

The surface elemental conditions of the various polyelectrolyte-grafted CLPE surfaces with a 90-min photoirradiation time were analyzed by X-ray photoelectron spectroscopy (XPS). The XPS spectra were obtained using an XPS spectrophotometer (AXIS-HSi165; Kratos/Shimadzu Co., Kyoto, Japan) equipped with a 15-kV Mg-K $\alpha$  radiation source at the anode. The take-off angle of the photoelectrons was maintained at 90°. Five scans were taken for each sample.

The static-water contact angles on the various polyelectrolyte-grafted CLPE surfaces obtained at various photoirradiation times were measured by the sessile drop method using an optical bench-type contact angle goniometer (Model DM300; Kyowa Interface Science Co., Ltd., Saitama, Japan). Drops of purified water (1  $\mu$ L) were deposited on the various polyelectrolyte-grafted CLPE surfaces, and the contact angles were directly measured with a microscope 60 s after dropping. Measurements were repeated 15 times for each sample, and the average values were regarded as the contact angles.

### 2.5. Friction test

Friction tests were performed using a ball-on-plate machine (Tribostation 32; Shinto Scientific Co., Ltd., Tokyo, Japan) with various lubrication conditions. Each of the various polyelectrolyte-grafted CLPE surfaces was used to prepare six sample pieces. A 9-mm-diameter ball of Co–Cr–Mo alloy was prepared. The surface roughness ( $R_a$ ) of the pin was less than 0.01  $\mu$ m, which is comparable to that of femoral head products. The friction tests were performed with a load of 0.98 N (contact stress roughly calculated by Hertzian theory is approximately 29 MPa), sliding distance of 25 mm, and a frequency of 1 Hz for a maximum of 100 cycles. The lubricants used were pure water at room temperature, acellular simulated body fluid (SBF, Kokubo solution, pH 7.4) with inorganic ion concentrations analogous to those found in human extracellular fluid at 37 °C [30], and a mixture of 25-vol% bovine serum at 37 °C. Before the friction tests, the specimens were pre-soaked in each lubricant for 24 h. The coefficient of dynamic friction for each specimen was determined by averaging five data points from the 100 (96–100) cycle measurements, and the average values for six sample pieces are reported as the mean coefficient of dynamic friction for each of the various polyelectrolyte-grafted CLPE surfaces.

### 2.6. Surface zeta potential measurement

The effective surface charge of the various polyelectrolyte-grafted CLPE surfaces was determined via an electrophoretic mobility method with an electrophoretic light-scattering spectrophotometer (ELS-800; Otsuka Electronics Co., Ltd., Osaka, Japan) equipped with a solid-plate sample cell. Before measurement, the specimens were equilibrated by soaking in a 0.01-mol/L sodium chloride aqueous solution for 1 h. The measurement was carried out in a 0.01-mol/L sodium chloride aqueous solution at 20 °C. Six samples for each of the various polyelectrolyte-grafted CLPE surfaces were prepared for the measurements, and the average values are reported as the surface zeta potential.

### 2.7. Characterization of protein adsorption by micro bicinchoninic acid method

The amount of protein adsorbed on the various polyelectrolyte-grafted CLPE surfaces was measured by the micro bicinchoninic acid (BCA) method. For each type of polyelectrolyte-grafted CLPE, ten sample pieces were prepared. Each specimen was immersed in Dulbecco's phosphate buffered saline (PBS; pH 7.4; ionic strength, 0.15 M; Immuno-Biological Laboratories Co., Ltd., Takasaki, Japan) for 1 h to equilibrate the polymer-grafted surface. The specimens were then immersed in bovine serum albumin (BSA;  $M_w = 6.7 \times 10^4$ ; Sigma–Aldrich Corp., MO, USA) solution at 37 °C for 1 h. The protein solution was prepared at a BSA concentration of 4.5 g/L, i.e., 10% of the concentration of human plasma levels. Then, the specimens were rinsed five times with fresh PBS, immersed in a 1-mass% sodium dodecyl sulfate (SDS) aqueous solution, and shaken at room temperature for 1 h to detach completely the adsorbed BSA from the polymer-grafted CLPE surface. A protein analysis kit (micro BCA protein assay kit, #23235; Thermo Fisher Scientific Inc., IL, USA) based on the BCA method was used to determine the BSA concentration in the SDS solution, and

this value was used to determine the amount of BSA adsorbed on the polyelectrolyte-grafted CLPE surface.

### 2.8. Hip simulator wear test

A 12-station hip simulator (MTS Systems Corp., Eden Prairie, MN) with untreated CLPE and the various polyelectrolyte-grafted CLPE cups, each with inner and outer diameters of 26 and 52 mm, respectively, was used for the wear test according to the ISO standard 14242-3. Three sample pieces were prepared for each untreated CLPE and the various polyelectrolyte-grafted CLPE cups. A Co–Cr–Mo alloy ball with a diameter of 26 mm (K-MAX<sup>®</sup> HH-02; Japan Medical Materials Corp., Osaka, Japan) was used as the femoral component. A mixture of 25-vol% bovine serum was used as the lubricant, which was replaced every  $0.5 \times 10^6$  cycles. Gait cycles were applied that simulated a physiological loading curve (Paul-type) with double peaks at 1793 and 2744 N (maximum contact stress roughly calculated by Hertzian theory is approximately 8 MPa) with a multidirectional (biaxial and orbital) motion at 1-Hz frequency. Gravimetric wear was determined by weighing the cups at intervals of  $0.5 \times 10^6$  cycles. Load-soak controls ( $n = 2$ ) were used to compensate for the fluid absorption by the specimens according to the ISO standard 14242-2. Testing was continued until a total of  $3.0 \times 10^6$  cycles was completed. When the gravimetric method was used, the weight loss of the tested cups was corrected by subtracting the weight gain in the load-soak controls; however, this correction was not perfect because only the tested cups were continuously subjected to the motion and the load. In addition, after  $3.0 \times 10^6$  cycles of the hip simulator test, the volumetric wear of the cups was measured using a three-dimensional coordinate measurement machine (BHN-305; Mitsutoyo Corp., Kawasaki, Japan) and reconstructed using three-dimensional modeling software (Imageware; Siemens PLM Software Inc., Plano, TX).

Wear particles were observed under a field emission scanning electron microscope (FE-SEM). The wear particles were isolated from the bovine serum solution used as lubricant in the hip simulator wear tests. To isolate the wear particles, the lubricant was incubated in a 5-mol/L sodium hydroxide solution for 3 h at 65 °C to digest adhesive proteins that were degraded and precipitated. To avoid artifacts, the contaminating proteins were removed by extraction with solutions having several densities: sugar solution, 1.20 g/cm<sup>3</sup> and 1.05 g/cm<sup>3</sup>; and isopropyl alcohol solution, 0.98 g/cm<sup>3</sup> and 0.90 g/cm<sup>3</sup>. This was followed by centrifugation at 25,500 rpm for 3 h at 5 °C (himac CP 70MX; Hitachi Koki Co., Ltd, Tokyo, Japan). The collected solution was sequentially filtered through a 0.1- $\mu$ m membrane filter, and the membrane was observed directly under an FE-SEM (JSM-6330F; JEOL DATUM Co., Ltd, Tokyo, Japan) at an acceleration voltage of 20 kV after gold deposition.

### 2.9. Statistical analysis

The mean values of the groups (untreated and various polyelectrolyte-grafted CLPE) were compared by one-factor analysis of variance (ANOVA) and the significance of differences in the static-water contact angle (static-water contact angle measurement), dynamic friction coefficient (ball-on-plate friction test), surface zeta potential (surface zeta potential measurement), and amount of adsorbed BSA (protein adsorption measurement by micro BCA method) were determined by post-hoc testing using Bonferroni's method. The statistical significance of gravimetric wear (hip simulator test) was judged by the Student's *t*-test. All statistical analyses were performed using add-on software (Statcel 2; OMS publishing Inc., Tokorozawa, Japan) for a computerized worksheet (Microsoft Excel<sup>®</sup> 2003; Microsoft Corp., Redmond, WA).

