Artificial heart: Hydrophobic coating to make it more natural

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Polyurethanes are biocompatible materials commonly used in medical applications, mainly as coatings for vascular grafts and artificial hearts. There are many methods to increase hemocompatibility of polyurethane surfaces. Presented study was undertaken to obtain hydrophobic polyurethane coatings by silicone and perfluorocarbohydrate modification. The aim of this work was to obtain smooth and homogenous coatings, as porous materials are likely to cause blood coagulation. Surface analysis (IR spectroscopy and scanning acoustic microscopy) revealed that the applied method allows to modify polyurethane surface successfully. An increase in material hydrophobicity was confirmed contact angle measurement.

Keywords and phrases: hydrophobic coatings, polyurethane vascular devices, blood coagulation.

Introduction

In modern world heart diseases have become a serious problem. A significant part of them are impossible to cure and the only solution is a heart transplant. Since the number of natural hearts is never sufficient, the major challenge for biomedical engineering is to construct a hemocompatible material for blood-contacting implants, such as artificial heart. Hemocopatibile material is compatible with blood, which means that it is non-thrombus and non-inflammable. Besides, other features are required, such as non-toxicity, non-carcinogenicity and leak-tightness [1].

Application of biomaterials is remarkably limited by blood coagulation. It is the most serious problem considering artificial organs. Thrombus reaction that happens naturally *in vivo* is extremely essential. There are two possible pathways to trigger the whole process (Fig. 1): extrinsic, activated by an injury, and intrinsic, initiated when blood contacts anomalous vessel surface. In each step an enzyme factor turns from an inactive form (zymogene) into an active one which in turn induces the next zymogene activation. The process is complex and consists of a number of interrelated steps, hence it is known as a blood coagulation cascade. The final product is a fibrin clot that stops bleeding until the vessel is reconstructed. Is has been reported that thrombus reaction also happens when blood contacts the surface of an implanted material. In that case it is undesirable and dangerous. The forming clots stick to the biomaterial surface, but they are likely to detach and disturb the circulation, e.g. cause stroke after reaching brain. Therefore, knowing the mechanisms that lead to clots forming is significant [2].

A natural healthy vessel, as well as heart, is covered inside by endothelium matrix. It is anti-trombous since it prevents proteins and cells from adsorption onto the vessel surface. An implanted material is void of those properties. The adsorption of proteins (known as Vroman effect) [3], platelets [4] and leukocytes [4] is often observed. These effects can cause an activation of the blood coagulation cascade. Therefore, there is a need to modify a surface of biomaterial to reduce the interactions between biomaterial and blood.

So far, a perfect implantable material is not yet discovered. Studies revealed that among all the polymers polyurethanes are characterized by the best hemocompatibility. Polyurethanes constitute a diverse group of chemical compounds. According to excellent mechanical properties they have been successfully brought into many branches of industry, e.g. as rubbers or foams. As biomaterials, they were first used in manufacturing early artificial hearts [5].

Hemocompatibility of polyurethanes is satisfying but not impeccable. Also, *in vivo* studies indicated degrada-

