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Adhesion Between Polymer Surface Modified by Graft Polymerization and Tissue During Surgery Using an Ultrasonically Activated Scalpel Device

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ABSTRACT: In the field of surgery, achieving adhesion between a polymer implant and tissue poses a challenge considering that suturing is not appropriate for the stability of such implants. An ultrasonically activated scalpel that generates heat by mechanical vibration and promotes adhesion between a polymer implant and native tissue by pressing the two materials together has very good potential for application in the field. To determine the type of polymer that is suitable for the purpose, we investigated polyethylene (PE) and polystyrene (PS) films, the surfaces of which were activated by corona discharge. Graft polymerization was then performed on the corona-treated surfaces to vary their properties. The corona-treated PE and PS films grafted with poly(acrylic acid), poly(methacrylic acid), poly(vinyl benzylacrylic acid), and poly(hydroxylethyl acrylate), respectively, adhered to the tissue when the ultrasonically activated scalpel was applied. The heat generated by the mechanical vibration and the applied pressure enabled the carboxyl or hydroxyl groups to bond with the proteins in the extracellular matrix. We therefore concluded that it was possible to integrate this technique in the development of new types of polymer devices that could be stably implanted in a living body. © 2014 Wiley Periodicals, Inc. J. Appl. Polym. Sci. **2014**, *131*, 40885.

KEYWORDS: biomedical applications; films; surfaces and interfaces

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INTRODUCTION

In the field of surgery, the achievement of bonding (or adhesion) between a polymer implant and native tissue poses a challenge considering that suturing is not appropriate for the stability of such implants. Whereas there are some polymeric implants that can be sutured to native tissue, there is also the requirement of flexibility and toughness. Not all materials that are used for implants possess these properties. Many polymers or metals are too rigid and almost impossible to suture, so adhesion between the rigid implant and the native tissue is achieved by cyanoacrylate glue or fibrin glue, which bond the tissue *in situ*.[1,2] However, clinical tissue adhesives are faced with challenges regarding adhesion strength, toxicity, and biocompatibility.[3,4] Hence, there is the need to develop a new adhesion technique that is nontoxic and produces strong adhesion.

We focused on achieving adhesion between an implant and a native tissue by means of an ultrasonically activated scalpel that applies heat and pressure to both components.[5] This technique, which was originally developed to stop bleeding by coagulating the proteins in the tissue,[6–8] is very effective for the adhesion of materials such as polymers and metals to tissue because the heat and pressure applied to the materials and the tissue enables bonding by a mechanism similar to that of protein coagulation as mentioned above.[9] We found that this principle could be used to develop a new device for achieving adhesion between materials and tissue.[9,10] However, not all materials can be bonded with tissue in this way. Considering that tissue is composed of extracellular matrix comprising collagen and elastin, the adhesion may rely on surface protein interaction. This means that the physical and mechanical properties as well as the chemical components of the surface are very important to

Additional Supporting Information may be found in the online version of this article.

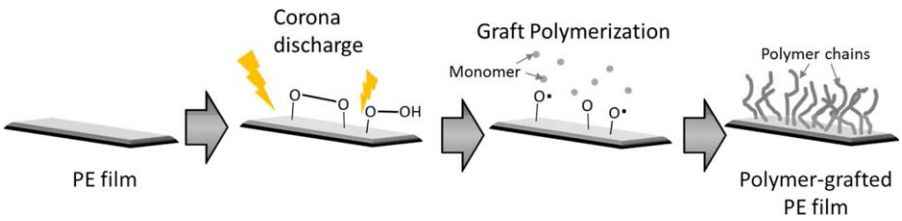

Figure 1. Schematic illustration of corona discharge on the surface of a polyethylene film, and graft polymerization. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

the adhesion. That is, the material should possess certain chemical and physical affinities for the proteins in the extracellular matrix.

In a previous work, using polymeric films as model materials, we found that functional groups and physical properties dominated material–tissue adhesion.[5] These findings were based on the structural and physical properties of the material, which had bonded to tissue upon the application of heat and mechanical vibration generated by the blade of an ultrasonically activated scalpel.[8,11] Pressurization of the tissue and polymer brought them together, and evaporation of the water by the heat promoted thermal cross-linking between the polymer and the proteins in the extracellular matrix, especially the collagens.[9] Polymers have nothing in common, including in their surface properties, beyond the presence of carboxyl or hydroxyl groups. However, the chemical structures of polymers are generally complex, which prevented a conclusion on the factor that dominated the adhesion to tissue.

