ISSN 2616-5880 Vol. 7, Issue 1: 121-129, DOI: 10.25236/AJMC.2026.070117

Injectable polyacrylic acid hydrogels with dual crosslinking networks for wound closure and healing

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Abstract: *This study fabricated polyacrylic acid hydrogels via UV-initiated radical polymerization. Two types were constructed: physically cross-linked P-PAA (via intermolecular hydrogen bonds) and chemically cross-linked C-PAA (using MBA as crosslinker). Both were prepared through simple, mild procedures suitable for scale-up. SEM showed P-PAA had a loose, large-pore network, while C-PAA exhibited denser architecture due to higher covalent crosslinking. Mechanical tests revealed compressive strengths of 0.21 MPa (P-PAA) and 0.54 MPa (C-PAA), indicating superior resistance for the chemically crosslinked system. Tissue adhesion tests confirmed that abundant carboxyl groups on PAA formed strong hydrogen bonds with skin surface amino groups, enabling tight adherence to porcine skin even under bending and torsion. Biosafety evaluation via CCK-8 and live/dead staining demonstrated no cytotoxicity and significant promotion of NIH-3T3 fibroblast proliferation for both hydrogels. In a rat full-thickness skin incision model, compared to sutures, PAA dressings remarkably reduced inflammation, promoted angiogenesis and fibroblast proliferation, accelerated wound closure, optimized collagen arrangement, and alleviated scar formation. Benefiting from reversible hydrogen-bond networks, P-PAA offered shear-thinning injectability, self-healing, and controllable pain-free debonding, avoiding secondary injury during dressing changes. In contrast, C-PAA provided higher mechanical strength and adhesion, suitable for wounds under high tension. The two hydrogels complement each other, covering applications from first aid and field care to clinical surgery. With convenient operation and low cost, they address the shortcomings of existing medical adhesives in adhesion, biocompatibility, and degradability, showing promising clinical translation prospects.*

Keywords: *hydrogel, microstructure, mechanical property, tissue adhesive, skin repair*

1. Introduction

Soft tissue injuries most commonly occur in organs such as skin and mucous membranes. Their repair mainly relies on suture closure or stapling to bring wound edges into apposition. Nevertheless, sutures and skin staples possess moduli far exceeding those of human soft tissues, inevitably causing secondary trauma and additional pain to surrounding tissues. Moreover, gaps between sutures or staples remain vulnerable sites on wounds. Some patients develop severe scarring due to prominent foreign body rejection reactions [1]. For injuries to softer or fragile visceral organs such as the liver, spleen and kidney, sutures or staples are not applicable. Only temporary compressive dressing can be used for hemostasis. Dressings represent one of the most widely adopted clinical interventions for treating chronic wounds, offering temporary wound protection, hemostasis, and anti-contamination functions [2]. To date, gauze-based products remain the most prevalently used dressings over the longest clinical history. However, traditional gauze dressings suffer from notable drawbacks: they impede epithelialization and delay wound healing; loose dressing fibers may detach and trigger foreign body reactions that interfere with healing; granulation tissue can grow into gauze meshes, producing pain and secondary damage during dressing removal; they provide insufficient bacterial barrier properties, allowing pathogen penetration and subsequent infection [3]. When injuries are complicated by special conditions, current therapeutic modalities often lack effective management strategies, especially for hemorrhage and infection. Compressive gauze hemostasis, the most common first line hemostatic approach, yields partial hemostatic effects yet carries risks of tissue damage and difficult removal. For injuries with high infection risks, internal fixation materials may provide favorable niches for microbial colonization [4-6]. Given the

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persistently high incidence of injuries complicated by severe hemorrhage or infection, existing therapies struggle to integrate hemostasis, anti-infection, and healing functions for complex wound management.

Tissue adhesives have been extensively investigated and developed as promising alternatives addressing the above-mentioned clinical challenges. Clinically available tissue adhesives mainly include fibrin glues and cyanoacrylate-based products. These adhesives enable convenient operation, avoid secondary injury, alleviate patient pain, and partially substitute internal fixation devices. Even so, numerous defects remain to be resolved for current medical tissue adhesives [7]. For direct contact with human tissues, medical adhesives must possess favorable biocompatibility and degradability: they should degrade completely or produce metabolically excretable degradation products to prevent immune rejection and guarantee safety [8]. Beyond these properties, low manufacturing costs and feasibility for mass production constitute key research priorities for clinical practicality.