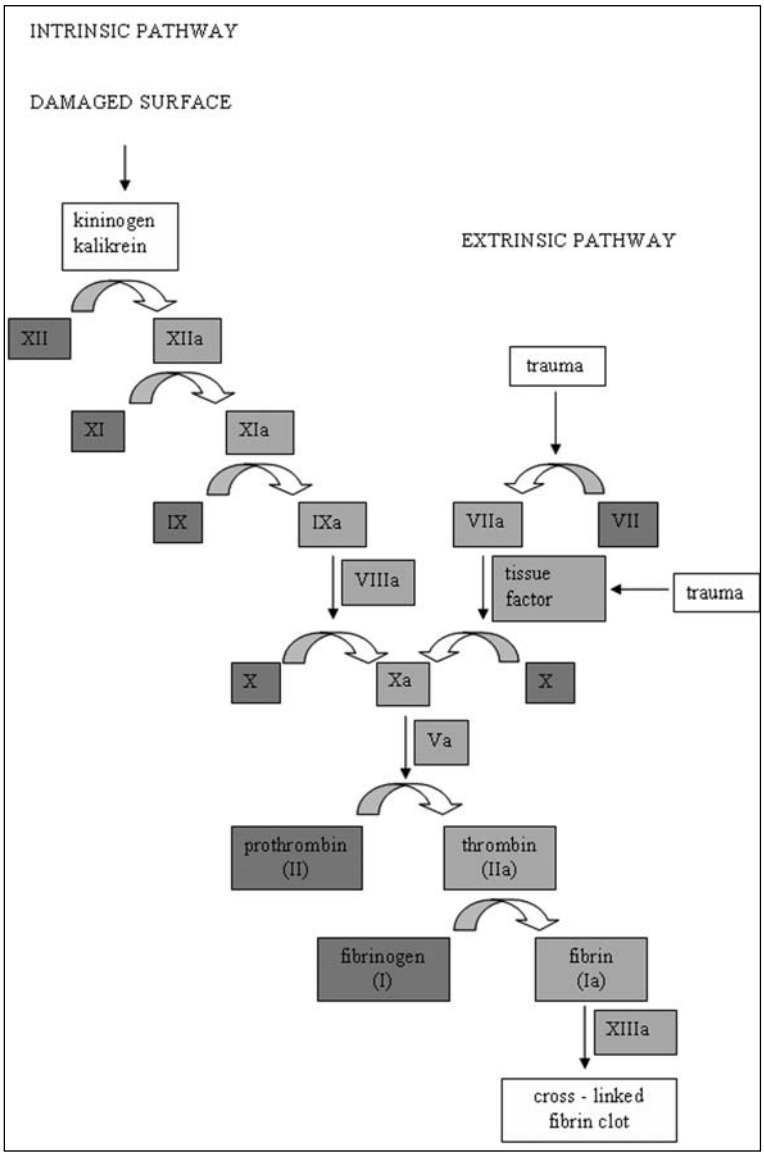

Fig. 1. Scheme of blood coagulation cascade.

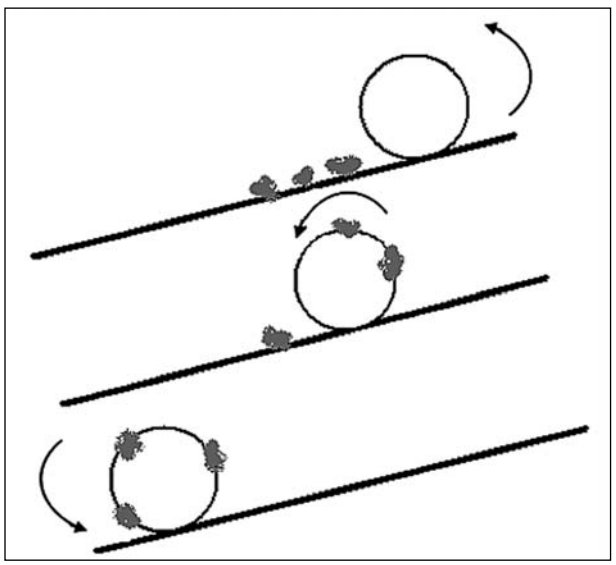

Fig. 2. The idea of a self-clearing surface.

tion of polyurethanes inside a body: oxidation, hydrolysis and radical processes so a surface modification is necessary to stop these effects [6]. A variety of methods has been proposed. Among them obtaining a hydrophobic surface by grafting a hydrophobic polymer, e.g. silicones [7] seems to be promosing. Hydrophobic surfaces are called self-cleaning. The idea is shown in the Fig. 2. Hydrophobic additives are likely to reduce adsorption of hydrophilic proteins and cells. The grafted compounds need to satisfy demands of good hemocompatibility and mechanical properties.

In this paper we present a simple method of surface modification of polyurethane with the aim of obtaining hydrophobic layers. Polyurethane was chosen as a polymer basis for further modification. As a modifying agent hydrophobic polymers — silicones (polydimethylsiloxane) and fluorocarbons (octafluoropentanol) were used. Diisocyanates were used as linker molecules. Polydimethylsiloxane as well as octafluoropentanol contains hydroxyl groups that are able to react with isocyanate groups. As a result, a chemical bond between polyurethane and a modifying polymer occurs.

Materials and methods

As backbones for coatings polyurethane (PU) was used. Namely: medical grade polyurethane Chronathane™ P–75A (CardioTech) in form of discs (40 mm diameter, 2 mm thick).

