POLYMERIC MATERIALS
AND THEIR APPLICATIONS

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STRUCTURE OF POLYMERIC MATERIALS

Polymer Chain or Backbone

End Group

\[ \text{Repeat Unit or 'MER'} \]

- Value of \( n \) determines molecular weight and properties
- Molecular weight range: A few hundred to millions

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STRUCTURE OF POLYMERIC MATERIALS

Polymers are long chain molecules built up from one or more repeating chemical units. The term MER is derived from the Greek word MEROS which means part or portion. A polyMER is made up of many MER units, an oligoMER contains a few MERs and a monoMER yields a single MER unit. (A MERCOR is a CORporation involved in the development and sale of specialty polymers and biomaterials.)

Monomers and oligomers are starting materials in the synthesis reactions that produce high molecular weight polymers. The n shown outside the brackets is known as the degree of polymerization. n multiplied by the molecular weight of the repeat unit or MER (and added to the MW of the end groups) gives the molecular weight of the polymer. Although polymers can be prepared with molecular weights of several million, most useful commercial polymers have molecular weights between about 10,000 and 1,000,000 grams per mole.

Often the end groups are different chemically than the repeating units that make up the polymer backbone. The identity and concentration of the end groups can affect polymer properties. The effect is most important when the molecular weight is low and end-group concentration is high per unit volume of polymer.
POLYMER BACKBONE MORPHOLOGIES

LINEAR

BRANCHED

CROSSSLINKED

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POLYMER BACKBONE MORPHOLOGIES

The properties and processability of polymers can vary widely depending on the shape of the backbone and the existence and number of intermolecular chemical bonds among the millions of molecules that make up the polymer sample.

Linear and branched polymer backbones with little or no bonding to their neighbors are most easily processed. They can be heated to reduce their viscosity and formed into useful shapes by extrusion and molding. Most linear polymers are also soluble in solvents. Polymer solutions can be used for fabrication by dipping, casting or spraying and as adhesives for bonding. Heat sealability and the ability to reprocess scrap are also common to most linear and branched polymer. Polyethylene, PVC and Dacron Polyester are thermoplastics with linear or branched polymer backbones.

Certain polymers must be crosslinked to attain useful properties. Epoxies and silicone rubbers, although quite different in physical properties, are both crosslinked "thermosets". During or after component fabrication the polymer undergoes a reaction that links adjacent polymer chains together. This can result in the creation of a single giant molecule which is no longer solvent-soluble or fusible at elevated temperatures. Adhesive and solvent bonding become difficult or impossible, as does melt processing.

At high crosslink densities some thermosets become hard and brittle. At lower extent of crosslinking other polymers become good elastomers. That is, they return to their original dimensions after being distorted by an external force.

So far we have seen that the chemistry or the repeating MER unit and the morphology of the backbone chain can both influence polymer properties. We can get even more variations in polymer properties by using more than one kind of repeating unit.
STRUCTURE OF COPOLYMERS

- Repeat Unit A
- Repeat Unit B

Random

Block

Graft

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STRUCTURE OF COPOLYMERS

Copolymers are synthesized in a variety of ways. They differ from homopolymers in that they contain more than one kind of MER unit. Copolymer, terpolymers and even multipolymers are possible. By combining the properties of more than one polymer type, many useful materials can be produced.

The arrangement of the repeat units gives further property variations to the copolymers. In random copolymers, many properties tend to be intermediate between the properties of the two homopolymers. The overall concentration of each monomer determines the final properties but unless one monomer is used in great excess over the other, the resulting properties can be quite different from either homopolymer.

In graft and block copolymers, particularly when graft or block length is high, the properties of the two homopolymers are retained. For instance a hard, high-melting block can be copolymerized with a soft, rubbery block. The copolymer will be a thermoplastic elastomer. At room temperature the liquid-like soft blocks are strengthened and reinforced by the hard blocks or segments. At elevated temperatures, the hard block melts and flows to permit thermoplastic processing. Upon cooling the original structure reforms. The thermoplastic polyurethanes have this structure.

Some graft and block copolymers give unexpected surface properties through preferential population of the surface region by one of the blocks. I'll show later how this can be used to advantage in the design of new biomedical polymers.

