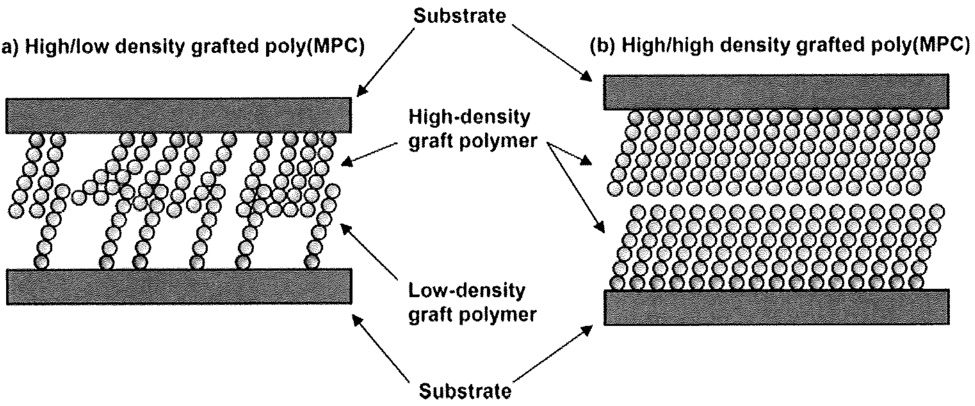

Figure 9. Hypothetical cartoons of high-density grafted poly(MPC)/low-density grafted poly(MPC) and high-density grafted poly(MPC)/high-density grafted poly(MPC) bearing interfaces.

gregate brush is bonded with the collagen network on the cartilage surface.[59] The bearing surface with poly(MPC) in artificial hip joints is assumed to have a brush-like structure similar to that of articular cartilage. CLPE-*g*-MPC/Co-Cr-Mo-*g*-MPC or cartilage/Co-Cr-Mo-*g*-MPC bearing couples can therefore be regarded to mimic the natural joint cartilage *in vivo.* The friction coefficient of cartilage/stainless steel (SUS) pin-on-plate ranges from 0.01 to 0.05,[61] and that of cartilage/cartilage pin-on-plate is 0.02,[62] as shown in Table I. In this study, it was found that CLPE-*g*-MPC/Co-Cr-Mo-*g*-MPC or cartilage/Co-Cr-Mo-*g*-MPC bearing couples mimic a natural joint, showed low friction (friction coefficient was ~0.01), as low as that of cartilage/SUS or cartilage/cartilage. Hence, it was considered that the Co-Cr-Mo-*g*-MPC surface is excellent for the femoral head articulating cartilage, because the cartilage/Co-Cr-Mo-*g*-MPC bearing couples showed a constant low friction coefficient of 0.006. We expect that the hemiarthroplasty with the Co-Cr-Mo-*g*-MPC femoral head bearing will be promising to preserve acetabular cartilage and extend the duration before THA in young patients.

On the other hand, in the previous study, the CLPE-*g*-MPC/Co-Cr-Mo-*g*-MPC prepared by the adsorption of the polymer to the substrate, termed as the "grafting to" method bearing couples showed high friction (friction coefficient was 0.12).[36] The poly(MPC) on Co-Cr-Mo used in this study might have a high density because the polymerization method used was surface-initiated graft polymerization, termed as the "grafting from" method, in which the monomers are polymerized from initiators or comonomers, whereas the poly(MPC) on Co-Cr-Mo prepared by the "grafting to" method might have a low density.[34,35] Figure 9 shows the hypothetical cartoons of high-density grafted poly(MPC)/low-density grafted poly(MPC) and high-density grafted poly(MPC)/high-density grafted poly(MPC) bearing interfaces. The high-density grafted poly(MPC)/high-density grafted poly(MPC) bearing interface shows a remarkably lower friction than the high-density grafted poly(MPC)/low-density grafted poly(MPC) bearing interface.[63] Fukuda and coworkers reported that the friction of the bearing couple was higher in low-density polymer brushes than in high-density ones.[64] Therefore, it is assumed that a bearing couple with low-density poly(MPC) brushes may cause high friction by stick-slip motion with interpenetration, as shown in Figure 9(a).[63,65] In contrast, high-density poly(MPC) fabricated by the "grafting from" method may attain low friction, such as that in the case of "superlubricity," owing to resistance to interpenetration by volume effects resulting from chain mobility. The reduction in friction may contribute to the improvement in antiwear properties.[19–21] Although a hip joint simulator test is necessary to examine tribological advantages in human body environments, a superlubricious metal-bearing material would enable the development of a novel biocompatible artificial hip joint system-artificial femoral head for partial hemiarthroplasty and metal-on-polymer/metal type for THA.

CONCLUSION

We prepared a superlubricious metal-bearing material for application as a novel artificial hip joint system: poly(MPC) was grafted onto the surface of a Co-Cr-Mo alloy by employing a MPSi intermediate layer and by using the photoinduced radical graft polymerization technique. The thickness and density of the grafted poly(MPC) layer increased with the MPC concentration and photoirradiation time, respectively. In conclusion, the grafted poly(MPC) layer successfully provided superlubricity to the

Co-Cr-Mo surface, and the CLPE-*g*-MPC/Co-Cr-Mo-*g*-MPC or cartilage/Co-Cr-Mo-*g*-MPC bearing interface, which mimicked a natural joint, showed an extremely low friction coefficient of 0.01, a value that is as low as that of a natural cartilage interface.

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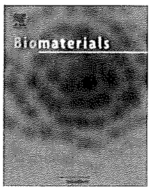

Wear resistance of artificial hip joints with poly(2-methacryloyloxyethyl phosphorylcholine) grafted polyethylene: Comparisons with the effect of polyethylene cross-linking and ceramic femoral heads

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ABSTRACT

Aseptic loosening of artificial hip joints induced by wear particles from the polyethylene (PE) liner remains the ruinous problem limiting their longevity. We reported here that grafting with a polymer, poly(2-methacryloyloxyethyl phosphorylcholine (MPC)) (PMPC), on the PE liner surface dramatically decreased the wear production under a hip joint simulator condition. We examined that the effect of properties of both PE by cross-linking and femoral head by changing the materials on wearing properties of PE. The PMPC grafting on the liners increased hydrophilicity and decreased friction torque, regardless of the cross-linking of the PE liner or the difference in the femoral head materials. During the hip joint simulator experiments (5×10^6 cycles of loading), cross-linking caused a decrease of wear amount and a reduction of the particle size, while the femoral head materials did not affect it. The PMPC grafting abrogated the wear production, confirmed by almost no wear of the liner surface, independently of the liner cross-linking or the femoral head material. We concluded that the PMPC grafting on the PE liner surpasses the liner cross-linking or the change of femoral head materials for extending longevity of artificial hip joints.

1. Introduction

Artificial hip joint replacement is one of the most effective treatments for severe arthritis patients whose number has been on the rise due to the growth of the elderly population [1–3]. The Osteoarthritis Research Society (OARSI) has given this treatment its highest recommendation and approved its cost-effectiveness [4]. Despite improvements in the implant design and surgical technique, the aseptic loosening secondary to periprosthetic osteolysis remains the most serious problem limiting their survivorship and clinical success. Up to 20% of patients implanted total hip replacement with conventional PE develop aseptic loosening within 10 years, and about half of them become disabled due to pain and loss of function [5,6]. The only therapeutic intervention is revision surgery, the number of incidents of which has been on the rise and is projected to double by the year 2026, leading to expansion of the social and economic impact [7,8], unless some limiting mechanism can be successfully implemented to prevent the periprosthetic osteolysis.

The most widely used bearing couple for artificial joint systems is the combination of a polyethylene (PE) acetabular liner and a cobalt–chromium (Co–Cr) alloy femoral head. Periprosthetic osteolysis is known to be triggered by the host inflammatory responses to the PE wear particles originating from the interface [9], which induces the phagocytosis by macrophages and the following secretion of bone resorptive cytokines [10]. Hence, aiming at reduction of the PE wear particles, various trials have been performed. Of these, highly cross-linked PE (CLPE) which underwent gamma-ray or electron beam irradiation experimentally exhibited a 42–93% reduction in the wear rate compared with the conventional PE in vivo [11–13]. Clinical results regarding wear resistance of CLPE are now in medium-term follow-up. The percent reductions in the rate of femoral head penetration have ranged

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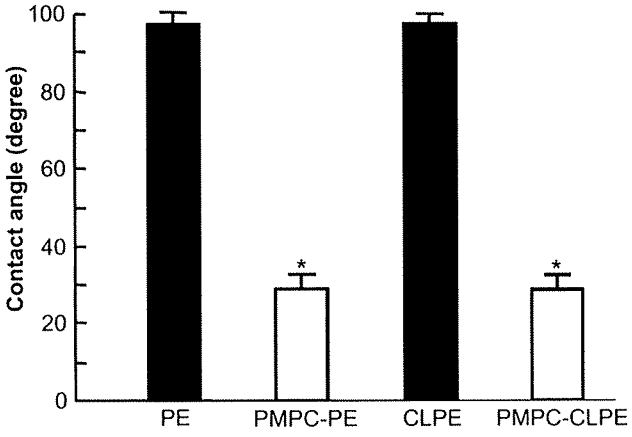

Fig. 1. Contact angles of a water drop on PE plates with or without cross-linking and PMPC grafting. Data are expressed as means (bars) ± standard errors (error bars) for 10 plates/group. *$p < 0.01$, significant effects of PMPC grafting.

from 23 to 95%, depending on which conventional PE was used as a control [14]. Another strategy for reduction of the wear particles is the change of the femoral head material from metal to ceramics to reduce the friction against the PE liner. Among them, alumina ceramic is prevalent with substantially lower PE wear rates than conventional metal heads [15–17].