Therefore, in the present study, we investigated the factors that determined adhesion between a material and tissue using polymers with simple but specific chemical structures as surface modifiers. Furthermore, we controlled the hydrophilicity of the polymers by altering their functional groups. The functionalization was done by grafting these polymers to polyethyelene (PE) and polystyrene (PS) surfaces that were activated by treatment with corona discharge.[12–17] By determining the factor that dominated adhesion between a material and tissue, it would be possible to develop a polymeric surface modifier that could be easily bonded to tissue using an ultrasonically activated scalpel and be used for a rigid polymer implant–tissue adhesion technique in clinics. In this paper, we report how the functional groups and surface properties affected adhesion to tissue.

EXPERIMENTAL

Graft Polymerization

Figure 1 shows schematic images of how the polymer was grafted on to the surface. We used PE and PS films because their chemical components are sufficiently simple for activation of their surface by corona discharge. Corona discharge treatment (15 kV in air at room temperature for a predetermined time) was used to alter the surface properties of the PE and PS films as described in our previous work and by other researchers.[12–16] The corona discharge was executed for 1 min on each side of the PE film to form coats of 1 cm × 4 cm area and ≈100 μm thickness. This was done by supplying oxygen into the chamber. The thickness of the polymeric films was 0.1–0.5 mm. After the corona discharge, the samples were placed in a glass ampule for graft polymerization.

The monomer was chosen based on the results of our previous study.[5] The purchased acrylic acid (AA) (Wako Chemicals, Tokyo, Japan), methacrylic acid (MA) (Wako Chemicals, Tokyo, Japan), 4-vinylbenzoic acid (VBA) (Kohjin, Tokyo, Japan), hydroxyethyl acrylate (HEA) (Wako Chemicals, Tokyo, Japan), and hydroxyethyl methacrylate (HEMA) (Wako Chemicals, Tokyo, Japan) were all distilled to eliminate moisture and the inhibitors and then kept in a cool and dark environment. To investigate the effect of the hydrophilicity of the polymer surfaces on the adhesion, we added similarly purchased hydroxylethyl acrylamide (HEAAm) (Sigma-Aldrich), sulfopropyl acrylate (SPA) (Sigma-Aldrich), sulfopropyl methacrylate (SPMA) (Sigma-Aldrich), 4-styrenesulfonic acid sodium salt (SSA) (Sigma-Aldrich), 2-[(acryloyl)amino]-2-methyl-1-propnanesulfonic acid (AMPS) (Sigma-Aldrich), and acrylamide (Wako Chemicals), which were distilled before use. The nomenclature of the monomers and polymers applied in this study is as presented in Table I, and the chemical structures are as shown in Figure 2. The polymers included either nonionic or negatively charged end groups. Polymers with positively charged end groups were not used in this study because of their inflammability *in vivo*.[18] The corona-treated PE or PS film was inserted into the distilled monomer in the glass ampule. After vigorous degassing for 10 min to eliminate oxygen, the glass ampule was sealed and kept at 70°C for 30 min for bulk polymerization. The glass ampule was then broken open to remove the film with the grafted polymers, which was immediately washed with water before being placed in hot water at 70°C for 48 h for thorough rinsing of the monomer.

The surfaces of the polymer were evaluated by attenuated total reflection Fourier transformed infrared spectroscopy (ATR-FTIR) (Spectrum 100, Perkin Elmer Japan) and the static contact angle (SCA) (FTA1000, JASCO International). The confirmed spectrum of the polymer comprised alkane (–CH, 2850–2960 cm^{-1}), aromatic ring (–CH and –C–C–, 3030 cm^{-1}, 1450–1600 cm^{-1}, amine (–NH, 3300–3500 cm^{-1}), carboxyl acid (COO, 1710 cm^{-1}; C═O, 1735 cm^{-1}), alcohol (–OH, 3400–3650 cm^{-1}), and a sulfoxyl group ($–SO_3$, 1100 cm^{-1}).[19,20] The sessile drop technique was used for the static contact angle test (1.1–1.2 μL), with the angle measured 30 s after the drop. The contact angle measurement was repeated five times. To evaluate the amount of polymer that was grafted onto the surface, we measured the weight of the film before and after the polymerization. We then calculated the graft density of the polymer by