Ideal wound repair materials should possess multiple desirable characteristics: rapid pro coagulant capacity, maintenance of a suitably moist wound microenvironment, oxygen permeability, absorption of wound exudates, accelerated wound closure, pain relief and infection prevention. Apart from conventional sterile gauze and absorbent cotton, multiple functional materials have been exploited for wound repair, including nanosilver, chitosan, alginate, collagen, and polyurethane. Although these materials exert certain beneficial effects on wound repair, their clinical applications are restricted by limitations such as single function, complicated fabrication workflows, high cost, unsatisfactory physicochemical properties, and potential toxicity. Accordingly, exploring and developing ideal wound repair materials remains a major hotspot and challenge in this research field.

Hydrogels feature three-dimensional network matrices resembling the extracellular matrix (ECM) and mechanical elasticity comparable to soft human skin tissues. They can mechanically induce cell adhesion, migration, and ECM deposition, thereby promoting wound repair. Microstructural parameters of hydrogels, such as water content, cross linking density and pore size, directly regulate cellular behaviors and consequently influence skin wound healing outcomes. As hydrophilic polymers forming stable cross-linked networks upon water absorption, hydrogels provide fundamental physical isolation and moisturizing effects for wound dressings. A moist wound microenvironment mitigates cellular dehydration, sustains cell proliferation and migration activity, optimizes the healing niche and retards progressive wound deterioration. Satisfactory moisturizing and hydration capacity constitutes one of the basic requirements for wound repair hydrogel products approved by the U.S. Food and Drug Administration [9,10]. Commercially available clinical hydrogel dressings (e.g., IntraSite® Gel, Flamigel® Gel, Purilon® Gel) outperform traditional dressings (gauze, absorbent cotton) in maintaining wound moisture balance, facilitating autolytic debridement, accelerating wound healing, and mitigating dressing change related pain. Studies indicate that protein-based polymers (gelatin, collagen, oligopeptides) suffer from limited solubility, compromising long term water retention of hydrogel scaffolds. By contrast, water soluble polysaccharides and their derivatives (hyaluronic acid, chitosan, sodium alginate) are widely applied in wound repair owing to superior long lasting moisturizing performance. The moisturizing efficacy of wound dressings is governed by water vapor transmission rate, which controls water evaporation at wound sites [11,12]. Therefore, tuning the water vapor transmission rate of hydrogels can modulate wound moisturizing performance. Appropriate hydrogels should be selected according to wound type, exudate volume and healing stages to achieve optimal therapeutic outcomes. Amorphous hydrogels composed of hydroxymethyl cellulose and propylene glycol have been directly applied clinically to deep partial thickness burn wounds after eschar shaving. Such hydrogels deliver prominent moisturizing effects, markedly reduce dressing wound adhesion, and associated patient pain, elevate wound healing rates, lower infection risk, decrease skin grafting necessity, reduce dressing change frequency, shorten complete wound healing time, and mitigate hypertrophic scar formation.

The most essential function of tissue adhesives is generating sufficient bonding strength across tissue interfaces to achieve wound closure without internal fixation devices. A greater practical challenge lies in adhesive performance under bleeding or exudate rich conditions in vivo, namely wet surface adhesion capability [13]. Despite substantial research efforts devoted to developing materials satisfying these requirements, few existing tissue adhesives simultaneously combine favorable biocompatibility and robust wet surface adhesion, highlighting an urgent demand for such biomaterials [14]. Accordingly, two injectable poly(acrylic acid) (PAA) hydrogel adhesives with strong tissue adhesive properties were designed and fabricated according to the physiological features of skin injuries (Figure 1). Abundant surface carboxyl groups readily form strong physical adhesive interactions with amino groups on tissue surfaces, endowing the hydrogels with favorable mechanical strength, prominent tissue adhesion and biocompatibility for rapid attachment and effective repair of injured rat skin wounds.

