Use of porous polyurethanes for meniscal reconstruction and meniscal prostheses


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In the past, porous materials made of an aromatic polyurethane (PU) were successfully used for meniscal reconstruction in dogs. Since aromatic PUs yield very toxic fragments upon degradation, a linear PU was synthesized by curing a poly(i-caprolactone) and 1,4-trans-cyclohexane diisocyanate based prepolymer with cyclohexanedimethanol. Porous materials of this polymer were also implanted for meniscal reconstruction. The results were comparable with the most successful implant series so far. Additionally, a porous meniscal prosthesis was developed to replace a total meniscus. Due to the very high shear stresses to which the prosthesis would be exposed, the stress hysteresis phenomenon linear PUs are known to exhibit could be of great consequence. Therefore an aliphatic PU network, synthesized by cross-linking poly(i-caprolactone) and 1,4-trans-cyclohexane diisocyanate with glycerol, was used. Dislocation caused by tearing out of the sutures was found to be a problem because the tear resistance of the material was relatively low. In this study the tearing problem has been partly circumvented by using a complex suturing technique. Meniscal prostheses turned out to induce fibrocartilage upon implantation, and degeneration of articular cartilage was less severe than after meniscectomy.

Keywords: Polyurethanes, porous material, meniscal reconstruction, tear strength

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Damage or tear of a meniscus usually occurs when fibrocartilage is exposed to abnormal shear stresses. This happens when the weight bearing joint is subjected to a combined flexion–rotation or extension–rotation motion. It has been known for years that healing of the meniscus is limited to the vascularized peripheral part of the meniscus. Only lesions through the synovial attachment of the meniscus and mid-substance lesions extending into the synovium may heal. After total meniscectomy the tissue that replaces the excised meniscus appears not to be fibrocartilaginous but fibrous tissue, which is functionally inferior. Therefore, nowadays a partial arthroscopic meniscectomy is the most frequently performed orthopaedic operation for a torn meniscus.

Nevertheless, partial absence of the meniscus also increases the stress on the articular cartilage and enhances development of arthritic changes. Accordingly, reconstruction of the meniscus or use of a meniscal prosthesis is of great importance. Much research has been carried out into the possibilities of surgical repair and use of a meniscal prosthesis. Suturing was found to be successful only when the lesion extended to the vascular part of the meniscus. In the case of synovial flap reconstruction, in which a flap of synovium was sutured into the defect, rabbit meniscus sometimes showed healing but dog meniscus exhibited only ingrowth of fibrous tissue and the ingrowth of tissue did not extend deeply. Using a cryopreserved allograft as a prosthesis is, apart from the potentiality of an immune response and the practical problems of transplantation, not very efficacious. After transplantation the viable cells are restricted to the surface and the areas immediately adjacent to the peripheral attachment. Collagen-based prostheses have a relatively short degradation time of about 6 weeks, so the toxic cross-link agent glutaraldehyde, which is released at degradation, can be of considerable consequence. In addition, the induction time of fibrocartilage formation was found to be about 20 weeks, which means that the degradation rate of the collagen-based prosthesis is much too fast for this application.
One of the most successful meniscal reconstruction methods seems to be the use of a porous biodegradable implant as a connection between the synovial lining and the lesion\textsuperscript{[13-15].} The implant stimulates the ingrowth of blood vessels and cells into the defect and healing of a lesion in the avascular part of the meniscus is possible. In the past, one series of implants made of an aromatic polyetherurethane was very successful. However, this polyurethane (PU) might yield aromatic diamines upon degradation, which are very toxic and carcinogenic\textsuperscript{[16].} Therefore, in this study, a poly(ε-caprolactone) and 1,4-trans-cyclohexane diisocyanate based prepolymer was cured with cyclohexanedimethanol in order to synthesize a linear aliphatic PU. Porous materials of this polymer were produced and implanted into menisci of dogs.

