

Soft Cardiac Patch Using a Bifacial Architecture of Adhesive/Low-Impedance Hydrogel Nanocomposites and Highly Conductive Elastomer Nanocomposites

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Soft implantable multichannel cardiac electrode arrays that establish direct monolithic interfaces with the heart are key components for advanced cardiac monitoring and electrical modulation. A significant technological advancement in this area is the development of stretchable conductive nanocomposites, fabricated through the integration of metallic nanomaterials and elastic polymers, aimed at achieving both high electrical conductivity and mechanical elasticity. Despite these advances, further progress in material performance and device designs is required to ensure seamless, reliable, biocompatible, and high-fidelity cardiac interfacing. Herein, the development of a soft multichannel cardiac patch based on a bifacial architecture of adhesive/low-impedance hydrogel nanocomposites and highly conductive elastomer nanocomposites is reported. The bifacial design facilitates the integration of the cardiac patch between the heart and other tissues/organs can be achieved. The hydrogel nanocomposite layer, positioned on the epicardial side, provides stable adhesion to the target cardiac tissue and enables low-impedance biocompatible interfacing with the heart, while the elastomer nanocomposite layer, positioned on the opposite side, offers high electrical conductivity for facile electrophysiological signal transfer and a low-friction surface minimizing unwanted interactions with surrounding tissues. The effectiveness of this bifacial patch in multiple applications involving various cardiac signal recordings and electromechanical modulation demonstrations is showcased.

1. Introduction

Despite significant advances in modern medicine, cardiovascular diseases remain a leading cause of mortality, accounting for ≈ 17.9 million deaths annually worldwide.^[1,2] A substantial proportion of these deaths are attributed to sudden cardiac death, often caused by myocardial tissue damage and the onset of unregulated/dyssynchronous heart rhythms, such as ventricular arrhythmias.^[3,4] To address the need for continuous heart rhythm monitoring and timely therapeutic interventions, a range of implantable cardiac devices (e.g., pacemakers, defibrillators, and resynchronization therapy devices) have been developed.^[5] These devices are designed to monitor cardiac rhythms closely and deliver electrical stimulation through direct contact with the heart tissues. However, such conventional implantable cardiac devices made of rigid electronic materials face challenges in terms of compatibility with soft dynamically moving heart, primarily due to the mechanical mismatch between the rigid materials of the device and the soft tissues

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of the heart.^[6,7] This mismatch can lead to a suboptimal tissue-device interface, potentially compromising device performance.

The stretchable electronics composed of photolithographically patterned metals deposited on ultrathin polymer substrates have been proposed as a solution.^[8–10] Unconventional form factor engineering to the device (e.g., kirigami structure, buckling, and island bridge) has been also investigated to provide sufficient stretchability.^[7,11] However, these types of devices still exhibit limitations in long-term durability under chronic mechanical stress and lack sufficient softness. To overcome these challenges, there has been a growing interest in the development of intrinsically stretchable electronics.^[12,13] Many types of materials, including polymer nanocomposites and liquid metals, have been suggested as a stretchable electrode. For example, liquid metal has been highlighted for their infinite stretchability and excellent conductivity.

Elastomeric nanocomposites, which are composed of conductive filler materials (e.g., metallic nanomaterials,^[13–20] carbon nanomaterials,^[21,22] and liquid metals^[23,24]) uniformly dispersed

in an elastomeric matrix, have also attracted significant attention. The elastomeric nanocomposites, featuring both high conductivity and elasticity simultaneously, can maintain effective functionality, even under the heart's repetitive motion. Despite these advances, several functional improvements are required. First, the mechanical softness of the electronics must be enhanced to more closely match that of the heart tissue. Commonly utilized elastomers, such as polydimethylsiloxane and EcoFlex, have a significantly higher Young's modulus than heart tissue, resulting in a mechanical mismatch.^[25–27] Second, the electronics must possess adhesiveness to ensure conformal contact with the tissue for long-term operation.^[28] The stable seamless interfacing between the device and the tissue could enhance the device performance, such as preventing the signal leakage,^[24,29,30] and yet, secure the device on the heart without disrupting its active multidirectional motion. Lastly, achieving optimal long-term biocompatibility is crucial for clinical translation. The use of highly biocompatible materials for epicardial electronics is required to minimize chronic immune responses (e.g., fibrous tissue formation).^[31,32]