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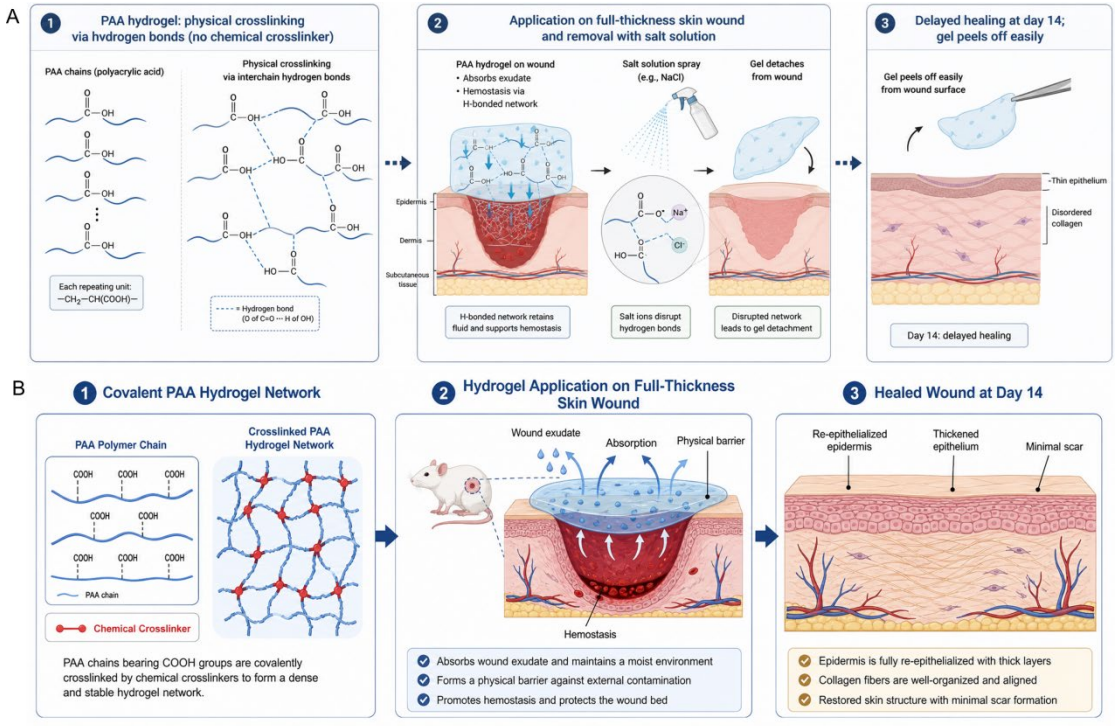

Figure 1: Schematic illustration of the fabrication route and wound sealing mechanism of injectable hydrogel adhesives.

2. Experimental Section

2.1 Materials

Acrylic acid (AA, 98%, Energy Chemical Reagent Co., Ltd.), manganese chloride (Macklin Reagent Co., Ltd.), 2-hydroxy-4'-(2-hydroxyethoxy)-2-methylpropiophenone (Irgacure 2959, Alfa Aesar), and N,N′-methylene-bis-acrylamide (MBA, 98%, Energy Chemical Reagent Co., Ltd.) were used as received. All other reagents were purchased from Concord Reagent Co., Ltd. and employed without further purification. Bone marrow mesenchymal stem cells (BMSCs) were obtained from the China Infrastructure of Cell Line Resource.

2.2 Fabrication of P-PAA and C-PAA Hydrogels

Both P-PAA and C-PAA hydrogels were synthesized via photoinitiated radical polymerization under ultraviolet (UV) irradiation for 4 hours, using identical monomer concentrations and mold geometries. For the crosslinked version (C-PAA), AA (1.8 g) was dissolved in 10 mL of deionized water together with Irgacure 2959 (57.6 mg, 1 mol% relative to AA) and MBA solution (118 μL of 10 mg mL^{-1} stock, corresponding to 0.03 mol% of AA). The mixture was then cast into molds and exposed to UV light to yield the C-PAA network. The non-crosslinked P-PAA hydrogel was prepared following the same protocol, except that the crosslinker MBA was omitted from the formulation.

2.3 Characterization

Morphological observations were performed on a JSM 6700F field-emission scanning electron microscope (FE SEM). Prior to imaging, freeze dried specimens were sputter coated with a thin platinum layer for 90 s to enhance electrical conductivity. Mechanical properties were evaluated at room temperature using an Instron 3365 universal testing machine. Cylindrical hydrogels were cast in plastic tube molds (6 mm inner diameter) for tensile tests and in glass vials (9 mm inner diameter) for compression tests; the actual diameter of each sample was carefully measured before testing. Tensile measurements were run at a crosshead speed of 50 mm min-1, while compressive tests were performed at 5 mm min^{-1}.

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2.4 Cytotoxicity and Cell Proliferation Assay

The cytotoxic effect of hydrogel extracts was assessed via the Cell Counting Kit-8 (CCK-8) method. BMSCs were seeded in culture dishes and incubated at 37 °C under 5% CO2 for 24 h. Thereafter, the culture medium was replaced with extracts from the hydrogels, and the cells were cultured for 1, 3, or 7 days. At each time point, the medium was exchanged for fresh DMEM supplemented with 10 μL of CCK-8 solution per well. Absorbance was recorded at 450 nm using a microplate reader to calculate cell viability. Control groups consisted of cells grown in standard DMEM containing 10% fetal bovine serum (FBS). All measurements were carried out in triplicate, and viability was determined by the formula:

$$\text{Cell viability (\%)} = [(A_{sample} - A\_blank) / (A\_control - A\_blank)] \times 100\% \quad (1)$$

2.5 Live/Dead Staining Assay

BMSCs cultured with hydrogel extracts were subjected to live/dead fluorescence staining. A working solution containing red fluorescent propidium iodide (PI) and green fluorescent AM stain was added to the cells and incubated at 37 °C for 1 h. Fluorescence signals were then visualized by confocal laser scanning microscopy (CLSM) with excitation at 568 nm (PI) and 488 nm (AM) to distinguish dead and live cells, respectively.