## 3. Results

As shown in the cross-sectional TEM images in Fig. 2, all graft polymerization processes afforded a uniform grafted polyelectrolyte layer on the CLPE surface with almost constant thicknesses of 100–150 nm.

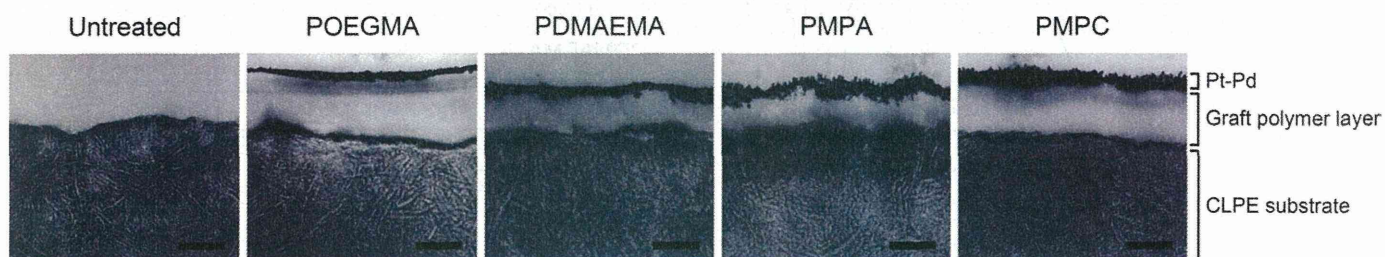


Fig. 2. Cross-sectional TEM images of untreated CLPE and various polyelectrolyte-grafted CLPE samples obtained with a 90-min photoirradiation time. Bar: 100 nm.

Fig. 3 shows the FT-IR/ATR and XPS spectra of the untreated CLPE and various polyelectrolyte-grafted CLPE samples subjected to photoirradiation for a 90 min. The FT-IR/ATR spectra after grafting contained new absorption peaks at  $1720\text{ cm}^{-1}$  (C=O);  $1090$  and  $1040\text{ cm}^{-1}$  (ether or carbohydrate group);  $1140$  and  $960\text{ cm}^{-1}$  (protonated or carbonated ammonium group); and  $1240$ ,  $1080$ , and  $970\text{ cm}^{-1}$  (phosphate group) for each of the polyelectrolyte-grafted CLPE samples. These peaks are chiefly attributed to the graft polymer on each CLPE surface. The XPS spectra of the binding energy region of the nitrogen (N) and/or phosphorous (P) electrons showed peaks for the PDMAEMA-, PMPA-, and PMPC-grafted CLPE samples, whereas peaks were not observed for untreated CLPE and the PEOGMA-grafted CLPE samples. The peaks at  $400$  and  $403\text{ eV}$  are attributed to N–H,  $-\text{NH}^+(\text{CH}_3)_2$ , and  $-\text{N}^+(\text{CH}_3)_3$ , respectively. The peak at  $134\text{ eV}$  is attributed to the phosphate groups. These peaks indicate the presence of a dimethylamino group in the DMAEMA units, phosphonoxy group in the MPA units, and phosphorylcholine group in the MPC units. Overall, these results from TEM observation, FT-IR/ATR, and XPS analyses indicate that the various polyelectrolytes were successfully grafted on the CLPE surface.

Next, as shown in Fig. 4, the static-water contact angle was  $95^\circ$  for untreated CLPE and this value decreased as the photoirradiation time increased. Thus, clearly, the photoirradiation time affected the hydration-kinetics of the polyelectrolyte graft chains. However, the decrease (or decrease rate) of the static-water contact angle differed for each of the polyelectrolyte-grafted CLPE surfaces. Nevertheless, after photoirradiation for more than 90 min (45 min for PEOGMA-grafted CLPE), the static-water contact angles of all polyelectrolyte-grafted CLPE samples reached the lowest values of  $20^\circ$ – $60^\circ$ .

Fig. 5 shows the coefficient of dynamic friction of polyelectrolyte-grafted CLPE samples obtained with a 90-min photoirradiation time in various lubricants. Control measurements

were carried out in water as the lubricant. From the figure, we see that the coefficients of dynamic friction for the polyelectrolyte-grafted CLPE samples were  $0.01$ – $0.05$ , representing a  $40\%$ – $85\%$  reduction compared with untreated CLPE. A calcification-like mineralized surface morphology was observed on the PMPA-grafted CLPE surface after SBF pre-soaking for 24 h before the friction test. The coefficient of dynamic friction of the PEOGMA- and PMPC-grafted CLPE surfaces did not differ significantly ( $p > 0.05$ ) between lubricants. The coefficient of dynamic friction of PMPC-grafted CLPE was significantly ( $p < 0.01$ ) lower than that of PEOGMA-grafted CLPE under each lubrication condition; the same relation was found for the static-water contact angles of the two types of polyelectrolyte-grafted CLPE surfaces. In contrast, the coefficient of dynamic friction of the PDMAEMA- and PMPA-grafted CLPE surfaces increased drastically ( $p < 0.01$ ) in SBF and BS, respectively. In addition, the untreated CLPE also showed increased friction in BS lubricant compared with that in water.

Fig. 6 shows the surface zeta potential and the amount of BSA adsorbed on the untreated CLPE and the various polyelectrolyte-grafted CLPE samples with a 90-min photoirradiation time. As seen in the figure, the surface zeta potential of the untreated CLPE and the PEOGMA- and PMPC-grafted CLPE surfaces with nonionic and zwitterionic grafted polymer layers was close to zero (Fig. 6A). In contrast, that of PDMAEMA-grafted CLPE with cationic grafted polymer layers was  $50.6\text{ mV}$ , which is strongly positive, and that of PMPA-grafted CLPE with anionic grafted polymer layers was  $-32.5\text{ mV}$ , which is strongly negative. The amount of adsorbed BSA on the PEOGMA-, PMPA-, and PMPC-grafted CLPE surfaces with nonionic, anionic, and zwitterionic grafted polymer layers significantly ( $p < 0.01$ ) decreased (Fig. 6B). In contrast, that of PDMAEMA-grafted CLPE with a cationic grafted polymer layer increased ( $p < 0.01$ ).

Fig. 7 shows how the polyelectrolyte hydrated layer characteristics affected the durability of artificial hips. Three-dimensional

