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Low Impedance, Durable, Self-Adhesive Hydrogel Epidermal Electrodes for Electrophysiology Recording

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by

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Acknowledgments

My journey at WashU has been unexpectedly delightful and full of surprises. Initially, after my first application was rejected, I thought that I would never be connected to this place. It was beyond my imagination that, six months later, I applied again and ecstatically received the offer letter. However, things did not go as smoothly as I had anticipated at the beginning. Transitioning to a different major from my undergraduate studies made my coursework particularly challenging. Starting in the spring semester also made it difficult to find research opportunities. Nevertheless, the support from those around me was immense, gradually helping to steer my life back on track.

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ABSTRACT OF THE THESIS

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by

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Master of Science in Electrical Engineering

Washington University in St. Louis, 2024

Professor Chuan Wang, Chair

Traditional electrodes used for electrophysiology recording, characterized by their hard, dry, and inanimate nature, are fundamentally mismatched with the soft, moist, and bioactive characteristics of biological tissues, leading to suboptimal skin-electrode interfaces. Hydrogel materials, mirroring the high water content and biocompatibility of biological tissues, emerge as promising candidates for epidermal electronic materials due to their adjustable physicochemical properties. However, challenges such as inadequate electrical conductivity, elevated skin impedance, unreliable adhesion in moist conditions, and performance decline from dehydration have significantly restricted the efficacy and applicability of hydrogel-based electrodes. In this thesis, we report a high-performance hydrogel epidermal electrode patch for electrophysiology signal recording and mechanical stimuli sensing applications. By integrating a polar solvent, PEDOT particles are clustered and realigned into extended chains to enhance charge transfer. KCl salt ions, serving as conductive agents, merge with the polyacrylamide crosslinked network to establish an interpenetrating polymer network (IPN), achieving superior electrical conductivity and reduced skin impedance. The hydrogel epidermal electrodes, fortified with catechol-rich tannic acid and an innovative suction cup design inspired by octopus tentacles, exhibit exceptional skin adherence

across diverse conditions. The incorporation of glycerol and a sealed hydrogel layer with the same mechanical properties as the hydrogel skin electrodes further solidifies the stability and functionality of the electrodes. This hydrogel epidermal electrode patch holds promising potential for capturing human electrophysiological signals, enabling high signal-to-noise ratio, high fidelity, durable and stable recordings in electrocardiograms and electromyograms without the need for external fixatives, and maintaining functionality in humid and submerged environments. In addition, this study provides insight into the potential of hydrogel epidermal electrodes as mechanical sensors, expanding their range of applications

Chapter 1: Introduction

1.1 Background

Over the past hundred years, interdisciplinary collaboration in modern medicine, biology, and biomedical engineering has led to significant breakthroughs in both the understanding and the engineering of the human body. In parallel, advancements in materials processing and wireless sensing technologies have been pivotal in the miniaturization of devices, paving the way for the widespread use of wearable technology[1-7]. The advantages of wearable devices like portability, outstanding flexibility, and cost-effectiveness are vital in enhancing tracking, health monitoring, and the functionality of clinical medical devices[8-11]. Among those advances, epidermal electronics that facilitates sensing and recording of a variety of physical, chemical signals and vital signs from human skin are of utmost importance and has already found ways into many clinically relevant applications[12, 13]. One of the most significant challenges associated with the development of epidermal electronics is that the rigid, dry and non-biological nature of traditional electronic materials such as metal, semiconductor, glass, ceramics and plastics is in essential contradiction with the soft, moist and active nature of biological tissue which often results in devices that are difficult to fit and uncomfortable to wear[14].

In recent years, stretchable electronic materials with significantly improved flexibility and stretchability have been fabricated by combining conductive materials with excellent electrical and mechanical properties (e.g., carbon nanotubes[15, 16], graphene[17], metal particles[18, 19], and conductive polymers[1, 20]) with elastomeric composite substrates. However, challenges persist due to the hard-to-bridge gaps and unstable adhesion between these conductive elastomers and tissues, coupled with high contact impedance. These problems are particularly acute in the real-time monitoring of

electrophysiological signals, leading to significant noise and motion artifacts, which complicate the signal analysis.

Hydrogel is a three-dimensional elastic cross-linked hydrated polymer network that resembles biological tissues with high water content[21, 22] and biocompatibility[23], making it a very promising material for epidermal electronics (Figure 1.1). The mechanical properties of hydrogels (toughness, stretchability and flowability) can be modulated during the synthesis process, and conductivity can be introduced by combining polymer networks with conductive fillers, or compositing hydrophilic matrix with ionic pendant groups or salts[8, 24, 25]. Crucially, the network composition, electrochemical characteristics, and biological functionalities of conductive hydrogels can be extensively and precisely adjusted. By varying elements such as the conductive filler, dopant, cross-linking agents, or hydration level, hydrogels can be engineered to exhibit distinctive properties and multifunctional capabilities, making them versatile in numerous promising sensing applications[26, 27].

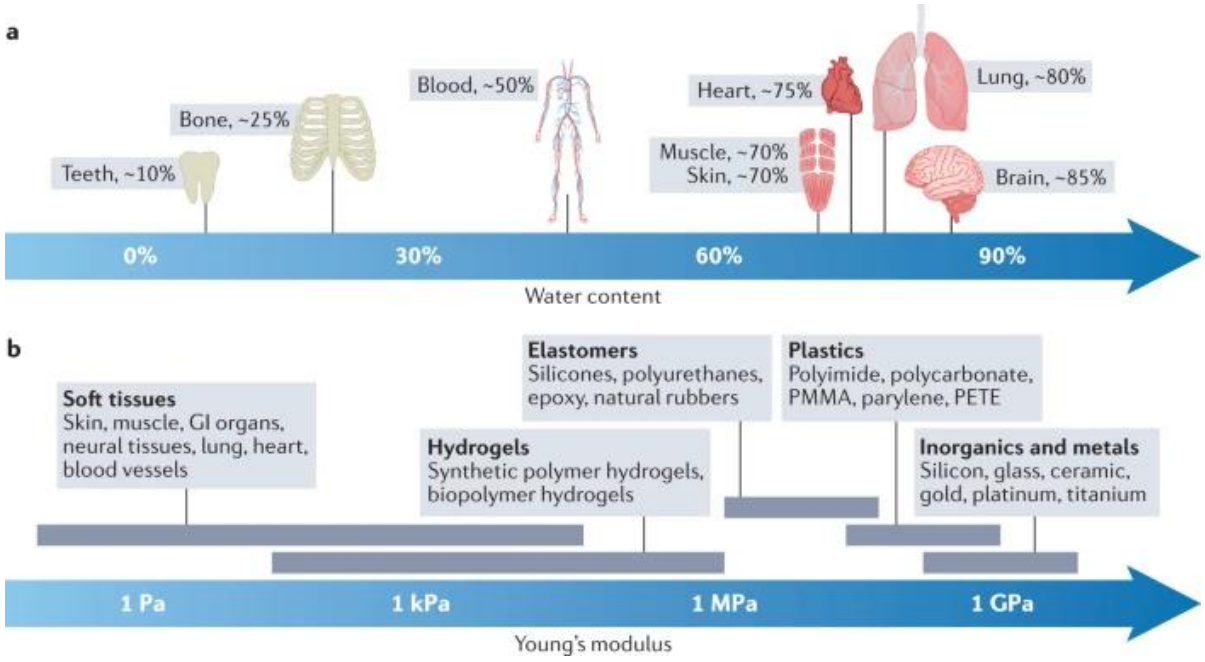

Figure 1.1 Similarities between hydrogels and biological tissues[14].

1.2 Basic Theory of Hydrogel

Hydrogel research was initiated in Germany during the mid-1930s, focusing on the dynamics of crosslinked polymers. In the 1940s, Paul Flory significantly advanced this field by laying the foundational understanding of hydrogels, including their crosslinked structures, swelling and hysteresis characteristics, and size deformation behaviors in pure water as well as in physiological fluids. Since then, hydrogel research has progressed swiftly, yielding a multitude of novel cross-linked structures and advancements in synthetic techniques[28].

1.2.1 Cross-linking Mechanisms of Hydrogels

For hydrogel formation, two essential conditions must be met. First, the polymer molecules, whether synthetic or natural, need to possess hydrophilic groups along their main or side chains. Second, there must be sufficient cross-linking strength among these molecules to establish a polymer network structure[29]. These polymer networks can be constructed through various bonding mechanisms, such as (1) proto-covalent cross-linking, (2) ionic forces, (3) hydrogen bonding, (4) affinity or "biorecognition" interactions, (5) hydrophobic interactions, (6) polymer crystallization, (7) physical entanglement of polymer chains, or (8) a combination of two or more of these types of interactions. This range of bonding options contributes to hydrogels' diverse properties and a broad spectrum of applications[28].

According to the cross-linking mechanism of gelling, hydrogels can be classified into four categories: physical hydrogels, chemical hydrogels, enzymatic hydrogels, and multiple cross-linking hydrogels[30, 31] (Table 1.1).

Classification basis	Hydrogel types	Classification basis	Hydrogel types	Classification basis	Hydrogel types
Polymer source	1. Natural hydrogels 2. Synthetic hydrogels 3. Hybrid hydrogel systems	**Configuration**	1. Amorphous (non-crystalline) hydrogels 2. Semicrystalline hydrogels 3. Crystalline hydrogels	**Anisotropy**	1. Homogeneous/random hydrogels 2. Oriented structural hydrogels
Polymeric composition	1. Homopolymeric hydrogels 2. Copolymeric hydrogels 3. Multipolymer hydrogels: a. *Interpenetrating network (IPN) hydrogels* b. *Double network (DN) hydrogels*	**Physical appearance and size**	1. Macroscopic hydrogels (matrix, film, etc., typically on the order of millimeters to centimeters) 2. Microgels (microspheres, 100 nm–10 μm) 3. Nanogels (nanoparticles, 10–100 nm)	**Electrical charge**	1. Nonionic (neutral) 2. Ionic (anionic or cationic) 3. Amphoteric electrolyte (ampholytic) species containing both acidic and basic groups 4. Zwitterionic (polybetaines) containing both anionic and cationic groups in each structural repeating unit
Crosslinking mechanisms	1. Physical crosslinking (noncovalent interactions): a. *electrostatic interactions* b. *hydrogen bonds* c. *crystallization* d. *metal-ligand coordination* e. *stereocomplex crystallization* f. *hydrophobic interactions* g. *conformation transformation* h. *host-guest interactions* i. *molecular-specific binding* j. *π-π stacking* 2. Chemical crosslinking (covalent junctions) a. *Via monomers* b. *Via polymers* 3. Enzymatic crosslinking (covalent junctions) 4. Multi-crosslinking	**Functions**	1. Smart hydrogels 2. Self-healing/self-recovery hydrogels 3. Injectable hydrogels 4. Strong adhesive hydrogels 5. High strength hydrogels 6. Superabsorbent hydrogels 7. Bionic hydrogels	**Stimuli response**	1. Conventional hydrogels (non-stimuli responsive) 2. Responsive (smart) hydrogels: a. *Physical stimuli: temperature, magnetic field, light, electric field, pressure, and sound* b. *Chemical stimuli: pH, solvent composition, ionic strength, and molecular species*

Table 1.1 Similarities between hydrogels and biological tissues[29]

Physical hydrogels are formed by non-covalent interactions (intermolecular interactions) such as (a) electrostatic interactions; (b) hydrogen bonding; (c) crystallization; (d) metal-ligand coordination; (e) stereocomplex crystallization; (f) hydrophobic interactions; (g) conformational transitions; (h) host-guest interactions; (i) molecular specific binding; and (j) π-π stacking[31, 32] (Figure 1.2). Physical gels can be further categorized into strong physical hydrogels (e.g., containing lamellar microcrystals, glassy nodules, or double and triple helices, etc.) and weak hydrogels (e.g., formed by hydrogen bonding, block copolymer micelles, hydrophobic interactions, and ionic bonding, etc.), depending on whether or not the physical bonds are permanent[33, 34].

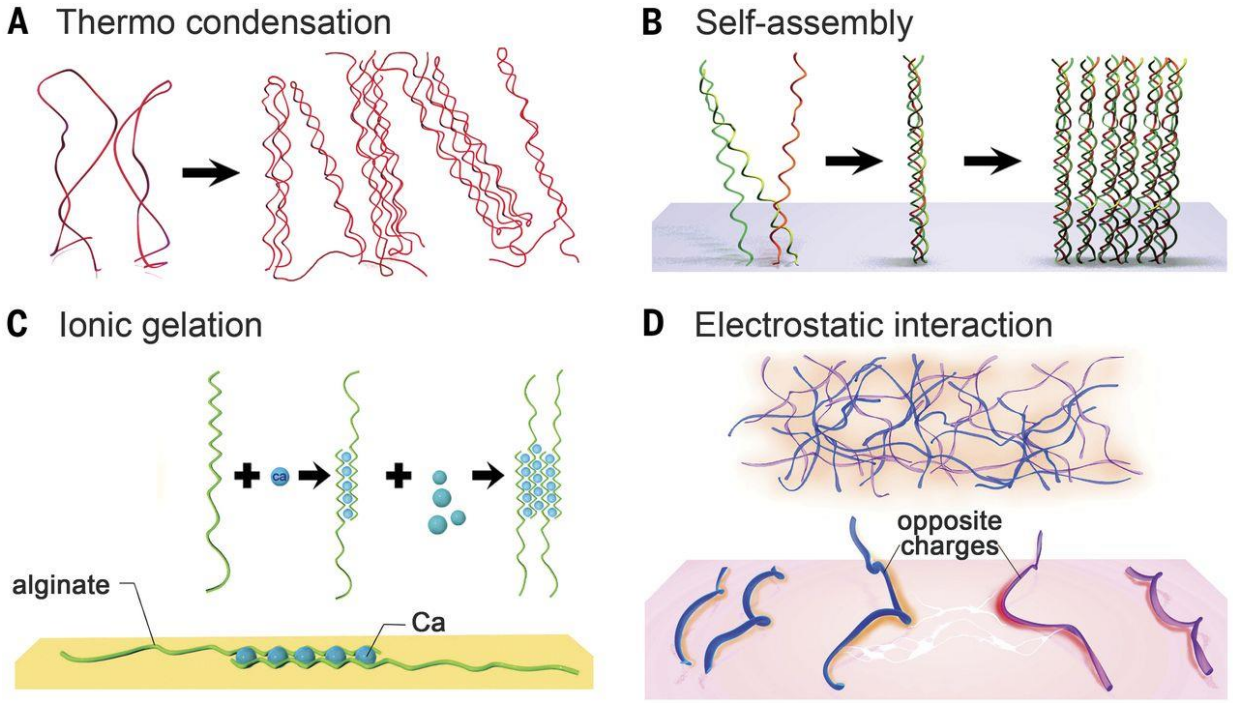

Figure 1.2 Different types of Physical cross-linking[31].

Chemically crosslinked hydrogels (also called true gels) are fabricated via the formation of covalent junctions between two polymer molecules so that the formed hydrogels are generally permanent, non-reversible, and stable. In the most succinct sense, a hydrogel is simply a hydrophilic polymeric network crosslinked in some fashion to produce an elastic structure. Thus,

any technique that can be used to create a crosslinked polymer can be used to produce a hydrogel[35]. There are two main types of covalent junction approaches used to prepare chemically crosslinked hydrogels: pathway via monomers and pathway via polymers[29].