2.6 In Vivo Skin Adhesion in Rats

Female Sprague-Dawley rats (4 weeks old, ~150 g) were employed as an animal model to evaluate the wound closure performance of the PAA hydrogel. After anesthesia with 2 wt% pentobarbital sodium, the dorsal hair was shaved and the skin disinfected with 75% ethanol. Two full thickness incisions (2 cm each) were made on both sides of the back using a scalpel. One wound received the hydrogel adhesive (approximately 0.6 mL per incision) applied via a dual channel syringe, while the contralateral wound was closed with conventional surgical sutures as a control. On postoperative days 3, 7, and 14, the rats were euthanized, and the incised tissues were harvested, sectioned, stained with hematoxylin and eosin (H&E), and examined under an Olympus microscope. In addition, healed skin samples from the 14 day group were excised and cut into rectangular strips of 7 × 2 cm, with the incision line centered in each strip. Tensile testing was then conducted at a strain rate of 50 mm min^{-1} to compare the mechanical recovery of the adhesive-treated versus suture-treated groups.

2.7 Ethical Statement

Animal experiments were conducted in accordance with the guidelines for the management and use of laboratory animals established by Beijing Feiyi Technology Co., Ltd. (Ethical approval numbers: FY-2024-YX-005 to 011).

3. Results and Discussion

3.1 Preparation of the Hydrogels

In this study, acrylic acid (AA) was used as the monomer to synthesize poly(acrylic acid) (PAA) polymer chains via ultraviolet (UV)-initiated free-radical polymerization. In the absence of any chemical crosslinker, such as N,N′-methylenebisacrylamide (MBA), the abundant carboxyl groups (-COOH) along the polymer chains interact through intermolecular hydrogen bonding, spontaneously forming a reversible physically crosslinked three-dimensional network and thereby yielding a physically crosslinked poly(acrylic acid) hydrogel (P-PAA). In contrast, when MBA was introduced into the polymerization system as a chemical crosslinker, the acrylic acid chains were crosslinked not only through hydrogen-bonding interactions but also through permanent covalent bonds, resulting in a chemically crosslinked poly(acrylic acid) hydrogel (C-PAA).

The two hydrogels exhibited distinct network architectures. In P-PAA, the physical crosslinking junctions are reversible and dynamically exchangeable, endowing the hydrogel with pronounced shear-thinning behavior and injectable processability. Consequently, P-PAA can be conveniently delivered through a syringe or dual-channel spray applicator and conformally applied to irregular wound surfaces, enabling in situ gel formation and immediate wound sealing. In contrast, the C-PAA network is stabilized by irreversible covalent crosslinks, resulting in greater structural stability but reduced injectability.

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Notably, regardless of the crosslinking mechanism, abundant free carboxyl groups remain along the PAA chains. These functional groups not only serve as sites for intermolecular interactions within the hydrogel network but can also form strong hydrogen bonds with reactive groups, such as amino and hydroxyl groups, present on the surface of skin tissue, thereby facilitating rapid and robust tissue adhesion. A schematic illustration of the preparation procedures for P-PAA and C-PAA and their application for wound adhesion is presented in Fig. 1. The prefabricated hydrogel adhesives can be stored for extended periods and require neither on-site formulation nor additional initiation prior to use. This simplifies handling and substantially shortens preparation time in emergency and surgical settings, making the materials particularly attractive for rapid hemostasis and wound closure in urgent-care applications.

3.2 Structural and Mechanical Characterization of the Hydrogels

To investigate the differences in the microstructures of the two hydrogels, freeze-dried P-PAA and C-PAA samples were examined by scanning electron microscopy (SEM). As shown in Fig. 2, both hydrogels exhibited loosely porous, foam-like morphologies with relatively uniform pore distributions, which may facilitate the exchange of nutrients and metabolic waste while providing physical space for cell migration. However, P-PAA exhibited substantially larger pores than C-PAA, which can be attributed to its lower effective crosslinking density. In P-PAA, crosslinking relies predominantly on hydrogen-bonding interactions, resulting in a relatively sparse population of crosslinking junctions and greater mobility of the polymer chains. Consequently, larger pores are formed following freeze-drying. In contrast, the densely distributed covalent crosslinks in C-PAA restrict polymer-chain extension and rearrangement, resulting in a more compact network structure and correspondingly smaller pores.