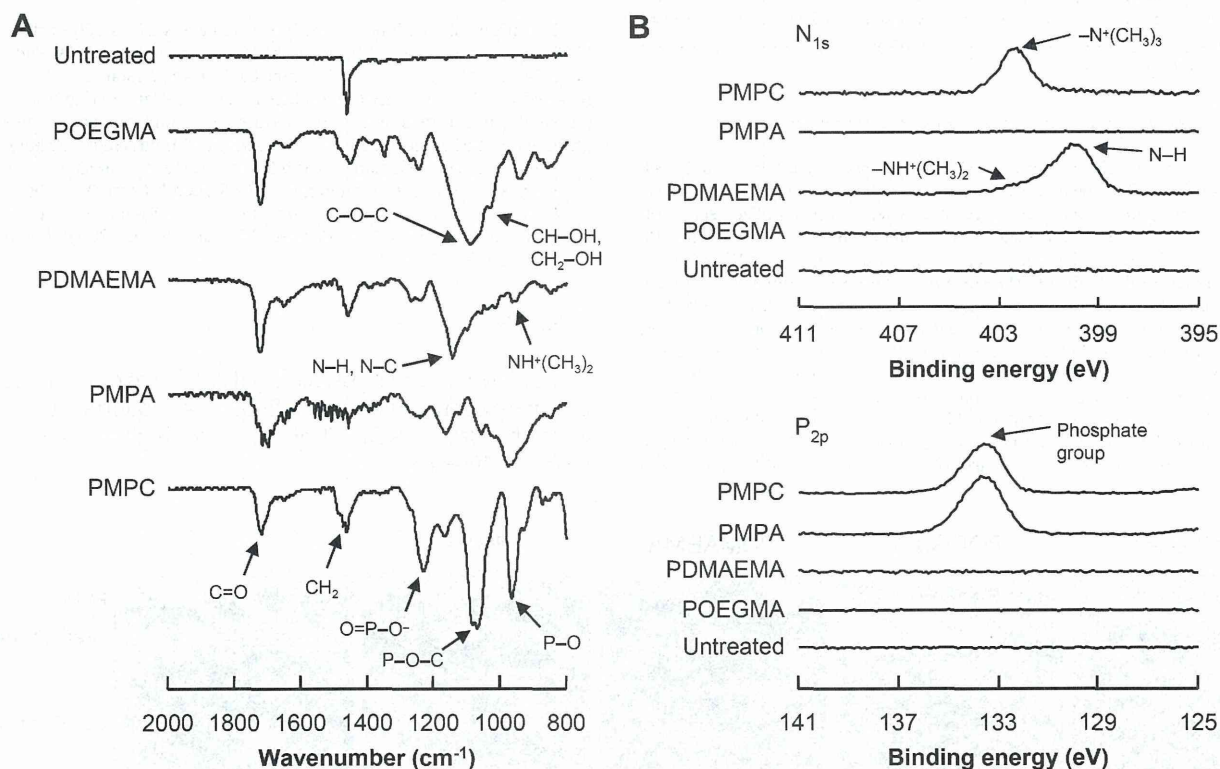
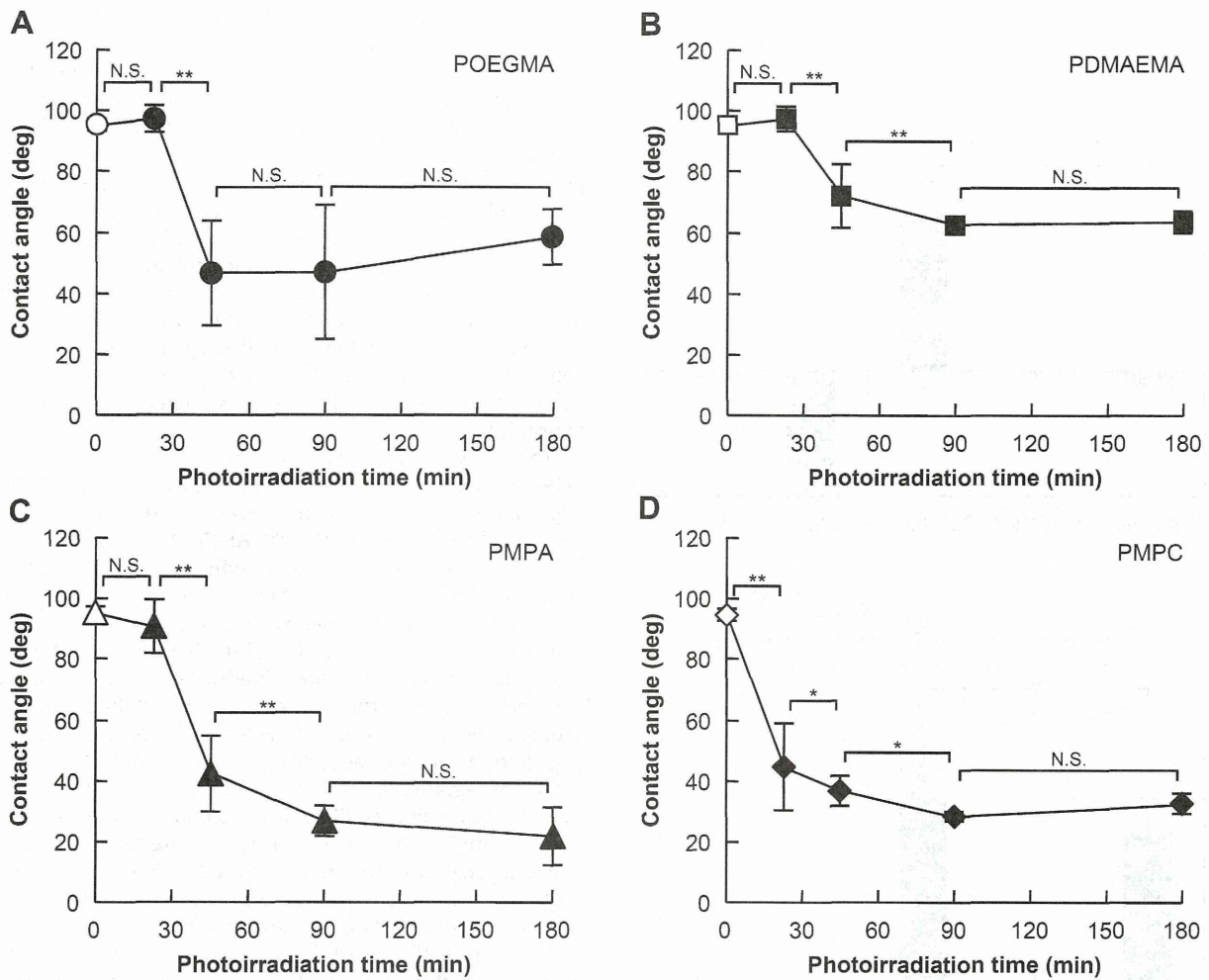


Fig. 3. (A) FT-IR/ATR and (B) XPS spectra of untreated CLPE and various polyelectrolyte-grafted CLPE samples obtained with a 90-min photoirradiation time.

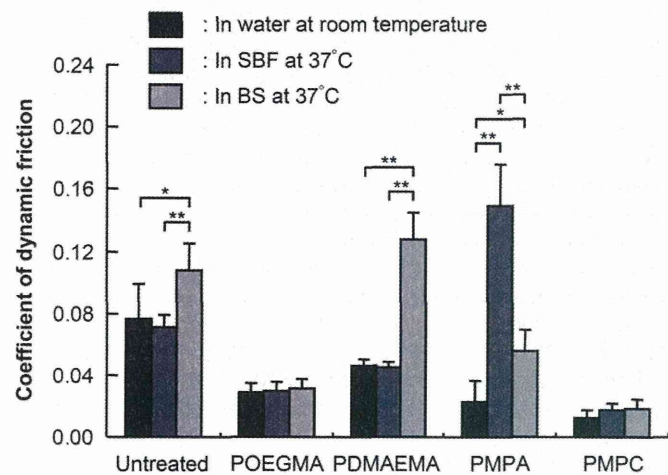


**Fig. 4.** Static-water contact angle on various polyelectrolyte-grafted CLPE surfaces as a function of the photoirradiation time. (A) POEGMA-grafted, (B) PDMAEMA-grafted, (C) PMPA-grafted, and (D) PMPC-grafted CLPE. Data are expressed as means  $\pm$  standard deviations. \* Indicates  $p < 0.05$ , \*\* indicates  $p < 0.01$ , and N.S. indicates no statistical difference.

coordinate measurements with the polyelectrolyte-grafted CLPE cups revealed little to no detectable volumetric wear, although substantial volumetric wear was detected in the untreated CLPE (Fig. 7A). The wear particles of untreated CLPE and polyelectrolyte-

grafted CLPE cups after  $0.5\text{--}1.0 \times 10^6$  cycles of the hip simulator test, as characterized by FE-SEM, were predominantly sub-micrometer-sized granules (Fig. 7B). In all cases, the morphologies of the wear particles exhibited no remarkable difference. However, remarkably fewer wear particles were found for POEGMA- and PMPC-grafted CLPE cups than for untreated CLPE cups.