Enzyme-catalyzed cross-linking reactions stand out because the enzymes involved don't integrate into the resulting molecular structure, a stark contrast to conventional cross-linking agents. Key enzymes in this domain include horseradish peroxidase (HRP), laccase, transglutaminase (TG), and tyrosinase. For example, HRP facilitates the breakdown of hydrogen peroxide, leveraging it as an oxidizing agent to promote the linkage of various phenolic and aniline compounds without becoming part of the structure itself. This method, especially when HRP is used, is often applied to create hydrogels in place using natural polymers like hyaluronic acid, dextran, gelatin, and poly(aspartic acid), along with poly(γ-glutamic acid) (γ-PGA), silk fibroin, and chitosan, due to their biocompatibility and functional properties[36-38].

The combinatorial crosslinking approach is frequently employed to create intricate hydrogels. Andrij Pich and Rienk Eelkema have conducted an in-depth analysis comparing macromolecular hydrogels (MHGs) and supramolecular hydrogels (SHGs), examining aspects such as the gelling mechanism, structure, functionality, and adaptability. They concluded that both MHGs and SHGs have their unique advantages, suggesting that combining the strengths of both types offers a promising strategy for developing adaptive, life-like soft hydrogel materials[29, 39].

1.2.2 Synthesis of Hydrogels

According to the traditional classification, hydrogels are synthesized by four main methods: (1) homopolymer hydrogels, which are networks formed from a single type of hydrophilic monomer; (2) copolymer hydrogels, created by intertwining chains of two different monomers, with at least one being hydrophilic to ensure water absorbency; (3) multipolymer hydrogels, which come into

being when three or more different monomers are combined; and (4) interpenetrating network (IPN) hydrogels, which can be crafted through two primary approaches — either within an existing network or from a solution, commonly by initiating the polymerization of a monomer within an already cross-linked hydrogel network, resulting in a secondary network that intertwines with the original[40, 41] (Figure 1.3).

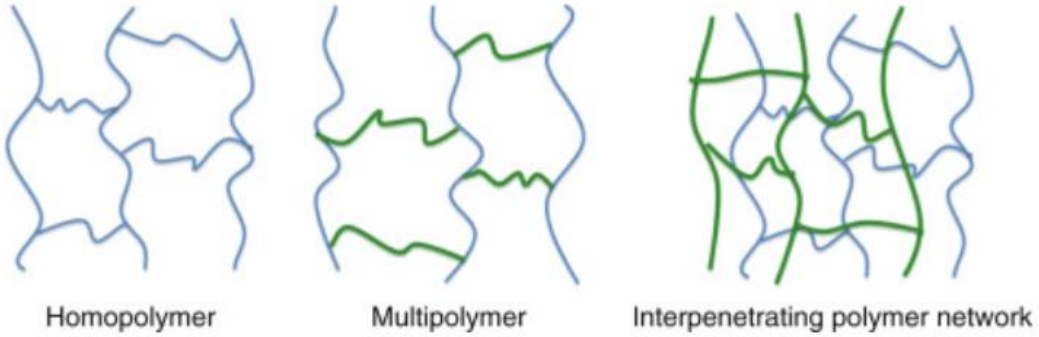

Figure 1.3 Illustration of three main types of hydrogels[41].

A broad range of vinyl monomers are suitable for crafting hydrogels, allowing for the creation of hydrogels using single or multiple monomer types. This diversity of monomers helps to customize hydrogels with specific physical properties to meet the needs of a particular application. Typically, a modest amount of a cross-linking agent is incorporated into hydrogel formulations. The polymerization process is generally triggered by radiation, ultraviolet light, or chemical catalysts, with the choice of initiator being dependent on the monomers and solvents utilized. The resulting polymerized hydrogel can take various forms, such as films, membranes, rods, particles, and emulsions. The process of polymer polymerization, which is relatively straightforward, involves the use of monomers and initiators that dissolve in the monomer. This method is characterized by rapid polymerization and a high degree of polymer formation due to the concentrated presence of the monomer[42]. However, as the reaction progresses, a significant increase in viscosity can occur, leading to heat generation. These issues can be mitigated by maintaining the reaction at lower

levels of conversion. Bulk polymerization of monomers yields a uniform hydrogel that forms a hard, glass-like polymer matrix. Upon exposure to water, this rigid structure softens and becomes pliable[35, 43].

Highly hydrated materials are often poorly resilient with low strength. IPN hydrogels offer ideas for overcoming the conflict between strength, resilience, and high water content (>90 wt%). The IPN hydrogels consist of entwined structures where multiple polymers weave together to form a network. At least one of the polymers in the IPN system is cross-linked, while the others remain non-covalently bonded to the main network[44, 45]. According to the IUPAC, an IPN is defined as "A polymer containing two or more networks that are molecularly entangled but not covalently linked to one another. They cannot be physically separated without breaking chemical bonds. This structure is distinct from a simple polymer blend because it is not merely a mixture of pre-existing polymer networks." This definition highlights the key distinction between IPNs and polymer blends, emphasizing the unique, intertwined nature of IPNs[46, 47]. IPN structures can be classified into semi-IPNs (SIPNs) and full-IPNs (FIPNs)[48]. SIPNs are composed of one linear polymer entrapped within the network of another crosslinked polymer, while FIPNs are composed of one crosslinked polymer interpenetrated within another crosslinked polymer network (Figure 1.4). Intermolecular entanglement and interpenetration can cause forced miscibility between components, which enables IPNs to integrate the excellent performance of the different participating components[49].

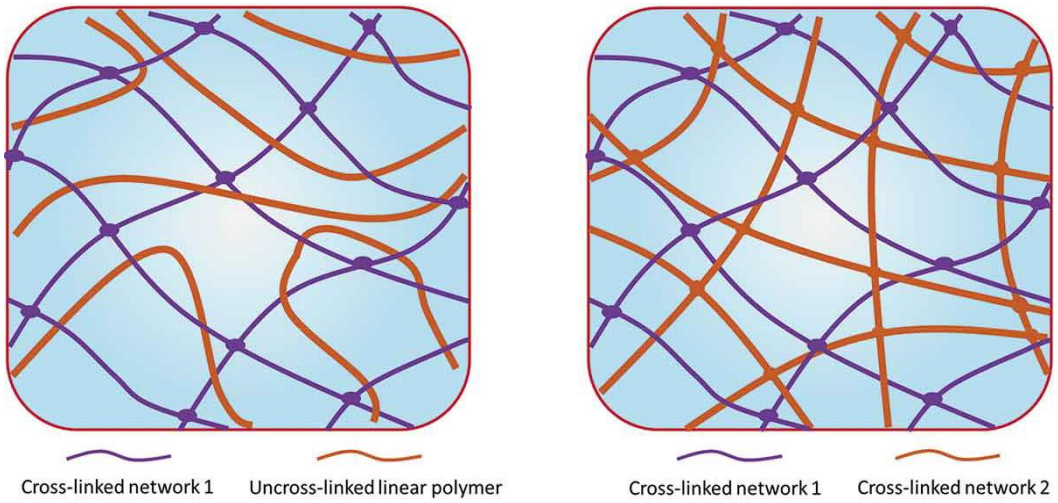

Figure 1.4 Schematic illustrations of SIPN and FIPN[47].

Excitation of photosensitive functional groups by UV radiation is one of the most common methods for synthesizing hydrogels. This method has significant advantages such as easy handling, fast preparation, and low production cost[50]. Ono et al. reported UV-light-irradiated chitosan hydrogels by introducing azide and lactose as light-sensitive moieties. The azide group converted into a nitrene group after UV irradiation, which bound to amino groups of chitosan to form a hydrogel in a short time[51]. Chitosan hydrogels were also created using UV irradiation, by first modifying chitosan and Pluronic acid with photosensitive acrylates[52]. Johannes et al. introduced the first example of UV transient hydrogels (Figure 1.5), which are covalently crosslinked gels designed to have a predefined lifespan under continuous UV light exposure. This unique property results from a process that simultaneously involves photopolymerization and photodegradation within poly(PEGMA-co-PEGDMA) hydrogels. These hydrogels undergo a self-driven transition from a precursor liquid state to a covalently crosslinked gel state, and eventually back to a liquid state, all initiated by a single UV light source. The ability to precisely control the reaction through spatial manipulation of the UV light enables the creation of two distinct hydrogel patterns from the same initial materials. The durability of these UV transient hydrogels can be finely tuned by

adjusting the intensity of the UV light and the concentration of the crosslinker (PEGDMA), allowing for meticulous control over their macroscopic characteristics within minutes[53].

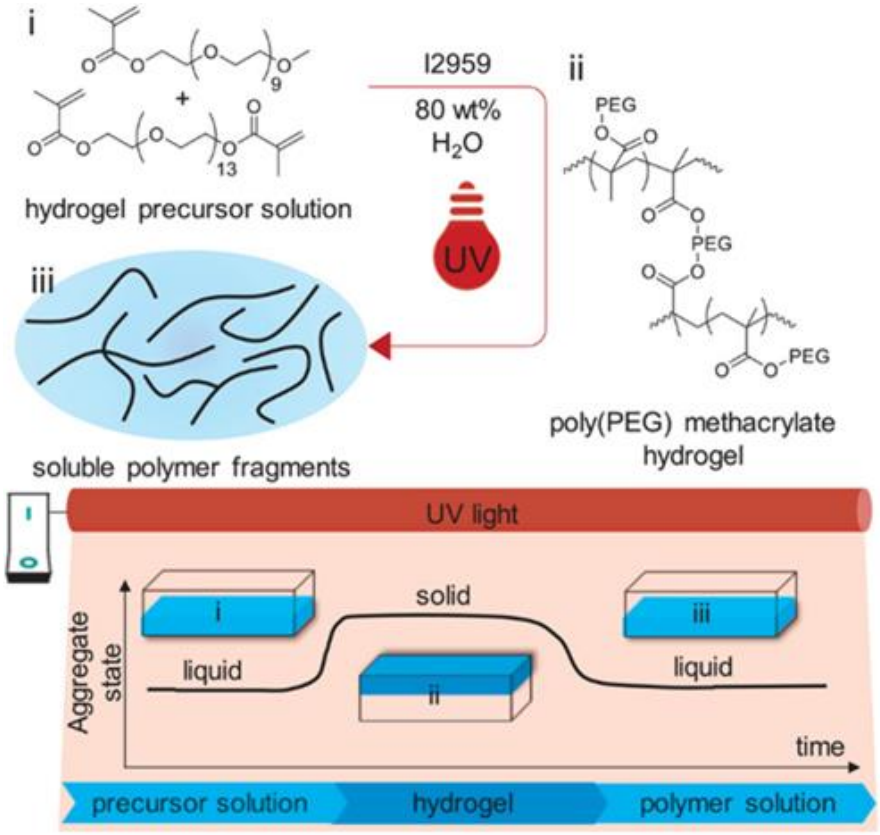

Figure 1.5 Reaction equation and schematic showing the UV-induced preprogrammed formation and degradation of hydrogel[53].

Another prominent method for hydrogel synthesis involves the initiation of free radical polymerization by cross-linking agents. The process begins with the use of an initiator like potassium persulfate (KPS), ammonium persulfate (APS), ceric ammonium nitrate, ferrous ammonium sulfate, 2-2'-azobisisobutyronitrile (AIBN), or benzoyl peroxide to kickstart the reactions. In this setup, vinyl monomers undergo radical polymerization in the presence of a crosslinker, leading to the formation of chemically crosslinked hydrogels. These vinyl monomers can vary widely and include substances such as acrylic acid, acrylamide, vinyl chloride, styrene, epoxide, N-vinyl-2-pyrrolidone, and 2-hydroxyethyl methacrylate. The crosslinking agents employed in this free radical process often encompass N, N'-methylenebisacrylamide (MBA),

ethylene glycol dimethacrylate (EGDMA), along melamine-based crosslinkers. Through this method, both crosslinked homopolymers like poly (2-hydroxyethyl methacrylate) (PHEMA) and polyvinylpyrrolidone (PVP), as well as copolymers combining N-vinyl-2-pyrrolidone and 2-hydroxyethyl methacrylate (HEMA), are produced, utilizing melamine-based crosslinkers and AIBN as the initiator.

1.3 Applications of Hydrogel

Hydrogels are pivotal in biomedical fields, attributed to their exceptional characteristics like biodegradability, compatibility with biological systems, affinity for water, high absorption capacity, viscoelastic nature, and soft and plush texture. Moreover, hydrogels are responsive to a range of stimuli including temperature changes, electric and magnetic fields, specific biomolecules, and variations in ionic strength[49]. Their adhesive and bioadhesive qualities further enhance their utility by extending the duration drugs remain effective within the body, positioning hydrogels as ideal materials for drug delivery systems[54].

Conventional energy storage and conversion systems are often limited by their bulkiness, stiffness, and lack of eco-friendliness. Moreover, the risk of leakage in these devices is heightened by the use of liquid electrolytes, which are not only costly but also potentially hazardous due to their organic composition[55, 56]. To address these concerns, hydrogel electrolytes have been introduced, offering a semi-solid, biocompatible, biodegradable, affordable, and eco-friendly alternative[57]. These attributes, coupled with their inherent flexibility, make hydrogel electrolytes particularly valuable for energy storage technologies like supercapacitors. Devices that incorporate hydrogels are characterized by their flexibility, stretchability, and resilience, maintaining functionality even when subjected to stressors such as stretching, bending, folding, and twisting, giving them a significant advantage over traditional electrochemical systems. Additionally, the self-healing

properties of hydrogels play a pivotal role in enhancing the durability and lifespan of wearable and portable electronics, making them ideally suited for the development of smart, lightweight electronic devices[58, 59].

1.3.1 Hydrogel as Coupling Agent

Due to its tissue-like mechanical and acoustic properties, hydrogels are widely used as a coupling agent at the tissue-ultrasound probe interface in ultrasound imaging[60]. Adrian et al. investigated a hydrogel acoustic coupling medium as a practical alternative to water for the clinical application of focused ultrasound (US) therapy. Conical couplers were designed and fabricated to fit a 3.5 MHz, spherically concave transducer for functional tests, including Schlieren imaging, power efficiency measurements and in vivo hemostasis experiments. Polyacrylamide was shown to have favorable acoustic properties that varied linearly with acrylamide concentration from 10% to 20% weight in volume. Attenuation coefficient, sound speed and impedance ranged from 0.08 to 0.14 dB/cm at 1 MHz, 1546 to 1595 m/s and 1.58 to 1.68 Mrayl, respectively. Intraoperative in vivo hemostasis experiments in a sheep model demonstrated that a gel-coupled transducer was able to induce hemostasis of hemorrhages in splenic and hepatic incisions (Figure 1.6), suggesting that polyacrylamide may be a promising coupling material for focused ultrasound therapy[61].

