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Surface modification of polypropylene surgical meshes for improving adhesion with poloxamine hydrogel adhesive

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Abstract

Tissue adhesive has notable clinical benefits in hernia repair fixation. A novel poloxamine tissue adhesive was previously shown to successfully bond collagen tissue with adequate adhesive strength. In application related to attachment of polypropylene (PP) mesh, the adhesive strength between the mesh and poloxamine hydrogel adhesive is limited by the hydrophobicity of PP monofilaments and lack of covalent bond formation. The purpose of this study was to compare two different surface modifications [bovine serum albumin (BSA) adsorption and poly-glycidyl methacrylate/ human serum albumin (PGMA/HSA) grafting] of PP mesh for improving the adhesive strength between poloxamine hydrogel adhesive and PP mesh. The PGMA/HSA surface modification significantly improved the adhesive strength for meshes attached with poloxamine hydrogel tissue adhesive compared with unmodified meshes and meshes modified by BSA adsorption. An area of 1 cm² adhesive provided for a maximum adhesive strength of 65–70 kPa for meshes modified by PGMA/HSA, 4–13 kPa for meshes modified by BSA, and 22–45 kPa for unmodified meshes. Optical microscopy and infrared spectroscopy (FTIR) confirmed the improved adhesive strength was achieved through mechanical interlock of the hydrogel tissue adhesive into the PP mesh pores and chemical bonding of the albumin after successful PGMA/HSA grafting onto the PP monofilaments.

Keywords

polypropylene (PP); poloxamine hydrogel adhesive; poly-glycidyl methacrylate/human serum albumin (PGMA/HSA); adhesive strength; surface modification

INTRODUCTION

Polypropylene (PP) is commonly used as sutures and grafts, such as surgical mesh, due to its long-term structural stability and low tissue response.¹ In hernia repair, PP meshes are usually fixed to abdominal tissue by sutures, staples, tacks, or tissue adhesive, such as fibrin glue and synthetic adhesive.^{2,3} Tissue adhesive has notable clinical benefits compared with other methods.³ However, fibrin glue has inadequate tensile and adhesive strengths compared with sutures or synthetic adhesives.⁴ In contrast, synthetic adhesives have

mechanical properties suitable for repairing defects in tissues exposed to high tensile loads, such as the bladder and abdominal wall.^{4–9}

Thermosensitive hydrogels are widely used in commercially available medical products, including drug delivery systems, wound dressings, and tissue engineering scaffolds.^{10–12} They can be fabricated from a variety of common polymers that provide beneficial properties such as biodegradation, flexibility, and fast gelation.¹² In the present study, a bifunctional poloxamine hydrogel adhesive that consists of a modified four-arm poly (propylene oxide)-poly (ethylene oxide) (PPO-PEO) block polymer and a thiol crosslinker was selected for testing.^{7,8} This hydrogel-based tissue adhesive was previously shown to exhibit adhesive strength that exceeds 70 kPa via mechanical interdigitation and covalent bond formation with tissue amines.^{7–9} However, when this hydrogel adhesive was tested with different types of PP mesh, the adhesive strength ranged from 10 to 61 kPa,¹³ which was lower than the adhesive strength for collagen tissues.⁷ It was speculated that the adhesive strength was limited by the hydrophobicity of PP monofilaments and lack of covalent bond formation.¹³ Thus, we hypothesized that surface modifications of PP mesh with introduction of serum proteins might improve the adhesive strength by achieving covalent bonds.

Two surface modification techniques have potential for this application. A common surface modification for PP is protein adsorption achieved through hydrophobic and van der Waals interactions.¹⁴ In this manner, hydrophobic regions in the proteins and the PP surfaces interact, which leaves the hydrophilic regions away from the PP surface, and it is suspected that the non-polar surface chemistry of PP monofilament will lead to the poor protein adhesion. Another surface modification technique involves grafting permanent covalent functional groups onto materials to form a protein coating.^{15,16} Poly-glycidyl methacrylate (PGMA), which contains an epoxy group in each repeating unit, can be used as an anchoring layer for grafting on the surface of medical devices.^{15,17,18} The PP surfaces of mesh can be activated with plasma to provide radicals and these radicals react with water, forming functional groups for depositing the PGMA layer. The PGMA layer has epoxy functionalities that reacts with human serum albumin (HSA) to form a three dimensional plastic albumin.