Considering that the healthy human joint cartilage surface is covered with a nanometer-scaled phospholipid layer [18], grafting a phospholipid-like layer on the liner surface may realize an ideal lubricity resembling the physiological joint surface. We found that the 2-methacryloyloxyethyl phosphorylcholine (MPC) polymers are biocompatible and hydrophilic due to phospholipid polar groups in the side chain [19,20]. The MPC polymers can suppress biological reactions even when they are in contact with living organisms [21–25]. Medical devices such as intravascular stents, soft contact lenses and the artificial lung are used clinically using the MPC polymers [26–28]. Aiming at the elimination of peri-prosthetic osteolysis, we have developed a novel hip CLPE liner with poly(MPC) (PMPC) grafting on its surface, and found that the graft dramatically decreased the production of wear particles during 3×10^6 cycles in the hip joint wear simulator using the Co-Cr alloy femoral head [29–31]. We also showed that the polymer particles grafted with PMPC were biologically inert and did not cause the subsequent bone resorptive responses [31]. For clinical application, we further confirmed the stability of the PMPC grafting on the liner surface even after the gamma-ray irradiation [32,33]. To compare the effects of the PMPC grafting, liner cross-linking, and the ceramic femoral head, the present study examined the wear resistance by the PMPC grafting on PE and CLPE liners, and against metal and alumina ceramic femoral heads.

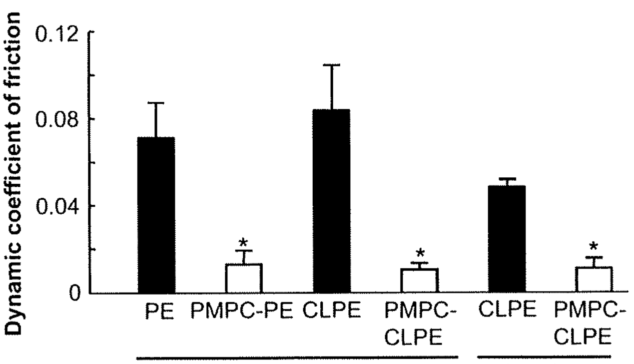

Fig. 2. Dynamic coefficients of friction for the various types of PE plates against Co–Cr alloy and alumina ceramic balls. Data are expressed as means (bars) ± standard errors (error bars) for 5 plates/group. *$p < 0.01$, significant effects of PMPC grafting.

2. Materials and methods

2.1. Materials

The MPC was obtained from NOF Co. Ltd. (Tokyo, Japan), which is synthesized by the same procedure as previously reported by Ishihara [19]. PE liners with and without cross-linking (CLPE and PE, respectively) (K-MAX® CLQC and K-MAX® QPIC), and Co–Cr and alumina ceramic femoral heads (K-MAX® HH-02 and K-MAX® HHA) were obtained from Japan Medical Materials Co. (Osaka, Japan). For PMPC grafting, PE and CLPE liners after coating with benzophenone as photosentisizer were placed in the MPC aqueous solution (0.5 mol/L) and photoinduced polymerization on the liner surface was carried out using an ultra-high pressure mercury lamp (UVL-400HA, Riko-Kagaku Sangyo, Funabashi, Japan), as previously reported

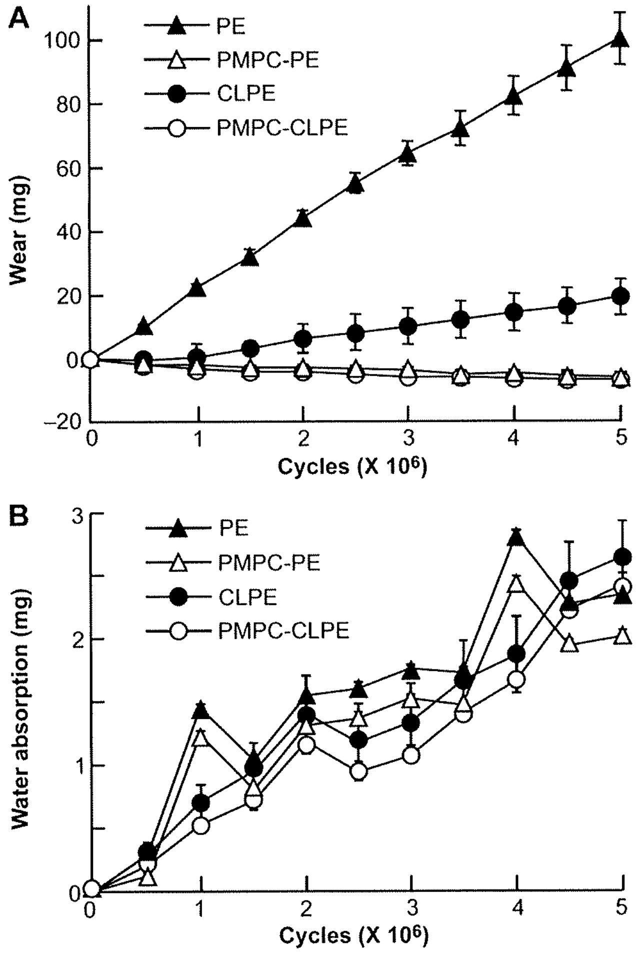

Fig. 3. Wear amount of PE liners with or without cross-linking and PMPC grafting in the hip joint simulator. (A) Time course of wear amount generated from PE, CLPE, PMPC-PE, and PMPC-CLPE liners during 5×10^6 cycles of loading against Co–Cr alloy femoral heads in the hip joint simulator. At intervals of 5×10^5 cycles the liners were removed from the simulator, and the weight loss was determined as the wear amount. (B) Water absorption of the liners which was only loaded axially to the femoral heads without rotational motion to determine the water weight absorbed by the liners during soak in the lubricant. Data are expressed as means (symbols) ± standard errors (error bars) for 8, 10, 5 and 10 liners for PE, CLPE, PMPC-PE, and PMPC-CLPE liners, respectively.

[30,31,34,35]. The wavelength of photopolymerization was selected by a glass color filter as 350 ± 50 nm.

2.2. Surface analyses

The surface elemental conditions of the PE and CLPE liners or plates with and without PMPC grafting were analyzed using highly sensitive X-ray photoelectron spectroscopy (XPS AXIS-HSi165, Kratos/Shimadzu, Kyoto, Japan) equipped with a 15-kV Mg-Kα radiation source at the anode. The functional group vibrations of the surfaces were examined by Fourier-transform infrared (FT-IR) spectroscopy with attenuated total reflection (ATR) equipment. The FT-IR/ATR spectra were obtained using an FT-IR analyzer (FT/IR615, JASCO, Tokyo, Japan) for 32 scans (1.2 s/scans) over the range of 800–2000 cm^{-1} at a resolution of 4.0 cm^{-1}. Hydrophilicity was evaluated by the static water contact angles on the surfaces using the sessile drop method with an optical bench-type contact angle goniometer (Model DM300, Kyowa Interface Science, Saitama, Japan). Drops of purified water (1.0 mL) were deposited on the liner surface, and the contact angles were directly measured with a microscope 60 s after dropping, according to the International Organization for Standardization (ISO) 15989 standard. At least ten contact angles were measured and averaged.

2.3. Dynamic coefficient of friction

The dynamic coefficients of friction between the plates and the Co–Cr alloy and alumina ceramic balls were measured using a ball-on-plate machine (Tribostation 32; Shinto Scientific, Tokyo, Japan) as a preliminary test for tribological properties [36]. A Co–Cr alloy and alumina ceramic ball with a diameter of 9.0 mm was prepared. The surface roughness of the ball was about 0.010 μm, which was comparable to that of the femoral head products. Each of the untreated CLPE and PMPC grafted CLPE surfaces were used to prepare five sample pieces. The friction tests were performed at room temperature with a load of 0.98 N, sliding distance of 25 mm, and frequency of 1 Hz. Pure water was used as the lubricant medium.

2.4. Hip joint simulator

A hip joint wear simulator test was performed under the conditions recommended by the ISO standard [37]. A 12-station hip simulator apparatus (MTS, MTS Systems, Minneapolis, MN, USA) with PE and CLPE liners with or without PMPC grafting in 46 mm acetabular cups that were coupled to 26 mm Co–Cr alloy or alumina ceramic femoral head were mounted on the rotating blocks to produce a biaxial or orbital motion. A Paul-type loading profile which is a physiological walking simulation with continuous cyclic motion and loading was applied (maximum force; 2744 N, frequency; 1 Hz) [38]. A diluted bovine calf serum (25 vol%) in distilled water was used as the lubricant. Sodium azide (10 mg/L) and EDTA (20 mM) were added to prevent microbial contamination and minimize calcium phosphate formation on the implant surface. The simulator was run up to 5×10^6 cycles for 3 months. At intervals of 5×10^5 cycles the liners were removed from the simulator and weighed on a microbalance (Sartorius GENIUS ME215S, Sartorius AG, Gottingen, Germany) to determine the wear amount. The lubricant was collected and stored at −20 °C for further analysis. Since the liners are known to absorb water during their soak in the lubricant, therefore we also measured the weight gain of the liners which were cyclically axial-loaded to the femoral heads without rotational motion (i.e., load-soak control) every 5×10^5 cycles to determine the amount of wear sustained.

After the total loading, morphological change of the liner surface was measured by a 3-dimensional coordinate measuring machine (XYZAX GS800B, Tokyo Seimitsu, Tokyo, Japan). To evaluate the actual removal of material caused by wear, melt-recovery experiments were performed and the liner surface was analyzed with a confocal scanning laser microscope (OLS1200, Olympus, Tokyo, Japan) as previously reported [39]. For the femoral head surface, in addition to field emission scanning electron microscopy (SEM, JSM-6330F, JEOL DATUM, Tokyo, Japan) evaluation, the surface roughness value R_a was measured using a roughness measuring instrument (SURFTEST-501, Mitutoyo, Yokohama, Japan) with a 5 μm diameter contact probe.

For isolation of wear particles, the lubricant after the loading was incubated with 5.0 N NaOH solution in order to digest adhesive proteins that were degraded and precipitated; the particles were then collected and underwent sequential filtrations (0.1 μm of minimum pore size), as previously reported [40]. Particle size was defined using the maximum dimensions determined by the SEM analysis.

