

NONWOVEN TECHNOLOGIES FOR MEDICAL IMPLANTS

Martin Dauner

Institut für Textil- und Verfahrenstechnik
Denkendorf, Germany

Introduction

The 20th century may be regarded as the century of endoprostheses. In preceding times only few people were treated with tissue replacements or even supporting structures inside the body. The application of “biomaterials” was almost limited to sutures, wound dressings and crutches. In some cases gold and ivory have been used for teeth and bone substitutes. In the late 19th, but mainly in the 20th century treatment of completely failed organs became possible by more or less self sufficient artificial implants. These implants require an integration to the surrounding tissue at the interface but take over the function of the replaced organ fully. As examples endoprostheses like hip or knee prostheses may be mentioned which transfer the applied load passively mimicking geometrically and mechanically the replaced structure. Besides the general demand on the biocompatibility the requirements on these prostheses are appropriate shape, stiffness and strength. Active implants like pacemakers are even more depart from the original organ as they function by electronic or biochemical ways. More or less inert materials like stainless steel and titanium, ceramics and high performance polymers have been introduced successfully as biomaterials. Despite the high degree of function and reliability one has reached today with these implants, there are a wide areas in which artificial organs cannot replace sufficiently the biological tissue.

At the end of the 20th century a new strategy for tissue replacement has been invented called tissue engineering where the rebuilding of the biological organs is promoted actively or passively with the aid of artificial “interactive” devices.

As an early approach the vascular prostheses, made of a porous knitted or woven tube, act as a scaffold for sometimes vascularised connective and scar tissue which penetrates the textile. Clotting blood makes the vessel tight and build the so called neointima as a quite fair substitute for the endothelium. Today one tries to promote the replacement of an injured tissue or organ by a new but identical tissue. Now the implants have to act as a housing and a scaffold rather than a self sufficient artificial organ. The bodies own cells are the main actor. For this application the “weak” textiles, preferably nonwovens, perform best by providing just space. Now resorbable polymers and ceramics are favorite materials, but in distinct cases non degradable polymers are required.

Where the replacement of tissues by endoprostheses failed, now in the 21th century new cartilage, skin and endocrine organs become realizable by tissue engineering with the help of nonwovens.

2. Nonwoven technology

The applications of nonwovens are characterized usually by their high porosity, large fiber surfaces and the absorbance capability for other substances. Regarding implants the following fields are covered by nonwovens:

- nonwovens as drug carrier or delivery system;

- the use of the (semi-) permeability of nonwovens: patches for defect covering or wound dressing
- nonwovens as scaffolds for tissue engineering.

What makes nonwovens suitable for tissue engineering? Cells need a well defined space as information how to organize themselves. Few as possible foreign material should claim this space. The cells shall arrange themselves in this space but most of them should not attach flat to any surface. Naturally they are generating such a space as the “extracellular matrix” which mainly consists of collagen. This extracellular collagen matrix prepared for scanning electron microscopy presents itself as a nonwoven structure.

For temporarily or long term substitution of the extracellular matrix high voluminous nonwovens are used. They are characterized by a high to very high porosity up to 98 %, i.e. only 2% polymer per volume. The pore size varies depending on the process and the processing parameters from 0,1 µm to 100 µm or even larger. The fiber size is about 1 to 15 µm, to low for flat cell adherence. The strength of these nonwovens is limited but usually sufficient for handling and mechanical stimulation. Their elasticity may be low to high depending on the used polymer, the fiber fineness and the processing.

Various processes for the manufacturing of nonwovens are known. Often the standard processes are to be modified for the specific need of implants. One main factor is the size of the device. Commercial nonwoven processes have high output rates. For example a typical melt blow equipment of 4 m width and 4000 capillaries extrude 400 kg polymer per hour. We have developed an one capillary melt blow tool for implants. Its production rate is 120 g/h.

2.1 Staple fiber nonwoven process

The staple fiber nonwoven process is the conventional process which is known from nonwovens of natural fibers as wool and cotton. For internal medical applications man made fibers are used. Polymers are melt or solution spun to filaments which are stretched to high orientation. The filaments are cut to staple fibers of 20 to 80 mm length, carded, crosslapped and finally strengthened. For implant applications where high volume nonwovens are required, calendaring as well as chemical or thermo-bonding are less suitable because they reduce the porosity. We prefer a needling process which interlace the fibers by barbed needles. This process is possible with flat sheets as well as with tubes.

For fiber stretching as well as for the carding usually spinning preparations are required which are not easily extracted from the final nonwoven. Thus we are washing the fibers after cutting. The use of non toxic spinning preparation is obligatory anyway. At the card we use an electrostatic discharging which allows the production at limited speed without spinning preparation.