The purpose of the current study was to use two different surface modifications of PP mesh to improve the adhesive strength between poloxamine hydrogel adhesive and PP mesh by achieving both mechanical interlock and covalent bonds. It was hypothesized that the adhesive strength between the poloxamine hydrogel adhesive and modified PP mesh is stronger than the adhesive strength between the poloxamine hydrogel adhesive and unmodified PP mesh.

MATERIALS AND METHODS

Materials and surface modification

Mesh samples.—Two commercially available heavyweight (HW) (Bard® Composix® Kugel® Hernia Patch) and lightweight (LW) (Bard Composix L/P Mesh, Davol Inc, Warwick, RI) warp knitted meshes were selected (Table I).^{19–21} The LW mesh has large

pore size to facilitate tissue ingrowth and incorporation. The meshes were composed of two layers: PP warp knitted mesh structure and submicron expanded polytetrafluoroethylene (e-PTFE) membrane. PP for surgical mesh is a hydrophobic material with desirable properties of flexibility, chemical resistance and thermal stability. The e-PTFE membrane acts as a barrier layer to minimize the tissue adherence to PP mesh. Since the tissue adhesive is only applied between the PP mesh side and the abdominal wall tissue during surgery, the e-PTFE layer was removed and only the PP meshes were investigated in this study.^{6,22,23} Sample meshes were cut at dimension of 1 cm × 3 cm, to match the aluminum holders used for lap shear testing (ASTM F2255–05).

Bifunctional poloxamine hydrogel adhesive.—Two polymers used for bifunctional poloxamine hydrogel adhesive were synthesized from Tetronic® 1107 (T1107, molecular weight: 15 k Da, HLB:18–23) (BASF corporation, Florham Park, NJ) following the published methods.^{7–9} The polymers were acrylated T1107 and acrylated T1107 with addition of N-hydroxysuccinimide (NHS). The acrylation (ACR) process was used to chemically crosslink the polymer within the hydrogel and the NHS process facilitated binding to the tissue amines.⁸ In the ACR process, the hydroxyl groups at the end of each four arms of T1107 were reacted with acryloyl chloride (Sigma-Aldrich) to form T1107-ACR with acrylate end groups (Figure 1). In the NHS process, the NHS groups (Sigma-Aldrich, St. Louis, MO) were added to partially acrylated T1107 with 50% acrylation to form T1107-ACR-NHS (Figure 1). The composition of T1107-ACR and T1107-ACRNHS was assessed by Proton NMR in d-chloroform.

The bifunctional poloxamine hydrogel solution was made with the final concentration of 30 wt % mixture of 75 wt % T1107-ACR (ACR conversion: 92%) and 25 wt % T1107-ACR-NHS (ACR conversion: 30%). The bifunctional poloxamine hydrogel adhesive was processed by crosslinking the hydrogel mixture with dithiothreitol (DTT) (Across Organics, NJ). This thiol donor crosslinker, was added using the process of Michel-Type addition reaction in 1× phosphate buffer saline.⁹ The molar ratio of thiol to acrylate in this adhesive solution was 1:1. Tetronic® 1107 is thermosensitive in water. The thermal gelation temperature of poloxamine hydrogel adhesive was at room temperature.⁷ When the concentration and/or the temperature are above its critical micellar concentration and critical micellar temperature, the aqueous solution becomes hydrogel. Therefore, this bifunctional poloxamine hydrogel adhesive was kept at 4C before use.