We have made many interesting polymers by combining one hard block with two or three different soft blocks. These polymers are block terpolymers or block multipolymers. They have interesting permeability properties and biocompatibility, both of which can be tailored over a wide range by varying block chemistry and concentration.
POLYMER MOLECULAR WEIGHT DISTRIBUTION SHOWING HETEROGENEOUS NATURE OF MAN-MADE POLYMERS

CONCENTRATION

low Molecular Weight Impurities/Additives

Polymer

high MOLECULAR WEIGHT low

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POLYMER MOLECULAR WEIGHT DISTRIBUTION

Unlike low molecular weight chemicals, man-made polymers are never produced with a single molecular weight common to all the polymer molecules in the batch. Instead, polymers have 'polydisperse' molecular weight distributions. That is, a typical polymer sample will contain a statistical distribution of chain lengths. Some much longer than the average and some much shorter than the average.

In addition to all the structural factors I have already mentioned, the shape of this distribution and its average molecular weight, can have pronounced effect on polymer properties. If we were to fractionate a typical polymer sample according to chain length, we might find that the low molecular weight homologues were waxes or even liquids, while the high molecular weight cuts were tough and viscous, even at elevated temperatures.

The macroscopic properties we measure and assign to polymers are really the weighted averages of the properties of the various polymer chains that are present in the sample.

In addition to polymer chains of various length, virtually all polymers contain nonpolymeric impurities. These can include unreacted monomers, catalysts used in the synthesis reaction or additives combined with the polymer to improve stability or processing.

Some of these nonpolymeric species can have a pronounced effect on surface properties and may, therefore, affect biocompatibility. I'll return to this subject later.
CURRENTLY-USED POLYMERIC BIOMATERIALS

INDUSTRIAL POLYMERS WHOSE BULK AND SURFACE PROPERTIES ARE "ACCEPTABLE" FOR USE IN BIOMEDICAL DEVICES

Dacron® Polyester
Lycra® Segmented Polyurethanes
Plexiglass® Polymethylmethacrylate
Tygon® Polyvinylchloride (PVC)
Teflon® Polytetrafluoroethylene

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CURRENTLY-USED POLYMERIC BIOMATERIALS

Virtually all of the polymeric biomaterials in clinical use today are industrial polymers that were adopted for use in medical applications.

Typically a polymer originally developed for industrial or fiber use is used to fabricate a medical device or prosthesis. As evidence is accumulated that the material is reasonably safe, it will be used in other devices. Eventually its use is accepted by the government regulators and the medical community and it becomes known as a biomaterial.

If the material manufacturer expects significant business to result, he may apply special care and additional testing to so-called medical grades of the polymer.

Silastic® silicone rubber from Dow Corning is a good example of this evolution. The silicones were originally developed for high-temperature electrical coil insulation. Following the success of the hydrocephalus shunt other medical applications were found and the medical grade silastics were introduced.

In general, the household names in biomedical materials were not developed to be used as biomaterials. The implication, of course, is that many of these materials may not be giving optimum performance in biomedical devices and prostheses.

We believe that biomaterials, particularly those used in critical or long-term implantable devices, should be designed to suit the application. It's only in this way that we can expect to maximize the safety and efficacy of devices of increasing complexity.

The Jarvik VII heart is constructed from Lycra® spandex elastomer (sold by Ethicon under the name of Biomer® segmented polyurethane). Lycra® was developed to replace natural rubber thread in textiles. It is widely use in pantyhose, underwear and stretch fabrics.

There is a very good reason for this state of affairs.
# Worldwide Consumption of Biomaterials

<table>
<thead>
<tr>
<th>Component</th>
<th>lb/year</th>
<th>MM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dialysis</td>
<td></td>
<td>18.0</td>
</tr>
<tr>
<td>Open Heart</td>
<td></td>
<td>1.1</td>
</tr>
<tr>
<td>Blood Collection &amp; Transfusion</td>
<td></td>
<td>2.6</td>
</tr>
<tr>
<td><strong>Total World Market</strong></td>
<td></td>
<td><strong>21.7</strong></td>
</tr>
</tbody>
</table>

Total World Market = $35 MM

Assumes 21.7 MM lb = 85% of total @ $1.00/lb, Remainder @ $5.00/lb

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WORLDWIDE CONSUMPTION OF BIOMEDICAL POLYMERS

This slide shows three of the largest uses for blood-contacting polymeric biomedical polymers. They were calculated from industry figures for the number of procedures in each category in the USA and doubled to get worldwide usage. The average weight of polymer used in each procedure was then used to get the total weight consumed. The majority of biomedical polymers in these categories are low-cost PVC compounds which sell for about $1.00/lb. The remaining 15% of the poundage may average about $5.00/lb.