Table I. The Nomenclatures of the Monomers and Polymers Used in This Study

Side chain	Monomer	PE polymer	Other functional group
Carboxylic group	Acrylic acid (MA)	PE-p(MA)	-
	Methacrylic acid (MA)	PE-p(MA)	$-CH_3$
	4-Vinylbenzoic acid (VBA)	PE-p(VBA)	$-C_6H_5$
Hydroxyl group	Hydroxyl ethyl acrylate (HEA)	PE-p(HEA)	—COO
	Hydroxy ethyl methacrylate (HEMA)	PE-p(HEMA)	$-CH_3$, —COO
	Hydroethyl acrylamide (HEAAm)	PE-p(HEAAm)	—CONH
Sulfoxy group	Sulfopropyl acrylate (SPA)	PE-p (SPA)	—COO
	Sulfopropyl methacrylate (SPMA)	PE-p(SPMA)	$-CH_3$, —COO
	4-Styrenesulfonic acid sodium salt (SSA)	PE-p(SSA)	$-C_6H_5$, —COO
	2-[(Acryloyl)amino]-2-methyl-1-propanesulfonic acid (AMPS)	PE-p(AMPS)	—CONH, CH_3
Amine group	Acrylamide (AAm)	PE-p(AAm)	—CO

measuring the amount of polymer on the surface and dividing it by the total surface area of the polymer film. The process was repeated five times for different PE films with different surface areas to determine the amount as accurately as possible.

Adhesion Method

Porcine aorta purchased from a local slaughterhouse (Shibaura Zoki) was used as the model living tissue. The aorta was trimmed and cut into pieces measuring 1×4 cm^2 in area and $\approx 400 \pm 100$ μm thick. A harmonic scalpel (Ethicon Endo-Surgery Japan, Johnson & Johnson, Tokyo, Japan), which is a type of ultrasonically activated scalpel, was used. A laparosonic coagulating shears (LCS) handgrip (Ethicon Endo-Surgery Japan) was used as the hand piece. A piece of porcine aorta facing the inner lumen, the polymer, and the polymeric film were layered and placed between the blades of the LCS, which generated heat by vibration (Figure 3). The samples were then gripped by a force of 20 kg for 1, 3, 5, and 7 s, respectively, to press the porcine aorta and the polymeric films together for adhesion. The applied conditions comprised an electric power output level (energy level Lv) of 1–5 and amplitudes of 50–100 μm, which were used to examine how the change in temperature produced by mechanical vibration affected the adhesion. It should be noted that the vibration generated heat that could produce temperature rises of up to 150°C or higher depending on the frequency. The increase in Lv was accompanied by an increase in the frequency (55.5–100 kHz), which increased the temperature.[11] The strength of the bond between the porcine aorta and the polymeric film was measured by a peeling test (180° peeling test method) using tensile test machine (Reoner II, Yamaden). The grip was used to hold the ends of the bonded polymer and tissue, respectively. They were then strained at a rate of 0.05 mm/s (PE films) or 0.1 mm/s (PS films) by a force of 20 N until the attached parts were separated (Supporting Information Figure 1). The ultimate strength was considered as the bonding strength, which is the similar term with peel strength. The test was repeated three times, and the average value was obtained. After the adhesion test, the samples were lyophilized overnight and plasma-coated with gold in preparation for scanning electron microscopy (SEM) (S-3400NK). The adhesion location was observed and compared with the native tissue (i.e., the nonadhered site and burnt site). In this study, we used bonding strength instead of peel strength as the unified term in order to avoid the confusion.

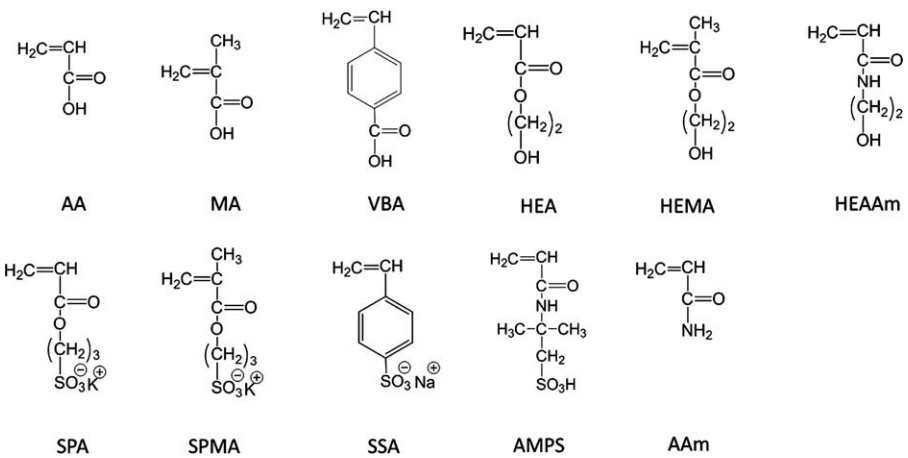

Figure 2. Chemical structures and abbreviations of the monomers used in this study. Refer to Table I for the complete nomenclature.