Surface modification.—PP for surgical mesh is a hydrophobic material with low surface energy and lack of functional groups, which make it difficult to strongly adhere layered coatings of another material. Two surface modification methods (BSA adsorption and PGMA/HSA grafting) were compared in this study. The BSA adsorption involved immersing PP meshes in 5% (w/v) bovine serum albumin (BSA) (Sigma-Aldrich) (pH = 7.4) in PBS buffer at 37C for 30 min.^{14,24} Samples were air dried at room temperature before applying poloxamine hydrogel adhesive. The PGMA/ HSA grafting involved fabrication of an albumin coating¹⁵ (Figure 2a). Mesh samples were treated under plasma for 10 min at 700 V DC, 15 mA DC, 10.5 W (Plasma Cleaner /Sterilizer, Harrick, Pleasantville, NY) followed by soaking in water for 30 min to activate the PP surface with hydroxyl, carboxylic

acid, and nitric oxide functional groups. After 30 min, samples were oven dried (~80C) and purged under nitrogen until fully dried. Plasma treated PP mesh samples were dip coated (Meyer Fientechnik, Gottingen, Germany; D3400) in 0.5% (w/v) PGMA ($M_n = 176,000$ g/mol) in chloroform at the speed of 300 mm/min. PGMA modified samples were annealed at 120C for 10 min followed by dip coating in 3% (w/v) human serum albumin (HSA) (Sigma-Aldrich corp., St. Louis, MO; CAS # 7024–90-7) solution in phosphate buffer for 2 h. The PGMA/HSA grafted samples were dried for 12 h and followed by annealing for 2 h at 120C. Unmodified PP meshes (HW&LW) were used as control groups (Table I).

Testing methods

Thermogravimetric analysis.—Unmodified LW and HW mesh samples were analyzed for thermal stability using thermogravimetric analysis (TGA) (TGA Q5000 V3.17 Build 265, TA Instruments, New Castle, DE) with a ramp rate of 10C/min from 25C to 600C in nitrogen gas. Samples were equilibrated at room temperature under nitrogen purge for 10 min prior to heating.

Infrared spectroscopy.—The bonding strength between BSA on the BSA modified PP and between the PGMA and HSA on the PGMA/HSA modified PP surface was confirmed by ATRFTIR analysis (Nicolet Magna 550 FTIR spectrometer equipped with a SpectraTech Endurance Foundation Series Diamond ATR, Thermo, Waltham, MA). Unmodified PP filaments from unmodified meshes were used as control groups. Prior to FTIR, the BSA modified PP and PGMA/HSA modified PP were rinsed in ultrapure water. In order to remove any HSA that was not covalently bonded to the PGMA, the PGMA/HSA modified samples were washed by immersing in phosphate buffer at 37C and pH 7.4 on an orbital shaker for 24 h.

Contact angle.—Static water contact angle of BSA modified, PGMA/HSA modified, and unmodified meshes was measured using optical contact angle goniometer (DSA-20E, FM40Mk2 EasyDrop, Krüss, Germany) to measure the hydrophilicity of the samples. In a controlled environment (humidity: 35%, temperature: 25.4C), droplets of water (2.0 μ L) ($n = 5$) were placed on the edge of meshes. Although the software could automatically measure the contact angle, the mesh knitting structures led to uneven surface, which decreased the accuracy of the automatic contact angle result. The final contact angle was measured manually using Image J (National Institutes of Health).

Lap shear testing.—Adhesive strength, defined as the peak load under uniaxial lap shear testing divided by the contact area between the mesh and adhesive, was measured consistent with ASTM F2255–05 completed at 37C using a 100 N load cell (Synergie 100, MTS, Eden Prairie, MN) at a 10 mm/min loading rate. The testing specimen was formed by two pieces of 1 cm $\times$ 3 cm aluminum holders adhered with collagen sheets (as tissue layers) and PP meshes ($n = 5$ each), separately. A 60 μ L volume of poloxamine hydrogel adhesive was applied over a 1 cm $\times$ 1 cm contact area between the mesh and collagen surface (Figure 3). The curing time for hydrogel adhesive was 1 h. Samples were covered by a cloth containing PBS to maintain moisture. Test groups included BSA modified mesh, PGMA/HSA modified mesh, and unmodified meshes and tissue layers as controls.