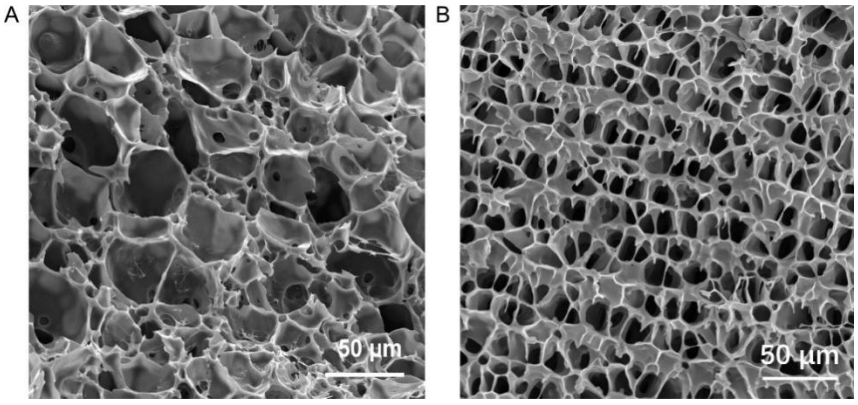

Figure 2: Microstructures of P-PAA and C-PAA hydrogels.

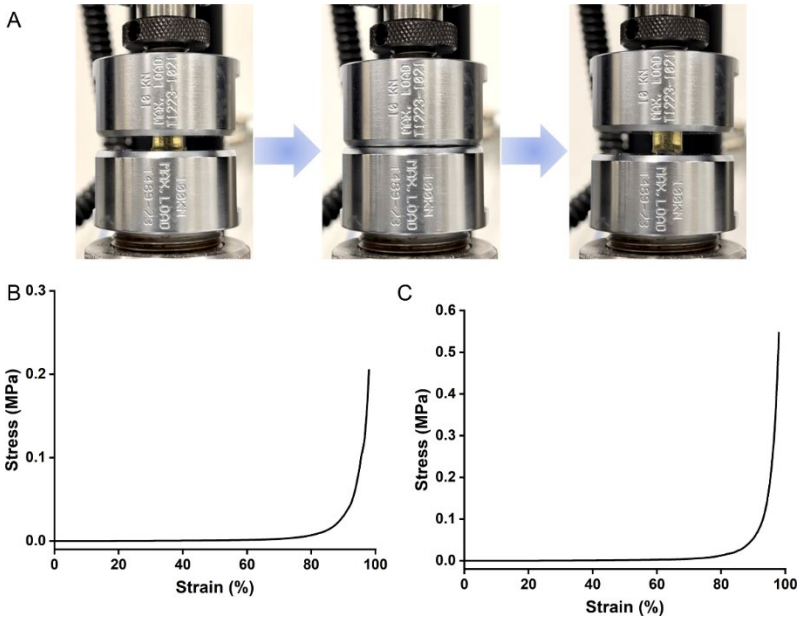

Figure 3: Mechanical property of P-PAA and C-PAA hydrogels.

Differences in crosslinking density also directly affected the mechanical properties of the hydrogels.

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As shown by the compressive stress–strain curves in Fig. 3, the compressive strengths of P-PAA and C-PAA reached 0.21 MPa and 0.54 MPa, respectively, with the latter being approximately 2.6-fold higher than the former. These results demonstrate that covalent crosslinking substantially enhances the compressive strength of the hydrogel, making C-PAA more suitable for wound sites subjected to relatively high external mechanical loads. Although physically crosslinked P-PAA exhibited lower mechanical strength, it possessed greater flexibility and deformability. Moreover, the reversible breaking and reformation of hydrogen bonds provided the material with efficient energy dissipation and self-recovery, conferring distinct advantages in dynamically deforming wound environments.

In addition, a wound adhesive must possess sufficient tissue adhesion strength to maintain stable attachment under exposure to body fluids and continuous mechanical deformation of the skin. Porcine skin was therefore used as an ex vivo tissue model, and PAA hydrogels were formed in situ on the skin surface. The abundant carboxyl groups within the hydrogel are theoretically capable of forming multiple hydrogen bonds with amino, amide, and hydroxyl groups present in proteins and glycoproteins on the porcine skin surface. These interactions may act synergistically with electrostatic and van der Waals interactions, thereby generating robust physical adhesion at the tissue–hydrogel interface.

As shown in the photographs in Fig. 4, the PAA hydrogel formed intimate contact with the porcine skin surface. Even under repeated bending, twisting, and stretching, the hydrogel layer remained firmly attached without visible peeling or interfacial slippage, indicating that the adhesive strength was sufficient to withstand substantial tissue deformation. These observations support the effectiveness of hydrogen-bond-mediated adhesion between PAA hydrogels and biological tissues and provide an experimental basis for subsequent evaluation of wound healing in animal models. Collectively, P-PAA combines favorable tissue adhesion, injectability, and appropriate mechanical properties, highlighting its potential as a promising material for wound sealing and tissue repair.