Ingrowth of fibrocartilaginous tissue into the implants is obtained by transformation of ingrowing fibrous tissue rather than by direct ingrowth\textsuperscript{[17].} For this reason the possibility of using porous meniscus prostheses for the replacement of complete menisci by the formation of a fibrocartilaginous meniscal replica was investigated. Due to the very high shear stresses to which the prosthesis will be exposed, the stress hysteresis phenomenon that linear PUs are known to exhibit\textsuperscript{[17]} could be of great consequence. During deformation, the physical cross-links can undergo a rearrangement which causes stress-softening and leads to permanent deformation. The phenomenon may shorten the time to material failure. This problem can be overcome by chemical cross-linking. Therefore, an aliphatic PU network, synthesized by cross-linking the poly(ε-caprolactone) and 1,4-trans-cyclohexane diisocyanate based prepolymer with glycerol, was used for this application.

### EXPERIMENTAL DETAILS

#### Materials and methods

Cyclohexane dimethanol (CHDM) and glycerol were vacuum distilled prior to use. 1,4-Dioxane was distilled from sodium. Elate 108 (Akzo Nobel Chemicals, Deventer, The Netherlands), a poly(ε-caprolactone) polyol and cyclohexane diisocyanate (CHDI) based prepolymer, was used directly from the supplier without further purification.

Estane 5701-F1, a 4,4'-diphenylmethane diisocyanate (MDI) based polyurethasterethane, was obtained from B.F. Goodrich Chemical Co. Estane films were obtained by evaporation of the solvent (1,4-dioxane). Elate and CHDM or glycerol were mixed ([OH/INCO] = 1) in 1,4-dioxane. Before curing, the solvent was removed by evaporation under a nitrogen atmosphere. Thin films were obtained by reaction in a Petri dish at 100°C under a nitrogen atmosphere for 48 h and post-curing at 110°C for 24 h using 0.001% stannous octoate as a catalyst.

Calorimeter studies were carried out on a Perkin Elmer DSC 7 calorimeter. The scanning rate was 10°C min\textsuperscript{-1} in the range of -100°C to 260°C.

Stress-strain and compression curves were determined at room temperature using an Instron (4301) tensile tester equipped with a 10 N or 100 N load-cell at a crosshead speed of 12 mm min\textsuperscript{-1}. For stress-strain measurements, specimens of 15 x 0.75 x 0.25 mm were cut from thin films. For compression, cylindrical specimens with a diameter of 10 mm and a length of about 6 mm were cut out of the foams by cooling them with liquid nitrogen. For determination of the compression behaviour of meniscal tissue, cylindrical specimens with a diameter of 6 mm and a thickness of about 2 mm were cut out of a meniscus of a dog. The menisci were not treated and the measurements were carried out under wet conditions.

For determination of the tear strength, test trouser specimens 3.75 cm long, 1.25 cm wide and with a longitudinal slit of 2.50 cm were used. The width of the tear path was 1.25 cm. The thickness of the test piece was 0.33 ± 0.04 mm. During testing the force was applied normal to the plane (Figure 1(a)) with a crosshead speed of 250 mm min\textsuperscript{-1} (ASTM D 1938-62). Mainly the trouser specimens tear along the direction of the cut (Figure 1(b)). When this was not the case (Figure 1(b2) and (b3)) the experiment was excluded.

An ISI-DS-130 scanning electron microscope was used for studying the pore structure of the porous materials.

For light microscopy, meniscal reconstruction implants and meniscal prostheses were fixed in formaldehyde and embedded in glycol methacylate\textsuperscript{[18].} Sections (2 μm) were stained with toluidine blue and Giesma.

Degradation of the polymer as a function of implantation time was determined upon stained slides using the Quantimet 520 Image Analysis System. The percentage polymer was measured using a magnification of two. The average value was assessed out of seven measurements. The apparatus correction factor was

![Figure 1](image-url)  
*Figure 1* Method for measuring tear strength (a) and different tear propagation on the trouser piece: correct (b1) and incorrect (b2, b3).
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determined by dividing the porosity before implantation, as calculated from density measurements, by the porosity assessed by the apparatus.

Light micrographs were used to estimate the percentage fibrocartilaginous tissue as a function of implantation time.