Optical microscope images of mesh surface before and after lap shear

testing.—Individual mesh surfaces before and after lap shear testing were observed under a stereo optical microscope (SMZ-168, Motic, Richmond, Canada) with an attached digital camera (Infinity 2, Lumenera, Ottawa, Canada). Images were captured under 12× with 0.243 pixel/cm at dimension of 1392 × 1040 with light projecting from bottom. Meshes modified with PGMA/HSA were investigated and unmodified meshes were used as control groups.

Statistical analyses

The effect of mesh type (LW, HW) and surface modification method (BSA adsorption, PGMA/HSA grafting, unmodified) on adhesive strength was compared with two-way analysis of variance (ANOVA) ($\alpha = 0.05$). The differences between the surface modification techniques and individual mesh types were identified by Tukey's post-hoc multiple comparisons ($\alpha = 0.05$). The effect of surface modification (PGMA/HSA grafting and unmodified) on contact angle was compared with paired t-test ($\alpha = 0.05$). The statistical analyses used OriginLab 9.0 (Northampton, MA).

RESULTS

Thermal property

Thermogravimetric analysis.—The thermal stability of the LW and HW PP meshes was well above the 120°C annealing temperature used in the PGMA/HSA grafting process, with similar behavior for both mesh types (Figure 4).

Surface properties

Infrared spectroscopy.—Chemical bonding between the albumin and PP was confirmed with Fourier-transform infrared spectroscopy (FTIR), with PGMA/HSA successfully grafted onto the PP monofilament (Table II). For PP mesh modified by BSA adsorption, the BSA was easily removed during routine rinsing, demonstrating poor protein adhesion on the PP monofilaments. This loss of BSA was confirmed by FTIR spectra having only a weak absorbance signal between 3300 and 3500 cm^{-1} and around 1600 cm^{-1} (Figure 5). In contrast, the epoxy groups in PGMA assisted HSA protein adhesion to the PP. The presence of PGMA is evident by absorbance around 1730 cm^{-1} (stretching of C=O groups).²⁵ The conversion of epoxy groups in PGMA is evident by the decrease of absorbance around 910 cm^{-1} . The presence of HSA is evident by absorbance at 1541 cm^{-1} (amide I C–H stretching and N–H bending), around 1653 cm^{-1} (bending of N–H groups) and absorbance from 3300 to 3500 cm^{-1} (stretching of amide A N–H groups), which was not observed in unmodified PP.^{26–30} There was no difference in wave number peaks comparing before and after washing in phosphate buffer (Figure 6).

Contact angle.—The hydrophilicity of meshes after surface modification was unchanged compared with unmodified meshes. The values of contact angles were larger than 110° for all types of meshes, consistent with the hydrophobic properties of the PP monofilaments (Figure 7). The meshes modified by PGMA/HSA (113°±6.02° for lightweight mesh and 129°±5.16° for heavyweight mesh) had smaller contact angles than unmodified meshes (128°±

9.14° for lightweight mesh and 138°±9.01° for heavyweight mesh). However, the difference was not statistically significant (paired t-test, $\alpha = 0.05$).

Mechanical property

Lap shear testing.—The PGMA/HSA surface modification improved the adhesive strength for HW and LW meshes attached with poloxamine hydrogel tissue adhesive compared with unmodified meshes (Figure 8, Table III). Mesh type was not a factor affecting the adhesive strength (two-way ANOVA, $p = 0.075$), but surface modification significantly affected the adhesive strength (two-way ANOVA, $p < 0.01$). The adhesive strength between unmodified meshes (44.95±20.16 kPa for heavyweight mesh, 21.69±8.642 kPa for lightweight mesh) and tissue layers was significantly lower than tissue layers alone (Tukey, $p < 0.05$). The adhesive strength of meshes modified by PGMA/HSA (69.63±30.93 kPa for heavyweight mesh, 65.25±16.30 kPa for lightweight mesh) was significantly higher compared with unmodified meshes and meshes modified by BSA (Tukey, $p < 0.05$) and equivalent to tissue layers alone (Tukey, $p > 0.05$). Therefore, the PGMA/HSA surface modification significantly improved the adhesive strength for meshes attached with poloxamine hydrogel tissue adhesive.