If we assume that dialysis, open heart and blood handling make up 85% of the total, the entire market for the unconfigured polymers is probably less than $40 MM. If we double this for non-blood-contacting applications, we can see that the total world biomedical polymer market may be less than $100 MM dollars.

On the other hand, just the US market for all biomedical devices and diagnostic products probably exceeds $10 BB.

The difference in the market for devices and the market for the materials from which they are made is measured in orders of magnitude. That's a lot of added value.

The incentive, then, is in the development of new devices. The materials producers often do better in concentrating their resources on higher volume specialty applications or commodity resins.
ADVANTAGES OF POLYMERIC BIOMATERIALS

Low Fabrication Cost
Low Materials Cost
Wide Range of Mechanical Properties Available
Relative Biocompatibility
Wide Range of Transport Properties Available
Optical Transparency
High Strength-to-Weight Ratio
Good Flex Life
Good Abrasion Resistance
Versatile Component Assembly
Ease of Compounding
Good Aesthetics

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DISADVANTAGES OF POLYMERIC BIOMATERIALS

Thrombogenicity
Tissue Reaction: Inflammation & Capsule Formation
Batch-to-Batch Variability
→ Lack of Understanding/Control of Surface Properties
Biodegradation
Mineralization
Shelf Life (e.g. natural rubber)
Retention of Sterilants/Outgassing
Thermal (Product) Stability
→ Disposal/Incineration (e.g. PVC & fluorocarbons)
Stress Cracking (e.g. acrylics, polycarbonate)
Permeability & Absorbivity

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BIOMATERIALS REQUIREMENTS vs DEVICE SOPHISTICATION

Material-Related Safety & Efficacy

Complex, Chronic Devices
( Artificial Heart )
Failure = Patient Death

Simple Acute Devices
( I.V. Catheter )
Failure = Patient Morbidity

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BIOMATERIALS REQUIREMENTS vs DEVICE SOPHISTICATION

Lack of biocompatibility is seldom implicated when complications occur with simple acute devices such as vascular catheters. If problems do occur, they are usually assumed to be the result of infection, reaction to infused fluids or improper use of the device. Since materials-related failures or complications may not be life-threatening, the need for optimum biomaterials performance is not universally appreciated.

In a more complex and/or chronically-implanted device like an artificial heart, the safety and performance of the device is highly dependent upon the performance of the biomaterials. Clotting, inflammatory response and infection in an artificial heart can result in sudden death or irreversible damage to the patient. Thus, there is a general awareness of the need for good biomaterials in these life-supporting devices.

In fact, all blood and tissue-contacting devices could probably benefit from optimized biomaterials. We believe that the use of superior biomaterials in even the most ubiquitous blood-contacting devices could significantly reduce morbidity and speed patient recovery. What is absolutely necessary for an artificial heart could reduce complications in the use of IV catheters.

We have to acknowledge that economic constraints exist and that new biomaterials should not add complexity to the production process or significantly increase raw materials cost. We also realize that basic studies of the effect of biomaterials on device-related morbidity and patient recovery are needed to document the improvements.
NEED FOR OPTIMIZED BIOMATERIALS IN CRITICAL DEVICES

PACEMAKER LEAD: Biodegradation Resistance, Flexlife, Thromboresistance

HEART VALUE: Biodegradation Resistance, Flexlife, Thromboresistance, Anticalcification

ARTIFICIAL HEART: Biodegradation Resistance, Flexlife, Thromboresistance, Anticalcification, Antimicrobial, Noninflammatory

MERCOR
NEED FOR OPTIMIZED BIOMATERIALS IN CRITICAL DEVICES

I've listed three critical biomaterials applications to show the complexity and the associated difficulty in designing biomaterials for critical biomedical devices.