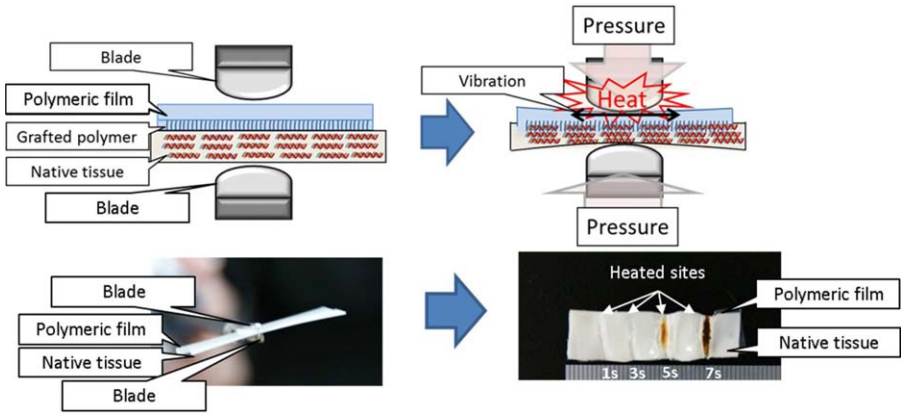

Figure 3. Schematic illustration of the adhesion process (upper) and the photographic images (below) between a polymer and native tissue using an ultrasonically activated scalpel. The mechanical vibration generates the heat and the blade presses the polymer film and native tissue together to enable adhesion. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

RESULTS

Confirmation of Graft Polymerization

Using ATR–FTIR (Supporting Information Figure 2), we confirmed the oxidation and graft polymerization on each film. It showed that the corona discharge produced carbonyl groups on the surface. After graft polymerization, we confirmed the existence of functional groups on the PE and PS surfaces depending on the chemical structure of each polymer. The graft density was measured to determine the effectiveness of the polymerization. Figure 4(a) shows the graft densities of the polymer on the PE and PS films. In the case of the PE films, the amount of grafted polymers on their surface was between 0.1 and 0.3 mg/cm^2, except for PE-p(VBA), which was 0.025 mg/cm^2). This suggests that general graft polymerization using our method produces grafted polymers in the range of 0.1–0.3 mg/cm^2. For PE-p(VBA), we tried to increase the amount of grafted poly (VBA) on the surface of the PE by increasing the polymerization time, but could not achieve significant difference by our standard method.

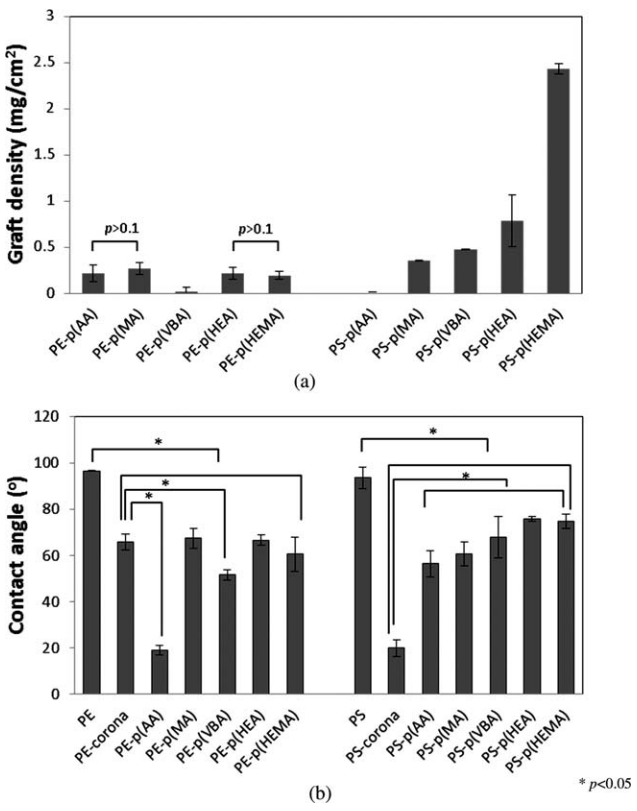

Figure 4. (a) Graft densities and (b) contact angles of PE and PS, corona-treated PE and PS films; all the polymers grafted onto the PE and PS films.