Preparation of reconstruction material and prosthesis

Porous reconstruction materials of Estane 5701-F1 were prepared as described before\textsuperscript{14}. For preparation of porous linear PU, the prepolymer was dissolved in 1,4-dioxane at a concentration of 25 wt% and mixed with cyclohexanedimethanol ([OH]/[NCO] = 1). The temperature was raised to 90°C and catalyst was added. During 45 min the solution was stirred and became more viscous. Just before the gel-point, 15 ml of the solution was mixed with 300 g NaCl (57%, 250–300 µm; 43%, 50–90 µm). Small NaCl particles were added to increase the viscosity and to prevent the crystals from sagging. The mixture was then frozen at —15°C. The solvent was removed by sublimation at 0.05 bar. The prepolymer/monomer/salt mixture was cured under nitrogen atmosphere at 100°C for 48 h and at 110°C for 12 h. After curing the salt was removed by washing the porous material with water. The porous materials were further extracted with ethanol for at least 1 week to remove the unreacted components and remaining solvent.

The method for making a total meniscal prosthesis was the same as that described above using glycerol as a cross-linker. The porous materials had the same porosity but a different ratio of macropores and micropores was used. Therefore the prepolymer was dissolved at a concentration of 30 wt% and mixed with glycerol. Fifteen millilitres of the solution were mixed with 43 g NaCl (65%, 250–300 µm; 35%, 50–90 µm).

Surgery

Meniscal reconstruction

The meniscal reconstruction technique is shown in Figure 2. Porous materials of Estane 5701-F1 and Elate 108/CHDM were used as implants into T-shaped lesions. T-shaped full-thickness lesions, occupying about 30% of the length of the meniscus, were made in the mid-portion of the lateral meniscus of dogs. All lesions had a longitudinal extension both into the anterior and posterior horns of the menisci. The standing leg of the T-lesion was excised. In the case of the Estane series, 15 implants were sutured into the defect\textsuperscript{15}. Due to dislocation, 12 menisci remained for study. In the case of the Elate 108/CHDM implants\textsuperscript{19}, in 19 menisci an implant was sutured into the defect. Due to infection 17 menisci remained for study.

Meniscal prosthesis

For the total meniscal prostheses, the porous Elate 108/glycerol network was used. Meniscal prostheses, shown in Figure 3, were cut out of porous materials after cooling with liquid nitrogen. The operative procedure for meniscal prostheses is shown in Figure 4. The meniscus was separated from its anterior and posterior attachments\textsuperscript{20}. Two drill holes were made in the lateral aspect of the proximal tibia, ending in the anterior and posterior area of the intercondylar eminence. Two sutures were attached to the prosthesis and were pulled through the drill holes. In the first six prostheses the sutures were attached to both prosthesis horns. However, dislocation due to tearing out of the sutures was observed. Therefore, in the remaining 10 prostheses, sutures were applied longitudinally through the entire prosthesis. The remains of the meniscal attachments were sutured to the appropriate meniscal horn. The prosthesis' periphery was sutured to the perimeniscal capsular and synovial tissues.

RESULTS AND DISCUSSION

Aromatic PU

In the past, an MDI-based polyurethane was used for meniscal reconstruction because of its good mechanical properties and minimal inflammatory and foreign body reactions\textsuperscript{3,15}. This PU elastomer was formed by the reaction of MDI with an adipic acid tetramethyleneglycol polyester, chain extended with...
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Figure 3 Total porous meniscal prosthesis.

Figure 4 Operative procedure meniscal prosthesis (with permission of the author). After separating the meniscus from its attachments, drill holes were made from the proximal lateral tibia (A) and ending on the anterior and posterior area of the intercondylar eminence (B). Two sutures were pulled longitudinally through the prosthesis (dotted line C) and attached to the proximal tibia. The remnants of the anterior and posterior meniscal attachments (D) were sutured to the appropriate meniscal horn.

tetramethylene glycol. The chemical structure of this aromatic PU is shown in Figure 5.

Since the mechanical properties of polyurethanes are largely influenced by the morphology, the thermal properties of the polymer were determined. DSC experiments, shown in Figure 6(a), were carried out in the temperature range of -100°C to 260°C. The DSC scan exhibits a $T_g$ at -32.2°C and three melting endotherms at 60°C, 85°C and 125°C with a total melting enthalpy of 5.2 J g$^{-1}$. These endotherms correspond to transitions in different phases of the PU. They arise from disruption of ordered non-crystalline hard segment aggregates for particular block lengths with a plasticizer effect of the soft segments. Apparently, in this system the hard segments and soft segments are to a certain extent phase mixed. In the
case of a phase separated system, crystalline hard domain disruption would take place at temperatures above 180°C\textsuperscript{23}. The non-crystalline morphology is also consistent with the transparent appearance of the film.

