when introduced into copolymer systems. It appears that a maximum disordering of polymer structure – i.e., random copolymers or terpolymers – is desirable. Polymers should be designed to exclude microblock regions that are large enough to phase

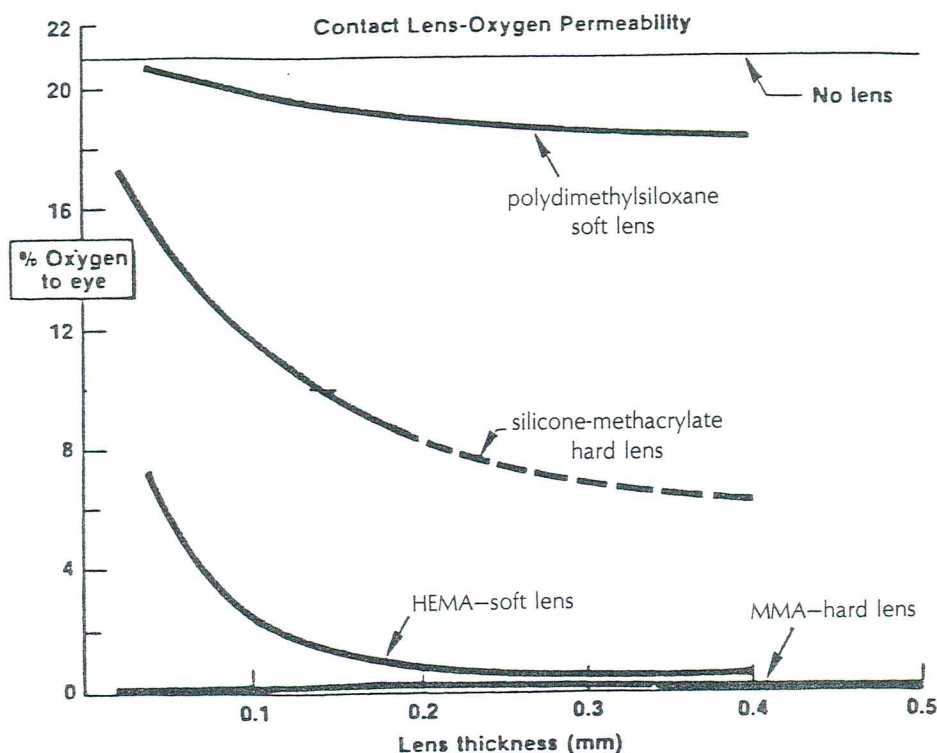


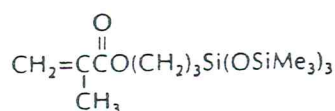
Figure 5.

separate. In order to provide satisfactory dimensional stability the systems are cross-linked.

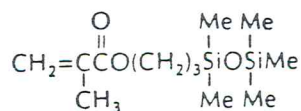
The earliest lens materials were dimethyl-phenylmethyl-siloxane copolymers containing sufficient phenyl groups (10–15 mole percent) to match the refractive index of the polymer to silica reinforcing fillers. These were peroxide cure elastomer systems [69-71]. Lenses based on this chemistry were extremely hydrophobic, causing discomfort and in many instances damage to the corneal epithelium. While the lenses had high oxygen permeability, the permeability of water was also high. Water permeated through the lenses from the surface of the eye. Since these unmodified siloxanes are extremely hydrophobic, water is unable to return to the tissue under the lens and dehydration of tissues along with adhesion of the lens takes place [72].

Approaches toward improving the hydrophilic properties of siloxane backbone lenses have included both plasma and radiation treatment [73-5]. Siloxane polymers can also be blended or grafted with hydrophilic polymer systems such as vinylpyrrolidone and glycidyl methacrylate [76-79]. Surface siloxane bonds have been hydrolyzed by titanates [80].

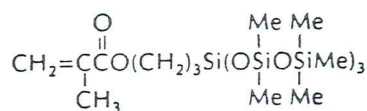
Probably the most significant new technology in silicone hard lens materials is that disclosed by Gaylord for the production of siloxane-modified methacrylate copolymers [81,82]. In the simplest example, methacryloxypropyltris(trimethylsiloxy)silane is copolymerized with other methacrylate monomers, by which means the oxygen permeability of siloxanes is combined with the mechanical and optical properties of the methacrylates. This technology is the basis of the Syntex Polycon® lens, which is said to employ a methacryloxypropyltris(pentamethyldisiloxanyl) silane [83]. Several research groups have found other siloxane-modified methacrylates not described in the Gaylord patent [84-6]. Toyo Contact Lens, for example, has



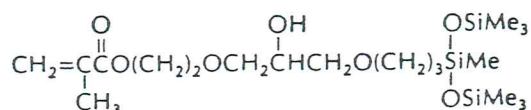
methacryloxypropyltris(trimethylsiloxy)silane



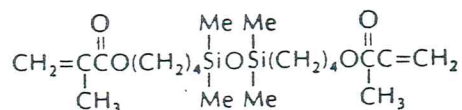
methacryloxypropylpentamethyldisiloxane



methacryloxypropyltris(pentamethyldisiloxanyl)-silane



methyl-di(trimethylsiloxy)silylpropylglycerol-methacrylate



bis(methacryloxybutyl)tetramethyldisiloxane

Figure 6.

prepared siloxane modified hydroxyethylmethacrylate monomers that have pendant hydroxyl groups which suggest improved hydrophilicity [87]. This chemistry has also been translated to soft contact lens systems [88].

Other siloxane-modified methacrylate technology which appears to have considerable obstacles for commercialization is that of Bausch and Lomb, in which a bis(methacryloxybutyl) end-blocking unit is prepared and then polymerized with D_4 [89-92]. This long, oxygen permeable group acts as a difunctional cross-linking unit in the polymerization of a variety of monomers. The result is a soft lens system that is both oxygen permeable as well as hydrophilic. An analogous result with a different technology, reported by Ciba-Geigy, employs a urethane bridge from a polysiloxane to hydroxyethylmethacrylate [49].

While not currently directed toward contact lens formulations, silicone-acrylate

copolymers develop properties that suggest future evaluation [93-5]. Hard contact lenses have been produced by application of high phenyl content silicone casting resin-technology [96]. It also should not be surprising to see some of the silicone-polycarbonate and other oxygen permeable membrane technologies translated to contact lens materials.

Many other thermoplastic silicones have been reported. Materials which appear promising for biomedical applications include silicone-polyamides IPNs [7], silicone-methylstyrene block copolymers [97], and silicone-polypeptide block copolymers [98]. An important review which discusses silicone rubbers from a biocompatibility and surgical approach has recently been published [98].

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