As we move toward the Bionic man, the requirements placed on the biomedical polymers increase dramatically. We feel that anyone in the business of manufacturing chronically-implanted devices and prostheses needs to maintain basic development expertise in the materials which make these devices possible.

Total reliance on vendors for critical biomaterials can lead to disastrous results. This is particularly true with complex, life-supporting implantables like the artificial heart.
ECONOMICS OF BIOMATERIALS DEVELOPMENT

1. Biomaterials Development Requires A Multidisciplined (Expensive) Approach

2. Total World Market Is < $40 MM / year and Divided Among Many Different Polymers

Result: Very Few New Biomedical Polymers Are Commercialized

MERCOR
ECONOMICS OF BIOMEDICAL POLYMER DEVELOPMENT

Unfortunately, the costs of maintaining a fully-staffed biomaterials development and manufacturing group is very high.

For the materials supplier who sees a relatively small market (and big potential liability) in biomedical uses of his materials, it often doesn't make sense.

As a result, very few new biomedical polymers are commercialized.

To the critical device manufacturer, the up-front expense of developing in-house biomaterials expertise is also high. However, in devices like vascular grafts, polymer heart valves and circulatory support devices we believe that in the long run the expense is justified.
(IN THE ABSENCE OF LEACHABLE COMPONENTS)
BIOCOMPATIBILITY IS DETERMINED BY THE CONSTITUTION OF
THE OUTER MOLECULAR MONOLAYERS

Interface

Surface Layer

Bulk Phase

MERCOR
BIOCOMPATIBILITY IS DETERMINED BY OUTER MOLECULAR LAYERS

Up to this point I've said very little about the development of new biomedical polymers and the factors which affect their in vivo response.

For the purposes of this discussion I'll restrict my remarks to so-called smooth, blood-contacting polymers. Unlike textured blood-contacting surfaces, the smooth surface is designed to have a minimal affect on the flowing blood in terms of thrombus formation and thromboadhesion.

There is good experimental evidence that the blood-materials interactions that occur at a smooth surface are only affected by the constitution of the outer one or two molecular monolayers of the polymer. That means that as long as the polymer doesn't contain any leachable impurities, the chemistry of the bulk polymer, which is distant from the biological interface, doesn't affect in vivo performance.

For this reason, investigators concentrate on the surface region with great interest.
LEVELS OF UNDERSTANDING IN APPLIED BIOMATERIALS SCIENCE

1. \textit{In Vivo} Response = f(Surface Chemistry)

2. \textit{In Vivo} Response = f(Bulk Chemistry)
   a. Level 1
   b. Surface Chemistry = f(Bulk Chemistry)

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LEVELS OF UNDERSTANDING IN APPLIED BIOMATERIALS SCIENCE

We like to think in terms of two levels of understanding in applied biomaterials science.

On one level we try to understand the relationship between the measured surface chemistry and the \textit{in vivo} response of the material. This task is complicated enough, due to limitations of existing surface analytical instruments.

For several reasons we feel a higher level of understanding is necessary in developing functional relationships between \textit{in vivo} response and the bulk chemistry of the polymeric material.

If we can understand how easily-made changes in bulk composition can change the surface chemistry of a polymer, we can use that ability to control \textit{in vivo} response, even if we have an incomplete understanding of \textit{why} a particular surface chemistry works.
SURFACE-ACTIVE CONTAMINANTS THAT MAY BE PRESENT IN POLYMERIC BIOMATERIALS

POLYMERIZATION SURFACTANTS
CATALYSTS
SOLVENTS
STABILIZERS
PLASTICIZERS
INTERNAL LUBRICANTS
ANTIBLOCKING ADDITIVES
DEGRADATION PRODUCTS
PROCESSING EQUIPMENT LUBRICANTS
SILICONES

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SURFACE-ACTIVE CONTAMINANTS IN POLYMERIC BIOMATERIALS

The study of polymer surface chemistry is complicated by the fact that many commercially-available polymers contain additives or impurities which are surface-active.

A surface-active agent is capable of migrating to an interface and populating that interface at a concentration that is much higher than its average concentration in the bulk phase. Extremely surface-active materials can have nearly 100% concentration in a surface, even if their initial bulk or average concentration in the polymer is in the parts per million range.