The stress–strain behaviour of this polymer is shown in Figure 7. The aromatic PU has a tensile strength of 63 MPa, a strain at break of 950% and a Young’s modulus (modulus at 0% strain) of 13.9 MPa. The large upturn at high strain is indicative of strain induced crystallization, which causes the high modulus at high strain and high tensile strength.

Porous materials of this aromatic PU were used for the meniscal reconstruction shown in Figure 2\textsuperscript{13,15}. Although the implant preparation and the implantation results have been described previously, the most successful implant series is included in this paper in order to compare it with results of new materials.

We concluded that for ingrowth of fibrocartilaginous tissue highly interconnected macropores of 150–300 μm are important. Therefore, a biporous structure containing macropores of 150–300 μm interconnected with micropores smaller than 50 μm was used. For preparation, a freeze-drying/salt-leaching technique was applied. The macropores, owing to the casting material, were dispersed in a matrix containing micropores which is a result of freeze-drying of the solvent.

A method was described to obtain a better interconnected porous structure. A small amount of water was added to the aromatic PU solution/saccharose mixture and resulted in highly interconnected porous materials. Adding amylalcohol and neopentylalcohol gave rise to the same open structure. Figure 8 shows a scanning electron micrograph of the aromatic PU implant in which the macropores are well interconnected by channel-like micropores. The morphological properties of the implants are presented in Table 1.

![Figure 8](image-url)

The compressibility, in addition to the porous structure, is an important property because the stresses on the implant due to motion of the knee joint probably induce the transformation of fibrous tissue into fibrocartilaginous tissue\textsuperscript{24}. It is very likely that not only the compression modulus, but also the compression behaviour is important, owing to the fact that the meniscus and implant are compressed to 10–20% during motion. In Figure 9 the compression behaviour of the implants and original meniscal tissue is shown. The compression moduli of the implants and meniscal tissue were 250 kPa and 1.2 MPa, respectively. The latter is a rough average of several tests because the modulus differs largely from dog to dog. Additionally, the modulus of the meniscus is also dissimilar for different parts of the meniscus\textsuperscript{25}.

Of the 15 inserted implants, dislocation of the implants was observed in three. Four of the remaining 12 menisci showed complete healing of the posterior and anterior lesions. Four menisci showed partial healing and four showed no healing. Eleven menisci were used for histological observation. In Figure 10 the percentage fibrocartilage in the implants as a function of time is shown. In the induction time up to about 20 weeks only fibrous tissue is observed. After 20 weeks the percentage fibrocartilaginous tissue increased up to 100% at 33 weeks. The percentage fibrocartilaginous tissue was between 70% and 100% for longer follow-up times.

Degradation of the polymer implants is important because the porous meniscal reconstruction materials act as temporary scaffold for the ingrowth of original tissue. In Figure 11 the degradation rate of the implants is shown. The percentage polymer in the implant decreased from 17% before implantation to 8% at 33 weeks. However, this polyurethane might yield 4,4’-diaminodiphenylmethane, which has been found to be mutagenic, carcinogenic, teratogenic and very toxic upon degradation. Therefore aliphatic diisocyanates are preferred\textsuperscript{19}.

**Linear aliphatic PU**

Unlike aromatic isocyanates, aliphatic isocyanates such as hydrogenated MDI are not very crystallizable due to the presence of stereoisomers\textsuperscript{26} and therefore possess poorer mechanical properties. However, a new trans-
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Table 1: Properties of meniscal reconstruction material and prosthesis

