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Biomimetic phosphorylcholine polymer grafting from polydimethylsiloxane surface using photo-induced polymerization

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Abstract

The biomimetic synthetic phospholipid polymer containing a phosphorylcholine group, 2-methacryloyloxyethyl phosphorylcholine (MPC), has improved the surface property of biomaterials. Both hydrophilic and anti-biofouling surfaces were prepared on polydimethylsiloxane (PDMS) with MPC grafted by surface-initiated photo-induced radical polymerization. Benzophenone was used as the photoinitiator. The quantity of the adsorbed initiator on PDMS was determined by UV absorption and ellipsometry. The poly(MPC)-grafted PDMS surfaces were characterized by XPS, ATR-FTIR and static water contact angle (SCA) measurements. The SCA on PDMS decreased from 115° to 25° after the poly(MPC) grafting. The in vitro single protein adsorption on the poly(MPC)-grafted PDMS decreased 50–75% compared to the unmodified PDMS. The surface friction of the poly(MPC)-grafted PDMS was lower than the unmodified PDMS under wet conditions. The oxygen permeability of the poly(MPC)-grafted PDMS was as high as the unmodified PDMS. The tensile property of PDMS was maintained at about 90% of the ultimate stress and strain after the poly(MPC) grafting. The surface-modified PDMS is expected to be a novel medical elastomer which possesses an excellent surface hydrophilicity, anti-biofouling property, oxygen permeability and tensile property.

Keyword: Polydimethylsiloxane; Phosphorylcholine; Protein adsorption; Wettability; Oxygen permeation; Friction

1. Introduction

Polydimethylsiloxane (PDMS)-based materials have been applied to various medical devices, such as ophthalmologic biomaterials, microfluidic devices, an artificial lung [1], and an artificial finger joint [2] due to its attractive properties of high oxygen permeability, good mechanical property, optical transparency, self-sealing property, convenient processability, and chemical stability. On the other hand, the native hydrophobicity and biofouling tendency of PDMS has been one of its biggest limitations for biomaterial applications. For example, nonspecific protein adsorption on a material is recognized as the first incident leading to subsequent events including thrombus formation, foreign body reaction, bacterial infection, and other undesirable responses. Owing the supersensitive analysis of microfluidic systems, the adsorption of biomolecules on PDMS significantly reduces the signal/noise ratio during detection [3]. As for an artificial lung, the blood activation on a large blood contact surface area is the critical problem [4]. Consequently, there are tremendous needs for methods to quickly and easily modify the surface properties of PDMS.

To modify the disadvantages of PDMS-based materials, oxygen plasma treatment is a simple and the most widespread technique to change its surface hydrophobicity [5–9]. However, the hydrophilic nature of oxidized PDMS is only temporary as the migration of PDMS chains leads to recovery of the native hydrophobic state [10–12]. Surface coating or grafting with hydrophilic polymers on PDMS is another technique [13]. Surface graft polymerization is better than coating due to the chemical stability of its covalent bonding with a substrate and lower risk for

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deposition. The grafting methods can be divided into two classifications known as "grafting-to" and "grafting-from" [14]. For the "grafting-to" method, the polymer chains carrying reactive anchor groups at the end or the side chains are covalently coupled to the PDMS surface using silane-coupling reagents. Plasma treatment can also introduce active species on a surface of the polymer, followed by polymerization of the monomers. However, the complication inherent in the "grafting-to" process is an intrinsic limitation of the number of functional groups per surface area for thermodynamic reasons [15]. On the other hand, the "grafting-from" method utilizes active species existing on a polymer surface with a high grafting density to initiate the polymerization of monomers from the surface. "Grafting from" can be usually accomplished by treating a substrate with plasma and glow-discharge to generate the immobilized initiators followed by polymerization [16]. However, immobilization of the initiator on the surface involves several steps that may lead to low graft densities of the initiator and tethered polymer if the reactions are not quantitative. In addition, a side reaction, which possibly exists in the initiator immobilization reaction, may introduce some undesired structures on the surface. Methods that circumvented these problems have also been reported to accurately attach initiators in one step on the surface of the substrate using self-assembly monolayer techniques [17]. Accurate characterization of the initiator is important for the "grafting-from" method. Recently, living polymerization techniques have extensively been investigated in order to grow high-density polymer brushes with a controlled length and narrow molecular weight distribution. Atom transfer radical polymerization (ATRP) is useful because of its versatility including monomer types, tolerance of impurities, and mild reaction conditions [18,19]. Although the ATRP grafting technique onto PDMS surfaces seems attractive for biomaterials, a simple scheme is preferable for practical use [20].