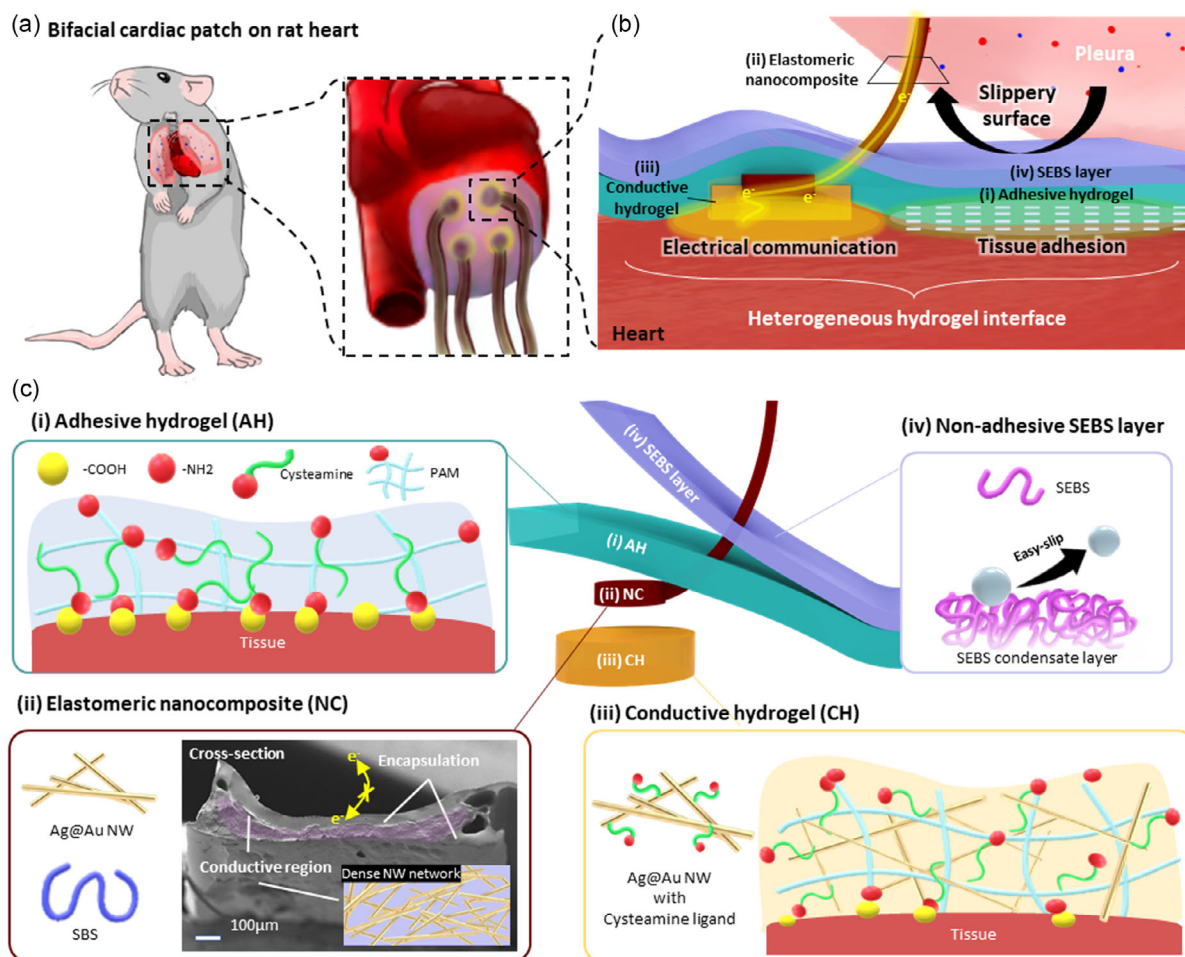


Figure 1. Design strategies of cardiac patch with heterogeneous hydrogel interface and elastomeric surface. a) Schematic illustration of the implemented cardiac patch on a rat heart. b) Bifacial cardiac patch design realized by integrating heterogeneous hydrogels in a single layer, layered with elastomeric nanocomposite outside. c) Exploded view of the cardiac patch, detailing each component: (i) cysteamine-integrated polyacrylamide adhesive hydrogel, (ii) elastomeric nanocomposite comprising conductive core and encapsulation, (iii) conductive hydrogel composed of cysteamine-modified Ag–Au NWs, and (iv) hydrophobic SEBS slippery layer.