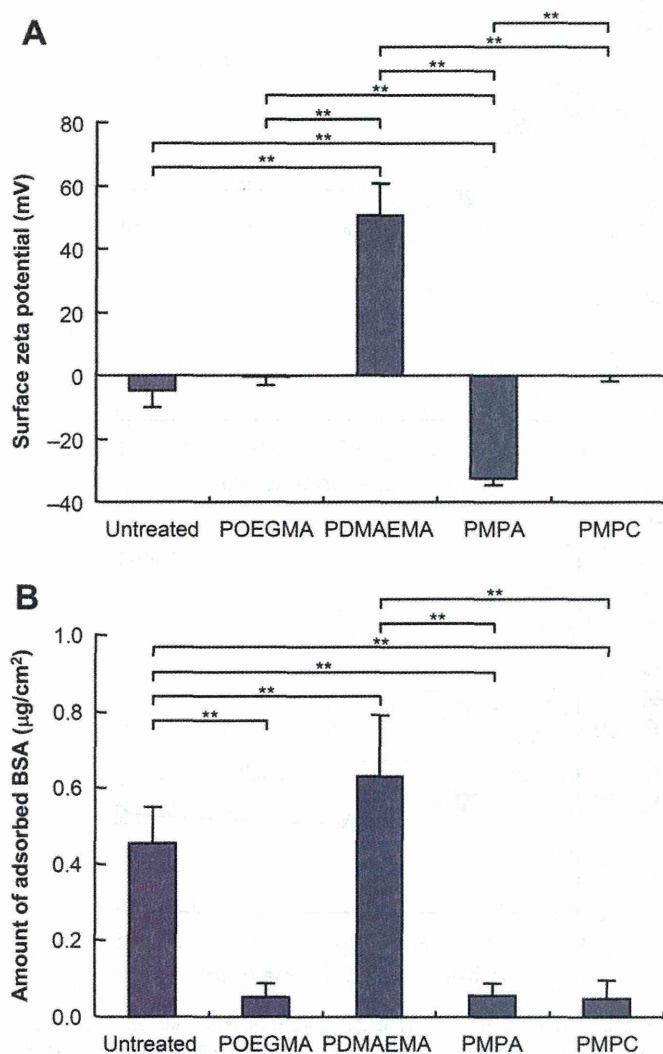
Fig. 8 shows the time course of gravimetric wear of the various polyelectrolyte-grafted CLPE cups during the hip simulator test. PDMAEMA-, POEGMA-, and PMPC-grafted CLPE cups were found to undergo significantly less gravimetric wear than untreated CLPE cups. These gravimetric wear results support the volumetric wear images in Fig. 7A. Furthermore, POEGMA- and PMPC-grafted CLPE cups showed a slight, gradual increase in weight during the hip simulator test. This is partially attributable to greater fluid (e.g., water, proteins, and lipids) absorption in the tested cups than in the load-soak controls. Note that when using the gravimetric method, the weight loss of the tested cups was corrected by subtracting the weight gain in the load-soak controls; however, this correction is not perfect because only the tested cups are continuously subjected to motion and load.



**Fig. 5.** Coefficient of dynamic friction of polyelectrolyte-grafted CLPE samples obtained with a 90-min photoirradiation time in the ball-on-plate friction test under various lubrication conditions. Data are expressed as means  $\pm$  standard deviations. \* Indicates  $p < 0.05$ , \*\* indicates  $p < 0.01$ , and N.S. indicates no statistical difference.

#### 4. Discussion

In natural synovial joints under physiological conditions, fluid film lubrication by the hydrated layer is essential for the smooth



**Fig. 6.** Surface zeta potential and the amount of adsorbed BSA for untreated CLPE and the various polyelectrolyte-grafted CLPE samples with a 90-min photoirradiation time. Data are expressed as means  $\pm$  standard deviations. \*\* Indicates  $p < 0.01$ .

motion of joints [31], and a nanometer-scaled hydrated polyelectrolyte layer that covers the joint cartilage surface provides hydrophilicity and works as an effective boundary lubricant [14–17]. Hence, grafting a cartilage-like hydrophilic polymer layer onto the bearing surface of an artificial joint may afford the same hydrophilicity and lubricity of the physiological joint surface [32]. With this viewpoint, we specifically investigated whether (1) the hydrophilic polymer or polyelectrolyte characteristics would affect the hydration- and friction-kinetics of the hydration layer and (2) the characteristics of hydration lubrication [24] of polyelectrolyte layers might assure the durability of artificial hips.

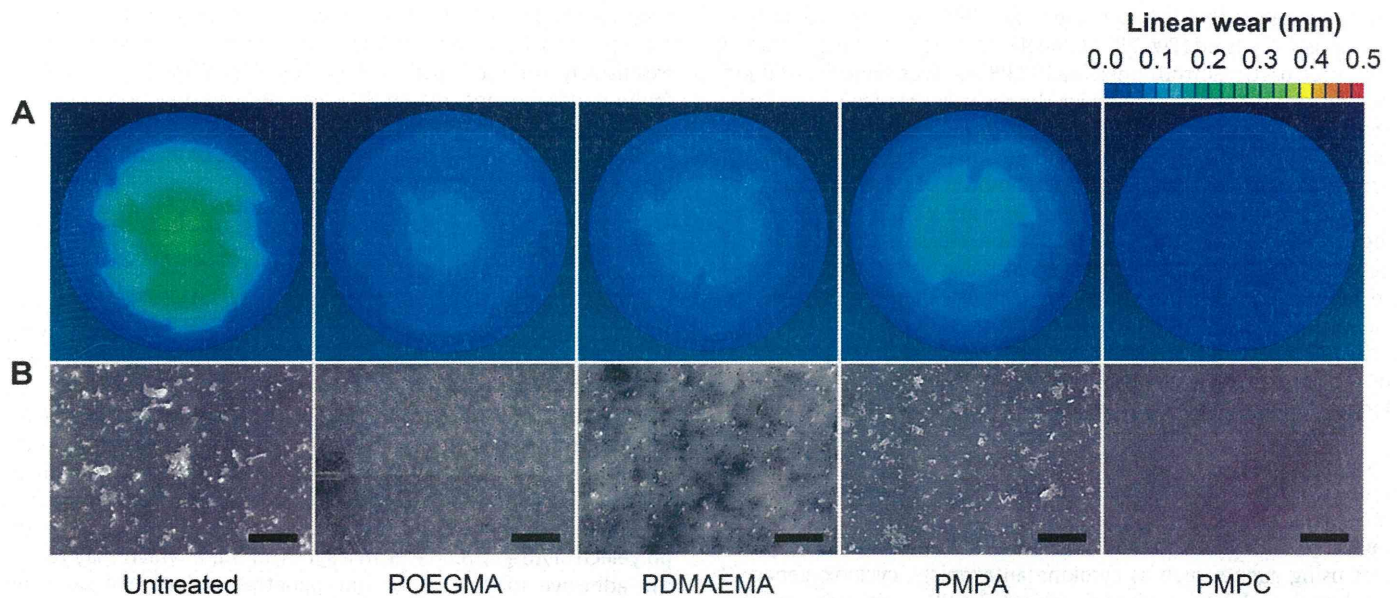
The results of the TEM, FT-IR/ATR, and XPS analyses confirmed successful synthesis of nanometer-scale hydrophilic layers with varying charge (nonionic/neutral, cationic, anionic, or zwitterionic properties) on a CLPE surface. The results also showed that the hydrophilicity of each of the various polyelectrolyte-grafted CLPE surfaces gradually increased with the photoirradiation time, as previously reported (Fig. 4) [23]. To obtain a uniform graft layer on a CLPE surface, the photoirradiation time during graft polymerization must be controlled [27]. Controlled 100-nm-thick, uniform graft layers with varying charge are essential for a proper comparison of friction and wear performance under various lubricant conditions.