Polymers used to form hydrophobic layer were: poly(dimethylsiloxane) with terminal hydroxylic groups (PDMS) and 1*H*,1*H*,2*H*,2*H*-perfluoro-1-oktanol (97%, PFC), both purchased from Sigma — Aldrich®. Aromatic polyurethane Estane 5715 P and 4,4′-methylenebisphenylisocyanates (MDI) were used as linker molecules. All organic solvents: anhydrous tetrahydrofuran (THF), anhydrous dimetyloformamid (DMF), both provided by Chempur, and toluene (purchased from Sigma — Aldrich®) were of analytical grade.

In this study the method of using PDMS to obtain a hydrophobic coating previously proposed by Dabagh, Abdekhodaie and Khorasani [7], modified by using aromatic polyurethane solution as indirect layer, was applied. Also, the same method was used to graft

PFC to the polyurethane surface. The whole process consists of two steps. At first the polyurethane discs are dipped in aromatic polyurethane solution (30% w/v in THF), mixed with diisocyanate (MDI) solution (20–30% w/v MDI in THF:DMF, volume ratio 4:1). After 3 minutes discs were dried and immersed in PDMS or PFC solution (for about 3h), then rinsed with toluene (30 minutes) and dried. The scheme presenting applied method in shown in Fig. 3. The chemical reaction between MDI and PDMS/ PFC in shown in Fig. 4.

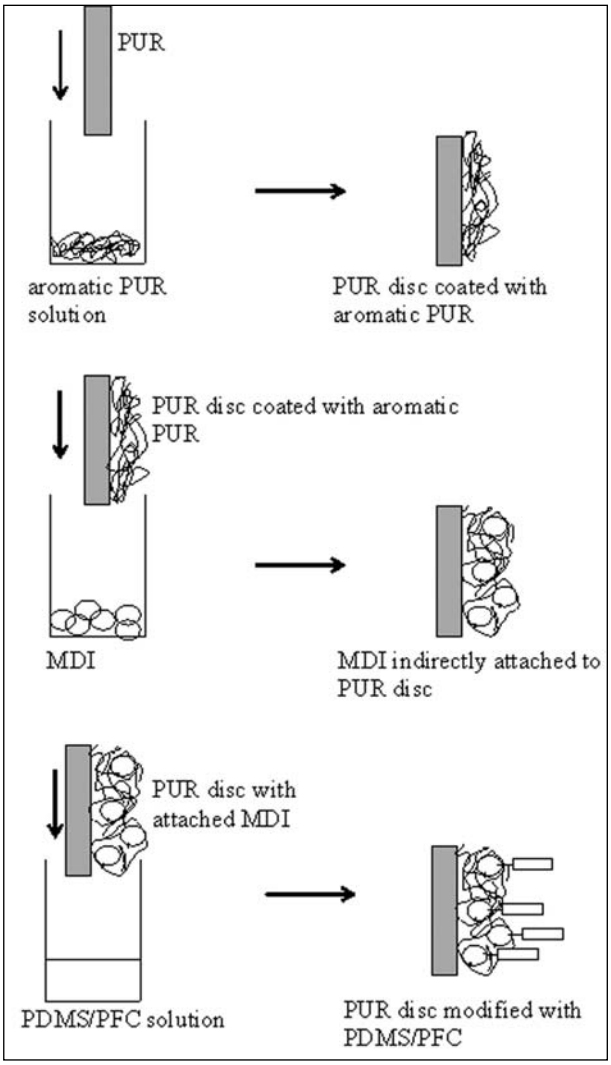

Fig. 3. Scheme of applied method.

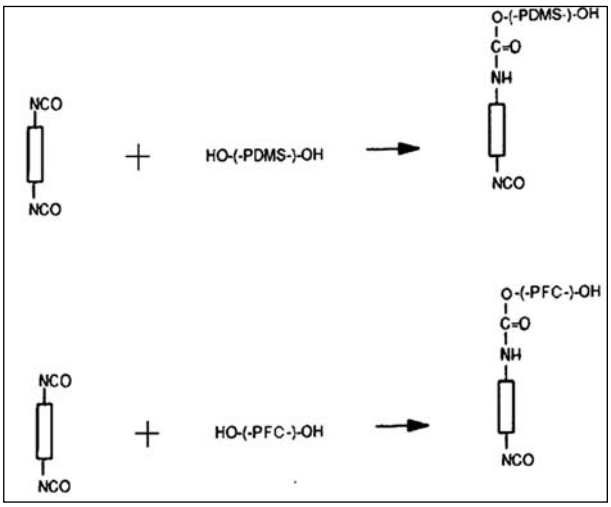

Fig. 4. Chemical reaction between MDI and PDMS/PFC.