Hydrogel nanocomposites, owing to their intrinsic ultrahigh softness, have been also actively explored as promising biomaterials for use in dynamic in vivo environment.^[33,34] Several approaches have aimed to enhance the functionality of hydrogel nanocomposites to meet the specific requirements of in vivo applications. For example, conductive hydrogels have been developed to serve electrical functions within biological systems by utilizing conductive polymer networks or integrating conductive nanomaterials. Additionally, grafting functional moieties onto the polymer network has been suggested for imparting adhesive properties to the hydrogel. However, hydrogel-based electronics are inherently limited in electrical performance due to their water-rich composition, which can lead to inaccurate measurements.^[35] Moreover, the interfacial toughness required for adhesive platforms to endure external strain demands both adhesiveness and toughness, rendering the use of bare hydrogel-based devices in biomedical applications particularly challenging.

To address these challenges, we present a cardiac patch with a bifacial architecture of adhesive/low-impedance heterogeneous hydrogel nanocomposites interface and highly conductive elastomer nanocomposites (Figure 1a,b). The tissue–device interface comprises adhesive hydrogel and conductive hydrogel, specifically designed to enhance biocompatibility and provide desired functionality (e.g., adhesiveness and low impedance) for long-term epicardial applications (Figure 1a). The adhesive hydrogel (Figure 1b,i) ensures the stable fixation of the cardiac patch at the application site, even with the heart's dynamic movements. The low-impedance conductive hydrogel (Figure 1b,ii) enables electrical interaction between the elastomeric nanocomposite and heart tissue, enabling effective electrical sensing and stimulation. Both hydrogels possess low modulus and high stretchability, ensuring biocompatibility and seamless contact between the device and the heart. The elastomeric nanocomposite electrode, located on the hydrogel and in contact with conductive hydrogel,