Specifically, the results of the present study show that the charge (nonionic, cationic, anionic, or zwitterionic) of the graft layer affects the hydration- and friction-kinetics of the CLPE bearing surface; the coefficient of dynamic friction in ball-on-plate friction test with a water lubricant depends on the hydrophilicity (static-water contact angles) as shown in Figs. 4 and 5. All of the polyelectrolyte-grafted CLPE surfaces exhibited considerably higher lubricity than the untreated CLPE surface in a water lubricant. This is because the water molecules in the hydration layers of the hydrophilic polymer or polyelectrolytes act as very efficient lubricants [32,33]. However, in other lubricant conditions, the polyelectrolyte-grafted CLPE surfaces with cationic and anionic polymer layer exhibited significantly different characteristics compared with neutral hydrophilic polymer- or zwitterionic polyelectrolyte-grafted CLPE surfaces, even though all showed high hydrophilicity.

The PDMAEMA-grafted CLPE samples exhibited a higher coefficient of dynamic friction in the BS lubricant containing proteins (e.g., albumin, globulin) than in water or SBF lubricants in the ball-on-plate friction test [34]. PDMAEMA has a positively charged  $-\text{NH}^+(\text{CH}_3)_2$  group at neutral pH, which in turn attracts negatively charged molecules (e.g., albumin molecules have negatively charged groups at neutral pH) (Fig. 6). This implies that the existence of protein molecules at the bearing interface increases the resistance to sliding motion. Since protein is adsorbed on the Co–Cr–Mo alloy counterface [35,36], the high resistance to sliding motion is interpreted as being the result of higher adhesive interaction or interpenetration of the adsorbed protein films formed on both PDMAEMA-grafted CLPE and Co–Cr–Mo alloy surfaces. However, the PDMAEMA-grafted CLPE cups exhibited high wear resistance in the hip simulator test at high load, despite the high coefficient of dynamic friction in the ball-on-plate friction test in the BS lubricant. This may be because the adsorbed protein film from the BS lubricant is squeezed out of the bearing interface at high load [37].

In contrast, the PMPA-grafted CLPE surfaces exhibited considerably poor lubricity in the ball-on-plate friction test with an SBF lubricant. The chemical structure of the negatively charged PMPA is characterized by the presence of a large number of trap sites for positively charged inorganic ions. Hence, the poor lubricity of PMPA-grafted CLPE surfaces is interpreted as being due to shrinkage or bridging of negatively charged polyelectrolyte chains, which reduced the mobility of the chains in a solution containing positively charged inorganic ions [32–34]. In a previous study, Kato et al. reported that PMPA-grafted poly(ethylene terephthalate) surfaces induced hydroxyl apatite deposition in SBF, mimicking biological mineralization [38]. In this study, a similar phenomenon was observed as the PMPA-grafted CLPE surfaces exhibited calcification-like mineralization after pre-soaking in SBF for 24 h before the friction test. (The chemical composition of the mineralized thin layer on the PMPA-grafted CLPE surfaces could not be detected by thin film X-ray diffraction analysis.) The BS lubricant in this hip simulator test contained not only protein molecules but also inorganic ions. Further, during the  $3.0 \times 10^6$  cycles of the hip simulator test, the test cups were soaked in the BS lubricant for over 30 days. It is therefore thought that the negatively charged PMPA in turn attracts only positively charged inorganic ions and repels the negatively charged molecules; the PMPA-grafted CLPE cups exhibited higher wear than the other polyelectrolyte-grafted CLPE cups due to abrasive wear by the mineralized hardened surface layer [39].

The POEGMA- and PMPC-grafted CLPE cups exhibited high wear resistance in the hip simulator tests (Figs. 7 and 8) as well as low coefficients of dynamic friction in the ball-on-plate friction tests (Fig. 5). In particular, the highly hydrated surface layer of PMPC-grafted CLPE provided extremely efficient lubrication under all



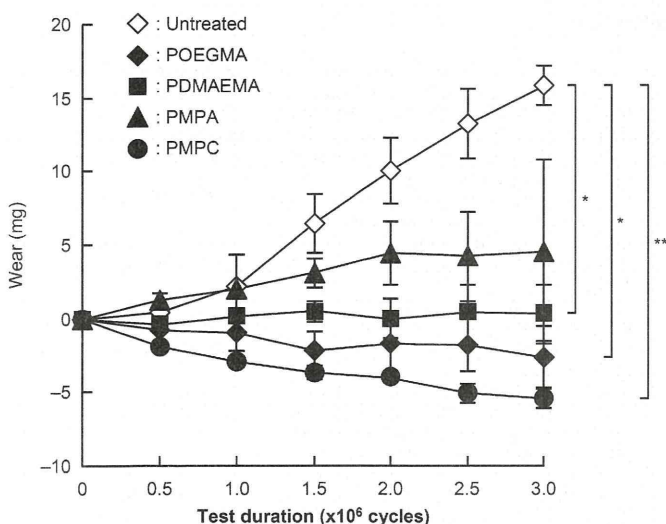
**Fig. 7.** Analysis of various polyelectrolyte-grafted CLPE cups after the hip simulator test. (A) Three-dimensional coordinate measurements of various polyelectrolyte-grafted CLPE cups and (B) SEM images of wear particles isolated from lubricants of the hip simulator test. Scale bar indicates 5  $\mu\text{m}$ .

conditions. Moreover, the zeta potential of the PMPC-grafted CLPE surface was close to zero because the ionic group in the MPC unit forms an inner salt and the electrostatic effects are diminished. Therefore, the zwitterionic grafted polymer layers attract water molecules only, and repel the protein molecules (Fig. 6B) and positively charged inorganic ions. This characteristic is the same as that of the nonionic grafted polymer layers. Chen et al. reported that the water molecules adsorbed on the surface of highly hydrophilic polyzwitterionic (i.e., PMPC) brushes act as lubricants to reduce the interaction between the brushes and the counter-surface [40]. Recent efforts to identify hydrophilic polymers or polyelectrolytes have focused on nanotribological studies of surface-attached molecules, seeking to emulate those at the cartilage surface [32,41,42]. However, the above results for PDMAEMA- and PMPA-grafted CLPE in the hip joint simulator tests demonstrate

that the mechanism of action of the polyelectrolyte-grafted layer cannot be explained simply in terms of fluid film lubrication. Therefore, we believe that the primary mechanism underlying the low friction and high wear resistance is the high level of hydration of the polyelectrolyte layer, such as the zwitterionic PMPC-grafted layer; water molecules in the hydration layers act as very efficient lubricants [32,42,43]. The secondary mechanism is attributed to a repulsion of protein molecules by the positively charged inorganic ions of the polyelectrolyte layer in a synovial fluid, which may reduce the adhesive interaction or interpenetration between opposing Co–Cr–Mo alloy surfaces or adsorbed protein films on the Co–Cr–Mo alloy [8,40].

After  $3.0 \times 10^6$  cycles of the hip simulator test, the POEGMA-, PDMAEMA-, and PMPC-grafted CLPE cups show a greater than 97% reduction in steady wear rate ( $-1.82$ – $0.12$  mg/ $10^6$  cycles) compared with untreated CLPE (Fig. 8). This suggests the approach may be promising for extending the longevity of THA prosthetics [44]. To explain this we assume that the hydrated bearing surface of the artificial joint modified with PMPC exhibited fluid film lubrication (i.e., hydration lubrication [24,45]) and suggest this artificial hip mimics the cartilage or SAPL layer on the cartilage in natural joints [23]. Although the PMPC layer has no direct analog to the cartilage surface containing SAPL, the results of the present study underline the possible importance at such surfaces of highly hydrated macromolecules in the chondroprotective and lubrication roles.