Trying to interpret the surface analysis of a polymer contaminated with an unknown substance is very difficult. In the absence of sensitive surface analysis, a sample's biological response may be wrongly assigned to the base polymer, when it is largely due to a contaminant. Processing or thermal history variations can lead to variability in in vivo performance if it produces differences in the amounts of additive or impurity in the surface. Finally, correlating surface chemistry to bulk chemistry is very difficult in the presence of unknown surface-active contaminants.

This subject may seem to be one of minor importance in the development of new biomedical polymers, but it is so common that it is often a major determinant of biological response.

The news is not all bad, however. It is possible to design surface-active molecules and intentionally add them to a base polymer for the purpose of improving biocompatibility.
SURFACE SEGREGATION OF LOW SURFACE ENERGY BLOCK IN BLOCK OR SEGMENTED COPOLYMER

AIR

(Low Surface Energy)

SOFT BLOCK

Surface

Bulk

HARD SEGMENT

(High Surface Energy)

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SURFACE SEGREGATION OF BLOCKS OR SEGMENTS

As I mentioned earlier, certain block and graft copolymers can add additional complexity to the relationship between surface chemistry and bulk chemistry.

Solids and liquids try to minimize their surface or interfacial energy. This is the same driving force that causes low energy surface-active impurities to migrate to the air-facing surface of a polymer. Since air is a low-energy fluid, the interface between air and the polymer will have the lowest energy when the polymer surface also has a low energy. Migration of the surfactant to the polymer surface succeeds in lowering polymer surface energy and therefore overall interfacial energy.

In many block and graft copolymers, another mechanism for interfacial energy minimization exists. By reorientation of the surface molecular layers, one of the blocks or grafts can preferentially populate the surface. Since the polarity of the block or grafts in a given copolymer are never identical, the block whose presence in the surface minimizes interfacial energy will often be found in highest concentration.

For instance, when brought to equilibrium in air, a block copolymer of the type shown in this slide will have a surface which is mostly comprised of the so-called soft block or low surface-energy block. It is even possible that none of the more polar hard segment will be present in the polymer. This conformation exist as long as the interface is the same.

If the polymer is put into the blood stream, it is exposed to the more polar environment of the blood. The polymer may then attempt to reorient its polar hard blocks toward the surface in order to minimize interfacial energy.

Thus, the design and manufacture of new biomedical polymers must take into account the possible migration of surface active species and the reorientation of the surface layer that can occur in different environments.

For most polymers, it is better to think of them as having dynamic, liquid-like surfaces, instead of rigid, unchanging surfaces that are fixed forever at the time they are fabricated.
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DILEMMA OF BIOMEDICAL POLYMER DEVELOPMENT

A. Bulk Properties \( = f \left\{ \begin{array}{l}
\text{Molecular Structure} \\
\text{Molecular Weight}
\end{array} \right\} \)

B. Surface Properties \( = f \left\{ \begin{array}{l}
\text{Molecular Structure} \\
\text{Molecular Weight} \\
\text{Impurities}
\end{array} \right\} \)

Optimum A = Optimum B

Solution: Coating
Grafting
Surface Treatment
Blending

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DILEMMA OF BIOMEDICAL POLYMER DEVELOPMENT

With the complexities I have just discussed, how do we go about developing new biomedical polymers?

All biomedical polymer applications have requirements which can be divided into bulk and surface property categories. An elastomer for an artificial heart, for instance, must have good bulk mechanical properties like flex life, toughness, appropriate softness and so forth. This same polymer must also have a surface which does not cause blood to clot or the adjacent tissue to become inflammed.

In many classes of polymers, the relationship between the molecular variables and bulk properties is fairly well understood. A reasonably systematic (if somewhat empirical) process may be used to achieve the desired bulk properties. In the case of surface properties, their relationship to the variables which can be manipulated by the materials scientist is less well known and is clouded by the influence of the ever-present impurities.

However, even if a precise functional relationship were known between surface properties and first-order molecular variables, another problem would exist. The chances are remote that an optimum in both surface and bulk properties could be found at one molecular structure and molecular weight.