<table>
<thead>
<tr>
<th>Material</th>
<th>Meniscal reconstruction</th>
<th>Meniscal reconstruction</th>
<th>Meniscal prosthesis</th>
</tr>
</thead>
<tbody>
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<td>Polymer</td>
<td>Aromatic PU</td>
<td>Aliphatic PU linear</td>
<td>Aliphatic PU network</td>
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<td>Porosity</td>
<td>83%</td>
<td>86%</td>
<td>86%</td>
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<td>Micropores</td>
<td>Channel-like, length: 100 μm diameter: 10-30 μm</td>
<td>Spherical-like, diameter: 10-30 μm</td>
<td>Spherical-like, diameter: 10-30 μm</td>
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<tr>
<td>Microporosity</td>
<td>59%</td>
<td>45%</td>
<td>34%</td>
</tr>
<tr>
<td>Macropores</td>
<td>Cubical shape 150-300 μm</td>
<td>Cubical shape 50-90 μm 150-300 μm</td>
<td>Cubical shape 50-90 μm 150-300 μm</td>
</tr>
<tr>
<td>Macroporosity</td>
<td>41%</td>
<td>24% 50-90 μm 31% 150-300 μm</td>
<td>23% 50-90 μm 43% 150-300 μm</td>
</tr>
<tr>
<td>Compression modulus</td>
<td>250 kPa</td>
<td>150 kPa</td>
<td>150 kPa</td>
</tr>
</tbody>
</table>

Figure 9: Compression behaviour of meniscal tissue of dogs (---), aromatic PU implants (-----) and linear aliphatic PU implants and prosthesis (-----).

Figure 10: Percentage fibrocartilage in aromatic PU implants as a function of implantation time.

diisocyanate, CHDI, has been used to prepare polyester- and polyether-based polyurethanes. These polymers are semicrystalline and possess good mechanical properties. The diisocyanates form hard blocks due to the symmetrical structure of the molecule and their rigid rod-like molecular shape. In this study, a poly(ε-caprolactone)/CHDI based prepolymers was cured with CHDM in order to synthesize a linear polyurethane. The chemical structure of this aliphatic PU is presented in Figure 12(a).

The DSC scan of aliphatic PU, presented in Figure 6(b), shows glass transition temperatures at –53°C and 60.2°C, a melting endotherm at 3.5°C and a melting endotherm at 223.8°C, with a melting enthalpy of, respectively, 11.0 J g⁻¹ and 5.4 J g⁻¹. The lower Tg and melting endotherm corresponds to the soft segments, whereas the higher Tg and melting endotherm corresponds to the hard segments. The presence of two melting endotherms indicates that the polyurethane is phase separated in mainly two phases. Otherwise one would expect intermediate transitions, which is the case for the aromatic PU described above. The films had a cloudy appearance due to light scattering by the crystallites. Apparently, the CHDI hard segments are less compatible with the poly(ε-caprolactone) soft segments than MDI is with poly(adipic acid/tetramethylene glycol). The DSC scan of aliphatic PU, presented in Figure 6(b), shows glass transition temperatures at –53°C and 60.2°C, a melting endotherm at 3.5°C and a melting endotherm at 223.8°C, with a melting enthalpy of, respectively, 11.0 J g⁻¹ and 5.4 J g⁻¹. The lower Tg and melting endotherm corresponds to the soft segments, whereas the higher Tg and melting endotherm corresponds to the hard segments. The presence of two melting endotherms indicates that the polyurethane is phase separated in mainly two phases. Otherwise one would expect intermediate transitions, which is the case for the aromatic PU described above. The films had a cloudy appearance due to light scattering by the crystallites. Apparently, the CHDI hard segments are less compatible with the poly(ε-caprolactone) soft segments than MDI is with poly(adipic acid/tetramethylene glycol). Both melting endotherms are shown over a broad temperature range, which is an indication of the wide distribution of the hard and soft domains.

The stress–strain behaviour of this aliphatic PU is shown in Figure 7. The PU has a Young's modulus of 46 MPa, tensile strength of 33 MPa and an elongation at break of 950%. The higher crystallinity of the hard blocks is responsible for the high Young's modulus. Although the tensile strength of this polymer is lower than the tensile strength of the aromatic PU, we decided to develop a porous material of this polymer.
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Figure 11 Percentage O, aromatic PU; and ●, linear aliphatic PU as a function of implantation time.