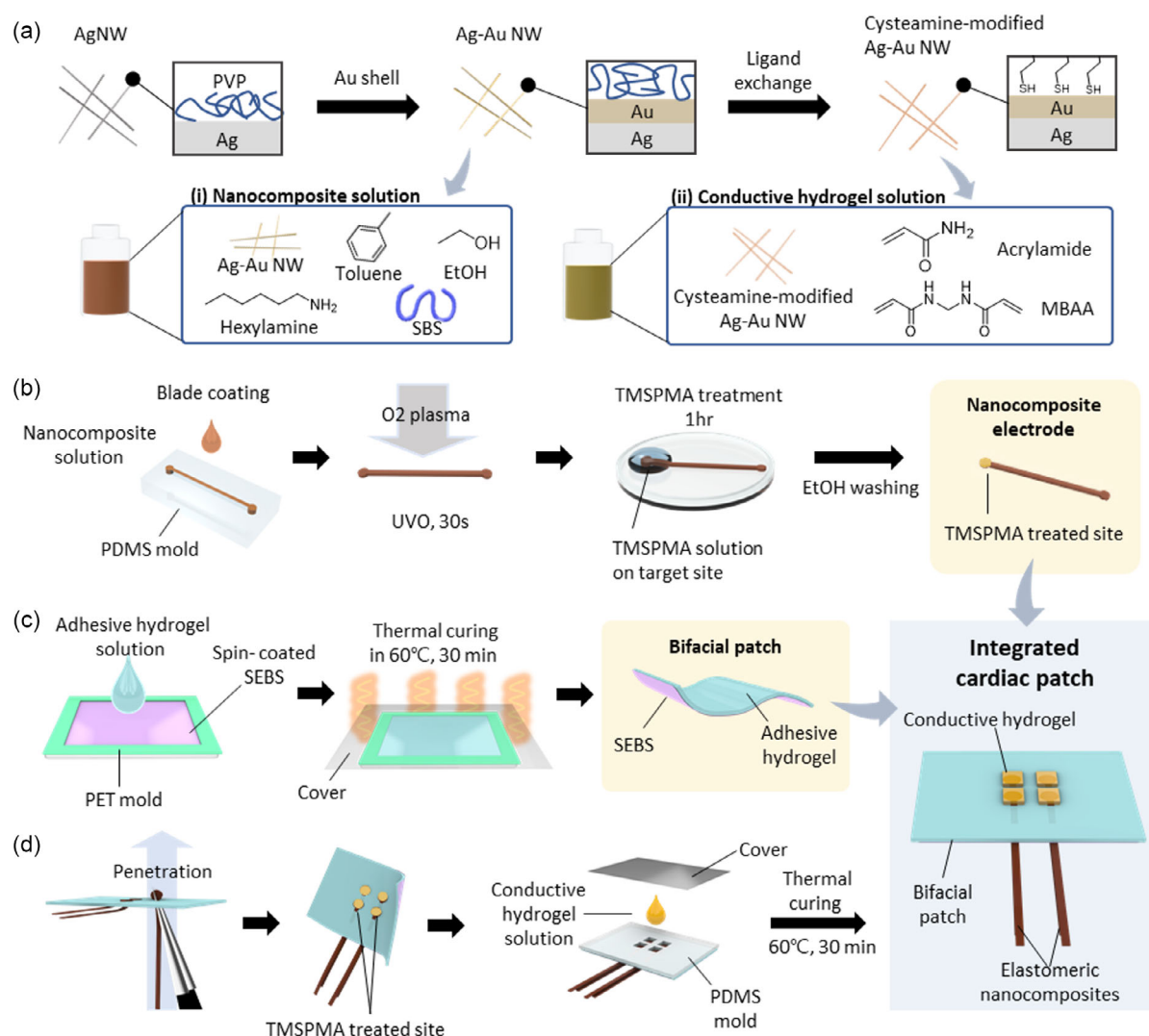


Figure 2. Schematic illustration detailing the fabrication process of the cardiac patch. a) Ag core-Au shell nanowire synthesis process and use of nanowires in preparation of (i) nanocomposite solution and (ii) conductive hydrogel solution. b) Fabrication of elastomeric nanocomposite by blade coating and TMSPMA posttreatment. c) Bifacial patch fabrication by curing adhesive hydrogel on the SEBS layer. d) Assembly of bifacial cardiac patch.

provides pathways for facile delivery of electrical signals (Figure 1b,iii). The top encapsulation layer consists of styrene-ethylene-butylene-styrene copolymer (SEBS) (Figure 1b,iv), which provides insulative low-friction surface and prevents current leakage and unwanted interaction with other organs. The proposed cardiac patch, integrated with four segments as aforementioned, has demonstrated stable cardiac electrogram acquisition and effective stimulation, underscoring its potential as a next-generation epicardial electronics.

2. Results and Discussion

2.1. Design Strategies and Fabrication of the Bifacial Cardiac Patch

The cardiac patch features a bilayer structure consisting of hydrogel nanocomposite interface with adhesive region at the bottom (Figure 1c,i) and conductive region (Figure 1c,ii), elastomeric nanocomposite electrode (Figure 1c,iii), and elastomer encapsulation layer at the top (Figure 1c,iv). The adhesive hydrogel is composed of a cysteamine-integrated polyacrylamide network. The cysteamine forms hydrogen bonds with the heart tissue,

which imparts adhesiveness to the hydrogel, while the soft and stretchable polyacrylamide network supports secure adhesion to the heart's surface. The conductive hydrogel consists of a polyacrylamide network including cysteamine-modified Ag-core-Au-shell nanowires (Ag-Au NWs), which serve as a contact electrode for elastomeric nanocomposites. The conductive hydrogel lowers contact impedance for facile bidirectional electrical signal delivery.

The elastomeric nanocomposite for the stretchable bioelectrode layer is formed by homogeneous mixing of styrene-butadiene-styrene (SBS) and Ag-Au NWs. The high electrical conductivity of the elastomeric nanocomposite endows the cardiac patch with facile electrical signal transport. The soft biocompatible hydrogel electrodes are in direct contact with the heart tissue (Figure 1c). These electrodes are connected to the elastomer nanocomposite for electrical recording/modulation functions. The outermost SEBS encapsulation layer acts as a hydrophobic backing (Figure S2, Supporting Information), preventing undesired interactions and adhesions between the cardiac patch and the surrounding tissues, such as the pleura.