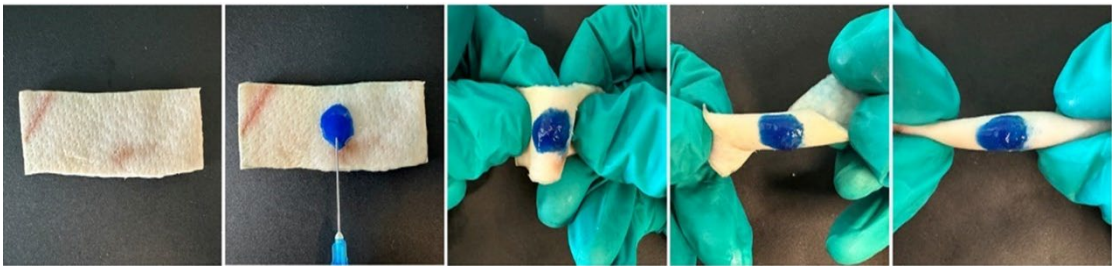

Figure 4: Adhesive performance of the PAA hydrogel on porcine skin.

3.3 In Vitro Cytotoxicity and Biocompatibility of Hydrogels

The biosafety and efficacy are core prerequisites for evaluating whether a biomaterial is suitable for in vivo wound repair. Prior to conducting animal experiments to assess wound adhesion effectiveness, we systematically verified the biosafety of PAA hydrogels from the perspectives of cytocompatibility and histocompatibility. Specifically, we prepared extracts of the PAA hydrogels and co-cultured them in vitro with mouse fibroblasts (NIH-3T3). After 24 hours of culture, cell morphology and adhesion status were observed using an inverted microscope. The results showed that, compared with the blank control group (PBS solution), cells in the PAA hydrogel extract groups exhibited plump morphology and good spreading, with no obvious cell shrinkage, floating, or disintegration. This preliminarily indicated that the PAA hydrogels lack acute cytotoxicity, and that residual trace monomers and initiators from the synthesis process had been essentially removed, posing no visible threat to cell growth.

To further distinguish the distribution of live and dead cells, we performed dual-fluorescence staining using Calcein-AM (green fluorescence for live cells) and propidium iodide (red fluorescence for dead cells). As shown in the confocal microscope images in Figure 5A, both the physically cross-linked (P-PAA) and chemically cross-linked (C-PAA) hydrogel extract treatment groups displayed dense and bright green fluorescence signals, while the red fluorescence signals representing dead cells were extremely weak and nearly invisible. This result intuitively demonstrated that PAA hydrogels from both cross-linking network systems do not compromise cell membrane integrity, and that the difference in crosslink model (hydrogen bond crosslink vs. covalent crosslink) does not affect the inherent cytocompatibility of the material, fully verifying the high safety of this material at the cellular level.

Beyond static cell viability, we dynamically monitored the effect of the hydrogels on cell proliferation activity using the CCK-8 assay. As shown in Figure 5B,C, compared with the PBS control group, both

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PAA hydrogel groups exhibited significant time-dependent increases in optical density (OD) values after 1, 3, and 5 days of culture, with proliferation rates significantly exceeding that of the control group. This indicated that PAA hydrogels not only lack cell-inhibitory effects but also significantly promote the proliferation of NIH-3T3 cells. Mechanistically, this may be attributed to the abundance of carboxyl groups (-COOH) in the PAA hydrogel backbone—these hydrophilic groups impart good water retention and a negatively charged microenvironment, effectively adsorbing serum proteins and various growth-promoting factors from the culture medium, thus providing a more favorable substrate interface for cell adhesion and division, thereby accelerating the cell cycle progression. This result strongly confirmed the excellent biocompatibility of the PAA material from the perspective of cellular metabolic activity.

Collectively, the in vitro cellular assays (including cell morphology observation, live/dead staining, and proliferation activity detection) demonstrate that PAA hydrogels exhibit excellent cytocompatibility, showing no significant toxicity while promoting cellular functional activity. This conclusion provides a solid theoretical foundation and safety assurance for subsequent in vivo animal implantation and wound adhesion effectiveness experiments. Based on this, we further established a rat full-thickness skin incision model to investigate the histocompatibility, in situ degradation characteristics, and actual wound closure efficacy of PAA hydrogels in a more complex physiological environment, thereby comprehensively evaluating their potential as medical adhesives.