![Graph showing percentage of polymer over degradation time (weeks)](image)

**Figure 11** Percentage O, aromatic PU; and ●, linear aliphatic PU as a function of implantation time.

**Figure 12** The chemical structures of linear aliphatic PU (a) and PU network (b).

**Figure 13** shows a scanning electron micrograph of the porous structure. The structure contains macropores as a result of leaching out the crystals and micropores caused by sublimation of the solvent. Due to the high viscosity of the prepolymer/curing agent melt and the high reaction rate, the microporous structure was not lost during curing. The macropores are interconnected with spherical micropores (10–30 μm). The isotropic structure is a result of liquid–liquid phase separation during cooling. The macropores are not highly interconnected with the micropores. Since previously described wetting agents (water, amyl- and neopentylalcohol) cannot be used for these materials due to their reactivity with the disocyanate, the resulting porous materials did not have an optimal interconnected pore structure. Nevertheless, they were used for meniscal reconstruction. The properties of the implants are presented in Table 1.

Compressive behaviour is shown in Figure 9. Despite the higher Young’s modulus of this polymer, the compression modulus of the implants was lower (150 kPa) due to the higher porosity.

After implantation, as shown in Figure 2, complete healing of the anterior lesions with meniscus-like tissue was observed in two-thirds of the implant menisci, whereas partial healing was found in 9% of the menisci. One-quarter of the menisci did not show healing. These results are comparable with the aromatic PU implants described above. In general, healing of the lesion only took place when the implant was attached to the lesion. Probably due to mechanical effects, sometimes the implant was disconnected at one side, mostly the posterior side. This was observed for both aromatic PU and aliphatic PU implants.

In a few implants at short follow-up time the middle of the implants was not completely filled with tissue. This trend may be attributed to the fact that the pores in the aliphatic PU implants were not optimally interconnected. The percentage fibrocartilage as a function of implantation time is shown in Figure 14. The induction time in which only fibrous tissue is observed is about 15 weeks. Although the lack of

The method for making porous polymers described previously could not be used for this PU because it was not soluble in a suitable solvent (DMF, DMSO, THF, chloroform, 1,4-dioxane). This was probably due to the highly ordered semicrystalline structure or as a result of cross-linking side-reactions forming aliphane, isocyanate dimers and trimers. Therefore, the porous structure was applied before curing of the prepolymer.
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Interconnectivity influences the ingrowth rate of fibrous tissue; it has no effect upon the transformation (metachromasia) into fibrocartilaginous tissue. These materials showed an even shorter induction time compared to the highly interconnected aromatic PU implants. After 15 weeks the percentage fibrocartilaginous tissue increases rapidly to 90% at 20 weeks. For the longer follow-up times the percentage is between 70% and 100%. Though the results are rather diffuse, these less interconnected implants showed a very quick ingrowth of fibrocartilaginous tissue. The percentage of 100% was first observed after 12 weeks, which was faster than for aromatic PU implants. In Figure 15 a light micrograph is presented of an implant 12 weeks after implantation. Chondrocytes can be observed lying in a metachromatic stained matrix, indicative of a chondroid matrix, with coarse collagen bundles. Morphologically, this tissue strongly resembles normal meniscal fibrocartilage. The lower tensile strength of the polymer and the different porosity and compressibility of the implants does not seem to influence the ingrowth of fibrocartilaginous tissue.

The degradation rate of the aliphatic PU is shown in Figure 11 and is slower than the degradation rate of the aromatic PU. The percentage polymer in the implant decreased from 14% before implantation to 10% at 52 weeks. The slow degradation rate is also shown by the macro pores, which still have the initial cubical form after 1 year. The soft segments cannot be responsible for this because they differ little for aromatic PU and aliphatic PU. It is likely to be the consequence of the different degree of phase separation and crystallinity. The aliphatic PU is phase separated to a higher degree and contains highly ordered crystalline hard blocks, whereas the hard domains of the aromatic PU are less ordered and non-crystalline.

Noteworthy is the linear behaviour of degradation. Apparently, immediately after implantation surface degradation takes place. This is unlike the in vivo degradation of semi-crystalline poly-L-lactic acid (PLLA). PLLA shows only a loss of mass after 20 weeks. It is known that semi-crystalline polymers are less susceptible to hydrolytic and enzymatic degradation. Furthermore, the surface of porous materials is much larger than the surface of solids.

Because presumably the ingrowth of fibrocartilage into the implants is obtained by transformation of ingrowing fibrous tissue rather than by direct ingrowth, the use of a porous meniscal prosthesis, for the replacement of complete menisci by the formation of a fibrocartilaginous meniscal replica, seems to be possible.