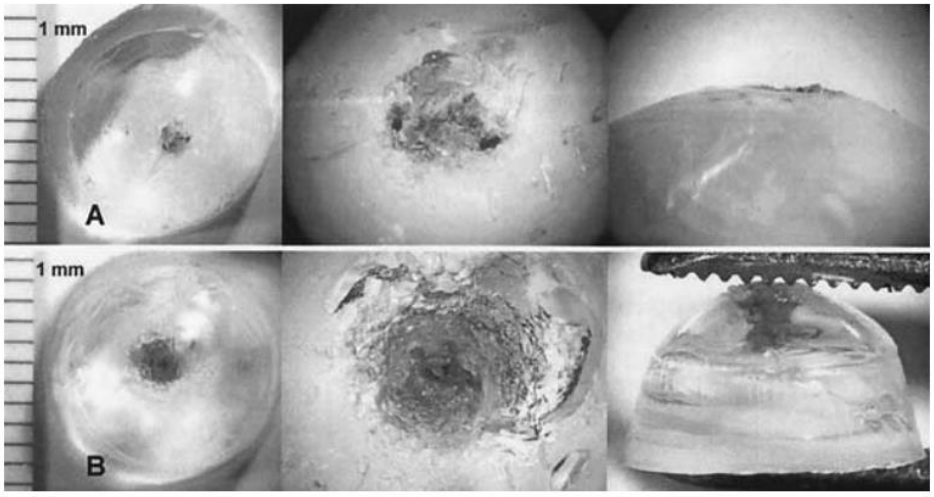

Figure 1.6 Gel cone tips after in vivo hemostasis experiment[61].

1.3.2 Hydrogel for Wound Healing

Hydrogels are extensively employed in the treatment of skin injuries as wound dressings or epidermal bandages, thanks to their distinctive capacity to enhance wound healing. They achieve this by preventing dehydration, offering antimicrobial protection, and facilitating transdermal drug delivery, thereby playing a pivotal role in skin recovery[62-65].

Jiang et al. developed a flexible bioelectronic system consisting of wirelessly powered closed-loop sensing and stimulation circuits with skin-interface hydrogel electrodes that can be adhered to and detached on demand (Figure 1.7). In mice, the wound care system continuously monitors skin impedance and temperature and delivers electrical stimulation based on the wound environment. Across preclinical wound models, the treatment group healed ~25% more rapidly and with ~50% enhancement in dermal remodeling compared with control[66].

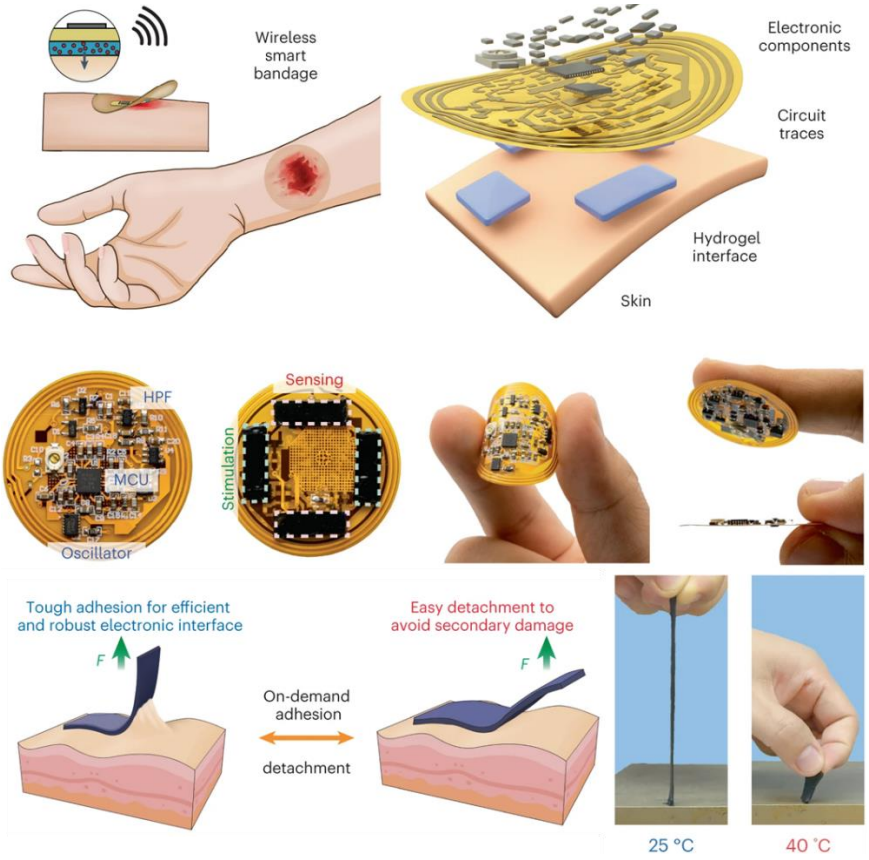

Figure 1.7 Wireless smart bandages with on-demand tissue-adhesion for chronic wound management[66].

Mirani et al. presented an advanced multifunctional dressing (GelDerm) capable of colorimetric measurement of pH, an indicator of bacterial infection, and the release of antibiotic agents at the wound site (Figure 1.8). This study demonstrates the ability of GelDerm to detect bacterial infections using in vitro and ex vivo tests with accuracies comparable to the commercially available systems. Wireless interfaces to digital image capture hardware such as smartphones serve as a means for quantitation and enable the patient to record the wound condition at home and relay the information to the healthcare personnel for following treatment strategies. In addition, the dressing integrates with commercially available patches is applied to the wound without chemical or physical irritation, and provides a sustained release of antibiotics to destroy bacteria[62].

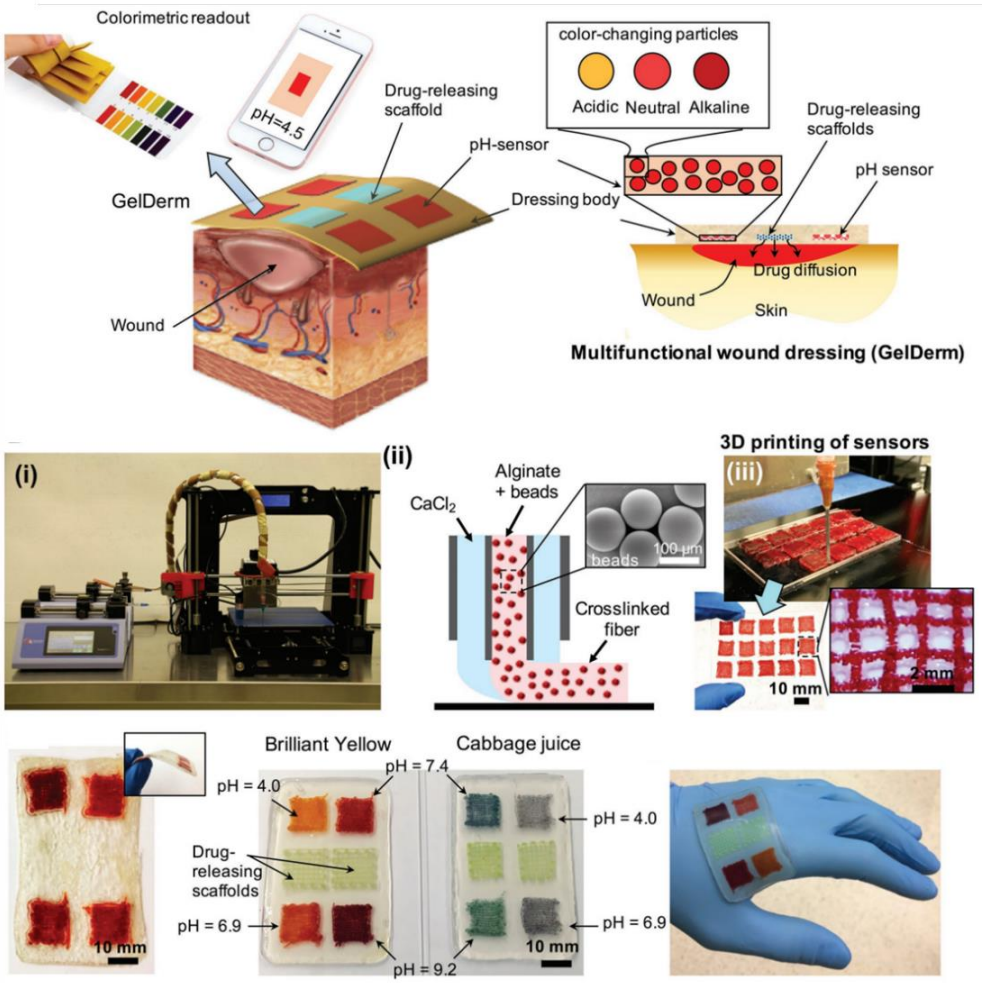

Figure 1.8 An advanced multifunctional dressing (GelDerm) for monitoring and management of wounds[62].

1.3.3 Hydrogels for Epidermal Health

Hydrogels are also widely used in many epidermal health monitoring and diagnostic applications[67]. For example, the high water absorption of various chemical sensing hydrogels allows them to be used as diagnostic devices based on body fluids (e.g., sweat) collected from the epidermis[9].

Koh et al. presented a collection of materials and device designs for soft, flexible, and stretchable microfluidic systems, including embodiments that integrate wireless communication electronics, which can intimately and robustly bond to the surface of the skin without chemical and mechanical irritation (Figure 1.9). The system is connected to several sweat glands, automatically directs sweat into a network of tiny channels and storage areas, and uses color changes to detect substances such as chloride, hydrogen ions, glucose and lactate in the sweat. Tests on people showed that this device works well both in controlled indoor cycling sessions and in outdoor long-distance bike races, even in dry conditions. The results include quantitative values for sweat rate, total sweat loss, pH, and concentration of chloride and lactate[68].

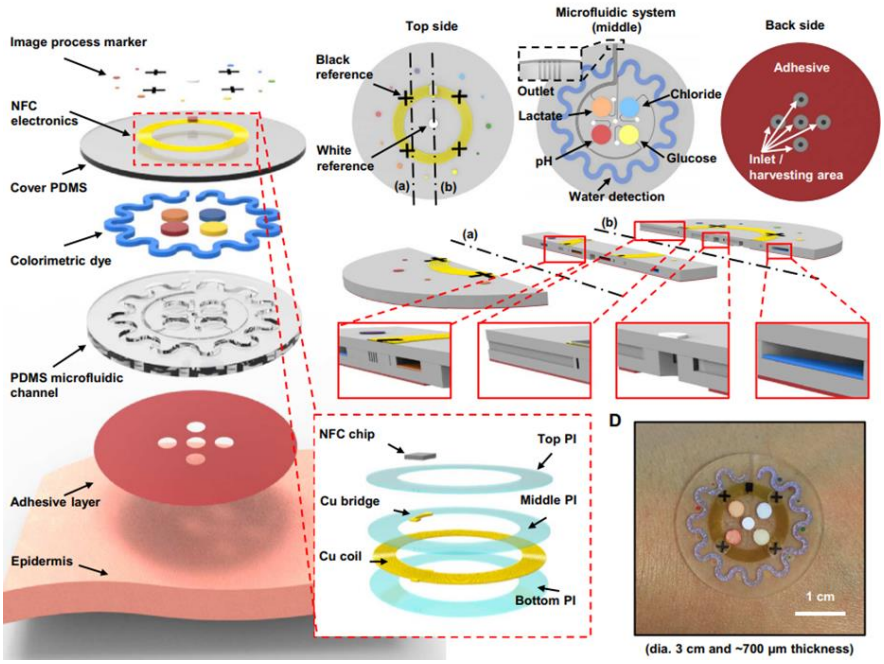

Figure 1.9 Schematic illustrations and optical images of an epidermal microfluidic biosensor integrated with flexible electronics for sweat monitoring[68].

Chung et al. developed a wireless, non-invasive technology that not only offers measurement equivalency to existing clinical standards for heart rate, respiration rate, temperature and blood oxygenation but also provides a range of important additional features, as supported by data from pilot clinical studies in both the neonatal and pediatric intensive-care units (Figure 1.10). These innovative methods include monitoring body movement and orientation, measuring the health benefits of direct skin contact, detecting unique sound patterns of heart activity, identifying vocal indicators related to the pitch and timing of crying, and tracking an alternative measurement method for systolic blood pressure, significantly improving the level of care in neonatal and pediatric intensive care settings[69].

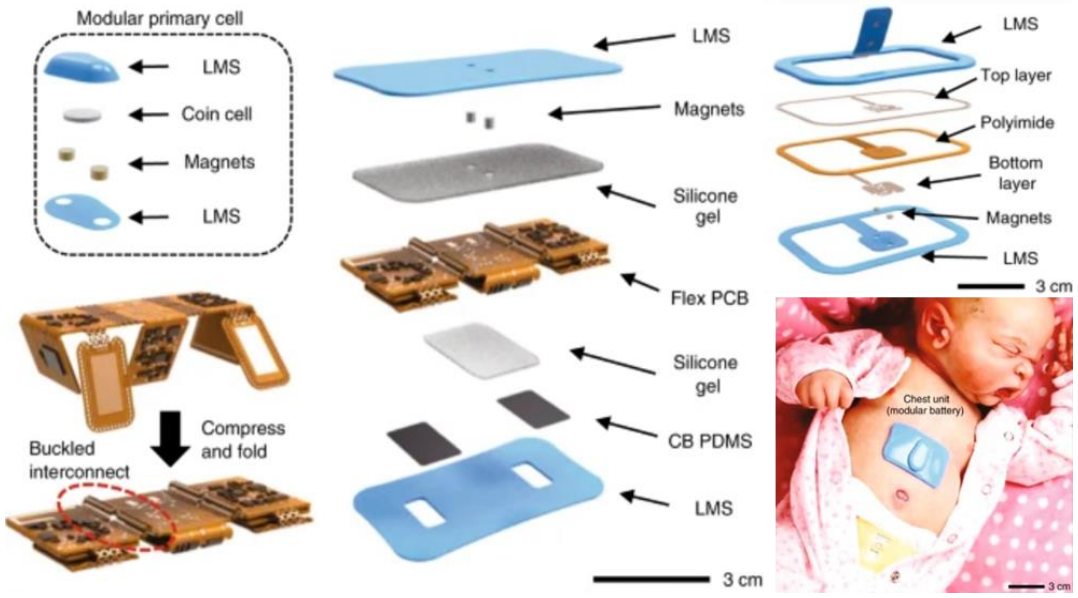

Figure 1.10 Design and characterization of a soft, wireless chest unit for physiological monitoring of neonatal and pediatric patients[69].

1.3.4 Hydrogels for Epidermal Electrodes

Epidermal electrodes are an essential component for various electrical sensing and stimulation devices in health monitoring, diagnostic, therapeutic and human-machine interfacing

applications[14]. Electrodes communicate with biological tissues in a bidirectional manner. In one way, electrodes stimulate excitable cells (e.g., neurons) by delivering electrical inputs to the tissues. In another way, electrical signals from electrically active cells propagate toward and are recorded by the electrodes. During this bidirectional communication, electrical signals are transmitted via ionic currents and electric potentials in the electrolytic tissue media and electronic currents and electric potentials in the conducting electrodes[25, 70, 71].