In the case of the graft polymerization on the PS film, we observed that the polymer graft density was greater than that on the PE films, except for AA [Figure 4(a)]. The amount of polymer grafted on to the PS film was high for HEA and HEMA, which had graft densities of approximately 0.8 and 2.5 mg/cm^2, respectively. The graft polymerization changed the PS film from being transparent to white, which was due to the large amount of polymer on the surfaces. Regarding VBA, which exhibited low adsorption on the PE film, the amount grafted on the PS film was 0.6 mg/cm^2.

Figure 4(b) shows the static contact angles of each graft on the PE and PS surfaces. It can be observed that the corona treatment decreased the contact angle compared to the much higher value on the PS film. The graft polymerization also decreased the static contact angle, which implied increased hydrophilicity of the film surfaces. The lowest static contact angles correspond to PE-p(AA) and corona-treated PS, which was due to the presence of the carboxyl group. The presence of the methyl group and benzyl group in the polymer containing the carboxylic acid side chain was the cause of the increased contact angle exhibited by the MA and VBA polymers (MA). Regarding the polymer with hydroxyl end groups, such as p(HEA) and p(HEMA), the contact angles were higher than those of the other samples. There was no significant difference between p(HEA) and p(HEMA). We could obtain polymer surfaces with the same end groups, but with different hydrophilicties, which showed that materials with the same functional groups, but different surface properties could exhibit different polymer–tissue adhesion properties.

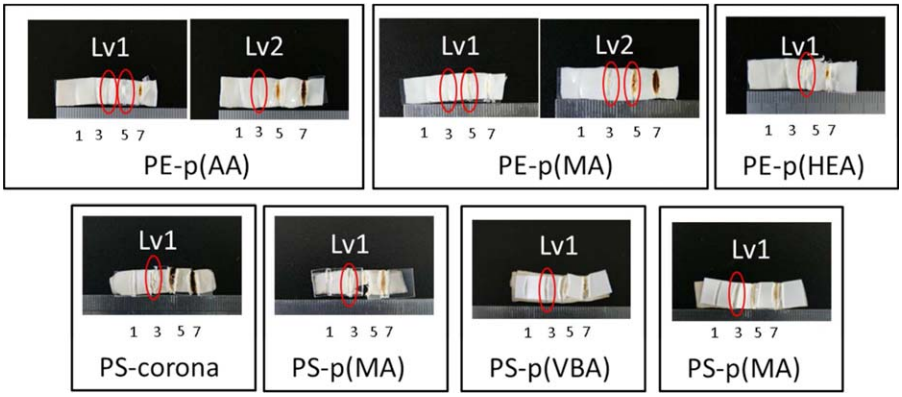

Figure 5. Photographic images of bonded polymer films and native tissues. The red circle indicates the part where the adhesion is sufficiently strong for measurement of the adhesion strength. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

Adhesion Between Polymer and Tissue

Photographic images of the adhesion between polymers and native tissue after application of heat and pressure are shown in Figure 5. Information about the specific adhesion tendencies with respect to the applied energy and the adhesion time are summarized in Table II for the PE-grafted polymeric films, and in Table III for the PS-grafted polymeric films. All the polymers were melted or burnt by the heat when the energy applied to the polymer–tissue was higher than Lv2, or when even Lv1 energy was applied for more than 7 s. Furthermore, at Lv 1, the required time for adhesion was within 3 and 5 s, except for one case. The polymeric films that had adhered to the native tissue were PE-p(AA), PE-p(MA), and PE-p(HEA). Regarding the PS films, the adhesion could be observed for PS corona, PS-p(AA), PS-p(MA), and PS-p(VBA). It should be noted that the vibration could generate heat that produced temperature rises of up to 150°C or higher depending on the frequency.[11] Whereas the range of adhesion conditions of the polymers on the PE films was wider (energy level of 1–2 and adhesion time of 3–5 s), the adhesion condition was limited to only Lv1 for 3 s. Interestingly, PS-p(HEA) and PS-p(HEMA) did not adhere to the tissue, although PS-corona did. Furthermore, PS-p(VBA) exhibited much higher adhesion to the tissue compared to PE-p(VBA).

The SEM images of the polymer–tissue adhesion sites (for Lv1 and 0, 1, 3, and 7 s, respectively) are shown in Figure 6. The pressurized tissue surface after 1 s of the adhesion test can be seen. No trace of polymeric film can be seen because no polymer had adhered to the tissue [Figure 6(b)]. For the adhered and burnt cases [Figure 6(c,d)], the remaining part of the polymer was detected. We could observe that the native tissue beneath the polymeric films had lost its fibrillar structure.