The fabrication process of the cardiac patch is presented in **Figure 2**. First, Ag-Au NWs were synthesized by coating Ag NWs with Au sheaths (Figure 2a). The silver core of Ag-Au

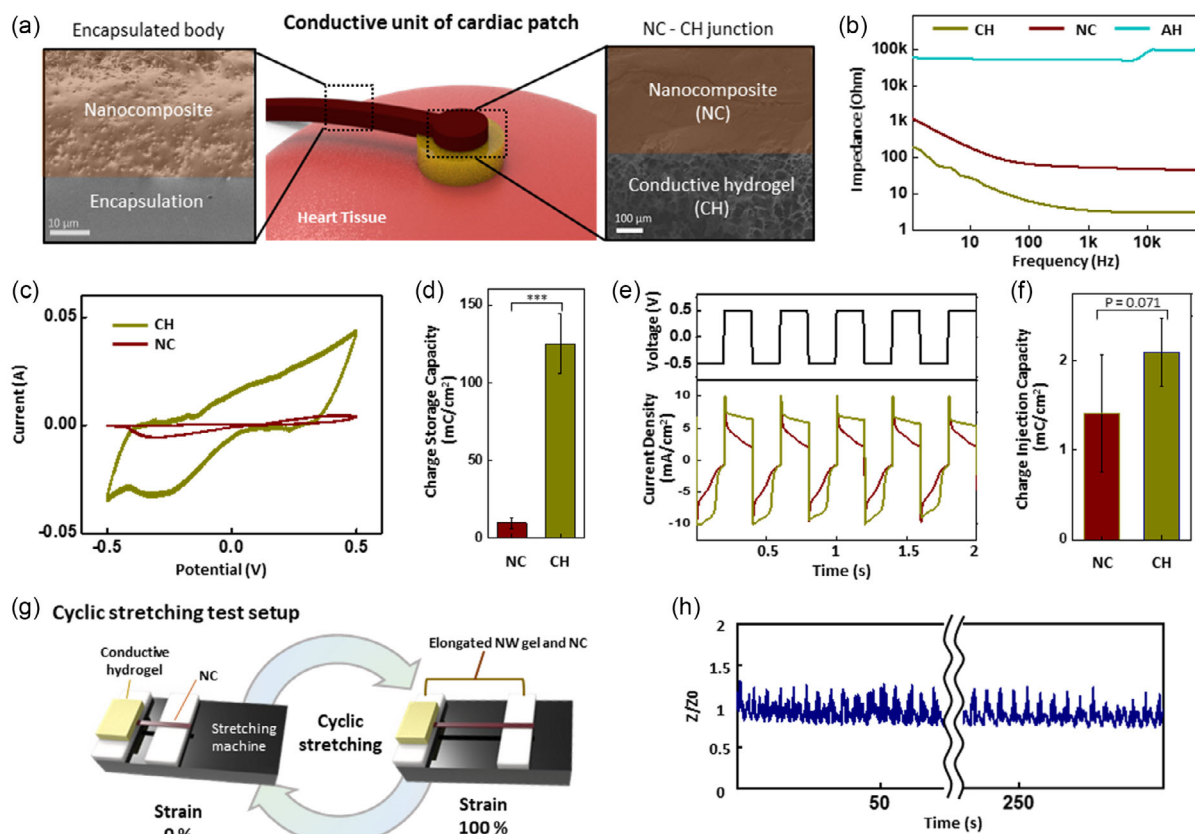


Figure 3. Electrochemical characterization of cardiac patch. a) SEM images of interfaces between nanocomposite and encapsulation (left), nanocomposite and conductive hydrogel (right). b) Impedance of conductive hydrogel, nanocomposite, and adhesive hydrogel. c) Cyclic voltammetry of conductive hydrogel and nanocomposite and d) charge storage capacity of each component ($n = 3$, $NC = 9.86 \pm 3.07 \text{ mC cm}^{-2}$, $CH = 125.21 \pm 19.30 \text{ mC cm}^{-2}$; $***p \leq 0.001$ for variables of significance). e) Current density of conductive materials induced by voltage pulse and f) charge injection capacity of each ($n = 3$, $NC = 1.42 \pm 0.66 \text{ mC cm}^{-2}$, $CH = 2.09 \pm 0.38 \text{ mC cm}^{-2}$). Error bars in bar graphs are the standard deviation of the measurements. g) Schematic illustration of cyclic stretching test setup and h) cyclic test results.