2.5. Statistical analysis

The results derived from each measurement in the static water contact angle measurement and friction test were expressed as mean values and standard deviation. Standard analysis of variance (ANOVA) was applied to the data of the existed among those groups, in this study. Two combinations (e.g., PE and PMPC–PE in the static water contact angle measurement) were especially made with Student's t-test ($p < 0.05$).

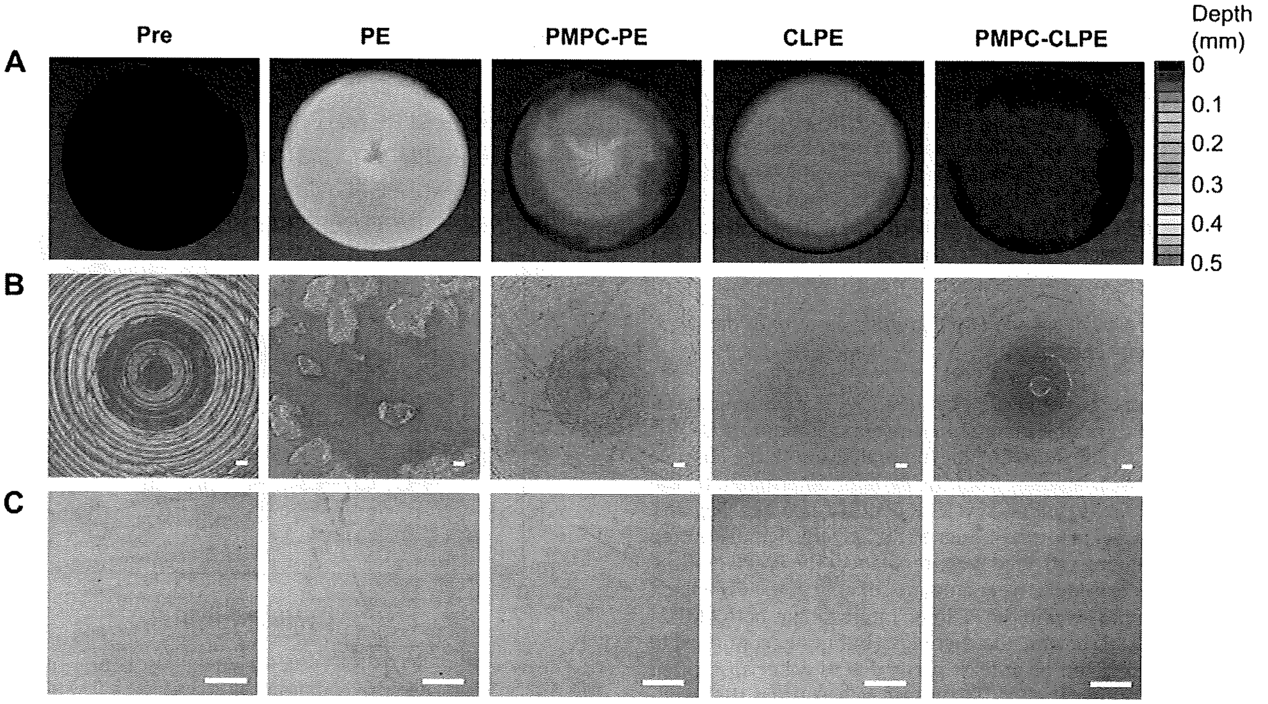

Fig. 4. Optical findings of the surfaces of the four liners and corresponding femoral heads in the hip joint simulator. (A) Three-dimensional morphometric analysis and (B) confocal scanning laser microscopic analysis of PE, PMPC–PE, CLPE, and PMPC–CLPE liners surfaces before (pre) and after 5×10^6 cycles of loading in the hip joint simulator. Scale bars 200 μm. (C) SEM images of the Co–Cr femoral head surfaces before (pre) and after 5×10^6 cycles of loading. Scale bars 5 μm.

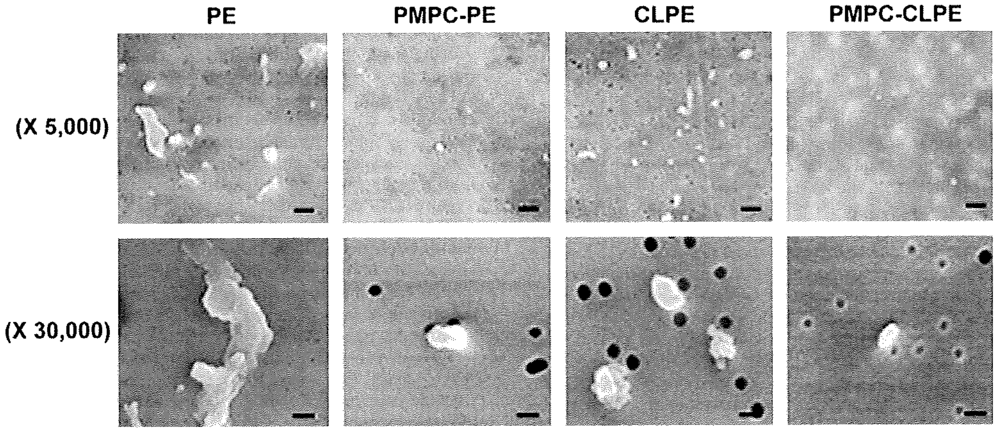

Fig. 5. SEM images of the wear particles isolated from lubricants of the simulators with four kinds of liners. Low (top; ×5000) and high (bottom; ×30,000) magnifications of SEM images of the wear particles isolated from lubricants of the simulators with four kinds of liners. Scale bars 1.0 and 0.1 μm, respectively.

3. Results

3.1. *Surface analyses of PE plates with or without cross-linking and PMPC grafting*

Successful photoinduced graft polymerization of tMPC on both PE and CLPE plates was confirmed by XPS and FT-IR/ATR spectroscopy. The XPS for nitrogen (N_{1s}) and phosphorus (P_{2p}) atoms showed the respective peaks at 403 eV and 133 eV only on the plates with PMPC grafting (PMPC–PE & PMPC–CLPE), but not on those without it (PE & CLPE). These peaks were characteristic of the phosphorylcholine present in the MPC units, since they were assigned to the $-N^+(CH_3)_3$ and phosphate groups, respectively. The FT-IR/ATR spectra representing phosphate group (P–O) at 1240, 1080 and 970 cm^{-1}, and ketone group (C=O) at 1720 cm^{-1} were also confirmed to be detected only on the plates with PMPC grafting. The contact angles of a water drop on the plates after PMPC grafting were 1/3–1/4 those of the original plates (Fig. 1), indicating that PMPC grafting increased hydrophilicity on the hydrophobic PE and CLPE surfaces. We then measured the dynamic coefficients of friction of the plates against Co–Cr alloy and alumina ceramic balls. The friction coefficient was about 80–90% lower in the liners with PMPC grafting than in those without it, regardless of the ball materials (Fig. 2). Cross-linking by itself did not alter both the water-drop contact angles and the friction coefficient (Figs. 1 and 2).

3.2. *Effects of cross-linking and PMPC grafting on wear of the liners in a hip joint simulator*

We then compared the wear of the liners with or without cross-linking and PMPC grafting using a hip joint simulator during 5×10^6 cycles of loading against Co–Cr alloy femoral heads. In the absence of the grafting, wear amount from the PE liner determined by the weight loss was increased in a cycle-dependent manner (Fig. 3A). This was suppressed in the CLPE liner by 80%, clearly demonstrating that the cross-linking provided wear resistance. The PMPC grafting dramatically decreased the wear amount not only in the CLPE liner, but also to a similar level in the PE liner. Interestingly, both PMPC–PE and PMPC–CLPE showed no weight loss, but rather continued to gain weight. This might be at least partly due to water absorption into the liners during the load–soak test. When using the gravimetric method, the weight loss in the tested liners is corrected by subtracting the weight gain in the load–soak controls; however, this correction cannot be perfectly achieved because only the tested liners are continuously subjected to load and motion. Water absorption in the tested liners is generally slightly higher than that in the load–soak controls. Consequently, the correction for water absorption by using the load–soak control as the correction factor leads to a slight underestimation of the actual weight loss. In fact, all four kinds of liners, which were only cyclically axial-loaded to the femoral heads without rotational motion in the simulator, gained weight, regardless of the presence or absence of cross-linking and PMPC grafting (Fig. 3B).

Three-dimensional morphometric analyses confirmed the wear resistance by the cross-linking and PMPC grafting (Fig. 4A). Although the CLPE liner showed less surface wear than the PE liner after 5×10^6 cycles of loading, both PMPC–PE and PMPC–CLPE liners exhibited almost no wear. Confocal scanning laser microscopic analyses of the liner surfaces revealed that the original machine marks that were clearly visible before the loading still remained on the PMPC–PE and PMPC–CLPE surfaces, while they were completely

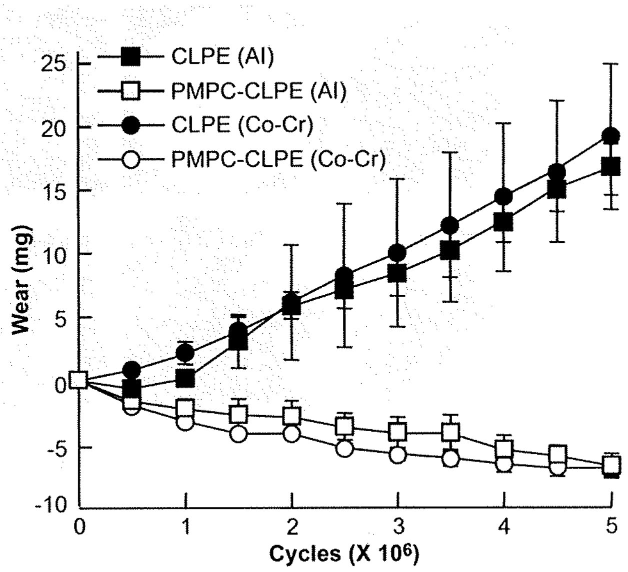

Fig. 6. Wear resistance of CLPE liners with and without PMPC grafting against Co–Cr alloy or alumina ceramic femoral heads in the hip joint simulator. Time course of the wear amount generated from CLPE or PMPC–CLPE liners during 5×10^6 cycles of loading against Co–Cr alloy or alumina ceramic femoral heads in the hip joint simulator. At intervals of 5×10^5 cycles the liners were removed from the simulator, and the weight loss was determined as the wear amount.

obliterated not only on the PE, but also on the CLPE surfaces (Fig. 4B). The XPS analysis also confirmed the remainder of the specific spectra of N_{1s} and P_{2p} at 403 eV and 133 eV, respectively, on the PMPC–PE and PMPC–CLPE surfaces, indicating that PMPC grafting remains even after 5×10^6 cycles of loading. The Co–Cr alloy femoral heads were free of visible scratches, and the surface roughness expressed by the R_a values was not different between before and after the loading in all groups ($R_a = 0.01$–0.02 μm), suggesting that there was no abrasive contamination with metal particles from the heads in the simulator (Fig. 4C).