The bonding strengths of the polymers that bonded to the tissue are shown in Figure 7. The bonding strengths were approximately 0.55–0.8 kPa for PE-p(AA), 0.26–0.56 kPa for PE-p(MA), and 0.68 kPa for PE-p(HEA). No significant increase in the bonding strength with increasing adhesion time or applied energy can be observed. The PS films [Figure 7(b)] exhibited relatively high bonding strength as compared to the PE films. Their bonding strength was approximately 4–6 kPa; no significant difference was observed among the samples.

The SEM images after the bonding strength test are shown in Figure 8. The part of the polymer that had adhered to the tissue remained after the bonding strength test. Magnification of the adhered part revealed integration of the polymer and the tissue.

DISCUSSION

Evaluation of Grafted Polymers on Polymeric Film

The corona discharge process deposited polar groups containing oxygen on the surfaces of the PE and PS films and increased their hydrophilicity [Figure 4(b)].[21] The radicals produced by the corona discharge allowed the monomers to initiate the polymerization from the surface, resulting in polymer brush (Figure 1). The polymers grafted to the polymeric films without difficulty, except in the case of VBA, for which the graft density on the PE film was very low. AA also exhibited very low graft density on the PS film (Figure 4(a)). These observations indicate that there were sufficient stably grafted polymers on the PE and PS surfaces, which confirmed the effectiveness of the proposed

Table II. Adhesibility of the Grafted Polymer Films to the Native Tissue According to the Energy Level (Lv) and the Adhesion Time

	PE-p(AA)				PE-p(MA)				PE-p(VBA)				PE-p(HEA)			
	Adhesion time (s)				Adhesion time (s)				Adhesion time (s)				Adhesion time (s)			
Lv	1	3	5	7	1	3	5	7	1	3	5	7	1	3	5	7
1	-	○	○	×	-	-	○	○	-	-	Δ	×	-	Δ	○	×
2	-	○	×	×	-	○	○	×	-	Δ	Δ	×	-	×	×	×

No adhesion, × Burnt, ○ Adhered, Δ Weakly adhered.

Table III. Adhesibility of the Grafted Polymer Films to the Native Tissue According to the Energy Level (Lv) and the Adhesion Time

	PS-corona treated				PS-p(AA)				PS-p(MA)				PS-p(VBA)			
	Adhesion time (s)				Adhesion time (s)				Adhesion time (s)				Adhesion time (s)			
Lv	1	3	5	7	1	3	5	7	1	3	5	7	1	3	5	7
1	-	○	×	×	-	○	×	×	-	○	×	×	-	○	×	×

	PE-p(HEA) Adhesion time (s)				PE-p(HEMA) Adhesion time (s)			
Lv	1	3	5	7	1	3	5	7
1	-	Δ	×	×	Δ	×	×	×

No adhesion, × Burnt, ○ Adhered, Δ Weakly adhered.

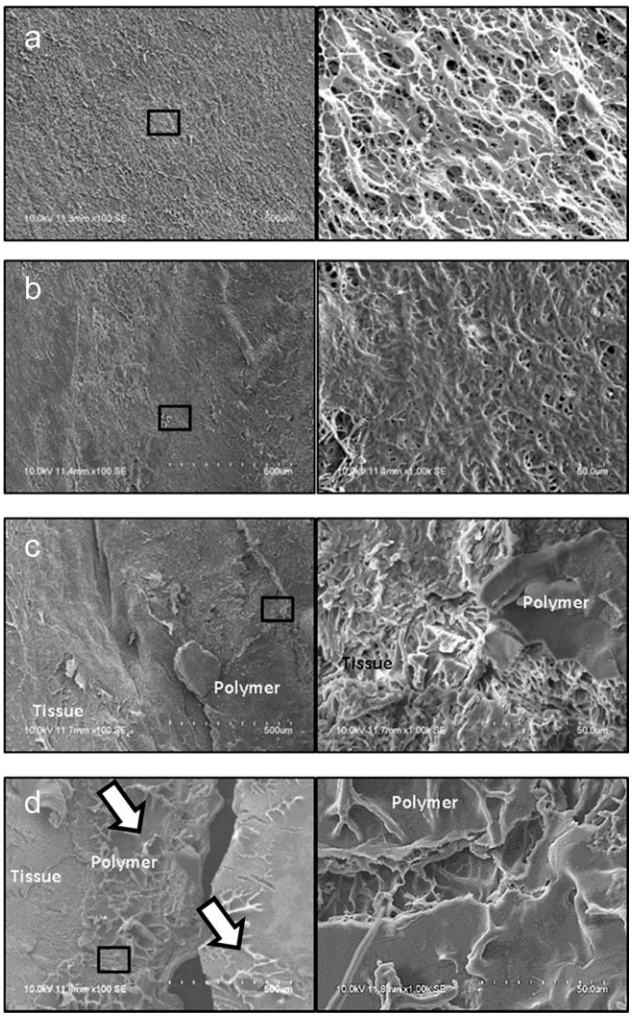

Figure 6. SEM images of adhered PS-p(MA) polymeric films and native tissues; the right images are magnifications of the small rectangles in the left images: (a) 0 s, (b) 1 s, (c) 3 s, (d) 7 s. Energy of Lv1 was applied. The arrow indicates the burnt site, and the crevasse in image (D) was produced by the cleavage of the polymer and the tissue due to too high temperature.