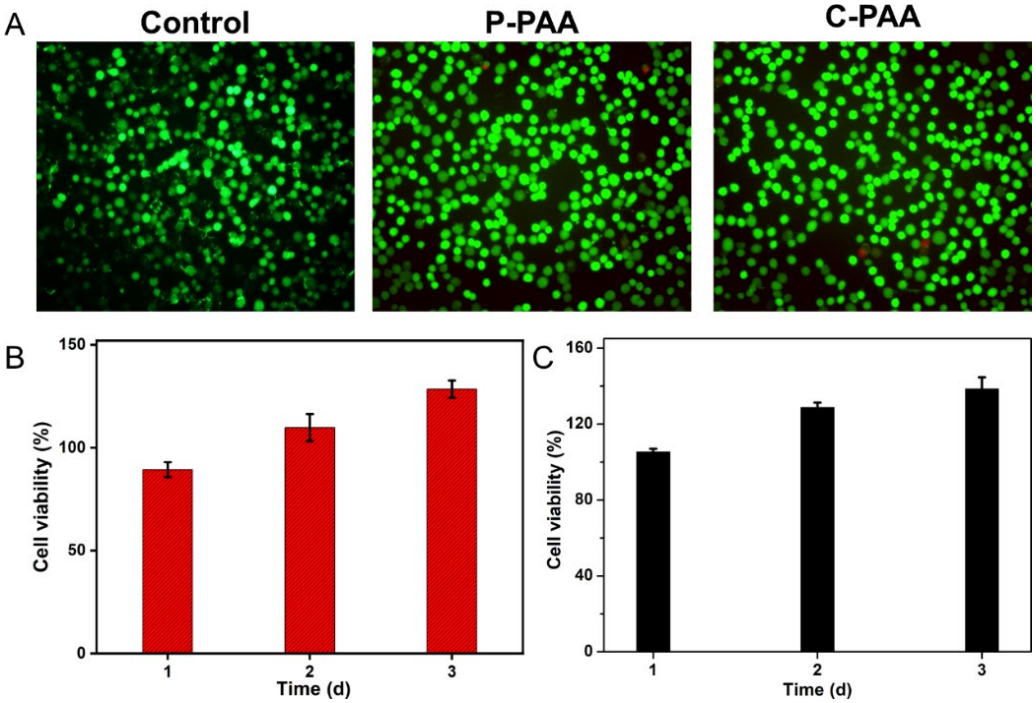

Figure 5: (A) Live/dead staining images of cells co-cultured with PAA hydrogel degradation products for 24 h; (B)CCK-8 assay for cytotoxicity of PAA hydrogel degradation products at various times.

3.4 Skin Wound Repair

After characterizing the mechanical adhesion and biosafety of the PAA hydrogels, and considering their favorable injectability and detachability, we selected the rat skin incision model to validate the effectiveness of P-PAA hydrogels as bioadhesives, investigating whether they could effectively close skin wounds and facilitate tissue regeneration. We made 2 cm long incisions symmetrically on both sides of the rat's back and closed the wounds using either P-PAA hydrogel or manual suturing. Wound healing was observed grossly on postoperative days 3, 7, and 14. As shown in Figure 6A, incisions were visible in both groups on day 3 post-surgery. By day 7, wounds closed with the PAA hydrogel had essentially healed, whereas the sutured group still exhibited residual scabbing around the incisions, likely due to secondary tissue damage caused by the suturing procedure. By the second postoperative week, all incisions in both groups had essentially healed. To further observe subcutaneous tissue healing, tissue samples near the incisions were collected on postoperative days 3, 7, and 14, sectioned, and subjected to H&E staining and examination. As shown in Figure 6B, on day 3 post-surgery, obvious tissue disruption was observed near the incisions in both groups, indicating that the wound had not yet been repaired. On day 7 post-surgery, the sutured group showed extensive fibrous hyperplasia near the incision, whereas

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the P-PAA adhesive group already exhibited hair follicle formation with milder fibrosis, suggesting that the adhesive effectively promotes wound repair. By day 14 post-surgery, histological fibrosis had alleviated in both groups, consistent with the gross observations.