Selection of the grafting polymer is the next parameter. Various types of polymers, such as poly(ethylene oxide) (PEO), poly(2-hydroxyethyl methacrylate), poly(2-hydroxyethyl acrylate), poly(acrylic acid), poly(acrylamide), and poly(*N,N*-dimethylacrylamide) were grafted onto the PDMS surfaces to make ideal biomaterial surfaces. Among these monomers, the PEO grafted surface has shown good anti-biofouling characteristics. The use of biomimetic materials is another promising approach to enhance the anti-fouling property and biocompatibility. The biomembrane-mimetic surface based on phosphorylcholine containing phospholipid polymers have shown an excellent resistivity of non-specific protein adsorption and cell adhesion [21–27]. These biomimetic polymers are included in 2-methacryloxloxyethyl phosphorylcholine (MPC), a methacrylate monomer having a zwitterionic phosphorylcholine headgroup in the side chain [28]. Poly(MPC) is known to possess a large amount of the free water fraction around the chain, which resists non-specific protein adsorption, and this also provides stabilization of biomolecules such as enzymes and proteins, even when the biomolecules are adsorbed on the surface [29–31].

In this paper, we reported the MPC "grafting-from" polymerization on PDMS using UV light. This polymerization has advantages in simplicity and efficiency. A graft polymerization was conducted in the MPC aqueous solution. This process made benzophenone remain on the PDMS surface after being placed in the monomer solution because benzophenone is insoluble in water. Therefore, the "grafting-from" polymerization can be performed on the surface attached benzophenone. The use of UV light to initiate a chemical reaction is applicable for the photolithographic micropatterning and the generation of multifunctional patterns at selected areas of the substrate. A similar poly(MPC) grafting was reported on an ultrahigh-molecular-weight polyethylene surface to decrease the friction and the amount of wear, and to improve the biocompatibility for an artificial hip joint [32,33]. Meanwhile, the main objective of the current study is to obtain a novel medical elastomer possessing a surface hydrophilicity, non-biofouling property, lubrication property, high-oxygen permeability and elasticity. Also, we checked the tensile properties of the poly(MPC)-grafted PDMS because the radicals may affect the elasticity of PDMS due to the formation of cross-links.

2. Experimental section

2.1. Materials

MPC was synthesized by a previously described method [28] and recrystallized from acetonitrile. Bovine serum albumin, bovine serum γ-globulin, and bovine plasma fibrinogen were purchased from Sigma-Aldrich Japan, and lysozyme from chicken egg white was purchased from Biozyme (Blaenavon, UK). The precursor of PDMS (Silpot 184®) and cross-linker of PDMS (Catalyst of Silpot 184®) were purchased from Toray-Dow Corning Co. (Tokyo, Japan). All other reagents and solvents were commercially available as extra-pure grade and were used as purchased. Distilled water was used in all the experiments. The nitrogen, oxygen, and argon gases were of high-purity grade.

2.2. Preparation of PDMS

The precursor of PDMS and cross-linker were fully mixed at the ratio of 10:1 by mass. The mixtures (5 mL) were evenly spread on a glass plate (10 cm × 10 cm) and degassed in a vacuum oven for 2 h at 25 °C. The curing reaction was then carried out at 60 °C for 6 h. The samples were cut into disks (10 mm diameter, 0.5 mm thickness) or the desired shapes.

2.3. Preparation of poly(MPC)-grafted PDMS

The PDMS was etched by oxygen plasma (300 W, 100 mL/min gas flow) for 1 min in advance. The MPC graft polymerization was able to be completed without the plasma treatment. The membrane was immersed in a 30 mL acetone solution containing benzophenone for 1 min. In the feed benzophenone concentration, "*x*", was varied from 0.1 to 10 g/L. The membrane was dried in vacuo under dark condition for 1 h at 25 °C. The concentration of the MPC aqueous solution, "*y*" (= 0.25 or 0.50 mol/L), was prepared in degassed pure water and then argon was bubbled for 3 min to eliminate any oxygen. As the sample nomenclature, the surface modified PDMS membranes are coded by "*x–y*" using a simplified