The POEGMA- and PMPC-grafted CLPE cups did not lose weight during the hip simulator test; instead, they gained weight even after correction for water absorption in the load-soak control, suggesting an underestimation of the load-soak control, as previously reported [12,46]. The cups showed comparable weight gains after  $3.0 \times 10^6$  cycles, irrespective of the presence or absence of the POEGMA- and PMPC-graft layers, confirming that the weight gain was the result of the water absorbed by the cup material (i.e., the CLPE substrate) and not the result of retention of extraneous materials on the surface POEGMA- and PMPC-graft layers. These cups decreased the amount of wear particles isolated from the lubricants. Because wear particles from POEGMA- and PMPC-graft CLPE surfaces were hardly observed as a result of their extremely



**Fig. 8.** Time course of gravimetric wear of various polyelectrolyte-grafted CLPE cups during the hip simulator test. Data are expressed as means  $\pm$  standard deviations. \* Indicates  $p < 0.05$ , \*\* indicates  $p < 0.01$  as compared with an untreated CLPE cup.

small amounts, they could not be classified according to size, as previously reported (Fig. 7B) [23,44,47]. In contrast, the majority of the wear particles from untreated CLPE surfaces were 0.1–1.0  $\mu\text{m}$  across. In addition to enhancing the wear resistance of the cups, reducing bone resorptive responses to generated wear particles is important for preventing periprosthetic osteolysis. Such responses are dependent not only on the total amount of wear particles, but also on the proportion of particles that are within the most biologically active size range. The CLPE cups release a large number of sub-micrometer and nanometer-sized particles which are known to induce a greater inflammatory response than larger particles. Hence, although CLPE indeed causes a reduction in the total amount of particles, it might not necessarily lead to the prevention of periprosthetic osteolysis. Clinical evidence regarding the longevity of artificial joints with CLPE is anticipated in the future.

A previous study has shown that polymer particles covered with PMPC are biologically inert with respect to phagocytosis by macrophages and subsequent bone resorptive actions [48]. An increasing number of studies are exploring potential pharmacological modifications for the adverse host response to wear particles using agents such as cytokine antagonists, cyclooxygenase-2 inhibitors, and osteoprotegerin, or anti-RANKL (receptor activator of NF- $\kappa\text{B}$  ligand) antibody. However, they may cause serious side effects because the agents must be taken for a long period after surgery and because they are not currently targeted to the site of the problem. Surface modifications of MPC polymers for other medical devices have suppressed biologic reactions when they are in contact with living organisms [49] and are now clinically used on the surfaces of intravascular stents [50], soft contact lenses [51], and artificial lungs and hearts [52] under the authorization of the Food and Drug Administration of the United States. Hence, PMPC grafting is superior to these developing pharmacologic treatments, because the absence of side effects has already been confirmed clinically by several medical devices. In addition, based on other related evidences and multicenter clinical trials, the Japanese government (Ministry of Health, Labor, and Welfare) approved the clinical use of PMPC-grafted CLPE acetabular liners (Aquala<sup>®</sup> liner; Japan Medical Materials Corp.) in artificial hip joints in April 2011.

Despite these promising results, our study is subject to a number of limitations. First, we did not entirely capture the stabilities of the polyelectrolyte-grafted layers. Surface modification was accomplished using same photoinduced radical polymerization technique as that for methacrylate monomers. Hence, we believe the present surface modification layers combined with the substrate by strong C–C covalent bonding to attain similar stability. Second, we used a confined period for the hip simulator test. Although experiencing  $3.0 \times 10^6$  cycles in the hip simulator is comparable to 3–5 years of physical walking, the duration may not be sufficiently long for young active patients. We are now running the hip simulator longer and thus far have confirmed almost no wear on the PMPC-grafted CLPE cups after  $1 \times 10^7$  cycles [44]. Third, we did not entirely capture the range of loading and motion conditions of the *in vivo* environment in terms of the variety of positions, the magnitude of loading, or the daily routine, although we believe this, according to the ISO standard 14242-3, can provide some indication of the wear performances. Fourth, the procedure for the isolation of wear particles in this study does not entirely capture the contribution of wear particles with a diameter of less than 0.1  $\mu\text{m}$ , as previously reported [53]. Cellular response to particles is thought to be dependent upon factors such as particle number, size, shape, surface area, and material chemistry. If nanometer-scaled particles are generated *in vivo*, it will be important to determine their biological activity in relation to that of micrometer-scaled particles. Fifth, the wear performance we report is only valid for this specific combination of Co–Cr–Mo alloy and

polyelectrolyte-grafted CLPE. It was well known that protein molecules such as albumin from the lubricant are adsorbed more extensively on the surface of Co–Cr–Mo alloy than on other femoral component materials such as alumina, zirconia, and zirconia-toughened alumina [35–37,54]. Hence, we believe this study can provide more effective indication of the wear performance.

## 5. Conclusions

We evaluated the friction and wear performance of controlled, 100-nm-thick, uniform hydrophilic grafted polymer layers with various types of surface charge in various lubricant conditions and obtained clear preliminary evidence for their effectiveness. The primary mechanism underlying the low friction and high wear resistance must be attributed to the high level of hydration of the grafted layer, where water molecules act as very efficient lubricants. The secondary mechanism is considered to be repulsion of protein molecules and positively charged inorganic ions by the polyelectrolyte-grafted layer in a synovial fluid, which may reduce the adhesive interaction or interpenetration between opposing Co–Cr–Mo alloy surfaces themselves or between protein films adsorbed on the Co–Cr–Mo alloy. Finally, we showed that the nanometer-scaled hydrophilic polymer or polyelectrolyte layer on the CLPE surface can confer high durability to acetabular cup bearings in THA. Our findings may have implications on future studies on surface modification with cartilage-like or SPAL-like layers, which are of great importance in the design of lubricated surfaces for artificial joints.

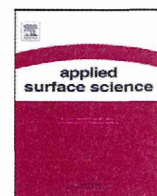
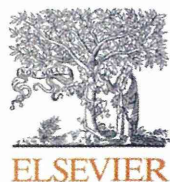
## Acknowledgments

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## Platelet adhesion-resistance of titanium substrate with mussel-inspired adhesive polymer bearing phosphorylcholine group

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### ABSTRACT

To make improvement of hemocompatibility on the titanium (Ti) substrate, mussel-inspired adhesive polymer (PMDP) with both catechol groups for surface anchoring and phosphorylcholine groups for obtaining biocompatibility was synthesized. Only 10 s-immersion of Ti substrate into an aqueous solution of PMDP resulted in reliable coating, because the coating was stable during one month rinsing. The PMDP-coated Ti substrate showed significant reduction of amount of proteins and number of platelets adhered on the surface compared with the non-treated Ti substrate. For this immobilization method, no chemical pretreatment of the substrates is necessary, and a stable and hemocompatible coating can be prepared by a simple, one-step procedure. Thus, we concluded that the proposed PMDP immobilization method has promising approach for improving hemocompatibility of Ti substrate for cardiovascular implant application.

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### 1. Introduction

Titanium (Ti) substrate is one of the biometals and has been widely used in biomedical devices and components since the late 1970s, especially in cardiac and cardiovascular applications (e.g., prosthetic heart valves, protective cases in pacemakers, implantable blood pumps, cardiovascular stents, and circulatory devices) [1]. The Ti substrate has many desirable properties such as a relatively low Young's modulus, good fatigue strength, formability, machinability, and corrosion resistance. However, the Ti substrates induce severe biological responses such as thrombus formation and tissue reactions [2]. As a result, anticoagulant therapy is necessary to minimize the risk of thromboembolic complications. Therefore, surface modifications of Ti substrates are indispensable for improving its hemocompatibility.