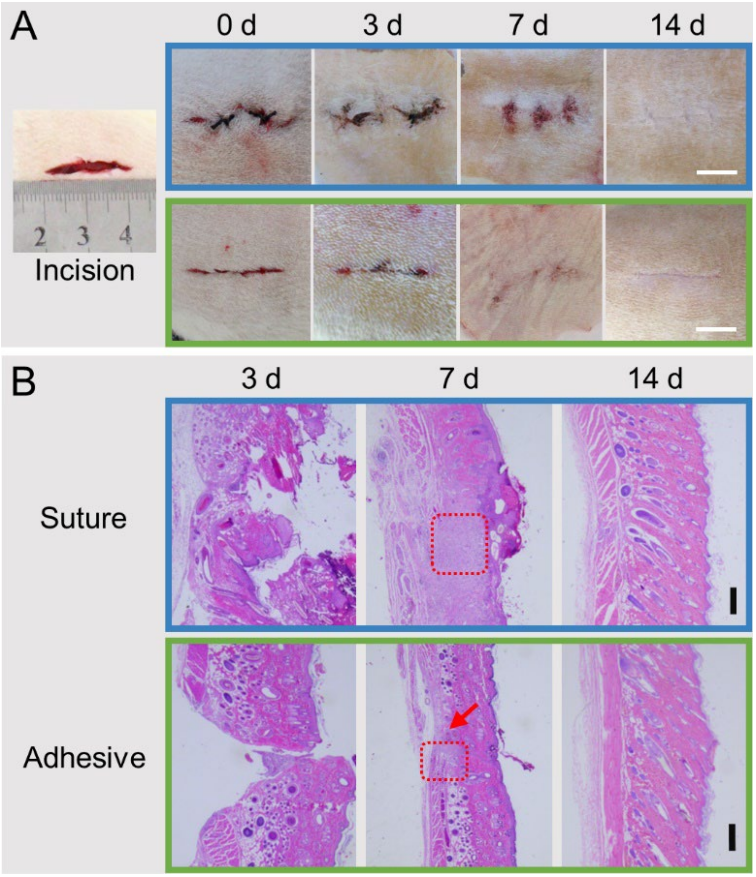

Figure 6: Closure experiment of rat's back skin incisions; (A) Representative photographs of incisions treated with sutures (blue dashed boxes) and adhesive (green dashed boxes), respectively, scale bar = 1 cm; (B) H&E staining results of skin tissue sections adjacent to the incisions. Fibrotic tissues are outlined by red dashed circles. Arrows indicate hair follicle formation, scale bar = 1 mm.

4. Conclusion

In summary, acrylic acid was used as the monomer to construct a physically cross-linked poly(acrylic acid) hydrogel (P-PAA) through UV-initiated free-radical polymerization. Without the addition of chemical cross-linking agents, the hydrogel was physically cross-linked through intermolecular hydrogen bonding. Meanwhile, a chemically cross-linked hydrogel (C-PAA) was prepared by introducing N,N′-methylenebisacrylamide (MBA). This preparation process is simple and conducted under mild conditions, requires no complicated post-treatment, and has good potential for large-scale production. Scanning electron microscopy (SEM) showed that P-PAA had a loose and porous three-dimensional network with relatively large pore sizes, resulting from its relatively low hydrogen-bonding cross-linking density. In contrast, C-PAA had a denser pore structure due to its highly cross-linked covalent network. Mechanical testing showed that the compressive strength of C-PAA reached 0.54 MPa, significantly higher than the 0.21 MPa of P-PAA, demonstrating its superior compressive performance and making it suitable for wounds that need to withstand external forces. Although P-PAA had a lower strength, its flexibility and energy-dissipation ability provided by reversible hydrogen bonds give it unique advantages in dynamic wound environments. Tissue adhesion experiments confirmed that the abundant carboxyl groups in both hydrogels could form multiple hydrogen bonds with amino groups and other active groups on the skin surface. The hydrogels remained tightly attached to porcine skin under bending and twisting conditions without detachment, demonstrating their reliable wound-sealing ability. In terms of biosafety evaluation, both the CCK-8 cell proliferation assay and live/dead fluorescence staining consistently showed that the extracts of P-PAA and C-PAA hydrogels exhibited no cytotoxicity toward mouse fibroblasts (NIH-3T3) and significantly promoted cell proliferation. This was attributed to the high-water content of the

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materials and the protein-adsorption microenvironment provided by their carboxyl groups, which were conducive to cell adhesion and growth. In a full-thickness skin incision injury model in rats, compared with the traditional surgical suture group, the PAA hydrogel dressing group showed significantly higher wound-healing rates on postoperative days 3, 7, and 14. Histological analysis using hematoxylin and eosin (H&E) staining and Masson's trichrome staining showed that inflammatory cell infiltration was significantly reduced in the hydrogel groups, while blood vessel density increased and fibroblast proliferation was active. Collagen fibers were more orderly arranged, and the ratio of type I to type III collagen was closer to that of normal skin, effectively reducing postoperative scar formation. Among them, the P-PAA hydrogel, based on its reversible hydrogen-bonding network, combines shear-thinning injectability, self-healing ability, and controllable painless detachment. During dressing changes, gentle stimulation can weaken the interfacial adhesion without damaging newly formed granulation tissue, thereby avoiding secondary injury caused by the removal of traditional gauze or sutures. In contrast, the C-PAA hydrogel exhibits greater mechanical strength and adhesion due to its covalent network structure and is suitable for surgical incisions or lacerations that are subjected to greater tension. The two types of materials complement each other and cover a variety of application scenarios, including home first aid, wilderness wound care, and clinical surgical healing. They are convenient to use and low in cost, effectively addressing the performance limitations of existing medical adhesives, such as the insufficient adhesion of fibrin glue and the toxicity and non-biodegradability of cyanoacrylate adhesives. These materials therefore demonstrate good potential for clinical translation and practical value.

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