DISCUSSION

This study evaluated two surface modification techniques (BSA adsorption and PGMA/HSA grafting) applied to PP mesh to improve the adhesive strength when attached to tissues using poloxamine hydrogel adhesive. The PGMA/HSA surface modification significantly improved the adhesive strength for meshes attached with poloxamine hydrogel tissue adhesive compared with unmodified meshes and meshes modified by BSA adsorption. The improved adhesive strength was achieved through mechanical interlock of the hydrogel tissue adhesive into the PP mesh pores and chemical bonding of the albumin after successful PGMA/HSA grafting onto the PP monofilament.

Surface modifications of PP surgical mesh, including biologic coating materials such as gelatin, purified collagen and extracellular matrix, are commonly pursued to improve various mesh properties.^{21,30–33} The coating should be applied on PP monofilament as a monolayer rather than allowing it to be trapped in mesh pores. A pore size of greater than 1 mm should be maintained to avoid scar plate formation on mesh instead of tissue ingrowth.^{34,35} The spontaneous driving force during coating PP mesh without any other interactions is van der Waals force, which leads to inadequate coating stability. In previous studies, TiMesh light, a PP mesh coated with 30–50 nm hydrophilic titanium coating, tended to achieve better adhesion to tissue without further fixation than other polypropylene mesh.³⁶ In contrast, the adhesive fixation strength of TiMesh with bioadhesive glue was lower than other polypropylene meshes with similar or larger pore size. This may be caused by a weaker interaction between polypropylene and titanium coating than the chemical bonds between titanium coating and the bioadhesive. In our study, the fixation strength with poloxamine hydrogel adhesive for BSA-modified PP mesh was also lower than unmodified PP mesh. In aqueous solutions, the core of the protein is hydrophobic, whereas the outer edges are hydrophilic. Therefore, the BSA coating was mainly adsorbed onto the PP surface

through van der Waals interaction. However, these interactions are notably weak, and adsorbed proteins can be separated from the PP surface when the interaction strength between the hydrophilic regions and the surrounding environment is larger than the hydrophobic and van der Waals interactions.

Key aspects of the PGMA/HSA surface modification provided for strong chemical bonds between the PGMA/HSA coating and the PP surface and helped to prevent disruption of the coating during lap shear testing. In fact, Luzinov et al.^{17,37–43} and others^{44–47} clearly demonstrated that PGMA reacts with polymeric surfaces treated by air plasma and that proteins are readily reacting with epoxy groups of PGMA via amino and carboxyl functionalities. For instance, it was demonstrated using XPS that after enzymatic (protease) removal of fibrinogen or bovine serum albumin layers anchored to PGMA significant amount of amino acids remain grafted to the PGMA layer.³⁷ During the surface modification process, the PGMA layer was first strongly chemically bonded through epoxy groups on the plasma activated PP surface. Further chemical bonds were formed between amino and carboxyl groups in albumin and remaining epoxy groups in PGMA, as confirmed by FTIR. The results from the current study support previous research using plasma treatment to improve the adhesion of functional groups and PGMA grafting to achieve strong covalent bonding between hydrophobic PP mesh surfaces and the hydrophilic coatings.^{26,30} Gorgieva et al. activated PP mesh with O₂ plasma to create functional groups on the PP surface, which further formed covalent bonds with a hydrophilic gelatin containing an antimicrobial-active agent while maintaining the mesh structural and mechanical properties.³⁰ The PGMA/HSA surface modification in the current study involved a “grafting to” method, with PGMA providing a uniform and homogeneous macromolecular anchoring layer (polymer brushes) for grafting polymers or proteins (e.g., HSA) to the PP surface.^{15–17} The PGMA epoxy groups are relatively stable at elevated temperature, which allowed for heat annealing and denaturation of HSA at 120°C. PGMA polymer layers also can be deposited on various polymers, such as polyethylene terephthalate (PET), polyethylene (PE), PP, polyvinylidene fluoride (PVDF), and nylon.¹⁷