NWs provides high conductivity, while the Au shell enhances biocompatibility by preventing silver ion leaching, achieving both biocompatibility and conductivity.^[36] The nanocomposite solution was prepared by mixing Ag–Au NWs and SBS in toluene (Figure 2a,i). The cysteamine-modified nanowires are produced by adding cysteamine to the Ag–Au NW solution, followed by sonication; these are then mixed with an acrylamide solution to form the conductive hydrogel (Figure 2a,ii). The elastomeric nanocomposite is fabricated by using a blade-coating method with a negatively patterned PDMS mold (Figure 2b). The electrode part is then treated with oxygen plasma to introduce hydroxyl groups, followed by 3-(Trimethoxysilyl)propyl methacrylate (TMSPMA) surface modification, thereby joining the conductive hydrogel and the elastomeric nanocomposite.^[37]

The bifacial patch, which enables the selective adhesion of the cardiac patch, is fabricated by curing an adhesive hydrogel onto a thin SEBS layer (Figure 2c). This SEBS layer is created by spin-coating a SEBS/toluene solution onto a glass substrate. A polyethylene terephthalate (PET) frame is then affixed to the SEBS layer, acting as a negative mold. An acrylamide precursor

solution, containing cysteamine, is poured into the mold and thermally cured, forming an adhesive hydrogel that is physically bonded to the SEBS. Once the bifacial patch is fully formed, it can be easily removed from the glass substrate. The elastomeric nanocomposite is then inserted through the bifacial patch with the TMSPMA-treated side of the electrode positioned on the hydrogel surface. A patterned PDMS frame is subsequently placed over the assembly, and the conductive hydrogel is thermally cured in the frame, resulting in the creation of the integrated cardiac patch (Figure 2d and S1, Supporting Information).

2.2. Material Characterization of the Cardiac Patch

The conductive hydrogel functions as a bio-interfacial layer between nanocomposite interconnects and cardiac tissue, facilitating bidirectional electric current transport (Figure 3a). To prevent current leakage to nontarget tissues, noncontacted regions of the nanocomposite interconnects were encapsulated with SBS.^[38] To evaluate the electrochemical performance of the cardiac patch, we measured the electrochemical impedance and