Protein adsorption is the first essential event followed by biological responses such as acute thrombus formation and inflammation and then fibrous encapsulation, bacterial adhesion, and infection

[3]. It is generally believed that reducing protein adsorption on the substrates can significantly attenuate these adverse biological responses. One well-known polymeric material used to prevent protein adsorption is hydrophilic poly(ethylene glycol) (PEG) [4]. Indeed, PEG functions well under both *in vitro* and *in vivo* conditions for a relatively short period. However, because PEG-based materials are susceptible to degradation by spontaneous oxidation under physiological conditions, these systems lack long-term stability, which reduces their effectiveness as a surface modifier [5]. In other words, PEG-based materials are not suitable for use in implantable cardiovascular devices.

Another promising and effective way of preventing protein adsorption to attain biocompatibility is to construct an artificial cell membrane surface on the substrates using phospholipid polymers. Such polymers have been synthesized using 2-methacryloyloxyethyl phosphorylcholine (MPC), which is a methacrylate monomer bearing the same polar group as that in the natural phospholipid molecules in the side chain [6,7]. In late 1980s, Ishihara developed a synthetic route for high-quality MPC and its polymers, which have been successfully applied for preparation of medical devices, implantable artificial organs, and cosmetics worldwide [8–12]. The MPC polymers show adequate stability both chemically and physically even when under *in vivo* conditions.

There are many reports of surface immobilization of MPC polymers on Ti substrates. However, these methods have many limitations for widespread practical use. Layer-by-layer assembly

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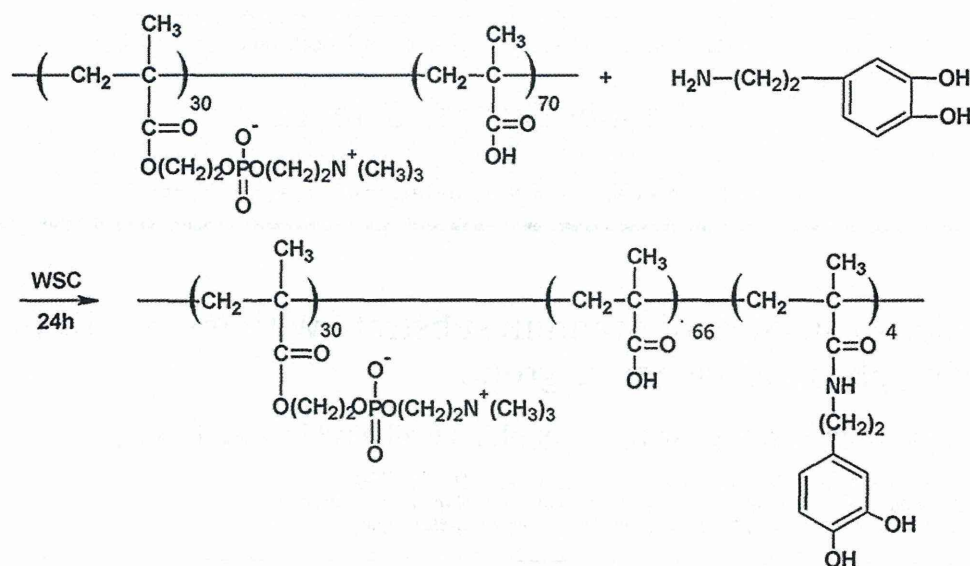


Fig. 1. Synthetic route of PMDP.

involves complex multistep procedures [13], the self-assembled monolayer technique requires surface-specific interaction [14], and surface-initiated atom transfer radical polymerization needs unstable polymerization conditions [15]. Thus, for practical applications, it is desirable to use a more simple, convenient, and versatile method to immobilize MPC units on Ti substrate surface [16].

Recently, to facilitate convenient adhesion of organic compounds to metal substrates, mussel-inspired chemistry has been widely investigated [17,18]. Mussels can rapidly and permanently adhere to all types of inorganic and organic wet surfaces in aqueous environments, which was explained by coordination bond on metal oxide, and covalent bond on organics [19], and that study was performed on Ti surface, the same as our research. It is widely speculated the catechol group is essential for adhesion [20,21]. This mussel-inspired chemistry can be used for surface modification using polymers. When a polymer with catechol groups is in contact with a metal substrate, the thin polymer film is spontaneously deposited on the surface. The functionalization of the polymers imparts new characteristics to the metal substrate. In fact, it has been reported that end-functionalized PEG-catechols could be immobilized on limited types of substrates such as titanium oxide (TiO<sub>2</sub>), gold, and stainless steel [22,23]. PEG-g-catechol with multiple catechols tethered on the backbone of PEG was able to modify versatile surfaces including noble metals, metal oxides and polymers, and greatly reduced protein and cell adhesion, but the synthesis of catechol-g-PEG is a difficult task requiring multi-step procedures [24]. Mussel-inspired heparin coating of polyurethane showed significant inhibition of blood coagulation and platelet adhesion [25].

In our previous work, water-soluble MPC polymers (PMDP) with catechol groups in the side chain were synthesized, and the stability and protein adsorption resistance of the PMDP coated Ti was discussed [16]. In this paper, we further explored the surface characteristics and stability of the coating polymer layer. In addition, to verify the improvement of hemocompatibility, we examined platelet adhesion on the Ti substrate after modification using PMDP.

## 2. Materials and methods

### 2.1. Materials

Water-soluble MPC polymer, poly(MPC-co-methacrylic acid(MAA)) (PMA), was obtained from NOF Co. (Tokyo, Japan),

which were synthesized by conventional radical polymerization of MPC and MAA [26]. The composition of the MPC units in PMA was 30 unit-mol%. The number average molecular weight (*M<sub>n</sub>*) of PMA was  $2.7 \times 10^5$ . Dopamine hydrochloride was purchased from Sigma-Aldrich (St. Louis, MO, USA). 1-Ethyl-3-(3-dimethylaminopropyl) carbodiimide, hydrochloride (WSC) was purchased from Dojindo (Kumamoto, Japan). Ti substrate with thickness of 1.0 mm was purchased from Sumitomo Metals, Ltd. (Tokyo, Japan). The Ti substrate was cut into 10 mm × 10 mm pieces and the mirror polished Ti substrates were cleaned using oxygen plasma apparatus (PR500 plasma reactor, Yamato Science, Tokyo, Japan) for 10 min before use.

### 2.2. Synthesis of PMDP

The PMDP was synthesized by condensation reaction between PMA and dopamine hydrochloride. The reaction scheme is shown in Fig. 1. Dopamine hydrochloride and WSC were dissolved in 4 mL of PMA aqueous solution (5.0 wt%), and 96 mL of pH 6.0 buffered solution (potassium dihydrogen phosphate and sodium hydroxide) was added. The reaction was carried out at room temperature for 24 h under Ar gas atmosphere to prevent the oxidation of the catechol groups. The molar ratio [dopamine hydrochloride]/[COOH] was 2.0. After the reaction, the polymer solution was filtered using ultrafiltration membranes (Millipore Co., USA; molecular size cut off:  $3.0 \times 10^4$ ) until there was no further release of unreacted dopamine through the membrane, which was confirmed by ultraviolet (UV, V-560, Jasco Co., Tokyo, Japan) adsorption. The polymer was freeze-dried.

### 2.3. Surface modification on Ti substrate with PMDP

Aqueous solution containing 2.0 mg/mL of PMDP was prepared using pure water (pH about 5.5). The Ti substrate was simply dipping in the PMDP aqueous solution at room temperature for 10 sec for preparing the PMDP-Ti substrate.