**Aliphatic PU network**

A totally porous meniscal prosthesis, however, will be exposed to much higher shear stresses than the implants used for meniscal reconstruction due to the larger surface and the absence of meniscal tissue. Therefore, the stress hysteresis phenomenon of linear PU can be of great consequence.

In order to avoid stress softening, an aliphatic PU network was synthesized by cross-linking the poly(epsilon-caprolactone)/CHDI-based prepolymer with glycerol. The chemical structure of this PU network is presented in Figure 12(b).

Cross-linking with glycerol has hardly any effect on \( T_g \), which is -53.5 and -53.0°C for the aliphatic PU and the PU network, respectively, as shown in Figure 6(c). It lowers the melting temperature for soft segments only from 3.5 to 1.5°C and decreases the melting enthalpy from 11.0 to 5.2 J g\(^{-1}\). It also decreases the melting enthalpy for hard segments from 5.4 to 2.6 J g\(^{-1}\) but it increases the melting temperature from 223.8 to 230.0°C. After cross-linking the \( T_g \) of the hard domains is no longer visible. Due to the fixation of the hard blocks during cross-linking, the entropy decreases and the melting point increases. Decrease of the melting enthalpy of the hard segments has hardly

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**Figure 14** Percentage fibrocartilage in linear aliphatic PU implants as a function of implantation time.

**Figure 15** Polarized-light micrograph of an aliphatic PU implant 12 weeks after implantation, showing chondrocytes (white arrow) dispersed in fibrocartilaginous tissue with collagen fibres (black arrow). The white areas represent the polymer.
any effect on the $T_g$ of the soft segments. The relative independence of the soft segment $T_g$ on the hard segment content is an indication that the system is phase separated and there is hardly any interphase mixing. It can be observed that cross-linking is only partially effective because the hard blocks do not completely disappear.

The stress–strain behaviour of this PU network is shown in Figure 7. The PU network has a Young's modulus of 25 MPa, tensile strength of 30 MPa and an elongation at break of 700%. Cross-linking causes the polyurethane to be more flexible. Due to the cross-links, hard block disruption cannot take place under stress, which results in a shorter strain at break. Because cross-linking does not have a large effect upon the tensile strength, this PU network seems suitable for use as a meniscal prosthesis.

The method for making porous PU network as total meniscal prosthesis was identical to the method used for the linear aliphatic PU. Because the interconnection of the aliphatic PU implants was not optimal and in order to accelerate the ingrowth of fibrous tissue, a slightly higher percentage macropore was used (Table 1). The porous structure shown in Figure 13, the compression behaviour shown in Figure 9 and the porosity of the prosthesis were identical to the respective properties of the meniscal implants.

Out of porous materials, total meniscal prostheses were cut with the dimensions shown in Figure 3. In the first six knees, prostheses were fixed using single sutures, which were placed through the anterior and posterior prosthesis horns and were pulled through the drill holes. However, in four of the six cases the meniscus prosthesis showed dislocation around the gel point) due to the tearing out of the sutures. Evidently the tear resistance of the prosthesis was not sufficient.

This problem was partly circumvented by applying sutures longitudinally through the entire prosthesis. Additionally, the remnants of the meniscal attachments were sutured to the appropriate meniscal horn and the prosthesis' periphery was sutured to the perimeniscal capsular and synovial tissues. Now only one out of 10 dislocated. The ingrowth of fibrous tissue starts from the synovium attachment to the inner part of the meniscus. No ingrowth took place from the articular cartilage of femur and tibia. The prosthesis was not connected to the femur and tibia, which is the identical configuration to a real meniscus.

Until 18 weeks ingrowth of tissue was not complete, the inner part of the meniscus remaining empty. After 10 weeks fibrous tissue began to transform into fibrocartilaginous tissue and after 18 the central areas also became filled and ingrowth was complete. Then the prostheses contained fibrocartilage only. This was determined by means of histological and immunological methods. In Figure 16 a light micrograph of a prosthesis after 20 weeks is shown. No difference of cell reaction in the porous aromatic PU, aliphatic PU and PU network materials could be detected. Degeneration of articular cartilage was observed, although it was less severe than in the meniscectomized control knees. Apparently the compression modulus of the implants of 150 MPa is not sufficient for a proper protection of the underlying articular cartilage.