Contact angle was measured by defining the contour of a sessile drop using the specialistic software (Images were analysed using Low-Bond Axisymmetric Drop Shape Analysis plug-in in ImageJ, free image-processing software). IR spectra of surfaces were obtained with Fourier Transform Infrared Attenuated Total Reflectance (FTIR-ATR). Coating surface was analyzed using scanning acoustic microscope (SAM).

Results and discussion

SAM analysis confirmed the acceptable smoothness and homogeneity of obtained coatings. The picture of PU surface grafted with PDMS is shown in Fig. 5. Figure 6 presents the picture of PUR modified with PFC.

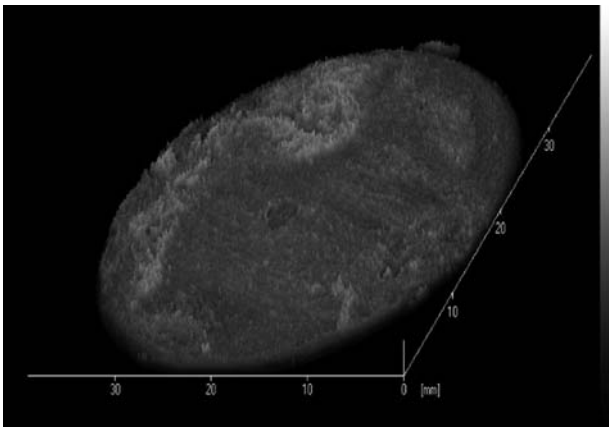

Fig. 5. SAM picture of PU surface modified with PDMS.

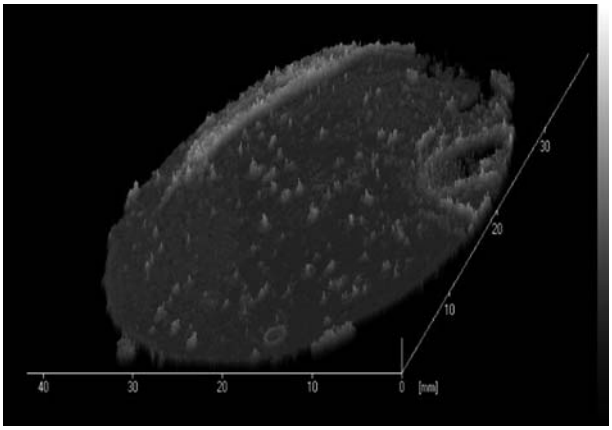

Fig. 6. SAM picture of PU surface modified with PFC.