Kim et al. presented a strategy for fabricating printable and highly stretchable conductors by using a water-soluble tape to transfer printed silver ink onto a stretchable substrate consisting of an Ecoflex elastomer and a tough hydrogel layer (Figure 1.11). Because the thickness of the Ecoflex elastomer film coated on the hydrogel is very thin (30 microns), the elastic modulus of the hybrid film produced is close to that of the hydrogel layer. In addition, the conductor fabricated on the hybrid film has a tensile strain of up to 1780%. The transfer method described is simpler compared to other techniques utilizing elastomer stamps or sacrificial layers, allowing the application of printable electronic components to substrates with a low elastic modulus (e.g., hydrogels)[72].

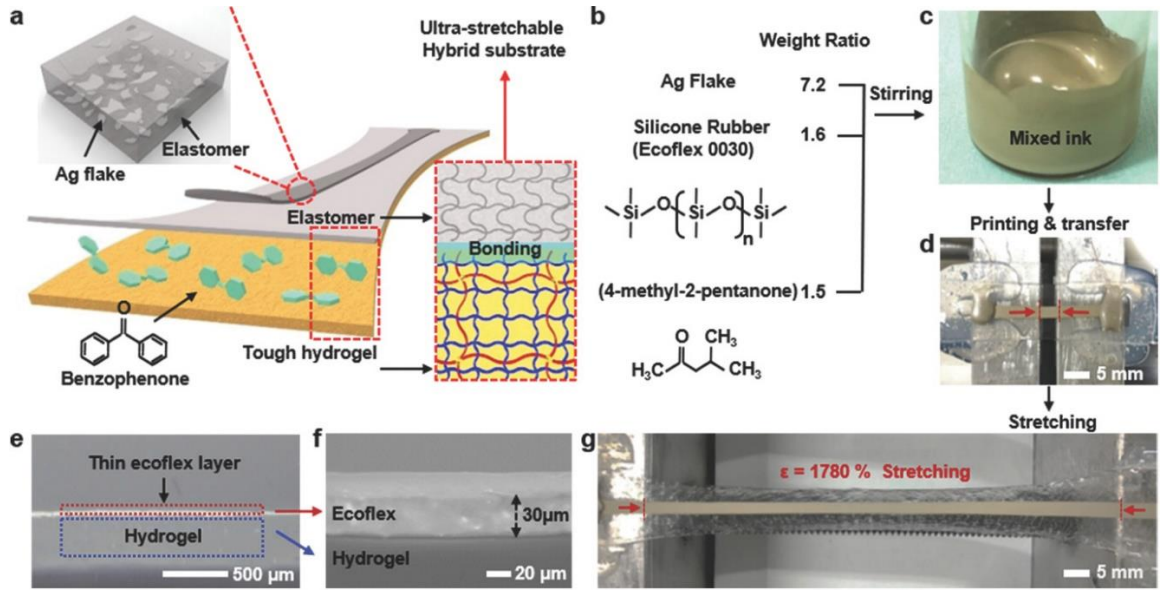

Figure 1.11 Fabrication of the stretchable conductor containing the hybrid substrate and conductive ink layer[72].

Pan et al. reported the synthesis of a scalable and multifunctional polyaniline (PAni) hydrogel with excellent electronic conductivity and electrochemical properties (Figure 1.12). The PAni hydrogel has a high specific surface area and a three-dimensional porous nanostructure, which can be used as a high-performance supercapacitor electrode with a high specific capacitance (~480 F g^{-1}), unprecedented rate capability, and cycling stability (~83% capacitance retention after 10 PAni hydrogels can also be used as active components of glucose oxidase sensors with fast response time (~0.3 s) and excellent sensitivity (~16.7 μA mM^{-1}). The scalable synthesis and excellent electrode properties of PAni hydrogels make them ideal candidates for use in bioelectronics and as future next-generation energy storage electrodes[73].

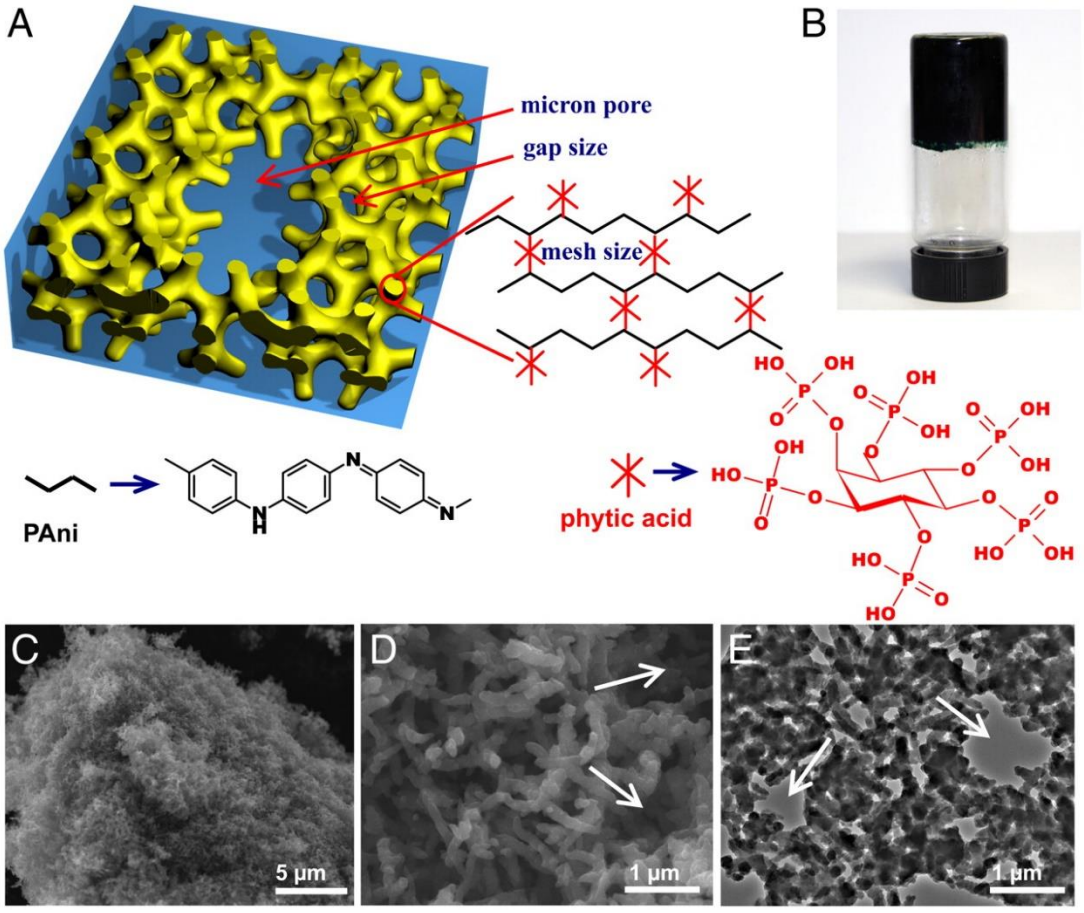

Figure 1.12 Chemical structure and morphology of phytic acid gelated and doped polyaniline hydrogel[73].

Lu et al. obtained high-performance pure PEDOT: PSS hydrogels by a simple method of designing interconnection networks of PEDOT: PSS nanofibrils (Figure 1.13). The method consists of mixing the volatile additive dimethyl sulfoxide (DMSO) into an aqueous solution of PEDOT: PSS, followed by controlled dry annealing and rehydration. The resulting hydrogels exhibit a range of properties required for bioelectronic applications, including high electrical conductivity (~20 S cm^{-1} in PBS and ~40 S cm^{-1} in deionized water), high tensile properties (> 35% strain), low Young's modulus (~2 MPa), excellent mechanical, electrical, and electrochemical stability, and tunable isotropic/anisotropic solubility in wet physiological environments[74].

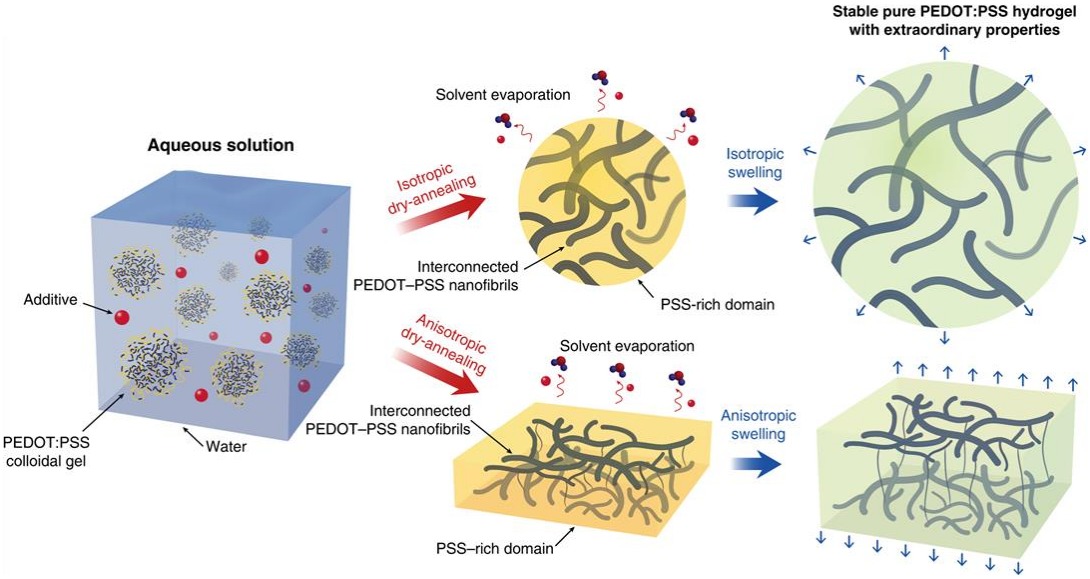

Figure 1.13 Dry annealing and swelling process of PEDOT: PSS with DMSO as additive[74].

1.4 Research Direction for This Thesis

Hydrogel interfaces have been introduced as coatings for metal electrodes and are widely used in various clinical diagnostic applications, such as electrocardiograms, electromyograms, and electroencephalograms, as well as in therapeutic applications like transcutaneous electrical nerve stimulation. These hydrogel epidermal electrodes, characterized by their softness and adhesiveness, maintain close contact with the skin, addressing the issue of poor interface contact between traditional dry and rigid metal electrodes and skin tissues. They are primarily categorized into ionic

hydrogel interfaces and polymer hydrogel interfaces. Most commercially available and clinically recognized epidermal electrodes are composed of ionic hydrogel interfaces, made of crosslinked hydrophilic polymers (such as polyacrylate copolymers) with high water content and conductive dissolved ions (like potassium chloride), providing low tissue-electrode impedance[22, 31, 75-78]. However, these hydrogels generally lack charge conductivity, and their ionic conductivity under physiological conditions is significantly lower (6-9 orders of magnitude lower than that of metals).

In contrast, conductive polymer hydrogels exhibit metal-like electronic conductivity, and their significantly enhanced conductivity makes them one of the most promising materials in the emerging field of hydrogel bioelectronics[77, 79]. In particular, hydrogels based on poly(3,4-ethylenedioxythiophene) polystyrene sulfonate (PEDOT: PSS) have attracted extensive research due to their excellent biocompatibility[80, 81]. Current methods for preparing PEDOT: PSS hydrogels primarily rely on mixing or in situ polymerization of PEDOT: PSS within a non-conductive hydrogel template to form an IPN[82-84]. However, this IPN-based conductive polymer hydrogel might increase tissue-electrode impedance, as the non-conductive hydrogel network acts as an electrical insulator.

Although conductive nanofillers like metal nanoparticles/wires, carbon nanotubes, and graphene have been added to IPN-based PEDOT: PSS hydrogels to enhance conductivity, the dispersion of nanofillers within the polymer chain in the hydrogel network (usually at the sub-nanometer level) may introduce potential issues, such as mechanical and electrical performance heterogeneity, as well as instability and cytotoxicity when in contact with moist biological tissues[84-87]. Moreover, the practical application of hydrogel epidermal electrodes is hindered by issues related to durability and reliability. For instance, sweat and environmental humidity can cause short circuits, while dehydration at high temperatures or even room temperature can lead to a decline in performance[88].

Polymer encapsulation or lamination has been validated as an effective approach for packaging the devices, but different mechanical properties lead to poor bonding between polymers and hydrogels, and most polymers have poor adhesion. Additionally, hydrogels tend to swell in water, which dilutes the network's polymer chains and causes the hydrogel to expand. This expansion reduces the adhesive contact points at the gel-substrate interface, leading to a loss of adhesive properties when immersed in water and rendering the tissue-electrode interface ineffective[89]. Thus, there's a pressing need to create hydrogel epidermal electrodes that maintain high conductivity, low tissue-electrode impedance, consistent adhesion, and stable properties.

This work successfully crafted a high-quality hydrogel epidermal electrode patch by meticulously selecting materials and designing crosslinking networks, further enhanced by integrating biomimetic structures. The addition of polar solvents enables the aggregation and reorientation of PEDOT grains into elongated chains, enhancing conductivity by facilitating charge hopping along these chains[20]. Salt ions, serving as conductive fillers, integrate with the polymer hydrogel to form an IPN structure, offering both high conductivity and low tissue-electrode impedance. Incorporating compounds rich in catechol groups enhances the adhesiveness of dopamine-like compounds, improving the adhesion between hydrogel epidermal electrodes and tissues[90]. The adhesion derived from chemical bonds, combined with biomimetic structures, ensures high levels of adhesion across various application environments (air, underwater e.g.)[91]. We also utilize an insulating hydrogel with identical cross-linking to the epidermal electrodes for a sealing layer, ensuring uniform mechanical properties and adhesion. In addition, adding glycerol enhances the water retention capability of hydrogel, ensuring that long-term storage does not compromise the performance[88]. Ultimately, we developed hydrogel epidermal electrode patches with high signal-

to-noise ratio, high fidelity, effective sealing, long-term stability, and versatility across a range of application scenarios.

Chapter 2: Synthesis and Characterization

2.1 Chemicals and Instruments

2.1.1 Chemicals

The names, specifications and manufacturer information of the chemicals required for the experiment are shown in Table 2.1.

Chemical	Specification	Manufacturer
2-Hydroxy-4'-(2-hydroxyethyl)-2-methylpropiophenone	Purity: 98%	SIGMA-ALDRICH
Poly(3,4-ethylene dioxythiophene)-poly(styrene sulfonate)	1.3 wt% dispersion in H_2O	SIGMA-ALDRICH
Potassium chloride	Purity: ≥99%	SIGMA-ALDRICH
Tannic acid	Analytical specification of USP	SIGMA-ALDRICH
Acrylamide	Purity: ≥99%	SIGMA-ALDRICH
N,N'-Methylenebisacrylamide	Purity: ≥99.5%	SIGMA-ALDRICH
Glycerol	Purity: ≥99.5%	SIGMA-ALDRICH
Sodium chloride	Purity: ≥99%	SIGMA-ALDRICH
PELCO conductive silver paint	Solid contents(silver): 73%	TED PELLA, INC.

Table 2.1 Chemicals, specifications, and manufacturers.

2.1.2 Instruments

The names of experimental instruments, models and manufacturer information are shown in Table 2.2.