method. The method used to calculate the graft density was a rather rough one, but it provided a good idea of how much polymer was on the surface of the PE and PS films. The graft density of most of the polymers on PE was between 0.1 and 0.3 mg/cm^2, and between 0.4 and 2.5 mg/cm^2 on PS. The higher graft density on PS is thought to be due to its higher oxidation ability by producing more radicals after corona treatment. For the same corona treatment time, the oxidation level on the PS surface was higher, which produced a higher hydrophilic surface

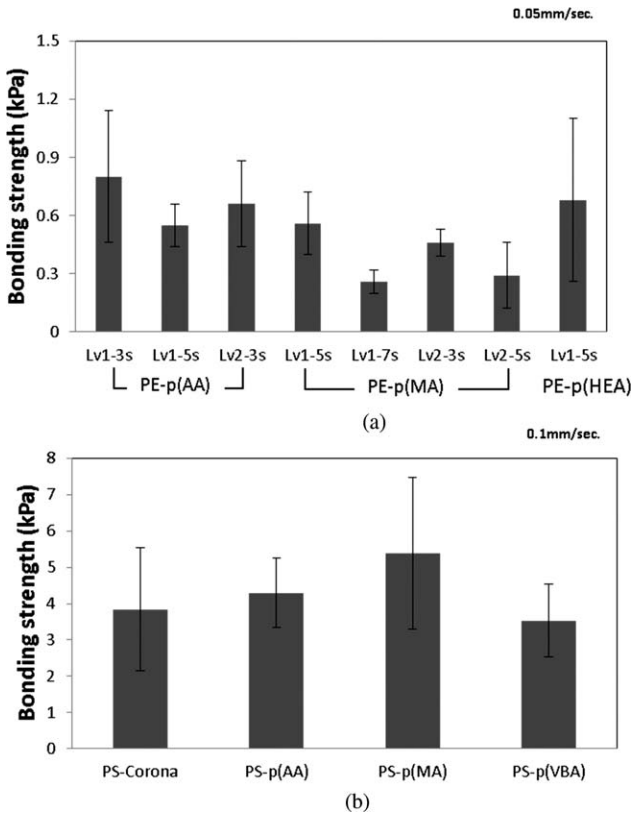

Figure 7. Bonding strength between (a) PE polymeric film and tissue, and (b) PS polymeric film and tissue. The experiment was repeated three times.

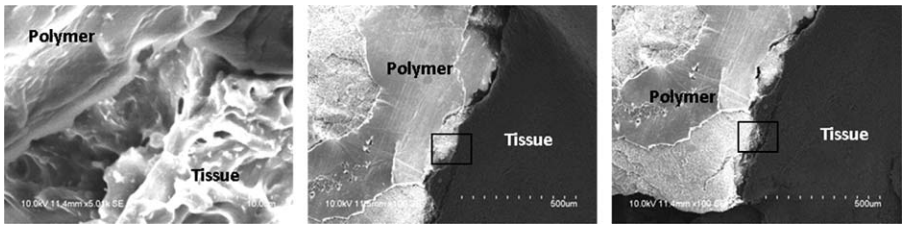

Figure 8. Polymer–tissue adhesion site [PS-p(MA) and tissue] after bonding strength test. The rectangle in each image indicates the area that was magnified to observe the integrated site.

than that of PE [Figure 4(b)]. This increased the polymer grafting on the PS film compared to the PE film.