PMPC grafting on both PE and CLPE liners decreased the amount of wear particles isolated from the lubricants (Fig. 5, top). Since the wear particles from PMPC–PE and PMPC–CLPE surfaces were hardly observed due to the extremely small amount, the wear particles could not be classified according to their size. In contrast, there was a substantial difference in the particle size between PE and CLPE: the majority of the particle length was 1.0–10 μm in PE while 0.1–1.0 μm in CLPE (Fig. 5).

3.3. *Effects of femoral head materials on wear of the liners in a hip joint simulator*

Next, we compared the wear of CLPE and PMPC–CLPE liners that underwent 5×10^6 cycles of loading against Co–Cr alloy and alumina ceramic femoral heads in the hip joint simulator. The wear from the CLPE liner increased in a cycle-dependent manner, and was little affected whether the material was Co–Cr alloy or alumina ceramic (Fig. 6). The PMPC grafting on the CLPE liner abrogated the wear production by femoral heads of both materials similarly. Three-dimensional morphometric analyses (Fig. 7A) and the confocal scanning laser microscopic analyses (Fig. 7B) confirmed a marked enhancement of wear resistance by PMPC grafting, regardless of the femoral head materials. The femoral heads of both materials were free of visible scratches (Fig. 7C), and the surface roughness expressed by the R_a values was not different between before and after the loading in all groups ($R_a = 0.01$–0.02 μm). SEM analyses showed that the number of wear particles of identical shape and size in both materials was decreased by PMPC grafting (Fig. 8).

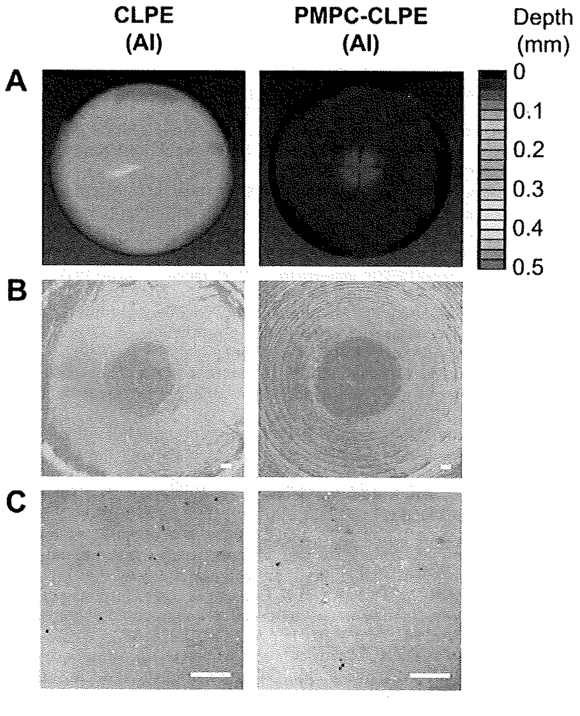

Fig. 7. Optical findings of the surfaces of the CLPE liners with and without PMPC grafting and the alumina ceramic femoral heads in the hip joint simulator. (A) Three-dimensional morphometric analysis and (B) confocal scanning laser microscopic analysis of the liner surfaces after 5×10^6 cycles of loading. Scale bars 200 μm. (C) SEM images of the alumina ceramic femoral head surfaces after 5×10^6 cycles of loading. Scale bars 5 μm.

4. Discussion

The present study demonstrated that surface grafting of MPC on the PE liner diminished the wear production during 5×10^6 cycles of loading in the hip joint simulator. This effect was independent of the presence or absence of cross-linking of PE, as well as the material of the femoral heads.

As a mechanism underlying the wear resistance by the PMPC grafting, we should consider an increase of lubrication between the liner and femoral head. Although phospholipids work as effective boundary lubricants, a study of natural synovial joints showed that fluid film lubrication by the intermediate hydrated layer is the predominant mechanism under physiologic walking conditions [41,42]. Because the present study showed that the PMPC grafting increased hydrophilicity and decreased the friction (Figs. 1 and 2) and our previous study showed that the free water fraction on the MPC polymer surface is kept at a higher level [43–45], the enhancement of wear resistance is likely attributable to the hydrated lubricating layer formed by the PMPC grafting. Contrarily, although cross-linking also suppressed the wear in the simulator, it affected neither the surface hydrophilicity nor the friction (Figs. 1 and 2). This means that cross-linking of PE did not alter the surface lubricity, but increases the hardness of the liner material, so that this procedure may not be essential once the liner surface was grafted with MPC, explaining the similar wear resistance between PMPC–PE and PMPC–CLPE liners (Fig. 3A).

In addition to enhancement of wear resistance of the liner, reduction of bone resorptive responses to wear particles generated is important for the prevention of periprosthetic osteolysis. The responses are dependent not only on the total amount of particles, but also on their proportion which are within the most biologically active size range [46]. The present SEM analysis of

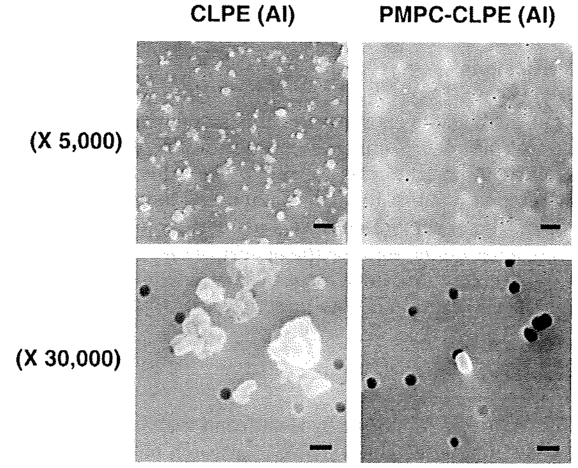

Fig. 8. SEM images of the wear particles isolated from lubricants of the simulators with CLPE liners with and without PMPC grafting against alumina ceramic femoral heads. Low (top; ×5000) and high (bottom; ×30,000) magnifications of SEM images of the wear particles isolated from lubricants of the simulators. Scale bars 1.0 and 0.1 μm, respectively.

the wear particles in the lubricants revealed substantial difference of the particle size distribution between CLPE and PE liners, as previously reported [47,48]. The CLPE liner releases a high number of submicrometer and nanometer-sized particles which are known to induce a greater inflammatory response than larger particles [46]. Hence, although cross-linking of PE indeed causes a reduction in the total amount of particles, it might not necessarily lead to the prevention of periprosthetic osteolysis. Another problem raised with the cross-linking of PE is that it may possibly deteriorate the mechanical properties of the liner causing fatigue cracking [12,49,50]. Clinical evidence regarding the longevity of artificial joints with CLPE is anticipated in the future.

Our previous report has shown that polymer particles covered with the PMPC are biologically inert with respect to phagocytosis by macrophages and subsequent bone resorptive actions [31]. An increasing number of studies address the potential pharmacologic modification of the adverse host response to wear particles [51–53] including cytokine antagonists, cyclooxygenase-2 inhibitors, and osteoprotegerin, or anti-RANKL (receptor activator of nuclear factor κB 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. Because the lack of side effects of the MPC polymer has already been confirmed clinically by several biomedical devices [26–28], this grafting exceeds the developing pharmacologic treatments.

A limitation of this study is the confined period of loading. Although the 5×10^6 cycles in the hip simulator is generally comparable to 5–10 years of physical walking, this may not be long enough for young active patients who show 2×10^6 daily steps/year by the previous study [54]. We are now running the hip simulator with longer loading up to three and a half years, and so far have confirmed almost no wear on the PMPC–CLPE liner after 7×10^7 cycles. Although the results of this study are promising, the third-body wear is a matter of concern. Additional testing needs to be done to determine the effects of third substance on the beneficial influence of PMPC grafting. Another limitation of the hip simulator experiment is that it does not entirely capture the range of loading conditions of a hip, in terms of either the variety of positions or the magnitude of loading, although we believe the simulator experiment can provide some indication of trends.

5. Conclusions

From the advantages shown in this study, we believe that the PMPC grafting for obtaining super-lubrication on the PE liner is more efficient than the cross-linking of the PE liner and the change of the femoral head materials for extending longevity of artificial hip joints. The development of this nanotechnology in the biomaterials science would improve the quality of care of patients having joint replacement and have a substantial public health impact. A large-scale clinical trial is now underway.

Acknowledgements

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Appendix

Figures with essential colour discrimination. Parts of Figs. 4 and 7 in this article are difficult to interpret in black and white. The full colour images can be found in the on-line version, at doi:10.1016/j.biomaterials.2009.02.020.