Instrument	Model number	Manufacturer
Electromechanical universal testing machine	5583 Load Frame	Instron
Mixed signal oscilloscope	MSO 2004B	Tektronix
Four-channel digital storage oscilloscope	TDS 2014C	Tektronix
Semiconductor device analyzer	B1500A	KEYSIGHT
Syringe pump	LEGATO 110	KdScientific
High-intensity UV lamp	UVP B-100A	Analytikjena
Magnetic hotplate stirrers	SP88857100	Thermo Scientific
Analytical balance	ME54TE/00	METTLER TOLEDO
Water purification system	Direct-Q$^{@}$3	Millipore Sigma

Table 2.2 Instruments, model number, and manufacturers.

2.2 Device Design and Synthesis Process

2.2.1 Design of Hydrogel Epidermal Electrode Patches

Ethylene glycol is known to improve the electrical conductivity of PEDOT: PSS. The involvement of polar solvents facilitates the separation and reorientation of PEDOT and PSS grains, leading to

the aggregation of PEDOT particles into elongated chains. This structure aids in the facilitation of charge hopping along the chains, enhancing conductivity[20]. After polarization, an IPN is formed with PEDOT: PSS long chains and polyacrylamide crosslinking networks, providing the hydrogel with exceptional conductivity and mechanical properties. Furthermore, the inclusion of sodium chloride (KCl) as the ionic conductive filler in the IPN introduces ionic conductivity alongside charge conductivity. This combination substantially lowers the tissue-electrode interfacial impedance, thereby enhancing the quality of human electrophysiological signal recordings.

Tannic acid (TA) is incorporated into the hydrogel system as an adhesion enhancer, providing the hydrogel with self-adhesive capabilities. The abundance of catechol groups in TA imparts the hydrogel with robust dopamine-mimicking adhesion, enabling secure attachment to tissue surfaces. To preserve the adherence of hydrogel epidermal electrode patches in moist conditions, we designed a suction cup structure inspired by octopus tentacles. This innovative design merges the mechanical grip of the suction cups with the chemical bond provided by TA, significantly improving adhesion and broadening potential applications.

Pure polyacrylamide and TA hydrogels, devoid of PEDOT: PSS and KCl, exhibit mechanical properties and adhesion akin to hydrogel epidermal electrodes, coupled with effective insulating qualities, positioning them as the optimal choice for a sealing layer. The consistent crosslinked network ensures a strong bond between the working electrode and the sealing layer. Ultimately, to further boost durability and water retention, the hydrogel was immersed in glycerol, significantly enhancing its longevity and reliability.

2.2.2 Synthesis Process of Hydrogel Epidermal Electrode Patches

Synthesis of hydrogel epidermal electrode solution: Start by blending 3 ml PEDOT: PSS solution with an equal volume of ethylene glycol and 4 ml deionized water. This mixture is then heated to

60 ℃ and stirred for 20 minutes. Following this initial step, add the chemical components in sequence: 2.13 g acrylamide, 0.01 g N,N'-Methylenebisacrylamide, 0.02 g tannic acid, 0.1 gram of 2-Hydroxy-4'-(2-hydroxyethoxy)-2-methylpropiophenone, and 0.45 g KCl, ensuring each is well integrated before adding the next. The epidermal electrode hydrogel solution is obtained (Figure 2.1 a).

Synthesis of hydrogel sealing layer solution: Start by blending 7 ml deionized water with 3 ml ethylene glycol. The following chemical components are then added in order: 2.13 g acrylamide, 0.01 g N,N'-methylenebisacrylamide, 0.02 g tannic acid, and 0.1 g 2-hydroxy-4'-(2-hydroxyethoxy)-2-methylpropiophenone, ensuring each is well integrated before adding the next. The hydrogel sealing layer solution is obtained (Figure 2.1 b).

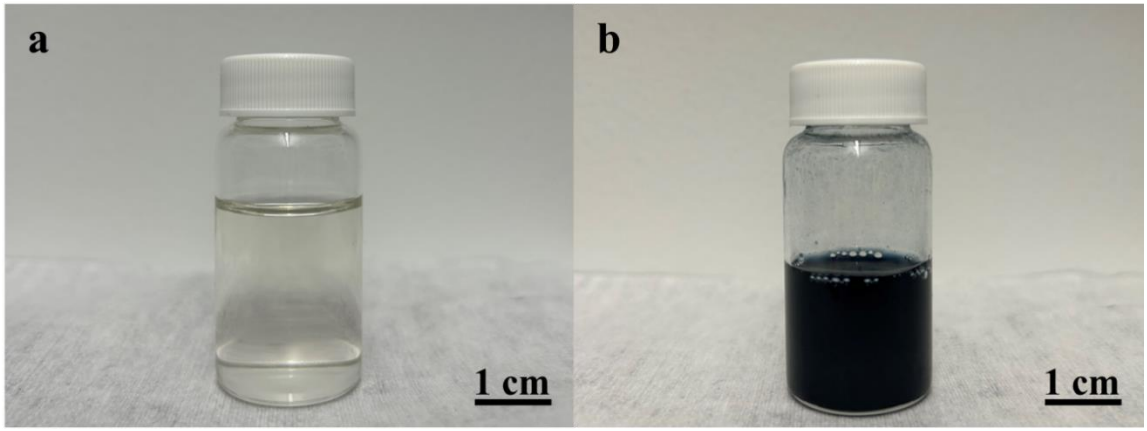

Figure 2.1 Photographs of (a) epidermal electrode hydrogel solution and (b) sealing layer hydrogel solution.

Fabrication of hydrogel epidermal electrodes: a circular Teflon mold with a radius of 1 cm is injected with 300 μl epidermal electrode hydrogel solution and cured for five minutes with UV light to obtain the epidermal hydrogel electrode (Figure 2.2 a). A silver wire is connected to the air-exposed side of the hydrogel epidermal electrode and fixed with silver glue. After the silver gel has cured spontaneously, it is transferred into a circular Teflon mold with a 2 cm radius. Then, a 2 ml sealing layer hydrogel solution is injected into the mold and cured for five minutes with UV

light. Finally, the epidermal hydrogel electrode patch is obtained by soaking it in glycerol for 30 minutes (Figure 2.2 b).

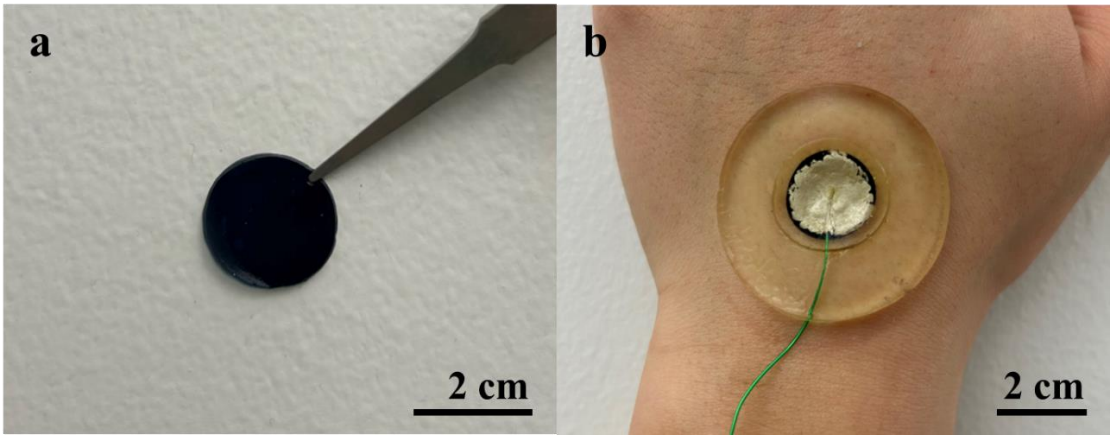

Figure 2.2 Photographs of (a) hydrogel epidermal electrode and (b) hydrogel epidermal electrode patch.

2.3 Characterization of Device

2.3.1 Water Retention Property

Glycerol, a humectant, attracts and retains moisture from the surrounding environment due to its hygroscopic nature. When a hydrogel is soaked in it, the glycerol forms hydrogen bonds with water molecules, resulting in reduced evaporation of water from the hydrogel. Moreover, glycerol interacts with the polymer chains in the hydrogel, increasing the mobility of these polymer chains which further enhances the water retention capacity of the hydrogel.

We compared the effects of different soaking times in glycerol on the water retention capability of hydrogel epidermal electrode patches (Figure 2.3). The results indicated that soaking in glycerol significantly reduced the dehydration rate of the hydrogels, and the water retention capability improved with increased soaking duration. After soaking for 30 minutes, the patches retained 95% of their weight after 20 days in an environment of 21℃ and 50% humidity, with minimal deterioration in electrical performance and self-adhesiveness. In contrast, patches soaked for 1

hour exhibited better water retention (losing only 3.5% of their weight) but showed a noticeable decrease in self-adhesion due to glycerol-induced swelling.

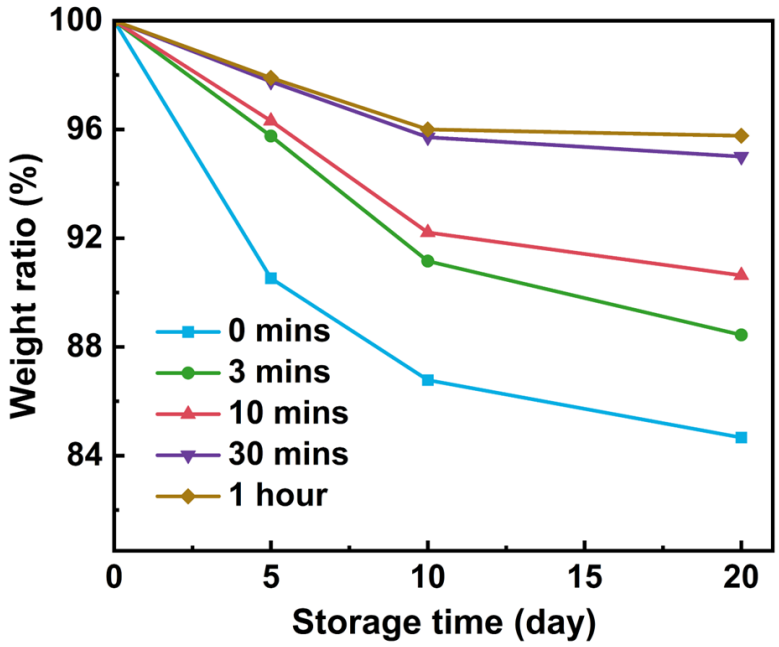

Figure 2.3 Effect of different soaking times in glycerol on water retention capacity of hydrogel epidermal electrode patches.

2.3.2 Electrical Properties

We prepared simple electrodes by attaching hydrogels with a 1cm radius and 1mm thickness onto the surface of copper tape (Figure 2.4). Two identical electrodes were placed 10cm apart on the arm and connected to an analyzer to measure the skin impedance of the sealing layer hydrogel (Pure), the hydrogel epidermal electrode without KCl (PEDOT: PSS) and with KCl (PEDOT: PSS & KCl) (Figure 2.5). The results showed that due to the insulating property of the polyacrylamide crosslinking network, the sealing layer hydrogel without conductive materials had a high skin-to-electrode impedance of around 10^6 Ω. After adding polarized PEDOT: PSS solution, the PEDOT long chains conducive to charge movement improved the conductivity of the hydrogel. However, due to the encapsulation by the insulating polyacrylamide crosslinking network, the contact between the PEDOT chains and the skin tissue was limited, resulting in a skin-to-electrode

impedance that is still greater than 10^5 Ω. To further reduce the skin-to-electrode impedance, KCl was introduced into the hydrogel system, which provides ions uniformly dispersed in the crosslinking network that act as a medium between the skin tissue and the PEDOT conductive long chains. Under the dual conductive system of charge and ions, the skin impedance of the hydrogel epidermal electrode was reduced to around 10^3 Ω, significantly enhancing its electrical performance as a tissue-machine interface. Compared to commercially-available electrodes widely used in the current market, the skin-to-electrode impedance of hydrogel epidermal electrodes has been reduced by a factor of 1000.

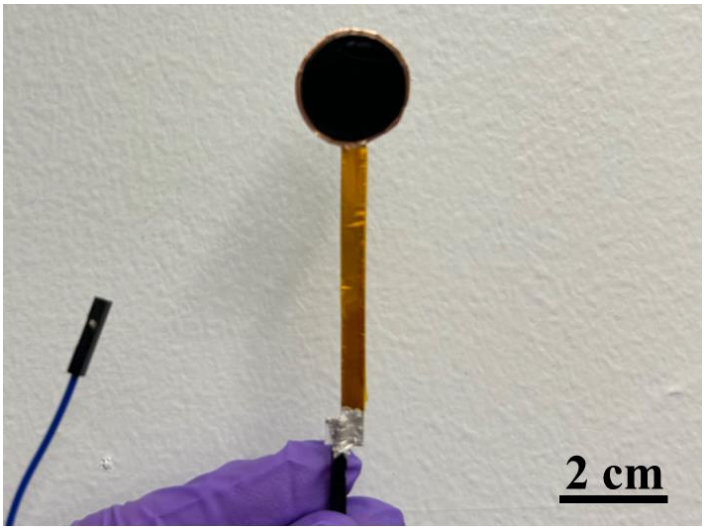

Figure 2.4 Simple electrode for skin impedance measurements.

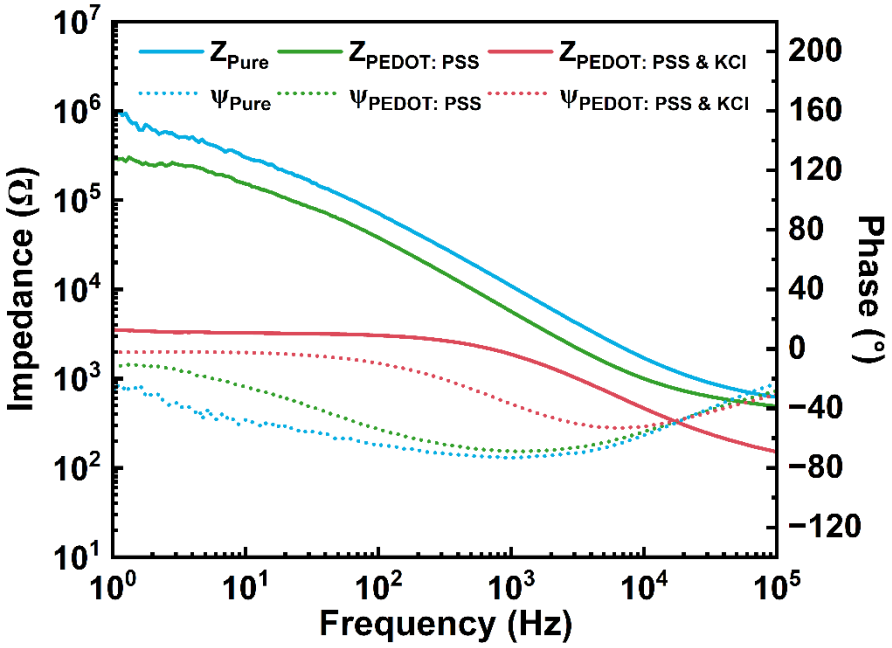

Figure 2.5 Skin impedance of the sealing layer hydrogel (Pure), the hydrogel epidermal electrode without KCl (PEDOT: PSS) and with KCl (PEDOT: PSS & KCl).