The staple fiber nonwoven process is an expensive and time consuming process with many production steps. As a special feature it offers to mix materials so as degradable with non degradable or different degradable polymers with graded degradation times.

2.2 Spunbonding process

To avoid the use of spinning preparations we have introduced a spunbonding process with online needling. Usually spunbonded nonwovens are strengthened by calendaring or melting adhesives and solved bonding agents respectively. These

processing aids reduce the porosity, change the surface properties of the nonwoven and impair the biocompatibility.

In the spunbonding process polymer is spun to filaments which are stretched online by a draw off jet and laid on a conveyor belt which delivers the nonwoven to the needling machine. If required cold calendering may provide the light nonwoven with some strength for the needle punching.

Flat high porous nonwovens can be produced very economically. Spinning preparations are not required. Any fiber forming thermoplastic polymer may be used; crystalline polymers are preferred.

Using degradable polymers like polyglycolic acid or polylactides we have found that they may shrink dramatically when they are brought into physiological environment in vitro or in vivo, i.e. 37°C in watery solution. This could be shown to be due to a depression of the glass transition of the polymers below 37°C. Shrinkage means not only reduction in size but more important a change in porosity and pore sizes. We are actually up to investigate the spinning parameters to eliminate this shrinkage.

2.3 Melt blown process

The melt blown process is the most simple way to produce nonwovens. A molten polymer is delivered through capillaries like at the filament spinning. A high speed hot air stream is pulling out the polymer from the capillaries which forms fine fibrils by that way. The fibrils are delivered to a support system by the air stream. They stick together due to residual heat and motion energy generating at once the almost finished nonwoven.

The melt blown system is unique to form complex shaped implants like tubular prostheses or even an auricle. For tubular prostheses the fibrils are wound on an rotating mandril. The auricle is formed on an accordingly shaped matrix which is to be moved by a 6 axes system to have an even distribution and orientation of fibrils.

All fiber forming polymers but especially elastomeric materials like polyurethanes can be processed. But as the orientation and crystallinity of the polymer molecules are generated only partially, shrinkage may occur as with the spunbonding process.

2.4 Solution spraying process

The solution spraying process have been developed mainly for small scale medical applications as the large amount of solvents is a limitation under economical and ecological views.

The process resembles the melt blown process. The solved polymer is delivered through a capillary and pulled out by an air stream which evaporates the solvent by vacuum produced by the jet injection effect. Fibrils are formed which stick together by residual solvent. The processing can be performed at ambient temperature.

The process has been developed for tubular prostheses but is extended now to flat sheets. It could be shown that amorphous and elastomeric polymers are best suitable for this process. But in any case they must be soluble in any reasonable solvent.

The size of the fibrils is about 1 μm and lower; the nonwoven is comparably tight with pore sizes of 0,1 μm to 10 μm and a porosity of 70 – 90%.

2.5 Wet laid nonwoven

One of the conventional nonwoven processes, the wet laid process, is rarely used for medical devices.

Fibers are dispersed and isolated in a liquid, usually water. A conveyor belt takes the randomly oriented fibers off the liquid. The fibers are dried and bound usually

thermally or chemically. This process may be used to produce nonwoven scaffolds from natural polymers.

3 Materials

General requirement on materials for implants is their biocompatibility which is to be determined for each application. Thermoplastic or soluble polymers can be processed to nonwovens. They are to be chosen depending on the intended application and the process. The spunbonded and the staple fiber process require crystalline polymers to avoid shrinkage during sterilization and at use. Solution spraying works best with amorphous polymers because residual solvent sticks in the amorphous phases and glue the fibrils together. Elastomeric polymers are preferably processed by melt blown or solution spraying.

3.1 Non resorbable polymers

Polyethyleneterephthalate (PET) and polypropylene (PP) are the most used polymers in the melt spinning processes as for technical applications.

High elastic structures are produced from segmented thermoplastic polyurethanes both by melt blown and by solution spraying.

3.2 Resorbable polymers

A great progress has been made with the synthesis of resorbable polymers first for suture materials.

Resorbable materials can dissolve from the body - or even from the in-vitro cell culture - when a new functional tissue has grown. Preferably the hydrolysable polyesters of the α -hydroxycarbonic acids are used: polyglycolic acid (PGA) and polylactic acid (PLA) and also a number of copolymers. All these polymers are melt processable but polyglycolic acid should not be processed by solution spraying.

The degradation rate (loss of strength) of these polymers range from 2 weeks (PGA) to over 52 weeks (P-L-LA). Typically the resorption – the complete loss of mass – takes three more times. The long degradation time of the poly-L-lactic acid (P-L-LA) can be reduced by γ -irradiation as it is used for sterilization.

3.3 Shrinkage

Shrinkage turned out to be an important problem of nonwovens for implants namely with resorbable polymers. Shrinkage reduces the porosity and the pore size and it changes the dimensions which must be avoided for shaped implants.