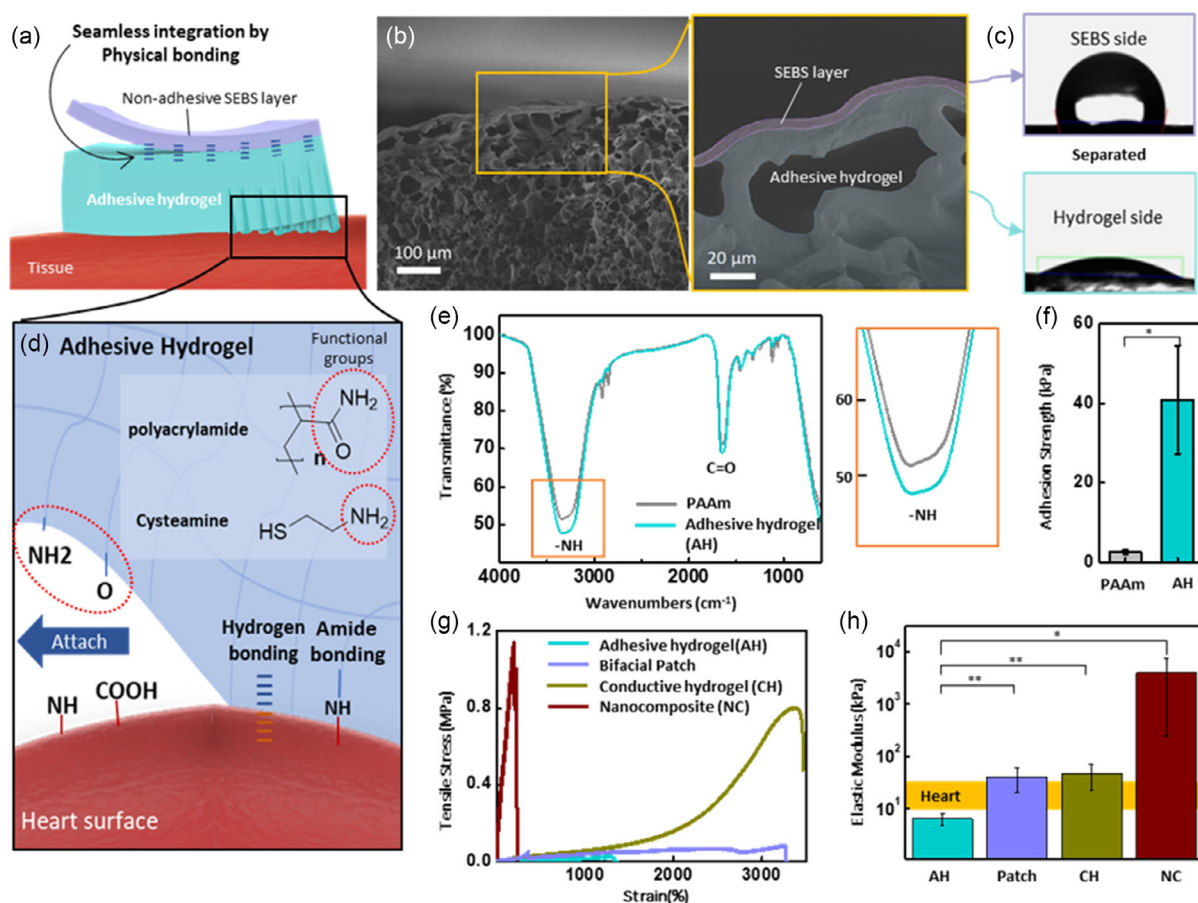


Figure 4. Mechanical characterization of the cardiac patch. a) Schematic illustration of bifacial patch and b) SEM images of seamless contact between SEBS layer and adhesive hydrogel. c) Contact angle analysis results of bifacial patch for each side. d) Illustration describing the tissue bonding mechanism of the adhesive hydrogel. e) FTIR spectrograms of polyacrylamide hydrogel (PAAm) and adhesive hydrogel. f) Lap-shear test results of PAAm ($n = 3$, 2.63 ± 0.48 kPa) and adhesive hydrogel ($n = 4$, 40.91 ± 13.6 kPa). g) Stress–strain curves and h) calculated elastic modulus of cardiac patch components: adhesive hydrogel ($n = 4$, 6.41 ± 1.6 kPa), bifacial patch ($n = 4$, 40.04 ± 19.3 kPa), conductive hydrogel ($n = 4$, 46.94 ± 23.9 kPa), and nanocomposite ($n = 3$, 3895 ± 3650 kPa). * $p \leq 0.05$, ** $p \leq 0.01$ for variables of significance.

conducted cyclic voltammetry on the conductive hydrogel. Across all frequency ranges, the impedance of the conductive hydrogel was lower than that of both the adhesive hydrogel and nanocomposites (Figure 3b). Additionally, the conductive hydrogel demonstrated a significantly higher charge storage capacity ($125.21 \pm 19.30 \text{ mC cm}^{-2}$) than the nanocomposite ($9.86 \pm 3.07 \text{ mC cm}^{-2}$) (Figure 3c,d). The charge injection capacity of the conductive hydrogel ($2.09 \pm 0.38 \text{ mC cm}^{-2}$) was also higher than that of the nanocomposite ($1.42 \pm 0.66 \text{ mC cm}^{-2}$) (Figure 3e,f), confirming the better electrochemical properties of the conductive hydrogel interface for transmitting bioelectric signals. The electrical property of the patch electrode, consisting of conductive hydrogel and elastomer, remained stable for 1 week under soaking to phosphate-buffered saline (PBS) solution, with little variation between electrodes (Figure S3, Supporting Information). Remarkably, the enhanced electrochemical performance was maintained even under repetitive mechanical deformation of $\approx 100\%$ strain (Figure 3g,h), suggesting its potential use on the dynamically beating heart.

The hydrogel patch exhibits seamless adhesion onto the curved and constantly moving epicardial surface while avoiding unwanted adhesion to the surrounding tissues (Figure 4a,b). The cardiac-facing side of the patch consists of cysteamine-modified

polyacrylamide hydrogel (Figure 4a, light blue; adhesive hydrogel). Owing to its material composition, this layer is hydrophilic (Figure 4c, bottom) and adhesive, facilitating seamless contact with the tissue surfaces. In contrast, the opposite side of the patch (Figure 4a, purple) is made of a hydrophobic (Figure 4c, top) nonadhesive SEBS layer, which prevents undesired adhesion to surrounding tissues. The top and bottom layers of the cardiac patch are seamlessly integrated through van der Waals interactions with this facile integration well maintained even after repeated stretching and recovering (Figure S4a,b, Supporting Information), ensuring a cohesive and functional structure.