To find the conductive filler with the lowest skin impedance, we compared three biocompatible salts commonly used as hydrogel conductive fillers (Figure 2.6): sodium alginate (SA), sodium chloride (NaCl), and KCl. The skin impedance of hydrogel epidermal electrodes was measured by adding the same molar amounts of SA, NaCl and KCl. The results indicated that the hydrogel with added KCl performed the best. Compared to sodium ions (Na+) and the larger, more complex sodium alginate molecules, potassium ions (K+) tend to have a higher mobility in the solution. This is attributed to the smaller size and lower degree of hydration of K+ ions, allowing them to move more freely within the solution, thereby endowing KCl with superior conductivity.

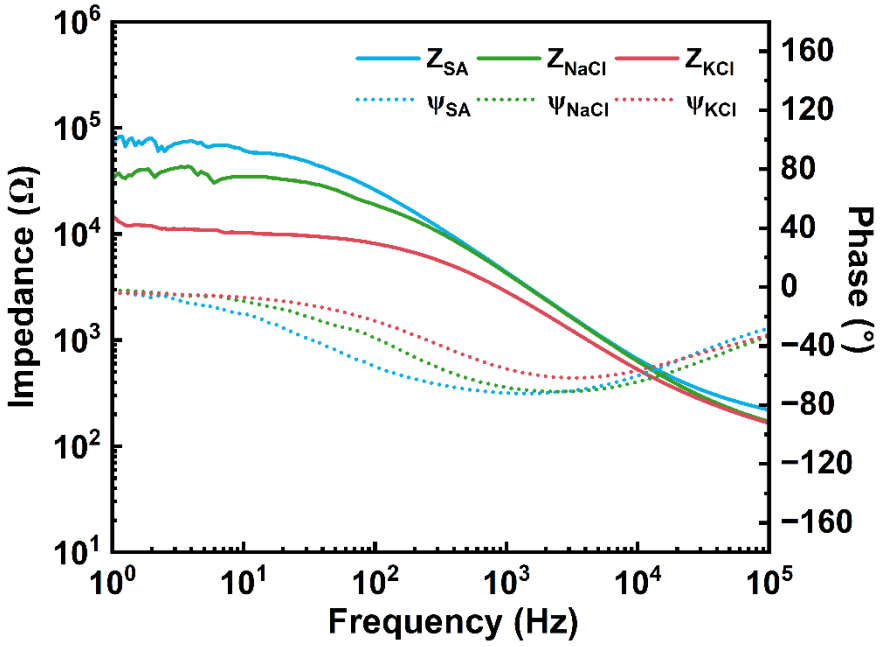

Figure 2.6 Skin impedance of hydrogel epidermal electrodes with different conductive fillers.

Furthermore, to determine the optimal concentration of the conductive filler KCl, we prepared hydrogel epidermal electrodes with concentrations ranging from 0 to 0.08 g/ml, increasing by 0.005 g/ml increments, and measured their skin impedance (Figure 2.7). The results showed that as the KCl concentration increased, the skin impedance gradually decreased. At a concentration of 0.045 g/ml, the hydrogel epidermal electrodes exhibited the lowest skin impedance across different test frequencies, indicating the optimal concentration. When the concentration continued to increase beyond this point, the skin impedance went up, which could be attributed to the amount of added KCL exceeding the solubility limit, leading to the precipitation of conductive material and thus increasing the impedance.

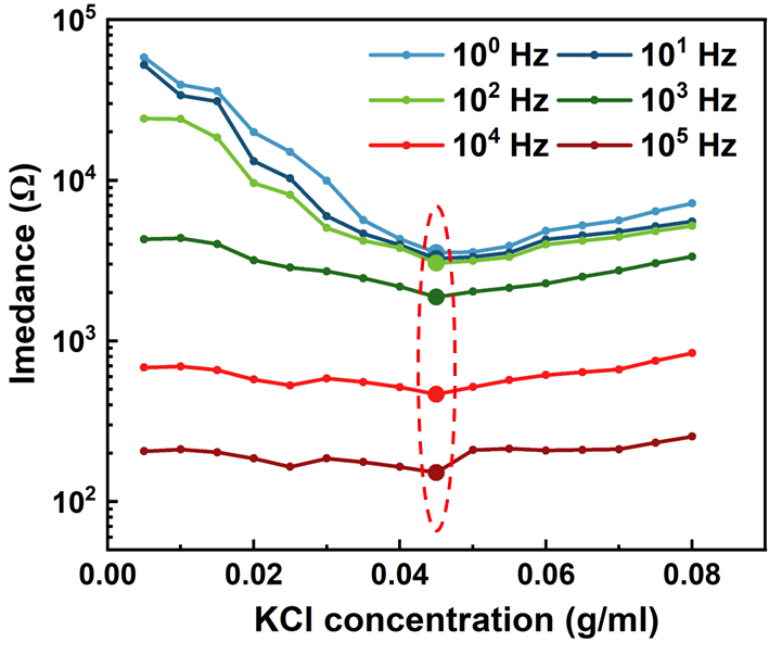

Figure 2.7 Skin impedance of hydrogel epidermal electrodes with different KCl concentrations.

2.3.3 Mechanical Properties

In general, human skin can usually be stretched to 200% to 300% of its flaccid length. Especially around joints, the skin is much more elastic. This requires hydrogel epidermal electrodes with high stretchability to accommodate a variety of movements. Polyacrylamide is renowned for its high elasticity and mechanical strength. When crosslinked, it forms a flexible and resilient network capable of enduring stretching and deformation. PEDOT: PSS also possesses a degree of flexibility due to its conjugated polymer backbone, which can undergo conformational changes without breaking. The intertwined network of these two polymers enhances the stretchability of the hydrogel. Additionally, water molecules occupy spaces within the polymer matrix, reducing internal friction and enabling the polymers to slide past each other more easily under stress, further enhancing the stretchability of hydrogel.

Here, we fabricated hydrogel skin electrode samples using a dumbbell-shaped mold and measured their tensile properties by Instron (Figure 2.8 a). To avoid the hydrogel skin electrodes from being damaged by the clamps, we first fixed them with tape, and then clamped the tape (Figure 2.8 b).

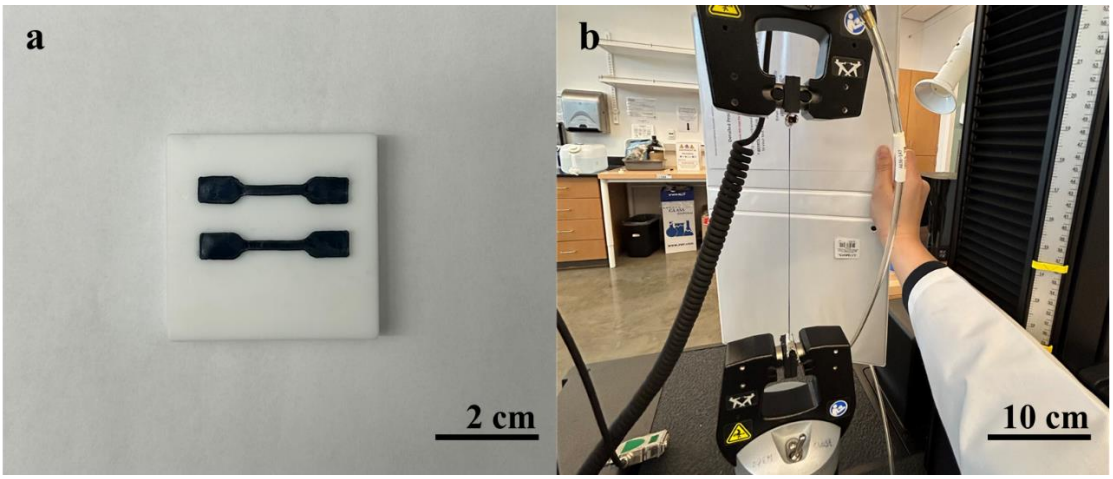

Figure 2.8 Photographs of (a) dumbbell-shaped hydrogel epidermal electrodes and (b) the stretchability measurement by Instron.

The tensile ratio of hydrogel epidermal electrode was calculated by:

$$Tensile = \frac{L-L_0}{L_0} \times 100\% \qquad (1)$$

The original length (L_0) is the length of the hydrogel between two tapes and the break length (L) is the length of the hydrogel between two tapes when it breaks.

We measured the stretchability of sealing hydrogel and epidermal electrode hydrogel at a stretching speed of 20 mm/min (Figure 2.9). The results showed that the sealing layer hydrogel could be stretched to 840%, far exceeding the stretch ratio of human skin and joints. In the hydrogel epidermal electrodes, the long chains of PEDOT: PSS intertwined with the polyacrylamide crosslinking network, increasing the overall density of the polymer network, thereby further enhancing its stretchability to 980%.

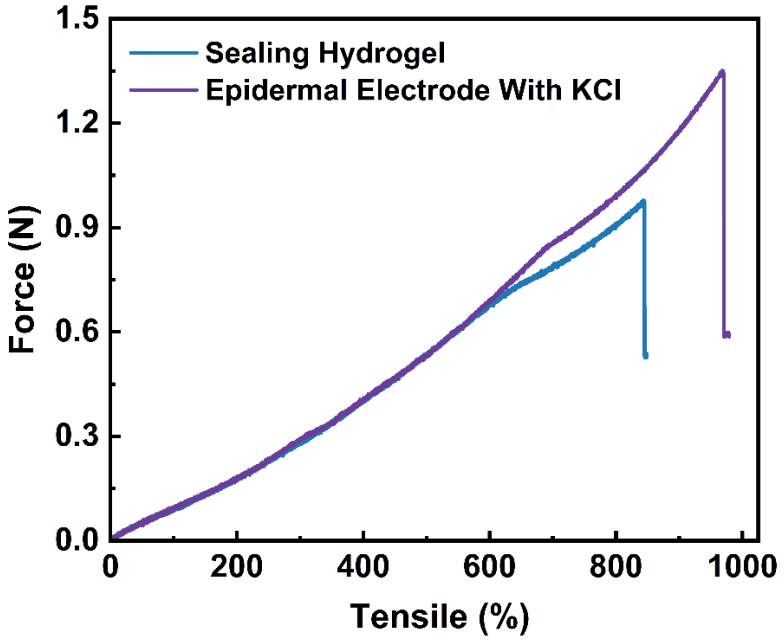

Figure 2.9 Stretchability of sealing hydrogel and epidermal electrode hydrogel.

We also measured the stretchability of the epidermal hydrogel electrode after soaking in glycerol for 30 minutes and then storing at room temperature (21℃ and 50% humidity) for 10 days (Figure 2.10). The results showed that they could still be stretched to 930%, demonstrating that the glycerol-enhanced hydrogel epidermal electrodes possess superior anti-aging properties.

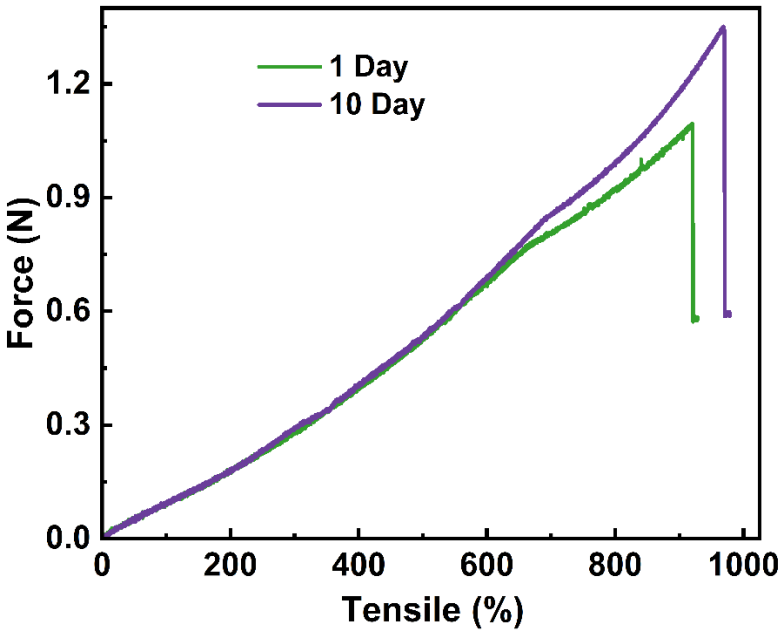

Figure 2.10 Stretchability of epidermal electrode hydrogel after room temperature storage.

In addition to tensile properties, the adhesion of the hydrogel skin electrode is also crucial in applications. Tannins contain many catechol groups, which can form strong and reversible hydrogen bonds with various surfaces (including biological tissues, synthetic materials, and metals, etc.), enhancing the adhesion of hydrogels. As a demonstration, a circular hydrogel epidermal electrode with a 1 cm radius can hold 1 kg weight without falling off (Figure 2.11).

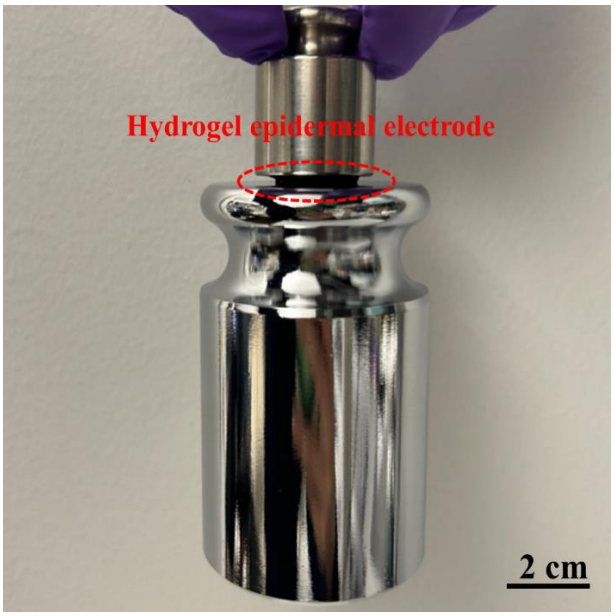

Figure 2.11 Circular hydrogel epidermal electrode with 1cm radius can hold 1 kg weight.

Furthermore, the phenolic hydroxyl groups in tannins can interact with hydroxyl groups on the surface of the substrate or tissue to enhance interfacial bonding. As a result, the wet adhesion of the hydrogel can be improved, making it more effective in humid environments, which are common in biological applications. However, this tannic acid-dominated chemical adhesion fails rapidly when the hydrogel is fully submerged in water. Swelling due to water absorption disrupts the intimate contact between the hydrogel and the substrate. Also, the presence of water dilutes the concentration of adhesive functional groups on the surface of the hydrogel, thereby reducing hydrogen bonding, van der Waals forces, and adhesive interactions with the substrate.

Octopuses can adhere to various surfaces underwater by creating a vacuum with suction cups aligned along tentacles, resulting in a strong adhesive force. The edges of the suction cups closely conform to the surface, ensuring a tight seal even on rough or irregular surfaces. Inspired by this, we designed a suction cup structure with mechanical adhesion for hydrogel epidermal electrode patches, based on the observation that hydrogel curves towards one side of the mold during the crosslinking process (Figure 2.11). This combination of chemical and mechanical adhesion endows the patches with the ability to maintain underwater adhesion for extended periods, preventing detachment for over an hour (Figure 2.12).