In the body environment the glass transition of these resorbable polymers is depressed to about the body temperature due to water uptake. Shrinkage is related to the amorphous phases in a polymeric material and to residual stresses, which are produced by the fiber spinning. It happens at heating above the glass transition because secondary intermolecular bonds become weak. It does not occur at crystalline materials when the binding forces in the crystalline phases are sufficiently strong. That means, shrinkage can not be avoided for amorphous polymers if they are under internal stresses; fully oriented (highly crystalline) yarns do not shrink; the shrinkage can be avoided for pre-oriented yarns by thermal treatment.

With the high oriented yarns used for staple fiber nonwovens shrinkage can be fully avoided. In the solution spraying process the low orientation of the polymer molecules and their possible complete relaxation during the processing allows the production of nonwovens free of shrinkage.

Nonwovens by the spunbonding process may show important shrinkage depending on the polymer and the process parameters. While with poly-L-lactide we succeeded to avoid the shrinkage by high orientation and crystallisation during the process; with polyglycolide up to 50 % shrinkage occurs. Additional thermal treatment is required to stabilize the nonwoven; yet it changes the nonwoven structure as well.

Melt blown nonwovens of resorbable polymers show the most dramatic shrinkage. Some copolymers shrink already during processing. The thermal treatment of complex shaped prostheses is rather complicated, sometimes impossible. So, hard work is still required to learn more about the shrinkage effect and how to avoid it.

4 Applications

Nonwovens may be used as implants at any application where the healing and rebuilding of the bodies own tissue shall be promoted. The process to be chosen depends on the application and the required way of interaction. The application of nonwovens range from patches from polyurethane for dura mater replacement which are produced by solution spraying and vascular prostheses as well from polyurethane by solution spraying over hemostyptic nonwovens from resorbable polymers to scaffolds for tissue engineering. Examples of the latter most fascinating field shall be given more in detail:

4.1 Tracheal prostheses

Tracheal prostheses were produced first by the solution spraying technology. The tubular nonwoven was reinforced by integrated horse-shoe shaped clasps according to the cartilage clasps of the natural trachea. The pore size on the outer surface allowed the ingrowth not only of tissue but of small blood vessels. The surface in the lumen had a porosity lower than $0,1 \mu\text{m}$ in order to avoid the penetration of bacteria from the contaminated breathing air.

In the 1990 the strategy was changed to a biohybrid tracheal prosthesis which shall have the natural layer of ciliated cells on the lumen surface. Bacteria tightness is no longer required. For hybridisation the prosthesis will be implanted in a well vascularized area of the patient's body. After ingrowth of tissue which is to cover the inner surface of the prosthesis, in-vitro cultivated ciliated epithelial cells are injected into the lumen where they seed on the inner surface. As soon as the cells have fixed themselves the prosthesis will be put in situ to replace the damaged tracheal segment.

For this application larger pore sizes are necessary to allow the complete penetration of body tissue. These pore sizes of about $150 \mu\text{m}$ could not be produced accordingly by solution spraying process. Thus the modified melt blowing process has been developed.

4.2 Cartilage repair: scaffolds for in-vitro hybridisation

In cell cultures cells grow only as a monolayer on the surface of the culture dish. If they are adapted to grow in the 3rd dimension they still lack of information of the intended shape and size for the transplant. The nonwoven offers this information besides a large (fiber) surface.

Today resorbable nonwoven scaffolds are mainly processed by staple fiber process which offer a high pore volume, fine fibers and no shrinkage. The needle punching assures the required strength. A sufficient number of cartilage cells is brought into the nonwoven structure, where they attach to the fibers and start to produce collagen and other components of the extracellular matrix. Specific articular cartilage can be

preferably generated by additional mechanical stimulation. When the new cartilage is formed the polymer should resorb prior to the implantation of the cartilage. A scaffold for the auricle shall be formed by melt blowing. Yet here the problem of shrinkage still limitates the use of resorbable polymers.

4.3 Biohybrid liver assist device

The liver has some ability to recover. At acute liver failure a temporary assist device bridges the time for recovery or the time to get a liver transplant. The purification of blood at the renal dialysis is accomplished extracorporally only by technical means. The more demanding tasks of the liver in blood purification, protein synthesis and other metabolic activities can not be simulated by artificial techniques. Here an extracorporal system was designed using liver cells which undertake the elimination and/or conjugation of metabolites.

In a dialysator housing a melt blow polyurethane nonwoven containing polypropylene capillaries is arranged. Hepatocytes will be cultivated in the nonwoven. Through the PP capillaries the hepatocytes are provided with oxygen. The blood flow is directed through the nonwoven passing the liver cells. The liver cells metabolize toxic substances of the blood. The blood will be delivered back purified to the patient's body.