Adequate adhesive strength for PP mesh fixation is critical for repairing defects in abdominal wall tissues that are exposed to high tensile loads. Adhesive strength should reach the maximum tensile strength of the abdominal wall of 11–27 N/cm and the maximum intra-abdominal pressure of 22.8 kPa in healthy adults.^{36,48} Using the approach of SchugPass et al. and assuming a defect size of approximately 1 cm² area, an abdominal pressure of 22.8 kPa would generate a force of approximately 2.28 N over the defect area.³⁶ The necessary adhesive strength to resist this physiological force can be calculated by dividing by the contact area between the mesh and adhesive. In the current study, an area of 1 cm² adhesive resulted in peak loads of 6.5–7 N during lap shear testing of meshes modified by PGMA/HSA (Figure 8), exceeding the maximum physiological levels. The mode of failure for this hydrogel adhesive is adhesive failure, caused by detachment between adhesive and tissue.⁸ However, when the adhesive strength between adhesive and mesh is lower than that between adhesive and tissue, detachment occurs between adhesive and mesh. The large pores of PGMA/HSA modified samples after lap shear testing were filled with much more poloxamine hydrogel tissue adhesive than unmodified mesh (Figure 9) confirming strong

chemical bonding and mechanical interlock are formed between PGMA/HSA modified mesh samples and hydrogel adhesive.

In previous studies, fibrin glue or semi-synthetic adhesives have met the above physiological requirements, but have used much larger areas of adhesive on the mesh surface.^{6,26,36} For example, an area of 134 cm² of fibrin glue was used to achieve a 4.6–7.2 kPa adhesive strength for lightweight meshes in a ball burst set-up,³⁶ which is below the 65 kPa average adhesive strength of LW meshes modified by PGMA/HSA in the current study (Figure 8). Moreover, the high viscosity of semi-synthetic adhesives prevents application over a large contact area during laparoscopic surgery.⁶ Synthetic cyanoacrylates adhesives have the advantages of fast fixation and adequate adhesive strength for hernia repair,^{4,5} but in vitro cytotoxicity due to formaldehyde release and in vivo tissue toxicity have been reported.⁴⁹

It is recognized that the nature of the abdominal wall affects the mechanical behavior of implanted surgical meshes in biaxial directions. A limitation of this study is that only uniaxial lap shear testing was used to assess adhesive strength. This method was selected because it is a common, highly repeatable method for generating adhesive strength results that are comparable to previous studies of this poloxamine hydrogel adhesive.^{7–9} Moreover, existing biaxial testing procedures (e.g., ball burst test, clamp-needle set-up) are insufficient for predicting the in vivo three-dimensional behavior of mesh. The ball burst test assumes uniform distribution of tension throughout all fibers within the mesh. However, the knitted structure of surgical mesh always behaves anisotropically.^{50,51} Another limitation is the use of collagen to represent the tissue layer, which does not fully capture the complex biological aspects of the abdominal wall. In future studies, in vivo applications of PGMA/HSA modified PP mesh and hydrogel adhesive will be explored to characterize adhesion to tissues and any potential negative effects.

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FIGURE 1.