Historically this basic dilemma of biomaterials development has been handled through a variety of surface treatments or coatings applied after the device or component is fabricated. The method we prefer involves a simple blending step before fabrication of the surface.
DISTRIBUTION OF SURFACE MODIFYING ADDITIVE (SMA™) IN BASE POLYMER

Before Surface Formation

After Surface Formation and Equilibration

In Blood

In Air

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DISTRIBUTION OF SMA IN BASE POLYMER

We do this by taking advantage of two mechanisms by which condensed phases of matter minimize their interfacial energy: the migration of species from the bulk to the surface and reorientation of surface molecules.

We synthesize novel copolymers and terpolymers we call Surface-Modifying Additives or SMAs. A small amount of SMA is blended with the base polymer before device fabrication. During and after fabrication, the SMA migrates to the surface in high concentration. This dramatically changes the outermost molecular monolayers, which we already said are the ones that determine biocompatibility.

The SMA is high in molecular weight and partially compatible with the base polymer so it remains permanently anchored to it. So little of the SMA is required to achieve the desired change in surface chemistry that the original bulk properties are preserved.
MOLECULAR STRUCTURE OF MERCOR POLYMERIC SURFACE MODIFYING ADDITIVES (SMA)

Surface-Active Hydrophobic Block

\[
\begin{align*}
\text{Nonpolar Block}^x & \quad \text{High CED (hard) Segment}^y \quad \text{Polar Block}^z \\
\end{align*}
\]

Hydrophilic Block

Vary \(x, y, z\) & \(n\) to vary polar/nonpolar balance, surface-activity, rheology etc.

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MOLECULAR STRUCTURE OF SMA

We design the SMAs to be amphipathic in structure. That is, they have both polar and nonpolar blocks connected by short hard blocks which permits them to reorient to minimize interfacial tension as their environment changes.

We believe that the SMAs' surface activity and polar/nonpolar structure are responsible for their ability to improve blood and tissue compatibility. The denaturing of plasma proteins has been implicated in surface-induced thrombosis. Plasma proteins are present in high concentration in the blood and adsorb onto surfaces immediately. We believe that if the interfacial energy between blood and the surface is minimized, that thromboresistance can be maximized.

Regardless of the strength of our hypothesis, the SMAs can impart a high level of thromboresistance to many different base polymers in a reproducible and cost-effective way.
SIMPLIFIED BIOMATERIALS DEVELOPMENT USING SURFACE MODIFYING ADDITIVES (SMA)

**Bulk Properties**
- Base Polymer Synthesis
  - Physicochemical Characterization
    - OK?
      - Yes: Base Polymer
      - No: No
  - 99%

**Surface Properties**
- SMA Synthesis
  - Surface Analysis and In Vivo Testing
    - OK?
      - Yes: SMA
      - No: No
  - 1%

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Simplified Biomaterials Development with SMA

The use of Surface Modifying Additives greatly simplifies the development of new biomedical polymers. Because the SMA determines surface properties independent of the base polymer, the development effort can be divided into two separate parts.

In one part, the base polymer is chosen or synthesized to satisfy the bulk property requirements. In another part of the development effort, the SMA is developed to maximize biocompatibility and may even be optimized in a different base polymer. Of course, in vivo testing is a big part of the SMA development process.

To complete the biomedical polymer development the SMA and base polymer are literally combined to obtain a material that satisfies both surface and bulk requirements. Often, the same SMA can be used in a variety of base polymers, giving each of them the same level of biocompatibility.
TECHNOLOGY OF SURFACE MODIFYING ADDITIVES IN THERMOPLASTIC BASE POLYMERS

~ 99% BASE POLYMER  ~ 1% SMA™

Extruder

Biomedical Polymer

MERCOR
TECHNOLOGY OF SURFACE MODIFYING ADDITIVES

The use of SMAs in the production of real biomedical devices is very simple. Before the device is made, the base polymer is blended with SMA and pelletized (or a base polymer solution can have SMA added to it).

The base polymer is then processed in the usual way to make the device. The surface modification generally develops spontaneously, but surface migration may be accelerated by storage at elevated temperature.

Since no post-fabrication coatings or surface treatments are required, rejects are eliminated and incremental expense is minimized. The process is extremely reproducible and can easily be monitored with simple contact angle measurements.
CONCLUSION

I freely admit that my predictions for the future of biomedical polymers are influenced by my own activities in the field.