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Effects of mobility/immobility of surface modification by 2-methacryloyloxyethyl phosphorylcholine polymer on the durability of polyethylene for artificial joints

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Abstract: Surface modification is important for the improvement in medical device materials. 2-Methacryloyloxyethyl phosphorylcholine (MPC) polymers have attracted considerable attention as surface modifiable polymers for several medical devices. In this study, we hypothesize that the structure of the surface modification layers might affect the long-term stability, hydration kinetics, wear resistance, and so forth, of medical devices such as artificial joints, and the poly(MPC) (PMPC) grafted surface might assure the long-term performance of such devices. Therefore, we investigate the surface properties of various surface modifications by using dip coatings of MPC-*co*-*n*-butyl methacrylate (PMB30) and MPC-*co*-3-methacryloxypropyl trimethoxysilane (PMSi90) polymers, or photoinduced radical grafting of PMPC and also the effects of the surface properties on the durability of cross-linked polyethylene (CLPE) for artificial joints. The PMPC-grafted CLPE has an extremely low and stable coefficient of dynamic friction and volumetric wear as compared to the untreated CLPE, PMB30-coated CLPE, and PMSi90-coated CLPE. It is concluded that the photoinduced radical graft polymerization of MPC is the best method to retain the benefits of the MPC polymer used in artificial joints under variable and multidirectional loads for long periods with strong bonding between the MPC polymer and the CLPE surface, and also to retain the high mobility of the MPC polymer. © 2008 Wiley Periodicals, Inc. J Biomed Mater Res 90A: 362–371, 2009

Key words: joint replacement; polyethylene; phosphorylcholine; surface modification; wear mechanism

INTRODUCTION

Polymeric biomaterials are widely used in the biomedical field for manufacturing artificial organs, medical devices, and disposable clinical apparatus.[1,2] Advancements in the biomedical field also demand substantial improvements in polymeric biomaterials. Conventional single-component polymer biomaterials cannot satisfy these requirements. Multicomponent polymer systems have therefore been designed and prepared for new multifunctional biomaterials.

Surface modification is one of the important means of preparing new multifunctional biomaterials. 2-Methacryloyloxyethyl phosphorylcholine (MPC) polymers have attracted considerable attention as surface-modifiable polymers for several medical devices.[3–11] MPC, a methacrylate with a phospholipid polar group in the side chain, is a monomer for preparing novel polymer biomaterials. An excellent synthetic route for MPC has been developed by Ishihara et al.[12] MPC can undergo conventional radical copolymerization with other methacrylate and styrene derivatives such as *n*-butyl methacrylate (BMA), *n*-dodecyl methacrylate (DMA), and 3-methacryloxypropyl trimethoxysilane (MPSi) to form poly(MPC-*co*-BMA), poly(MPC-*co*-DMA), and poly (MPC-*co*-MPSi), respectively.[5–11] These MPC polymers are some of the most common biocompatible

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and hydrophilic polymers studied thus far. They have potential applications in a variety of fields such as biology, biomedical science, and surface chemistry, because they possess unique properties such as good biocompatibility, high lubricity and low friction, antiprotein adsorption, and cell membrane-like surfaces. Several medical devices have already been developed by utilizing the MPC polymers and used clinically; therefore, the efficacy and safety of the MPC polymers as biomaterials are well established.[9–11]

When a natural joint in the human body ceases to function, for example, due to disease, trauma, or overuse, an artificial joint replacement often becomes necessary. There is a substantial increase in the number of artificial hip and knee joints used worldwide each year for primary and revised hip and knee joint replacements.[13] This indicates that a higher quality and longer lifetime have been increasingly desired for artificial joint replacements. Normally, artificial joints allow the body to regain mechanical and biological functions. Medical implants must be adapted to the dynamic loads experienced during use, and they must have the desired long-term biological interaction with the surrounding tissue. A typical artificial joint replacement system used as a medical device comprises a metallic surface made of a cobalt-chromium-molybdenum (Co-Cr-Mo) alloy that articulates against an ultra-high molecular weight polyethylene (UHMWPE) polymeric component. However, the artificial joint replacements are subjected to adhesive and abrasive wear, and both metallic and polymeric debris. These are known to produce a variety of cytokines and tumor necrosis factors that progressively resorb the bone by osteolysis, leading to aseptic loosening of the artificial joint after a number of years, which is recognized as a serious problem.[14,15] Different combinations of bearing surfaces and improvements in bearing materials have been studied with the aim of reducing the number of UHMWPE wear debris that induce osteolysis.[16–18]

Recently, we have developed an artificial hip joint by using poly(MPC) (PMPC) grafted onto the surface of cross-linked polyethylene (CLPE; PMPC-grafted CLPE); this device is designed to reduce wear and suppress bone resorption.[19–24] MPC has also been directly grafted from biomaterial surfaces through photoinduced radical polymerization.[25,26] This photoinduced radical polymerization facilitates the direct grafting of MPC onto biomaterial surfaces. The following are the expected advantages of this technique: (1) controllable graft polymer density and length and grafting site,[21,24] (2) covalent bonding between the graft polymer and biomaterial surfaces (as high immobility), which assures the long-term stability of graft chains, (3) high mobility of the graft polymer chain and/or free end groups of the polymer, and (4) occurrence of grafting only on the surface, and no effect of grafting on the bulk properties.[22] In particular, strong bonding between the surface modification and the surface is an important issue, which is associated with the long-term retention of the benefits of the surface modification used in artificial joints under variable and multidirectional loads, for a promising long-term performance of artificial joints.

In this study, we hypothesize that the structure of surface modification layers might affect the surface density of the phosphorylcholine group, long-term stability and mobility of the polymer chain, hydration kinetics, and so forth, and the PMPC-grafted surface might assure the long-term performance of artificial joints. Therefore, we investigate the surface properties of various surface modification layers with the MPC polymer and the effects of the surface properties on the durability of the CLPE for artificial joints. The results reveal that the structure of the PMPC-grafted layer on the CLPE surface plays an important role in reducing the wear of the orthopedic bearing surface in the long term.

MATERIALS AND METHODS

Materials

MPC was industrially synthesized using a previously reported method.[12] Poly(MPC-*co*-BMA) (PMB30; MPC:BMA unit mole fraction = 0.3:0.7)[12] and poly(MPC-*co*-MPSi) (PMSi90; MPC:MPSi unit mole fraction = 0.9:0.1)[8] were synthesized in ethanol using 2,2′-azobisisobutyronitrile as initiator by a conventional radical copolymerization method. A compression-molded UHMWPE (GUR1020 resin; Poly Hi Solidur, IN) sheet stock was irradiated with 50 kGy γ-rays in N_2 gas and annealed at 120°C for 7.5 h in N_2 gas in order to achieve cross-linking. The CLPE specimens were machined from this sheet stock after cooling.

MPC polymer coating

The preparation of the MPC polymer coated CLPE is schematically illustrated in Figure 1. The physical coating of PMB30 was carried out by the solvent evaporation method, where the CLPE specimens were dipped into ethanol solution containing 0.2 mass % PMB30 for 10 s for coating, and then placed in an ethanol vapor atmosphere at room temperature for 1 h. The coated CLPE specimens were again dipped for 10 s and placed in the ethanol vapor atmosphere at room temperature for 1 h (PMB30-coated CLPE).

The chemical coating of PMSi90 was also carried out by the solvent evaporation method. Before the PMSi90 coating, the CLPE specimens were irradiated with O_2 plasma at a 200 W high-frequency output and 150 mL/min O_2 gas flow for 2 min by using an O_2 plasma etcher (PR500,

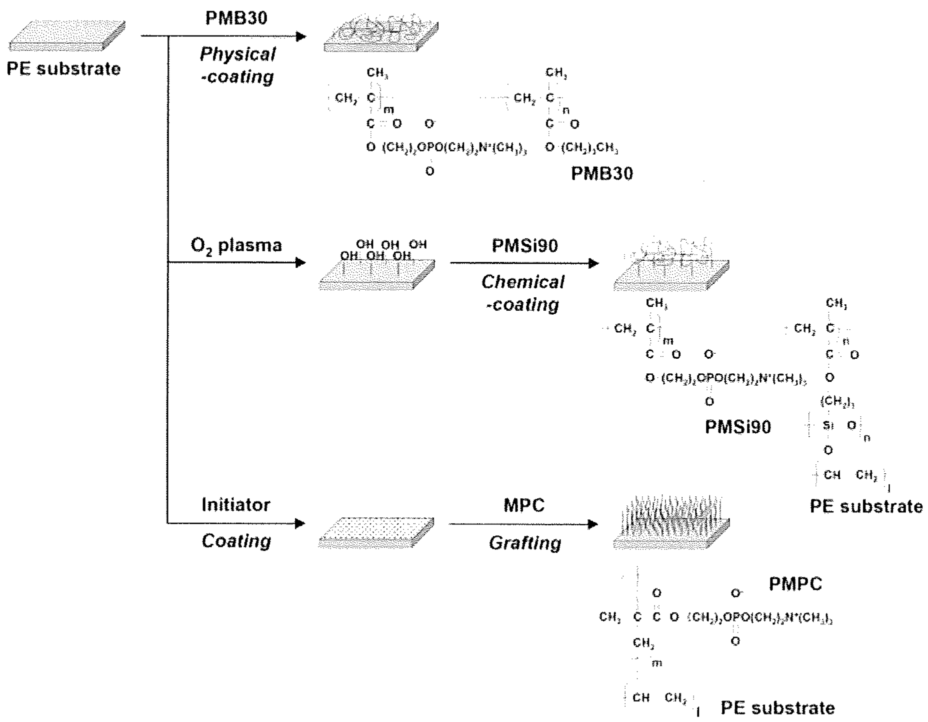

Figure 1. Scheme for the preparation of MPC polymer-coated CLPE and PMPC-grafted CLPE.

Yamato Scientific Co., Tokyo, Japan). The O_2 plasma irradiation formed the surface hydroxide layer. The CLPE specimens were dipped into ethanol solution containing 0.5 mass % PMSi90 and 0.063 mg/mL succinic acid (Kanto Chemical Co., Tokyo, Japan) for 10 s for the silanization of trimethoxysilane group of PMSi90 and placed in the ethanol vapor atmosphere at room temperature for 1 h. The coated CLPE specimens were annealed in air at 70°C for 3 h for dehydration (PMSi90-coated CLPE). These PMB30- and PMSi90-coated CLPE specimens were then sterilized by 25 kGy γ-rays in N_2 gas.