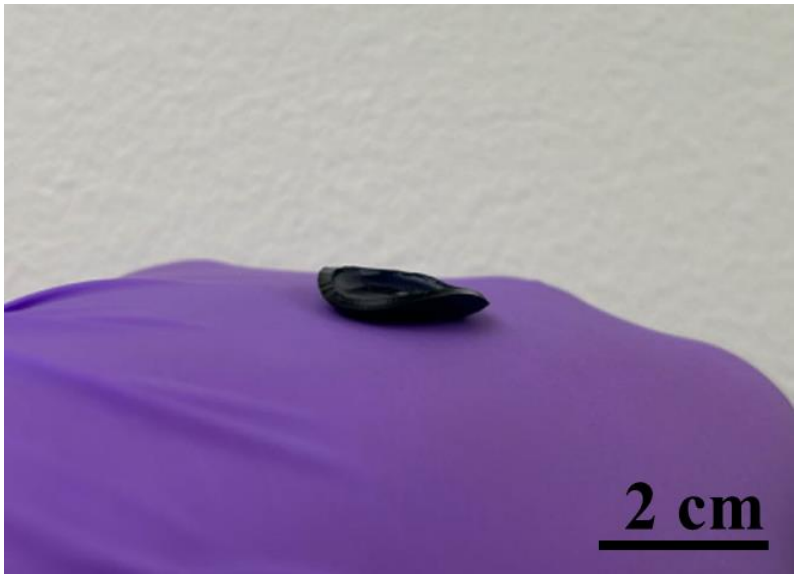

Figure 2.12 Photograph of suction-cup shaped hydrogel epidermal electrode.

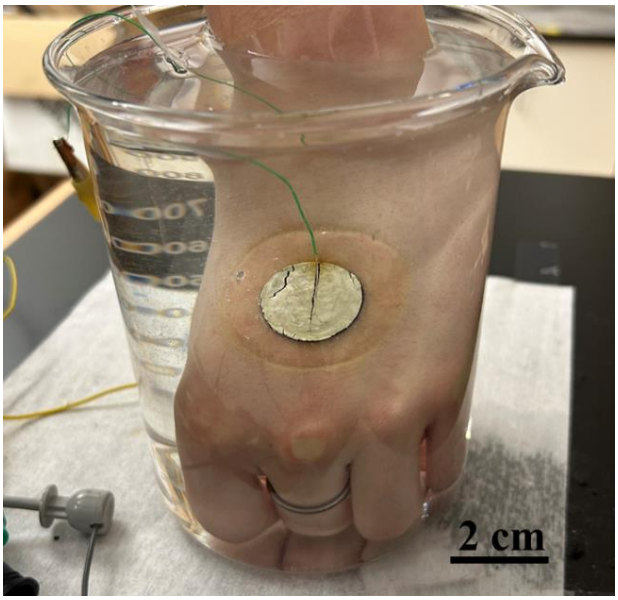

Figure 2.13 Photograph of hydrogel epidermal electrode patches adhering to the skin underwater.

Chapter 3: Applications

Hydrogel epidermal electrode patches boast low skin impedance, outstanding adhesion, superior stretchability, along with remarkable resistance to aging and stability. These properties enable them to reliably record electrophysiological and mechanical signals from the human body in a variety of scenarios.

3.1 Electrophysiological Recording

To illustrate the enhancement in electrical performance afforded by the addition of KCl salt to hydrogel epidermal electrodes, we conducted electrocardiogram (ECG) tests comparing commercial electrodes with hydrogel epidermal electrode patches with and without KCl. For the tests, one electrode was placed on each side of the left and right forearms, connected to an analyzer, with the scale set at 200 mV and 1 s. Using MATLAB, we calculated the signal-to-noise ratio (SNR) for the different electrodes based on the ECG waveforms and raw data collected (Figure 3.1). The results show that, compared to traditional commercial electrodes, hydrogel skin electrodes exhibited a significant increase in SNR, from 17.33 dB to 23.54 dB, indicating superior signal quality. The hydrogel skin electrode with KCl showed another 1.5 dB improvement in SNR over the hydrogel skin electrode without KCl. This improvement was attributed to the ionic conductivity provided by KCl, which further reduced skin impedance, consistent with prior results.

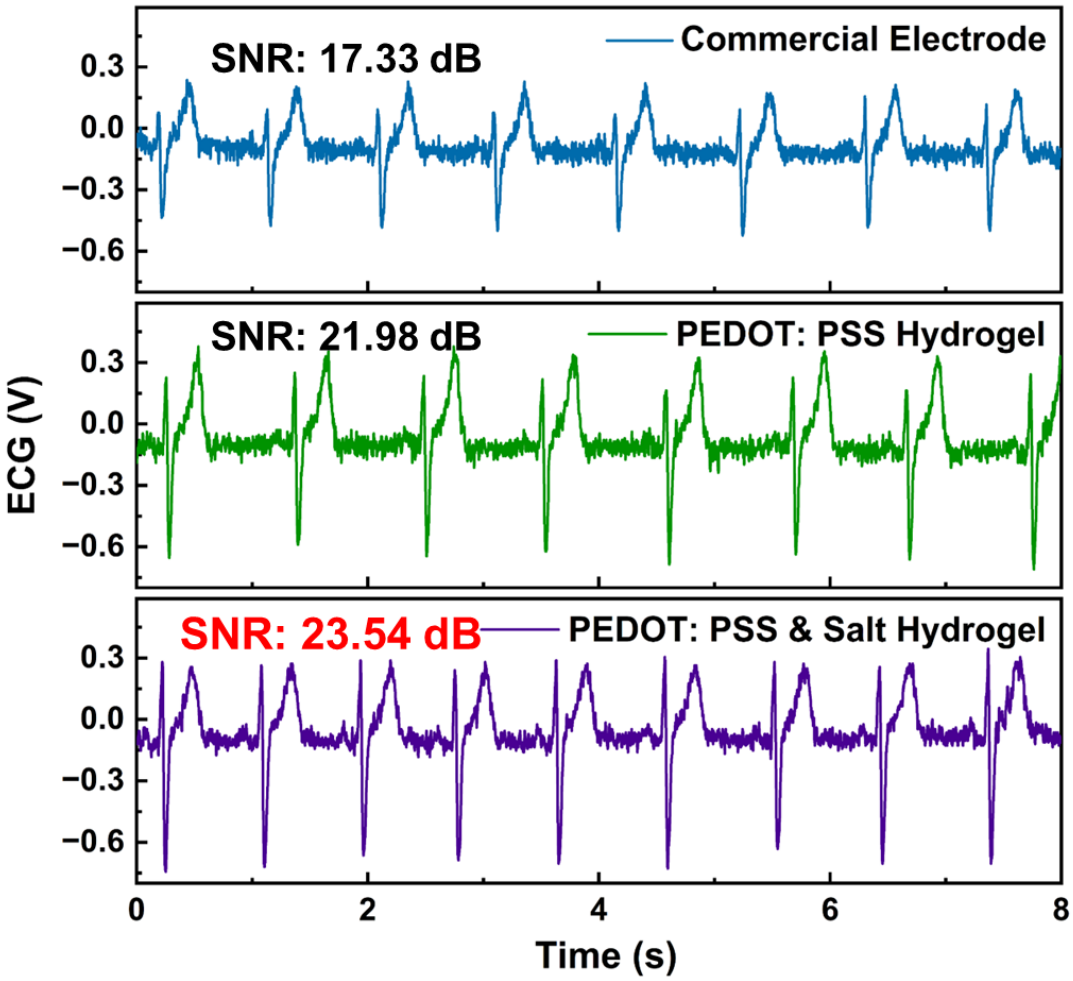

Figure 3.1 ECG of commercial electrodes, hydrogel epidermal electrode patches with and without KCl salt.

The MATLAB code for calculating SNR is shown in Figure 3.2.

a

```
%% SNR_ECG
%% Noise
t1 = 2.538:0.002:3.112;
plot(t1',Noise)
[up1,lo1] = envelope(Noise,500,'analytic');
hold on
plot(t1',up1,'-',t1',lo1,'--')
hold off
NoisePeak=up1-lo1;
NoiseAverage = mean2(NoisePeak)


%% Signal
SignalAverage = 0.378-(-0.646)
SNR = 20*log10(SignalAverage/NoiseAverage)
```

b

```
%% SNR_EMG
%% Noise
t1 = 38.79:0.02:48.52;
plot(t1',Noise)
[up1,lo1] = envelope(Noise,500,'analytic');
hold on
plot(t1',up1,'-',t1',lo1,'--')
hold off
NoisePeak=up1-lo1;
NoiseAverage = mean2(NoisePeak)


%% Signal
t2 = 69.43:0.02:78.76;
plot(t2',Signal)
[up2,lo2] = envelope(Signal,500,'analytic');
hold on
plot(t2',up2,'-',t2',lo2,'--')
hold off
SignalPeak=up2-lo2;
SignalAverage = mean2(SignalPeak)
SNR = 20*log10(SignalAverage/NoiseAverage)
```

Figure 3.2 MATLAB code for calculating SNR for (a) ECG and (b) EMG.

To confirm the self-adhesive capability of hydrogel epidermal electrode patches, we performed ECG tests comparing the use of tape-secured hydrogel epidermal electrode patches versus relying solely on patch self-adhesion, and subsequently calculated SNRs (Figure 3.3). The results show that the SNR achieved through the self-adhesive one was nearly identical to that obtained with tape fixation. These findings indicate that hydrogel epidermal electrode patches can deliver high-quality ECG recordings without the need for external pressure fixation, significantly enhancing user comfort.

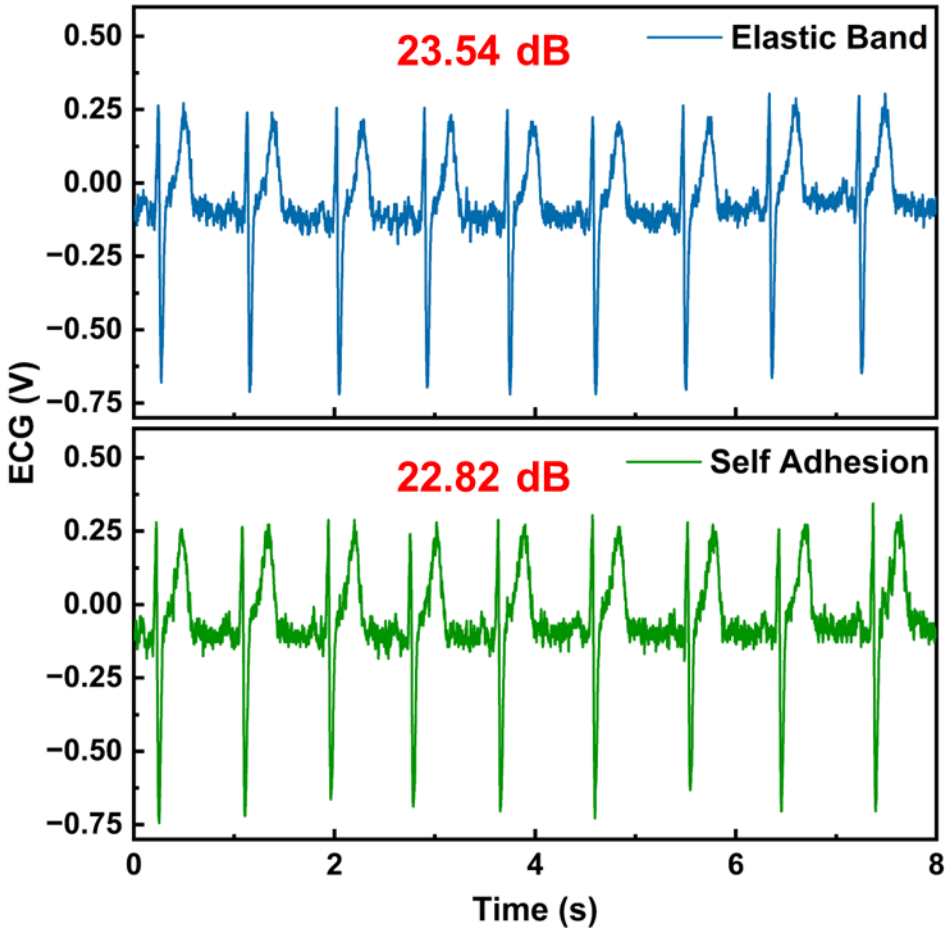

Figure 3.3 ECG of hydrogel epidermal electrode patches secured with tape and self-adhesive.

The hydrogel epidermal electrode patch proved capable of monitoring the electrical activity of muscles during nerve stimulation as well. A series of hydrogel epidermal electrode patches, positioned 5 cm apart on the arm, were used to capture electromyogram (EMG) signals from the forearm muscle groups. As the grip strength of the forearm muscle group intensified, there was a corresponding increase in the EMG amplitude, further demonstrating the superior signal quality offered by our hydrogel epidermal electrode patches (Figure 3.4).

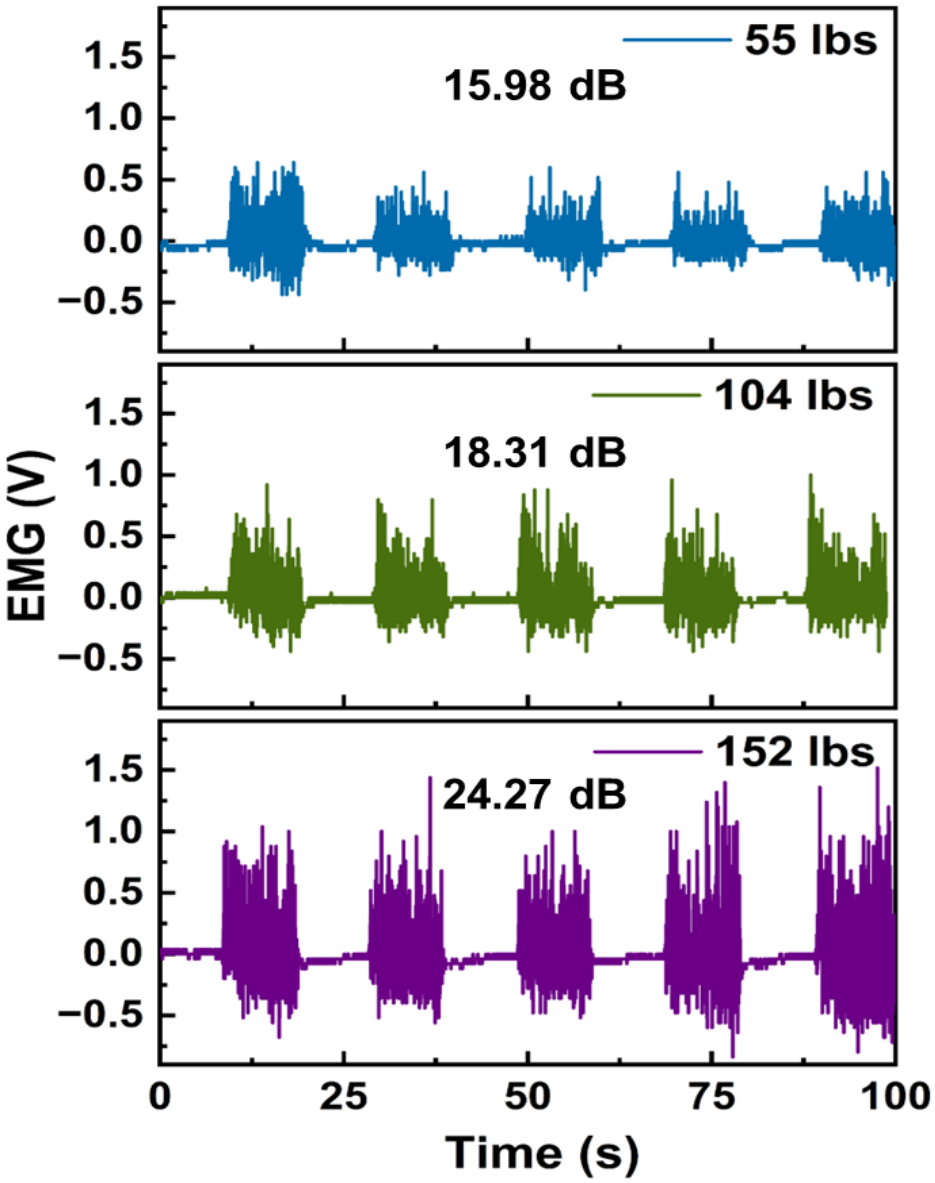

Figure 3.4 EMG generated by different grip strengths in forearm muscle groups.