Tetronic® T1107 (A) acrylation reaction, (B) N-hydroxysuccinimide (NHS) reaction. T, Tetronic® T1107; DCM, Dichloromethan (HPLC grade); TEA, trimethylamine; THF, tetrahydrofuran; DMAP, 4-dimethylaminopyridine; EDC, 1-(3-dimethylaminopropyl)-3-ethylcarbodiimide hydrochloride.⁹

FIGURE 2. Schematic diagram for surface modification methods. (a) PGMA/HSA surface modification method. Modified mesh samples were annealed at 120°C. (b) BSA surface modification.

FIGURE 3.
Specimen for lap shear testing.

FIGURE 4.

The weight loss (TGA curves, mass %) and the rate of weight loss [derivative thermogravimetry curves (DTG), mass %/°C] of unmodified HW and LW meshes. Green solid line: TGA curve for lightweight mesh, red solid line: TGA curve for heavyweight mesh, blue dashed line: DTG curve for lightweight mesh, black dashed line: DTG curve for heavyweight mesh.

FIGURE 5.
FTIR spectra of heavyweight and lightweight meshes modified by BSA.

FIGURE 6.

FTIR spectra of unmodified mesh, heavyweight, and lightweight meshes modified by PGMA/HSA before and after phosphate buffer washing.

FIGURE 7.

Contact angle images of heavyweight and lightweight meshes before and after PGMA/HSA modification. U, unmodified; H, heavyweight; L, lightweight. Paired *t*-test of contact angle before and after surface coating: $p = 0.05$.

FIGURE 8.

Adhesive strength between mesh and tissue under uniaxial lap shear testing. UH, unmodified heavyweight mesh; UL, unmodified lightweight mesh; H-BSA, heavyweight mesh modified by BSA; L-BSA, lightweight mesh modified by BSA; H-PGMA/HSA, heavyweight mesh modified by PGMA/HSA; L-PGMA/HSA, lightweight mesh modified by PGMA/HSA.

FIGURE 9.

Hydrogel tissue adhesive residue on unmodified and PGMA/HSA modified mesh samples after lap shear testing. Unmodified mesh with minimal attached adhesive have pores without residue (bright white) and modified mesh with attached adhesive have pores with residue (gray). U, unmodified; L, lightweight mesh; H, heavyweight mesh.

TABLE I.

Polypropylene Mesh Information and Surface Modification Techniques

Mesh Type	Classification	Structure	Weight (g/m ²)	Thickness (mm)	Porosity	Surface Modification
HW	Standard Class III		95	0.57	Area: 57% Weight: 83%	PGMA/HSA grafting BSA adsorption None
LW	Light Class III		41	0.48	Area: 64% Weight: 90%	PGMA/HSA grafting BSA adsorption None

Classification: (16,17)

Porosity: (18)

TABLE II.

Functional Groups in FTIR Wavelength

Bond	Wavenumber (cm ⁻¹)	Reference
PP		
C—H ₃ and —CH ₂ —	2990–2850	(13)
	1380–1370	
	1475–1450	
	1465–1440	
PGMA		
C=O	1530–1830	(22,27)
C—O	1240,1270	
C—O—C stretching	1189, 1141	
Albumin		
Amide A NH stretching	3300	(24–26)
Amide I band	1600–1700	
β sheet	1610–1640	
Random coil	1640–1650	
α helix	1650–1658	
β-turn structure	1660–1700	
Amide II band	1500–1600	
Amide II CH stretching and NH bending	1541	

TABLE III.

Statistical Analysis of Lap Shear Testing Strength with Factors of Mesh Type and Surface Modification Method (*: Tukey Post-Hoc, $p < 0.05$)

Adhesive Strength (kPa)	Surface Modification	Collagen	Unmodified	BSA	PGMA/HSA
Collagen	Collagen				
UH	Unmodified	66.59 ± 25.26	*	*	$p > 0.05$
UL		44.95 ± 20.16			*
H-BSA	BSA	21.69 ± 8.642	—	*	
L-BSA		13.29 ± 4.378			*
H-PGMA/HSA	PGMA/HSA	4.053 ± 0.8463		—	
L-PGMA/HSA		69.63 ± 30.93			—
		65.25 ± 16.30			