MPC graft polymerization

The preparation of the PMPC-grafted CLPE is schematically illustrated in Figure 1. The CLPE specimens were immersed in acetone (Wako Pure Chemical Industries, Osaka, Japan) solution containing 10 mg/mL benzophenone (Wako Pure Chemical Industries) for 30 s, and then dried in the dark at room temperature in order to remove the acetone. In previous studies, using ultraviolet spectroscopy, the amount of benzophenone adsorbed on the surface was reported to be 3.5×10^{-11} mol/cm^2.[3] MPC was dissolved in degassed pure water to obtain a concentration of 0.5 mol/L. Subsequently, the benzophenone-coated CLPE specimens were immersed in the aqueous MPC solutions. Photoinduced graft polymerization was carried out on the CLPE surface using ultraviolet irradiation (UVL-400HA ultra-high pressure mercury lamp; Riko-Kagaku Sangyo Co., Funabashi, Japan) with an intensity of 5 mW/cm^2 at 60°C for 90 min; a filter (model D-35; Toshiba Corp., Tokyo, Japan) was used to restrict the passage of ultraviolet light to wavelengths of 350 ± 50 nm. After the polymerization, the PMPC-grafted CLPE specimens were removed, washed with pure water and ethanol, and dried at room temperature. These specimens were then sterilized by 25 kGy γ-rays in N_2 gas.

Surface analysis

The functional group vibrations of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE surfaces were examined using attenuated total reflection (ATR) by Fourier transform infrared (FTIR) spectroscopy. The FTIR/ATR spectra were obtained in 32 scans over a range of 800–2000 cm^{-1} by using an FTIR analyzer (FT/IR615; Jasco International Co., Tokyo, Japan) at a resolution of 4.0 cm^{-1}.

The surface elemental contents of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE were analyzed using X-ray photoelectron spectroscopy (XPS). The XPS spectra were obtained using an XPS spectrophotometer (AXIS Hsi 165; Kratos Analytical Ltd., Manchester, UK) equipped with an Mg-Kα radiation source by applying a voltage of 15 kV at the anode. The take-off angle of the photoelectrons was maintained at 90°. Each measurement was scanned five times; five replicate measurements were performed for each sample, and the average values were considered for the surface elemental contents.

The static water-contact angles of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE were measured using an optical bench-type contact angle goniometer (model DM300; Kyowa Interface

Science Co., Saitama, Japan) by the sessile drop method. Drops of purified water (1 μL) were deposited onto the surface modified CLPE with MPC polymer, and the contact angles were directly measured after 60 s by using a microscope according to the ISO 15989 standard.[27] Subsequently, 15 replicate measurements were performed for each sample, and the average values were considered as the contact angles.

Cross-sectional observation by transmission electron microscopy

A cross-section of the PMB30, PMSi90, and PMPC layers on the CLPE surface was observed using a transmission electron microscope (TEM). The specimens were first embedded in epoxy resin, stained with ruthenium oxide vapor at room temperature, and then sliced into ultra-thin films (~100-nm thick) by using a Leica Ultra Cut UC microtome (Leica Microsystems, Wetzlar, Germany). A JEM-1010 electron microscope (JEOL, Tokyo, Japan) was used for the TEM observation at an acceleration voltage of 100 kV.

Characterization of protein adsorption by micro bicinchoninic acid method

The amount of protein adsorbed on the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE surfaces was measured by the micro bicinchoninic acid (BCA) method. Each specimen was immersed in Dulbecco's phosphate-buffered saline (PBS, pH 7.4, ion strength = 0.15*M*; Immuno-Biological Laboratories Co., Takasaki, Japan) for 1 h to equilibrate the surface modified by the MPC polymer. The specimens were immersed in bovine serum albumin (BSA, $M_w = 6.7 \times 10^4$; Sigma-Aldrich Corp., MO) solution at 37°C for 1 h. The protein solution was prepared in a BSA concentration of 4.5 g/L, that is, 10% of the concentration of the human plasma levels. Then, the specimens were rinsed five times with fresh PBS and immersed in 1 mass % sodium dodecyl sulfate (SDS) aqueous solution, and shaken at room temperature for 1 h to completely detach the adsorbed BSA on the surface modified by the MPC polymer. A protein analysis kit (micro BCA protein assay kit, No. 23235; Thermo Fisher Scientific, IL) based on the BCA method was used to determine the BSA concentration in the SDS solution, and the amount of BSA adsorbed on the surface modified by the MPC polymer was calculated.

Friction test

A friction test was performed using a ball-on-plate machine (Tribostation 32; Shinto Scientific Co., Tokyo, Japan). Each of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE surfaces was used to prepare six sample pieces. A Co-Cr-Mo alloy ball with a diameter of 9 mm was prepared. The surface roughness of the ball was Ra ≥ 0.01, which was comparable to that of femoral ball products. The friction tests were performed at room temperature with various loads in the range of 0.49–9.80 N, sliding distance of 25 mm, and frequency of 1 Hz for a maximum of 100 cycles.[28] Pure water was used as a lubricant medium. The mean coefficients of dynamic friction were determined by averaging five data points from the 100 (96–100) cycle measurements.

Hip joint simulator wear test

A 12-station hip joint simulator (MTS Systems Corp., MN) with the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE cups ($n = 2$), both having inner and outer diameters of 26 and 52 mm, respectively, was used for the hip joint simulator wear test. A Co-Cr-Mo alloy femoral ball component with a size of 26 mm (Japan Medical Materials Corp., Osaka, Japan) was used as the femoral component. A mixture of 25 vol % bovine serum, 20 m*M*/L of ethylene diamine tetraacetic acid, and 0.1 mass % sodium azide was used as a lubricant, according to the ISO 14242-1 standard.[29] The lubricant was replaced every 0.5×10^6 cycles. Walks that simulated a physiologic loading curve (Paul-type), with double peaks at 1793 and 2744 N loads with a multidirectional (biaxial and orbital) motion of 1 Hz frequency, were applied. The wear was determined by weighing the cups at intervals of 0.5×10^6 cycles. Load-soak controls ($n = 2$) were used to compensate the fluid absorption by the specimens.[30] The testing was continued until a total of 3.0×10^6 cycles were completed.

Statistical analysis

The results derived from each measurement in the water-contact angle estimation, friction test, and protein adsorption test were expressed as mean values and standard deviation. The statistical significance ($p < 0.05$) was estimated by Student's *t*-test.

RESULTS

Figure 2 shows the FTIR/ATR spectra of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE. An absorption peak was observed at 1460 cm^{-1} for all test specimens. This peak is mainly attributed to the methylene (CH_2) chain in the CLPE substrate and the MPC polymer chain. However, absorption peaks at 1240, 1080, and 970 cm^{-1} were observed only for the CLPE, whose surface was modified by the MPC polymer. These peaks corresponded to the phosphate group (P—O) in the MPC unit. Similarly, an absorption peak at 1720 cm^{-1} observed in the surface modified CLPE corresponded only to the carbonyl group (C=O) in the MPC unit. The absorption peak intensity of the P—O group of the PMPC-grafted CLPE was the highest in the CLPE, whose surface was modified by the MPC polymer.

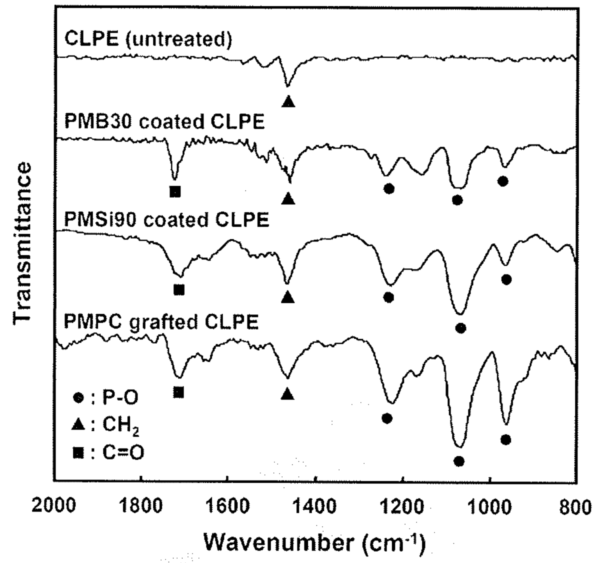

Figure 2. FTIR/ATR spectra of untreated CLPE, MPC polymer-coated CLPE, and PMPC-grafted CLPE.

Table I summarizes the surface elemental compositions of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE. The nitrogen (N) and phosphorous (P) contents in all the CLPE specimens, whose surface were modified by the MPC polymer, were observed. The surface elemental compositions of both N and P in the surface modified CLPE increased with an increase in the MPC composition in the polymer for surface modification. In particular, the elemental compositions of N and P in the PMPC-grafted CLPE surface were 5.2 and 5.3 atom %, respectively. The elemental composition of the PMPC-grafted CLPE surface was almost equivalent to the theoretical elemental composition (N = 5.3, P = 5.3 atom %) of PMPC.

Figure 3 shows the cross-sectional TEM images of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE. For the PMB30 and PMSi90 coatings, and PMPC grafting, a 100-nm thick MPC polymer layer was clearly observed on the surface of the CLPE substrate. No crack for poor adhesion and/or delamination was observed at the interface between MPC polymer layer and CLPE substrate. These results indicate that each surface modification layer on the CLPE substrate is uniform and cover closely, regardless of the binding conditions: the surface modification layers by the PMB30 and PMSi90 coatings, and PMPC grafting are combined with the substrate by physical adsorption and covalent bonds of Si—O—C and C—C, respectively. In the PMB30-coated CLPE, a bilayer structure attributed to dipping twice was clearly observed on the surface modification layer.

Figure 4 shows the static water-contact angles of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE. The static water-contact angles of the untreated CLPE and PMB30-coated CLPE were 90° and 100°, respectively, and they decreased markedly to ~10° (i.e., 8°–13°, $p < 0.001$) by the PMSi90 coating and PMPC grafting.

Figure 5 shows the amount of BSA adsorbed on the surfaces of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE. The amount of BSA adsorbed on the CLPE surface modified by the MPC polymer was considerably lesser ($p < 0.001$) than that of the untreated CLPE, that is, 0.05–0.10 μg/cm^2. These results imply that the surface modification by the MPC polymer results in good biocompatibility.