Toluidine blue method was used to confirm that the substrate was coated with the solution. Samples with a diameter of 10 mm were immersed into a 500 mmol/L toluidine blue solution with pH 10 (set by sodium hydroxide). After shaking for 12 h at 37 °C, samples were washed (3 times for 15 min) with sodium hydroxide solution (pH 10). Samples were shaken for 30 min at 37 °C in acetic

acid solution (50%) to dissolve the adsorbed toluidine blue on the surfaces of the samples. The toluidine blue concentration (which was similar to the surface carboxyl group concentration of the film) of the solution was colorimetrically determined with microplate reader (uQuant, Bio-tek instruments Inc., Vermont, USA) at 631 nm.

To follow the polymer adhesion process, a quartz crystal microbalance (QCM) sensor was used with dissipation monitoring (QCM-D, Q-Sense, Gothenburg, Sweden) and a fundamental resonant frequency of 5.0 MHz. The Ti-coated (Ti/Au) QCM sensor obtained from Q-Sense was cleaned using oxygen plasma for 10 min before use. The QCM sensor was exposed to the water solution until a stable baseline of the QCM signals was obtained. The QCM cell was then filled with 2.0 mg/mL of PMDP aqueous solution. After the PMDP solution was retained for 20 min in the QCM cell, phosphate buffered saline (PBS) solution was flowed to replace the PMDP solution and wash away the weakly adsorbed PMDP from the surface. The QCM signals were monitored throughout the procedure. All the measurements were performed at 37 °C and repeated at least three times.

#### 2.4. Surface characterization and stability evaluation of the coating polymer layer

After coating, the PMDP-Ti substrate was rinsed in PBS at 37 °C for a month, and the chemical structure, surface element and morphology of the film before and after rinsing were tested.

The chemical structure of the film was investigated by Fourier transform infrared spectroscopy (FT-IR, NICOLET 5700, Nicolet Biomedical, Bloomfield, USA) over the range from 400 to 4000  $\text{cm}^{-1}$ .

The surface elemental analysis was carried out by an X-ray photoelectron spectroscope (XPS, XSAM800, Kratos, Manchester, UK) with a Mg-K $\alpha$  radiation source at the anode. The applied voltage was 10–20 kV, and the electric current was 45 mA. The take-off angle of the photoelectron was maintained at 90°. The binding energy scale was referenced by setting C<sub>1s</sub> peak at 284.6 eV.

The surface morphology of the PMDP-Ti substrate was examined by scanning electron microscopy (SEM, JSM-7001F, GEOL, Tokyo, Japan).

#### 2.5. In vitro hemocompatibility

To test the resistance of the substrate surface to protein adsorption, bovine serum albumin (BSA, Sigma–Aldrich) was used without further purification. The amount of BSA adsorbed on the PMDP-coated surface was quantified using the QCM-D. First, a Ti-coated QCM sensor was used as a QCM-D. After flowing 2.0 mg/mL of PMDP aqueous solution through the QCM cell, the sensor was exposed to a PBS (pH 7.4) solution until a stable baseline of QCM signals was obtained. Then, 1.0 mg/mL of BSA in PBS was flowed to fill the QCM cell. After the BSA solution was retained for 20 min in the QCM cell, the PBS solution was flowed to replace the BSA solution and wash away the weakly adsorbed BSA from the surface. The QCM signals were monitored throughout the procedure. All the measurements were performed at 37 °C and repeated at least three times.

Fresh human whole blood used in our experiments was obtained legally from the central blood station of Chengdu, China. The analysis was performed within 12 h after the blood donation. The amount of the samples used for statistical count was not less than four.

*In vitro* platelet adhesion experiment was carried out to evaluate the surface thrombogenicity of the samples and to examine the interaction between blood and the materials *in vitro*. Fresh human whole blood was centrifuged at 1500 rpm for 15 min to separate the platelet from blood corpuscles, and the resulting platelet rich plasma (PRP) was used for the platelet adhesion experiment. After incubation in PRP 37 °C for 2 h, samples were rinsed three times

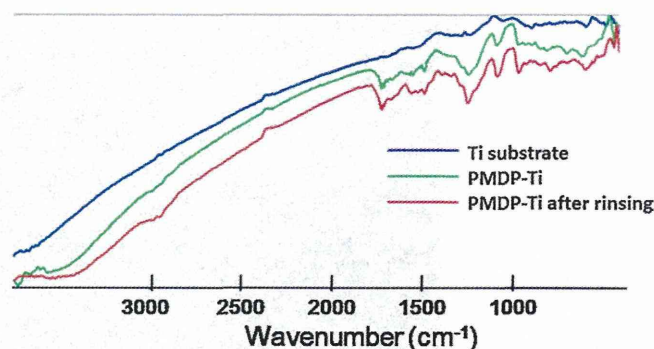


Fig. 2. IR spectra of Ti substrate, PMDP-Ti substrate, and PMDP-Ti substrate after one month rinsing in PBS at 37 °C.

with phosphate buffered solution (PBS) and treated by 2.5% glutaraldehyde overnight, and were dehydrated and dealcoholated, and were dried by critical point dryer (BAL-TEC, CPD030, Liechtenstein, Switzerland). Finally, the samples were gold deposited and examined by SEM. The size of all of the samples applied to experiments was 10 mm in diameter.

The ratio of platelet adhered on the sample surface was determined by measuring the lactate dehydrogenase (LDH) activity of the cells lysed with Triton X-100. Human fresh platelet rich plasma (PRP, 50  $\mu\text{L}$ ) were placed on each sample and incubated at 37 °C for 45 min. Samples were subsequently rinsed three times in PBS. After rinsing, 40  $\mu\text{L}$  Triton X-100 (diluted to 1%) was dropped onto the surface of the sample. 5 min later, 25  $\mu\text{L}$  of the lysates was taken from each surface and mixed with a substrate solution of 200  $\mu\text{L}$  NADH and sodium pyruvate in a 96-well plate. Lastly, the absorbance of above solutions was examined using microplate reader at 340 nm. The slope of the curve was evaluated.

### 3. Results and discussion

#### 3.1. Surface modification on Ti substrate with PMDP and stability evaluation of the coating polymer layer

##### 3.1.1. Amount of adhered PMDP

After immersing in the aqueous solution of the PMDP for 10 s, the Ti substrate was pulled out and dried under vacuum. This is very simply method for modifying the Ti substrate with the PMDP [17]. Toluidine blue test indicated that the concentration of carboxyl groups at the Ti substrate was  $15.4 \pm 1.1 \text{ nmol/cm}^2$  after coating, indicating PMDP adhered on Ti immediately.

The adhesion of the PMDP on the Ti substrate was further followed with QCM-D. A small amount of PMA was deposited on the substrate (little change in frequency); on the other hand,  $354 \text{ ng/cm}^2$  of the PMDP immediately adhered to the substrate (20 Hz change in frequency), and the amount of adhered polymer was stable during 20 min [16].

These results show that the PMDP adhered to the Ti substrate immediately from its aqueous solution *via* the dipping procedure.

##### 3.1.2. Chemical structure and components

As shown in Fig. 2, after coating, the catechol group in PMDP was verified by the appearance of an absorbance peak at  $1553 \text{ cm}^{-1}$ , which is attributed to the aromatic ring of dopamine hydrochloride. In addition, the presence of the ester carbonyl group of the methacrylate units in PMDP was verified by the appearance of an absorbance peak at  $1715 \text{ cm}^{-1}$ . After one month rinsing, the peaks of catechol groups and ester carbonyl group in the FTIR spectra indicated the presence of PMDP on the Ti substrate.

The XPS signals of phosphorus atom and carbon atoms were observed at 133 eV and 285–288 eV, respectively. The data