Figure 16 Scanning electron micrograph of a meniscal prosthesis after 20 weeks. The pores are filled with active chondrocytes (white arrows) making fibrocartilage (dark areas).

**Tear strength**

As concluded above, the tear resistance of the implanted materials is very important. Unlike properties such as tensile strength, crystallinity and degradation rate, the tear properties of polymers used for biomedical applications have, as far as the present authors are aware, never been discussed. We decided to investigate the tear strength of the polymers we used for meniscal reconstruction and meniscal prosthesis so far.

The basic definition of tear strength is the force per unit thickness required to propagate a tear. The tearing energy can be determined using the following relationship:

$$G_t = 2F/w$$

where $F$ is the applied force and $w$ is the width of the torn path. So the tearing energy is the tear strength multiplied by two because tearing gives rise to two new surfaces.

In the majority of the literature written about tear strength, the threshold tearing energies are determined and examined in terms of the theory of rubber elasticity. The threshold tearing energy can be obtained under swollen conditions at high temperature and low tearing rate. Under these near-equilibrium conditions a threshold tearing energy of about 40 J/m$^2$ for urethane elastomers has been found.

Under normal conditions, however, much more energy can be dissipated during tearing. This loss of energy arises from viscoelastic effects and strain induced crystallization. Values of about 10$^2$ to about 10$^3$ J/m$^2$ are determined depending upon the rate of tearing, test temperature and elastomer composition. It was shown by Bhowmick that the tear strength of cross-linked rubbers decreases with increasing cross-link density and elasticity modulus. This is ascribed to the fact that stress relaxation mechanisms at long times are most effective when the strain modulus is small (i.e. around the gel point) due to relaxation of entanglements. Furthermore, a high degree of cross-linking will hinder strain induced crystallization.
The maximal tear strength and maximal tearing energy of aromatic PU, aliphatic PU and PU network are presented in Table 2. Noteworthy is the large influence cross-linking has upon the tearing energy, which decreases from 114 kJ m⁻² for the linear aliphatic PU to 20.8 kJ m⁻² for the PU network. In addition to the effects of cross-linking on the tear strength as described above, cross-linking also causes partial disruption of the hard segments, which lowers the viscoelastic energy dissipation. Consequently, in order to synthesize polymers with a high tear strength and to prevent tearing out of the sutures in the future, polymer networks are not suitable. Segmented PUs like the linear aliphatic PU are more appropriate.

### CONCLUSIONS

In addition to aromatic PU implants, porous materials of a trans-1,4-cyclohexane disiocyanate and poly(ε-caprolactone) based prepolymer cured with cyclohexanedimethanol can also be used for meniscal reconstruction. The different chemical structure, porous structure, porosity, compressibility and degradation rate of these materials compared to porous aromatic PU implants does not affect the ingrowth of fibrocartilage. The lack of interconnectivity, however, decreased the ingrowth rate of fibrous tissue, but had no effect upon the transformation into fibrocartilaginous tissue. These materials showed an even shorter induction time of 15 weeks, whereas the induction time of highly porous aromatic PU implants was 20 weeks.

A porous aliphatic PU network was synthesized for total meniscal prostheses because networks are less susceptible to stress hysteresis. This PU network was formed by cross-linking the trans-1,4-cyclohexane disiocyanate and poly(ε-caprolactones) based prepolymer with glycerol. Tearing out of the sutures was found to be a serious problem. It appeared that cross-linking had a large effect upon the tearing energy of the polymer, which decreased from 114 to 20.8 kJ m⁻² for the linear PU and the PU network respectively. In this study tearing out of the sutures has been partly circumvented by using a more complex suture technique, but improving the tearing resistance of the materials might be more effective. In order to do so, PU networks are not suitable since cross-linking leads to disruption of the phase separated system and therefore decreases the viscoelastic energy dissipation during tearing.

Despite the suturing problems, a meniscal replica was developed after implantation of a total porous PU network prosthesis. After 18 weeks the prostheses contained fibrocartilage only. Degeneration of articular cartilage decreased compared to meniscectomy but could still be observed. Therefore, in future the use of a prosthesis with a higher compression modulus is necessary.

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