Adhesion Between Polymer and Tissue

As mentioned above, the adhesion was induced by the heat generated by the mechanical vibration of the blade.[11] The pressure pressed the two different materials together and the heat evaporated water from the interface. The heat and vibration were transmitted by pressing the polymer and the tissue together, which also caused the polymer chains to chemically interact with the proteins in the tissue, resulting in the formation of a crosslink (Figure 3).[5,9] When the plain PE and PS films were used for the adhesion test, none adhered to the tissue. However, the corona treatment enabled the adhesion of the PS film to the native tissue. This was due to the formation of carboxyl or hydroxyl groups on the surface after the decay of the radicals.[16] Measurements revealed that the decrease in the static contact angle for PS was greater than that for PE, which indicated that the formation of radicals on the PS film was much higher than that on the PS film for one minute of corona treatment. The actual adhesion between the polymers containing carboxylic acid or hydroxyl groups and the tissue when using the ultrasonically activated scalpel were shown for four cases (AA, MA, VBA, and HEA). The adhesion was very weak for PE-p(VBA), PS-p(HEA), and PS-p(HEMA). The adhesion of the hydroxyl group to tissue was confirmed using a poly(vinyl alcohol) (PVA) film (Supporting Information Figure 3).The successful adhesions required only 3–5 s, which agrees with the results of our previous study.[5] It is thought that a complex but highly stable polymer–protein layer was formed between the polymer graft and the tissue. When an implant contains carboxylic acid or hydroxyl groups, it is possible to stabilize it in a living body. The adhesion is thought to be triggered by the dehydrothermal crosslinking between the polymeric film and the tissue. The significant heat immediately evaporates the water and the pressure enables the polymer surface and the tissue to form a peptide bond.[22] However, too much heat or a too long period of heating may seriously burn or cleave the sample (Figure 5 and Tables II and III). It is thought that too much heat damages the film or tissue, resulting in their detachment after burning. Too low energy or a too short period of energy application is also not recommended because there would be insufficient energy or time to stimulate adhesion.

Interestingly, part of the polymer remained on the tissue after the adhesion test as shown in Figures 6 and 8. The polymer–tissue interface was integrated (Figure 6, arrow), which indicated that the adhesion between the polymer and native tissue was strong. During the bond strength test, the polymer was not peeled from the native tissue, but the part of the polymer damaged by heat that did not adhere to the native tissue was broken. The damage for the PS films was much less than that for the PE films owing to the higher melting temperature of PS, which contributed to its higher bond strength. We are therefore of the opinion that strong adhesion between an implant and tissue can be achieved if the implant is sufficiently strong to withstand heat.

The alkyl chains served another important purpose in the side chains. As shown for the HEA and HEMA polymers, no adhesion occurred, except in the case of PE-p(HEA). It should be noted that there was no adhesion, even for those grafted on the PS films, although significant amounts of poly(HEA) and poly (HEMA) were grafted. As was observed for the PVA film (Supporting Information Figure 3), the presence of the hydroxyl group induced strong polymer–tissue adhesion. Zhang et al. noted that the hydroxyl group promoted adhesion of the polymer through hydrogen bonding.[16] In the cases of p(HEA) and p(HEMA), the effect of the hydroxyl group on the side chain was hindered by the presence of the alkyl chains. This implies that although the hydroxyl group has a strong bonding affinity for tissue, it is mitigated by the other functional groups in the polymer chains.

Regarding the surface properties, we further investigated how the hydrophilic polymer affected the adhesion by grafting anionic sulfoxy groups on to the PE films, which produced contact angles in the range of 20°–40° (Supporting Information Figure 4). Although sufficient polymers were grafted on to the films, none adhered to the tissue (Supporting Information Table I). This indicates that the adhesion ability is not so much as dependent on the hydrophilic nature of the polymer on the surface as it is on the functional groups in the side chains. In the case of the corona-treated samples, the decrease in the contact angle was due to the presence of carboxyl groups after the decomposition that produced –OH or –COOH. That is, the increase in the carboxyl groups after oxidation was accompanied by an increase in the hydrophilicity, which is related to the adhesion between polymer film and tissue. This increased the polymer–tissue adhesibility.

CONCLUSION

Our findings on polymer–tissue adhesion using an ultrasonically activated scalpel are summarized as follows: (1) carboxyl groups and hydroxyl groups are required for the adhesion. (2) A sufficient amount of the polymer is required for bonding with the

tissue. (3) Alkyl chains should be avoided. (4) The temperature and time allowed for the adhesion should be controlled. (5) The hydrophilicity/hydrophobicity of the polymer surface is not a dominant factor of the adhesion. We showed that it was possible to control the adhesion to tissue, which opens new opportunities for polymer–tissue adhesion technology. Furthermore, we were able to change the bonding strength by changing the base film used for the graft polymerization.

Although the bonding strength was not sufficiently high to withstand the severe conditions in living organisms and still not enough to be used for the clinical application that can replace the conventional adhesives, we were able to provide basic knowledge about polymer–tissue adhesion. We are conducting further study to design polymers that can form sufficiently strong bonds with tissue by the application of heat and vibration for direct application to long-term polymer implants.

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