Diving and underwater activities can significantly stress the cardiovascular system. The combination of water pressure, the need to control breathing, and the physical exertion associated with swimming or operating equipment can increase heart rate and blood pressure. ECG monitoring helps to evaluate an individual's heart's ability to cope with these stress factors. Underwater environments, especially during deep-sea diving or when there are issues with the breathing apparatus, can pose a risk of hypoxia (low oxygen levels). Monitoring the heart's

electrical activity can help detect signs of oxygen deprivation, which is crucial for timely intervention.

Currently, despite the availability of commercial products like swimming watches, chest straps, and armbands designed for monitoring heart rate (HR) during aquatic sports, their functionality is limited to recording heart rate at discontinuous intervals. Achieving real-time, continuous recording of raw electrophysiological signals in underwater environments remains an unresolved challenge. The significant interference caused by water makes the use of electrodes impractical underwater. The ingress of water into the interface between the skin and the electrodes can easily lead to short-circuiting and poor contact, complicating the accurate recording of biopotentials.

Hydrogel epidermal electrode patches can maintain adhesion and sealing underwater for long periods, we utilized them for underwater ECG recordings (Figure 3.5).

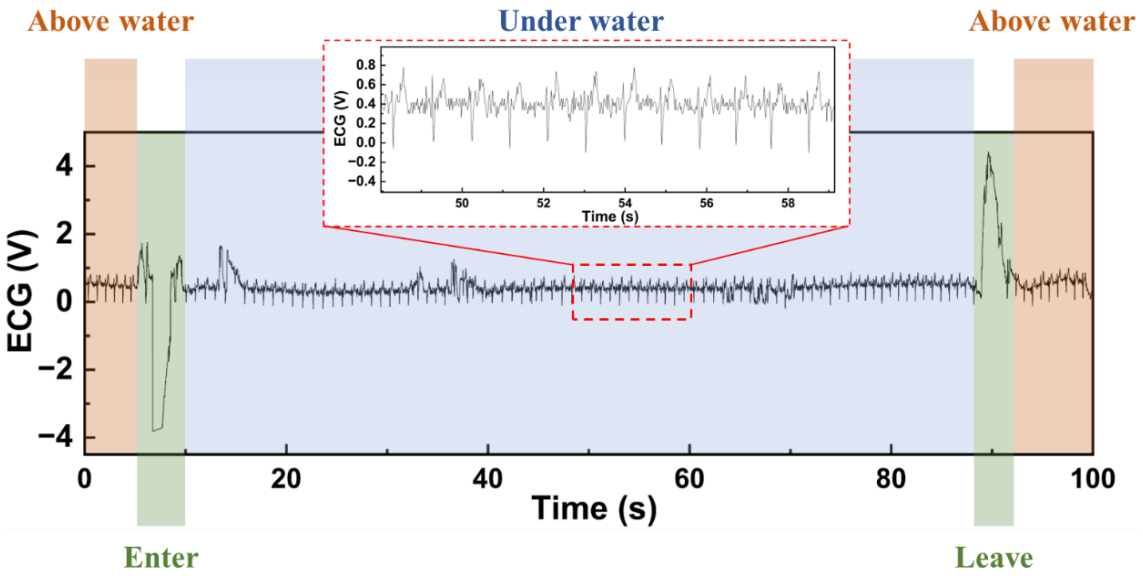

Figure 3.5 Underwater ECG recording of hydrogel epidermal electrode patches.

While there is some noise observed during entry and exit from the water, likely attributed to the movement of the wires, the ECG signals recorded underwater were nearly identical to those

captured before and after immersion. This consistency underscores the potential of hydrogel epidermal electrodes for real-time ECG monitoring in aquatic environments.

3.2 Mechanical Sensor

Hydrogel epidermal electrodes are inherently elastic and flexible, enabling them to withstand substantial deformation such as stretching, bending, and compression without permanent damage. This flexibility ensures that the conductive network within the hydrogel can be reversibly stretched and revert to its original state, making it well-suited for mechanical sensor applications that require adaptation to dynamic surfaces or movements.

As the hydrogel epidermal electrodes are stretched, the spacing between the conductive particles within them alters, which in turn modifies the electrical pathways in the gel. These alterations in pathway length or direction can influence the overall electrical resistance of the hydrogel. Here, we fixed strips of hydrogel epidermal electrodes at both ends of the syringe pump, stretched it at a rate of 20 mm/min and recorded the resistance in real-time (Figure 3.6).

Figure 3.6 Measurement device for tensile resistance change of hydrogel epidermal electrode.

The change of the hydrogel epidermal electrode resistance as a function of the stretch length is shown in Figure 3.7. The resistance of the hydrogel skin exhibited a linear increase across an extensive stretching span from 200% to 500%, with the rate of change remaining consistent during both stretching and relaxation phases. This consistency indicates that the conductive material within the hydrogel is evenly distributed, and the structural integrity of the hydrogel skin electrode is maintained throughout the stretching process.

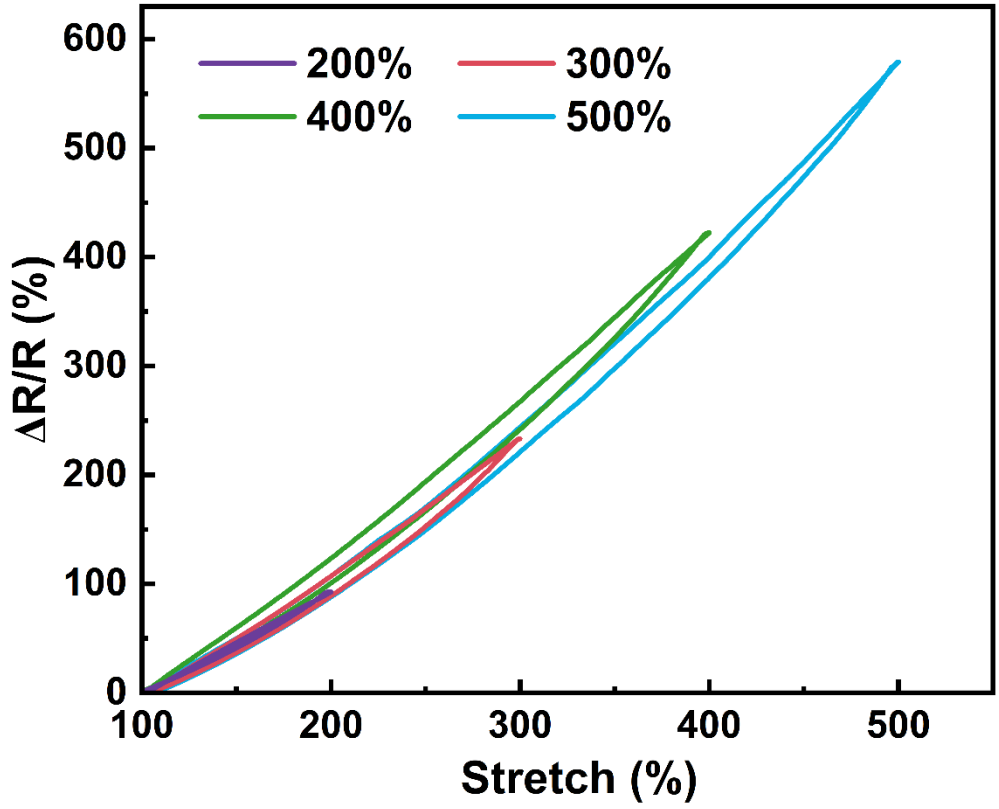

Figure 3.7 Changes in hydrogel epidermal electrode resistance by different stretch lengths.

To validate the stability of the hydrogel epidermal electrode as a strain sensor, we conducted cyclic tensile tests, capturing the resistance changes in real-time (Figure 3.8). The results indicated that the hydrogel epidermal electrode's resistance change was highly consistent at stretches up to 400%. When stretched up to 500%, the hydrogel exhibited gradual increase in resistance cycle after cycle, potentially due to permanent alterations in the local hydrogel

structure induced by stretching. However, a 400% stretch capacity adequately fulfills the epidermal electrodes' needs for accommodating the diverse deformations experienced by human skin and joints.

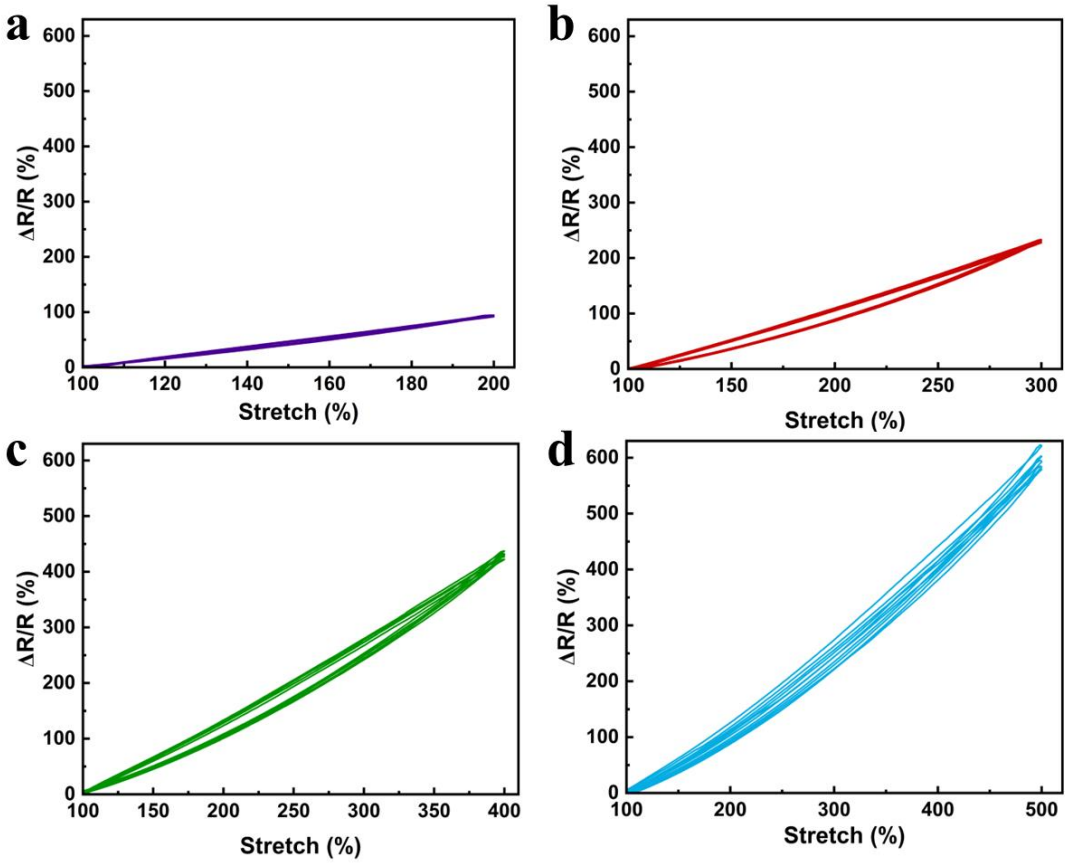

Figure 3.8 Changes in hydrogel epidermal electrode resistance by cyclic stretching.

Chapter 4: Summary and Prospects

In this study, we have adeptly engineered and developed hydrogel epidermal electrode patches characterized by their exceptional electrical conductivity, minimal skin impedance, superior stretchability, robust adhesion, and consistent performance. The introduction of ethylene glycol, a polar solvent, causes the PEDOT particles to cluster and align into extended formations, thereby increasing the electrical conductivity through enhanced charge mobility. KCl salt ions, serving as conductive agents, blend with the polymer hydrogel to create an IPN, merging charge and ionic conductivity for heightened electrical conduction while substantially reducing skin impedance, which in turn augments the fidelity of human electrophysiological signal capture. The inclusion of catechol-rich elements bolsters adhesion, replicating dopamine-like bonding to secure firm attachment to tissue surfaces and minimize noise from electrode-skin gaps. A sealing hydrogel, mirroring the cross-linking structure of the epidermal electrodes, acts as a reliable sealing layer, preserving consistent mechanical attributes and adhesion. Drawing inspiration from the adhesive capabilities of octopus tentacles, we have conceptualized a suction cup design that ensures sustained adhesion in moist environments, marrying mechanical adherence with chemical bonding for expanded application versatility. Furthermore, the incorporation of glycerol significantly boosts its hydration retention, thereby extending the longevity and efficacy of the hydrogel's properties.

Through conducting ECG and EMG recordings, it was observed that the hydrogel epidermal electrode patch demonstrated a notable > 6 dB enhancement in SNR compared to commercial electrodes. This significant increase in SNR, coupled with the nearly equivalent quality of ECG signals without the necessity for external fixation, markedly enhanced diagnostic accuracy and

user comfort. Furthermore, we have achieved high-quality, real-time underwater ECG recordings, paving the way for enhanced safety and health monitoring for divers and underwater workers. Additionally, the exploration of hydrogel skin electrodes as mechanical sensors, leveraging their exceptional stretchability and uniformity, broadens their potential applications.

However, this research is presently in its initial exploratory phase, with numerous challenges yet to be addressed. For example, the current methodology for recording human electrophysiological signals requires wired connections to the equipment, which introduces significant noise due to wire movement, consequently restricting user mobility. Moving forward, we aim to enable signal acquisition via wireless transmission, facilitating the real-time recording of electrophysiological signals during user activity, thereby enhancing usability and mobility.

Moreover, despite our design of the suction cup structure influenced by the mold's effect on the hydrogel's curving orientation, the underlying mechanisms remain largely unexplored. This lack of a comprehensive understanding results in the structure's unpredictability. Future efforts will be directed toward conducting in-depth research to develop a systematic and human-manipulable structural design.

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