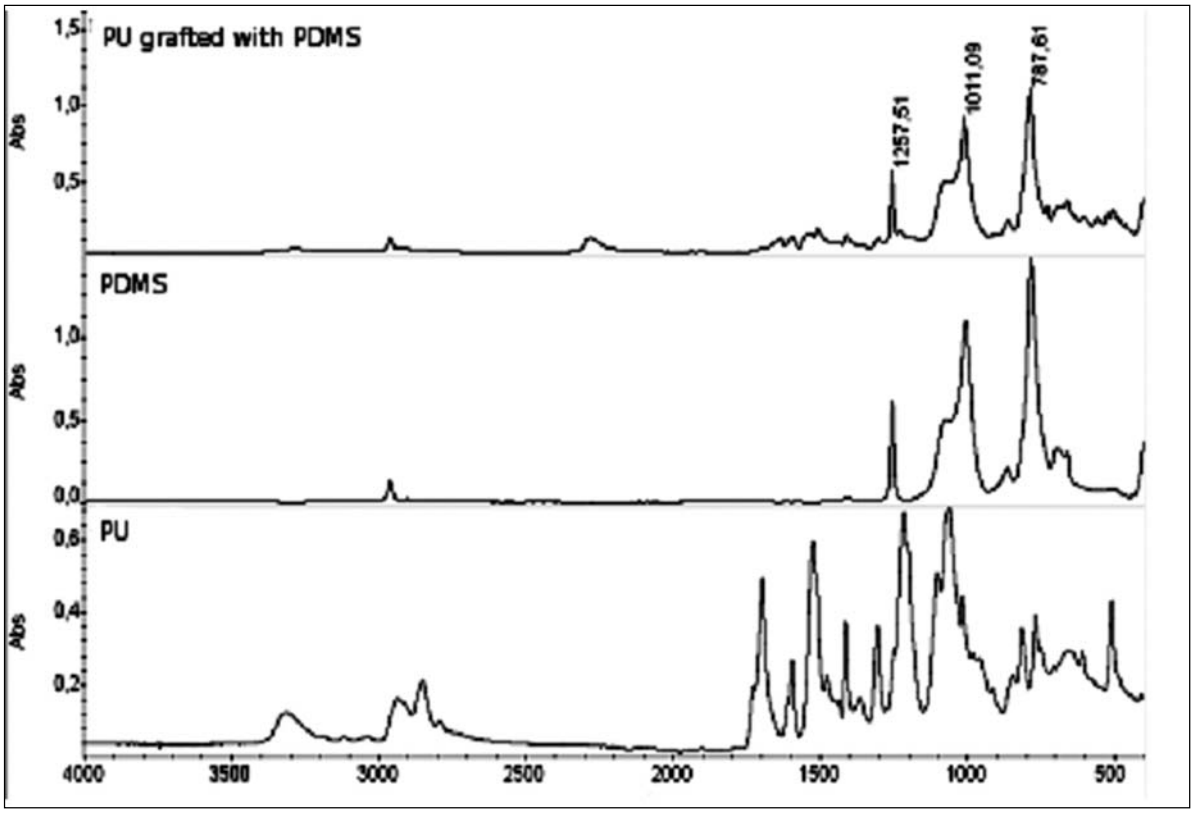

Fig. 7. FTIR spectra of PUR, PDMS and PUR grafted with PDMS.