Figure 6 shows the coefficients of dynamic friction of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE. As compared to the untreated specimens, the PMB30-coated and PMSi90-coated CLPE specimens showed a reduction of ~30% (i.e., 25–30%, not significant) in their coefficients of dynamic friction. Further, as compared to the untreated CLPE specimens, the PMPC-grafted CLPE specimens showed a reduction of ~84% ($p < 0.005$) in their coefficients of dynamic friction.

Figure 7 shows the coefficients of dynamic friction of the untreated CLPE, PMB30-coated CLPE, PMSi90-coated CLPE, and PMPC-grafted CLPE as a function of the loads in the ball-on-plate friction test. The untreated CLPE showed the highest coefficient of dynamic friction, that is, ~0.075. This value was almost constant throughout the experiment. The coefficients of dynamic friction of the PMB30-coated

TABLE I

Surface Elemental Composition (atom %) of CLPE, MPC Polymer Coated CLPE, and PMPC Grafted CLPE (n = 5)

Sample	C	O	N	P	Si
CLPE (untreated)	99.8 (0.3)[a]	0.2 (0.3)	0.0 (0.0)	0.0 (0.0)	0.0 (0.0)
PMB30 coated CLPE	69.9 (1.0)	25.5 (0.6)	2.1 (0.2)	2.5 (0.3)	0.0 (0.0)
PMSi90 coated CLPE	60.5 (0.7)	30.4 (0.4)	4.1 (0.2)	4.0 (0.2)	1.0 (0.0)
PMPC grafted CLPE	58.0 (0.2)	31.5 (0.2)	5.2 (0.1)	5.3 (0.1)	0.0 (0.0)
PMPC[b]	57.9	31.6	5.3	5.3	0.0

[a] The standard deviation is in parentheses.
[b] Theoretical elemental composition of PMPC.

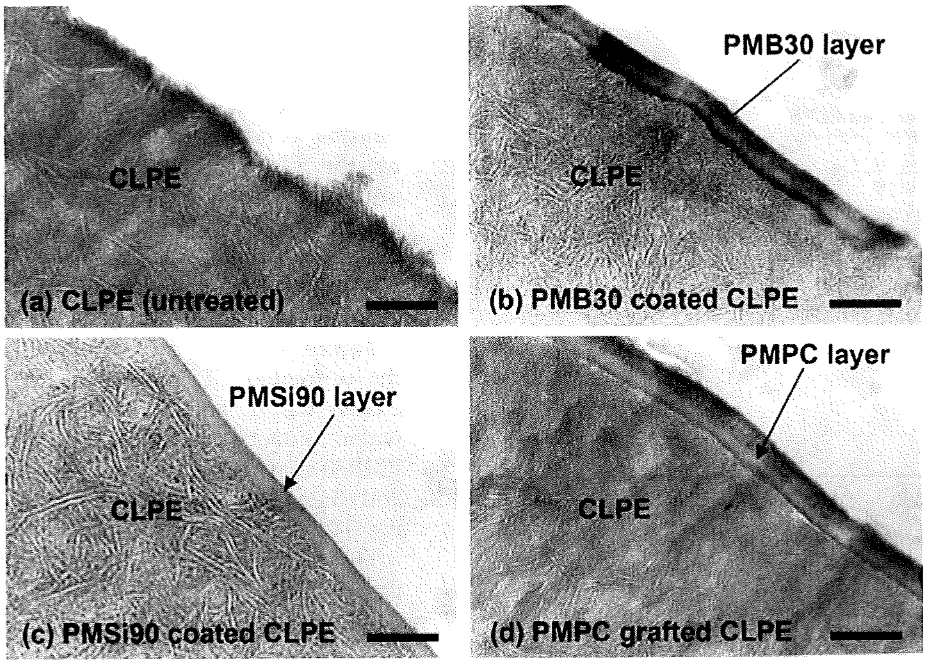

Figure 3. Cross-sectional TEM images of untreated CLPE, MPC polymer-coated CLPE, and PMPC-grafted CLPE. Bar: 200 nm.

CLPE and PMSi90-coated CLPE were smaller (~0.055) than those of the untreated CLPE at loads of up to 0.98 N; then, these coefficients increased to the level of the coefficients of the untreated CLPE at loads above 1.96 N. For both PMB30-coated CLPE and PMSi90-coated CLPE, the MPC polymer layer coating showed almost the same coefficients of dynamic friction. The PMPC-grafted CLPE showed a remarkably low friction coefficient of ~0.026 at a load of 0.49 N; this value decreased gradually and reached ~0.005 at a load of 9.80 N.

Figure 8 shows the gravimetric wear of the untreated CLPE, PMB30-coated CLPE and PMSi90-coated CLPE, and PMPC-grafted CLPE cups in the hip simulator wear test. It was observed that the wear in the PMPC-grafted CLPE cups was significantly lower than that in the untreated CLPE cups. There was no significant difference in the wear of the untreated CLPE and PMB30-coated CLPE cups. The PMSi90-coated CLPE cups showed slightly lower wear than the untreated CLPE cups; however, the weight change varied for each cup (standard

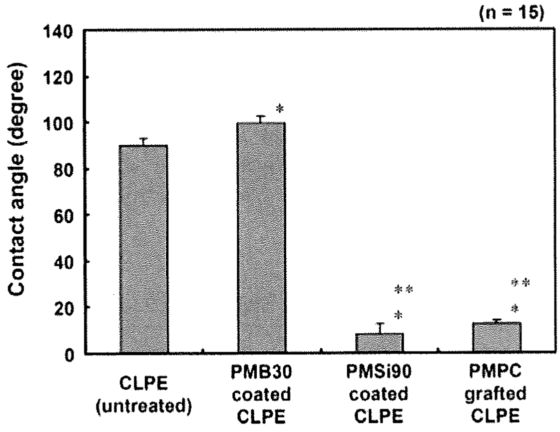

Figure 4. Static-water contact angle of untreated CLPE, MPC polymer-coated CLPE, and PMPC-grafted CLPE. Bar: Standard deviations. *: Significant difference ($p < 0.001$) as compared to the untreated CLPE, **: significant difference ($p < 0.001$) as compared to the PMB30-coated CLPE.

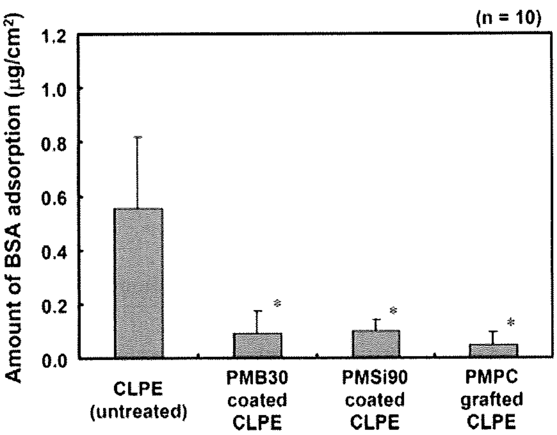

Figure 5. Amount of BSA adsorbed on the surfaces of the untreated CLPE, MPC polymer-coated CLPE, and PMPC-grafted CLPE. Bar: Standard deviations. *: Significant difference ($p < 0.001$) as compared to the untreated CLPE.

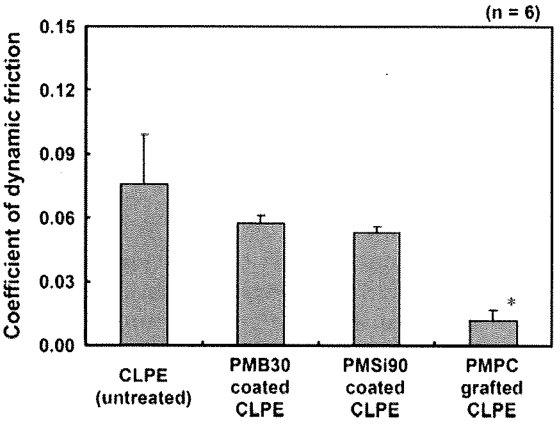

Figure 6. Coefficients of dynamic friction of untreated CLPE, MPC polymer-coated CLPE, and PMPC-grafted CLPE. Bar: Standard deviations. *: Significant difference ($p < 0.005$) as compared to the untreated CLPE.

deviation = ±9.0 mg). The PMPC-grafted CLPE cups showed a slight increase in weight. This was partially attributable to the enhanced fluid absorption in the tested cups as compared to that in the load-soak controls. While applying the gravimetric method, the weight loss in the tested cups is corrected by subtracting the weight gain in the load-soak controls; however, this correction cannot be achieved perfectly because only the tested cups are continuously subjected to motion and load. Usually, the fluid absorption in the tested cups is generally slightly higher than that in the load-soak controls. Consequently, the correction of the fluid absorption by using the load-soak data as the correction factor leads to a slight underestimation of the actual weight loss.[22,29] In this study, a steady wear rate was calculated using data from 2.0×10^6 to 3.0×10^6 cycles; in the untreated CLPE, PMB30-coated CLPE, and PMSi90-coated CLPE cups, these rates were 6.1, 5.9, and 4.5 mg/10^6 cycles, respectively. In contrast, the wear rate of the PMPC-grafted CLPE cups was markedly lower, that is, −1.5 mg/10^6 cycles.

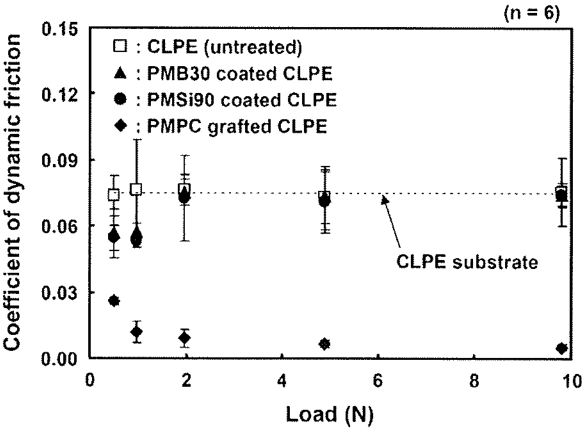

Figure 7. Coefficients of dynamic friction of the untreated CLPE, MPC polymer-coated CLPE, and PMPC-grafted CLPE as a function of loads in the ball-on-plate friction test. Bar: Standard deviations.