I do think that despite a dramatic increase in biomaterials research in recent years, industrial polymers will continue to be used in the majority of acute biomedical devices. It will take a lot of study to establish the benefits of improved materials in these devices (as measured by reduced patient morbidity). In the meantime, short-term economic considerations will rule and low cost will win out over superior biocompatibility.

For example, after years of discussion of the possible harmful effects of leachable plasticizers in PVC blood bags and tubing, polyvinylchloride disposables may finally lose favor due to environmental problems. The incineration of PVC produces toxic and corrosive by-products. Because it is the most commonly-used biomedical polymer, pollution is considered a major drawback to its continued use. In Europe there is talk of banning PVC disposables in the near future. If this occurs, there will be a major shake-up as device producers move to more environmentally-safe polymers.

Bioactive coatings applied to device components after their fabrication continue to be developed. Although they add an expensive step to the manufacturing process, they have the advantage of drug-like activity. The covalent bonding of heparin or other anticoagulants can actively prevent thrombus formation at the blood-contacting surface. Improvements have been made in preserving the activity of surface-bound heparin. If permanence of the coating can be improved, these heparin-bonded surfaces could be used in cardiopulmonary bypass to reduce complications associated with extended use of the 'heart-lung machine'.

Many new copolymers have been evaluated for use as biomedical polymers. Those with mixed polar/nonpolar structure permit the systematic tailoring of composition, in the search for an optimum in biocompatibility.

For nearly twenty years there has been interest in a copolymer which could combine the strength and easy processing of thermoplastic polyurethanes with the biocompatibility and in vivo stability of silicone rubber. New work with reactive silicone
oligomers has led to practical methods of synthesizing these very interesting biomedical polymers. The silicone-urethanes should find uses in both acute and chronically-implanted devices, but they are not yet commercialized.

Finally, we hope to see the widespread use of our Surface Modifying Additives in a variety of blood and tissue-contacting devices. Our first clinical application of these materials is in the Pierce-Donachy Ventricular Assist Device, which has shown excellent thromboresistance. The SMAs can be used in a variety of existing base polymers and so represent a cost-effective way to improve biocompatibility of a variety of existing devices without the expense of developing entirely new materials.

Although we need more studies to prove it, I firmly believe that an across-the-board improvement in the biocompatibility of commonly-used and more exotic biomedical devices will lead to dramatic reductions in patient morbidity and significantly reduce the cost of health care.
BIOGRAPHY

Robert S. Ward is a Chemical Engineer and President of MERCOR Incorporated, a wholly-owned subsidiary of Thoratec Laboratories Corporation located in Berkeley, California. MERCOR is a developer and manufacturer of high-performance specialty polymers and polymer compounds. MERCOR was formerly the Biomaterials Division of Thoratec Laboratories and biomedical polymers remain a major area of concentration within the company.

Mr. Ward holds several foreign and U.S. patents covering (biomedical) polymer compositions and process technology. At MERCOR, he has supervised the development of a family of high flex-life polyurethaneureas, high moisture-vapor-permeable wound dressings and textile coatings, shape-memory thermoplastics, and a novel method of polymer surface modification which significantly reduces platelet uptake of blood-contacting biomaterials. He has directed the development and production of a number of polymers and biomedical devices including the Pierce-Donachy VAD and a family of polyurethane vascular grafts. Mr. Ward was Principal Investigator for the NHLBI-DTB contract, Procurement of Primary Reference Materials, under which Thoratec produced polymer reference standards for worldwide distribution to investigators of blood-materials interactions.

Prior to joining Thoratec in 1979, Mr. Ward was Director of Research at AVCO Medical Products, the first manufacturer of Intraaortic Balloon Pumps. While at AVCO, Mr. Ward performed structure property/studies on Avcothane-51® which led to his development of several new biomaterials. From 1971 to 1976, Mr. Ward supervised the engineering scale-up and production of Avcothane-51® and produced biomedical blood-contacting devices and prototypes which employed various thermoplastic biomaterials.

He is the author of many publications in the field of biomedical polymers and is a member of the Society of Biomaterials and the American Chemical Society.