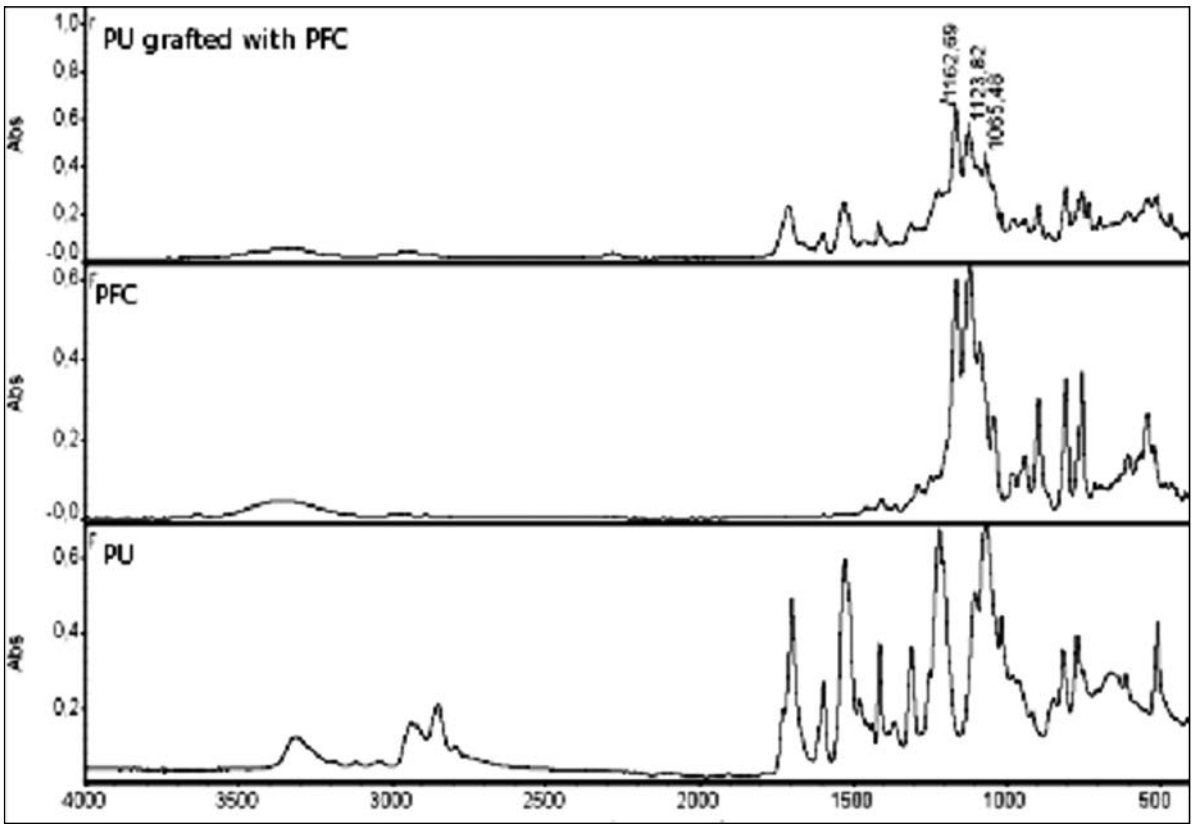

Fig. 8. FTIR spectra of PUR, PFC and PUR grafted with PFC.

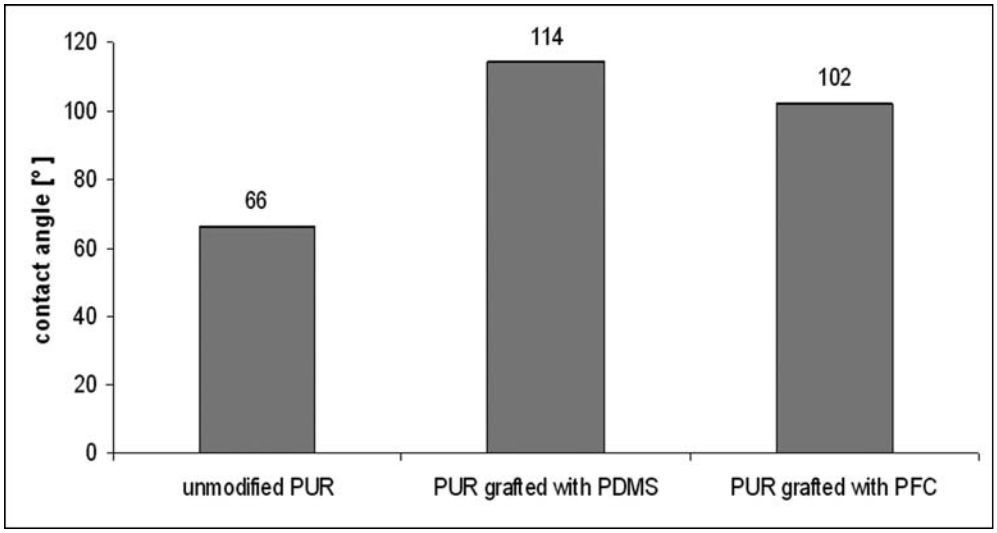

Fig. 9. Differences in contact angle measurement before and after grafting.

Figure 7 presents the spectra of unmodified PU surface, pure PDMS and PU grafted with PDMS. After grafting characteristic bands for silicones can be observed: 787 (Si-C stretching), 1011 (Si-O-Si stretching) and 1257 cm^{-1} (CH_3 symmetric deformation of Si-CH_3). The spectrum of PU grafted with PFC is shown in Fig. 8. Bands at 1085, 1123, 1162 cm^{-1} (C-F stretching) indicate the presence of PFC onto PUR film.

In Fig. 9 the results of contact angle measurement are presented. After the modification values increased from 66° (unmodified PUR) to 114° (PDMS graft) and 102° (PFC graft).

Conclusions

Presented results indicate that applied method allows to obtain hydrophobic layer on polyurethane surface. FTIR — ATR analysis revealed that the process of grafting was successful. That was also proved by contact angle measurement since values of contact angle increased after modification. SAM analysis indicates that obtained materials are acceptably homogenous, but PDMS coating is slightly smoother. Presented method is fast and could be automated. Fabricated coatings should be put to further research to study *in vivo* interactions with blood.

Acknowledgements

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