DISCUSSION

In this study, we have investigated the surface properties of various surface modification layers formed on the CLPE surface by the physical and chemical coating of the MPC polymers or photoinduced radical grafting of MPC. Here, we discuss the durability of the CLPE whose surface is modified by the MPC polymer in terms of the characteristics of the nanometer-scale layer of the MPC polymer.

The layer, whose surface contains the physically coated PMB30, is combined with the substrate by physical adsorption.[11,12] The layer, whose surface contains the chemically coated PMSi90, is combined with the substrate by physical adsorption and/or slight Si—O—C covalent bonding ascribed to the 10% MPSi in the PMSi90 composition; a hydrolyzed silane molecule of PMSi90 has three —OH groups that react with the —OH groups of the surface oxide layer of the CLPE substrate induced by the O_2 plasma irradiation, and they form covalently siloxane bonds.[27] On the other hand, the surface modification layer obtained by PMPC grafting is combined with the substrate by strong C—C covalent bonding.[19–24]

In Figure 7, the coefficients of dynamic friction of the PMB30-coated CLPE and PMSi90-coated CLPE increased to the level of the coefficient of dynamic

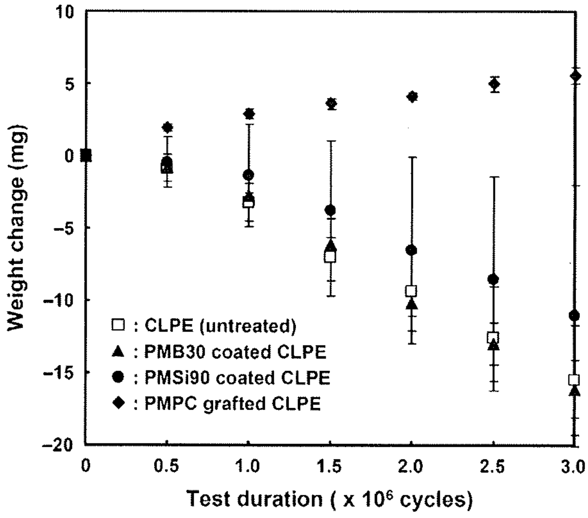

Figure 8. Weight change (volumetric wear) of the untreated CLPE, MPC polymer-coated CLPE, and PMPC-grafted CLPE in the hip joint simulator wear test. Bar: Standard deviations.

friction of the untreated CLPE at loads above 1.96 N. In addition, as shown in Figure 8, there was no significant difference in the wear of the untreated CLPE and the MPC polymer coated CLPE cups in the hip joint simulator tests. In contrast, the PMPC-grafted CLPE showed an extremely low and stable coefficient of dynamic friction and volumetric wear as compared to the untreated CLPE and MPC polymer coated CLPE, as shown in Figures 7 and 8. These results indicate that the PMB30 and PMSi90 surface modification layers are removed from the CLPE surface. The surface modification layers by the PMB30 and PMSi90 coatings are combined with the substrate by physical adsorption and chemical bonds of Si—O—C, respectively. Therefore, the physical adsorption and chemical bonds of Si—O—C are ineffective in the case of large and multidirectional loads. The chemical bonds of Si—O—C are probably insufficient, because the MPSi composition in the PMSi90 polymer is 10%. Therefore, it is thought that a sufficient number of strong bonds between the surface modification layer and the CLPE surface are essential for the long-term retention of the benefits of the MPC polymer used in artificial joints under variable and multidirectional loads.

Since MPC is a highly hydrophilic compound, the MPC polymers PMSi90 and PMPC are water-soluble. The water-wettability of the PMSi90-coated and PMPC-grafted CLPE surfaces with a high MPC unit mole fraction (90 and 100%) was considerably greater than that of the untreated CLPE surface, as shown in Figure 4. Kobayashi et al. reported that the water molecules adsorbed on the surface of the highly hydrophilic PMPC act as lubricants and reduce the interaction between the PMPC and the counter-bearing face.[31] Therefore, it is thought that the artificial hip joint bearing with the PMSi90-coated and PMPC-grafted surfaces exhibited considerably higher lubricity than that with the untreated CLPE. The amount of water molecules absorbed on the surface of the CLPE (water thin-film) is expected to play an important role with regard to the property of low friction.

However, in reality fact as shown in Figure 6, the PMSi90-coated CLPE specimens exhibited a maximum reduction of ~30% in their coefficients of dynamic friction as compared to the untreated CLPE specimens. In contrast, the PMPC-grafted CLPE specimens exhibited a reduction of ~84%. It was previously reported that the polymer concentration (i.e., viscosity) increased with the friction coefficient in the mixed lubrication regime.[32] Therefore, It is assumed that an ultra-low friction of PMPC-grafted CLPE that occurs during sliding is related to the effective viscosity of the PMPC in the mixed lubrication of the intermediate hydrated layer. The high-density PMPC graft chains in the PMPC-grafted CLPE are assumed to exhibit a brush-like structure.[25,33] The viscosity of the PMPC layer reflects the mobility of the free end groups of the MPC polymer or MPC polymer chains themselves.[34,35] In contrast, the mobility of PMSi 90 is limited by the cross-linking and network entanglement of the gel structure of PMSi90. Hence, we think that the polymer brush-like structure of the PMPC-grafted CLPE with high mobility of polymer chains can function as a considerably better surface hydration lubrication system of artificial joints than the gel structure of the PMSi90-coated CLPE. These considerations are based on the previous studies on charged polymers (polyelectrolytes) by Klein and coworkers.[34,35]

The significant reduction in the coefficient of friction of the grafted PMPC resulted in a substantial improvement in wear resistance. A previous study reported that the hydrogel cartilage surface was assumed to have a brush-like structure: a part of the proteoglycan aggregate brush bonds with the collagen network in the cartilage surface.[36] It is thought that the bearing surface with high-density PMPC in artificial hip joints has a brush-like structure similar to an articular cartilage. We assume that the bearing surface of the artificial hip joint combined with a 100-nm thick MPC polymer layer results in a fluid film lubrication (or mixed lubrication) of the intermediate hydrated layer; which suggests that this novel artificial hip joint mimics the natural joint cartilage.

As shown in Figure 7, the coefficients of dynamic friction of the untreated CLPE surface were constant for all load values. In contrast, those of the PMPC-grafted CLPE surface decreased with increasing load. These results show that the PMPC-grafted layer does not follow Amonton's law of $F = \mu N$. The maximum contact stress of ~62 MPa at a load of 9.8 N is higher than the yield tensile strength of the CLPE (~23 MPa). The maximum contact stress is roughly calculated by the Hertzian theory. The elastic CLPE substrate is deformed slightly by the loads, the low friction coefficient may have been shown in order to get much more amount of water thin-film with the larger contact area of concave surface. We also think that these results imply that the lubrication of the PMPC-grafted CLPE is dominated by the hydrodynamic lubrication mechanism.

As shown in Figure 4, the static water-contact angle of the PMB30-coated CLPE was ~100°. Sibarani et al. reported that the PMB30-coated polymer surfaces showed high advancing (~100°) and low receding (~20°) contact angles.[5] Moreover, Yamasaki et al. reported that the PMB30-coated polymer surfaces required more than 30–300 min to achieve complete equilibrium.[37] This indicates that the PMB30 cannot be hydrated easily, due to the low MPC unit composition of the copolymer and low mobility of the polymer chain. However, as shown in Figure 5, the PMB30-coated CLPE surface, which could form a phosphorylcholine-enriched surface after equilibrating for 1 h, showed excellent biocompatibility as an antiadsorption surface for BSA.

The adsorption of the representative plasma protein, BSA, on the CLPE surface decreased to 9–18% due to the surface modification by the MPC polymer, as shown in Figure 5. It is hypothesized that the mechanism of protein adsorption resistivity on the surface modified by the MPC polymer is based on the water structure resulting from the interactions between water molecules and phosphorylcholine groups.[26,37,38] The large amount of free water around the phosphorylcholine group is considered to detach proteins easily and even prevent conformational changes in the adsorbed proteins.[26,38] Hence, we expect that the protein adsorption will decrease with increasing MPC unit composition in the MPC copolymer. However, in this study, there was no significant difference in the protein adsorption of the CLPE, whose surfaces was modified by the MPC polymer (the MPC unit compositions were 30, 90, and 100%). The reduction in protein adsorption is also considered to be caused by the presence of a hydrated layer around the phosphorylcholine group.[26,39] The latter consideration is consistent with the results of the water contact angle measurement, friction test, and cross-sectional TEM observations of the CLPE, whose surface is modified CLPE by the MPC polymer (Figs. 3, 4, and 6). The previous studies reported that the protein concentration of lubricants such as bovine serum considerably affected the wear rate of the UHMWPE cups in the joint simulator test: the protein concentrations in the synovial fluids of both normal and diseased joints (~20–40 mg/mL), including joints after total arthroplasty, were associated with the highest wear rates.[40,41] We think that the antiprotein adsorption surface on the CLPE prepared by the MPC polymer will prevent the highest adhesive wear rates *in vivo* caused by the protein adsorption. Moreover, the CLPE whose surface is modified by the MPC polymer is expected to exhibit tissue and blood compatibility as biocompatibility, because previous studies have reported that the MPC polymer modified surfaces exhibit *in vivo* biocompatibility.[3–11]

CONCLUSION

In this study, we systematically investigated the surface properties of the various surface modification layers formed on the CLPE surface by the MPC polymer by dip coating or photoinduced radical grafting. It is concluded that several important issues are involved in the long-term retention of the benefits of the MPC polymer used in artificial joints under variable and multidirectional loads, for example, strong bonding between the MPC polymer and the CLPE surface and high mobility of the free end groups of the MPC polymer. We should employ the photoinduced radical graft polymerization to create strong covalent bonding between the CLPE substrate and the surface modification layer, and also to retain the high mobility of polymer chains of that layer.

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