The adhesiveness of the hydrogel interface is primarily attributed to the presence of cysteamine (Figure 4d), which forms hydrogen bonds and amide bonds with the tissue,^[39,40] enhancing adhesion (Figure 4d, red dotted circle). Fourier transform-infrared (FTIR) spectroscopy confirmed that the infrared peaks corresponding to the carbonyl group (1702 cm^{-1} ; Figure 4e)^[41] in polyacrylamide and the primary amine group (3300 cm^{-1} ; Figure 4e, orange box)^[42] in both polyacrylamide and cysteamine were amplified by the incorporation of cysteamine. These functional groups promote hydrogen bonding and amide bonding with tissue, leading to improving adhesion (Figure 4f). Adhesive property of this

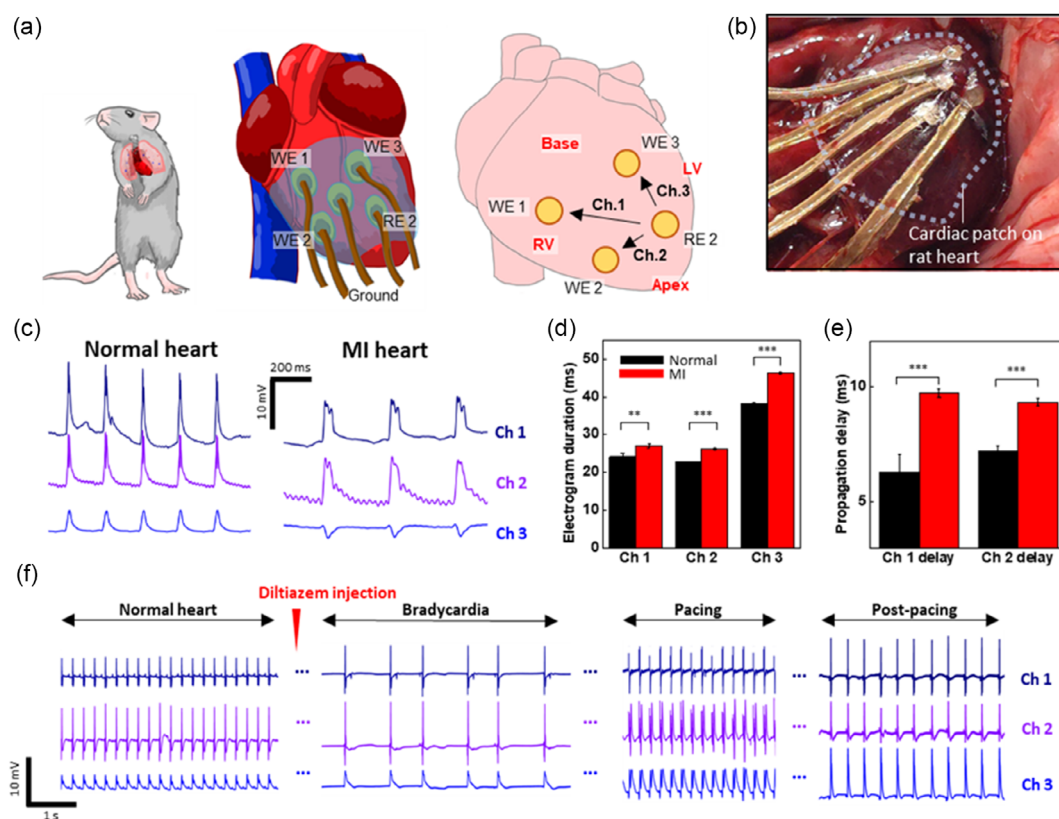


Figure 5. Cardiovascular application of cardiac patch for various disease models. a) Schematic illustration demonstrating electrode array mounted on the rat heart. The right image describes the anatomical location of each electrode. b) Optical image of the cardiac patch on the epicardial surface. c) 3-Channel epicardial electrogram of the normal heart (left) and myocardial infarction (MI) heart (right). d) QRS duration of the rat heart before (black) and after MI induction (red) ($n = 5$; $**p \leq 0.01$, $***p \leq 0.001$ for variables of significance). e) Propagation delay of the epicardium with Channel 3 at the left ventricle (LV) apex as the reference ($n = 5$; $***p \leq 0.001$ for variables of significance). f) 3-Channel epicardial electrogram of the heart under bradycardia, pacing, and post-pacing.

cysteamine hydrogel remained even after repeated stretching and recovering (Figure S4c, Supporting Information). Additionally, the low modulus and high stretchability of the adhesive hydrogel, which closely resemble those of cardiac tissue, ensure stable application of the cardiac patch on the soft, dynamic epicardial surface (Figure 4g,h).

2.3. Diagnostic and Therapeutic Applications of the Cardiac Patch

The potential in vivo cardiac application of the patch was evaluated using an animal model (Figure 5a). An epicardial patch comprising three-channel electrodes (including one reference

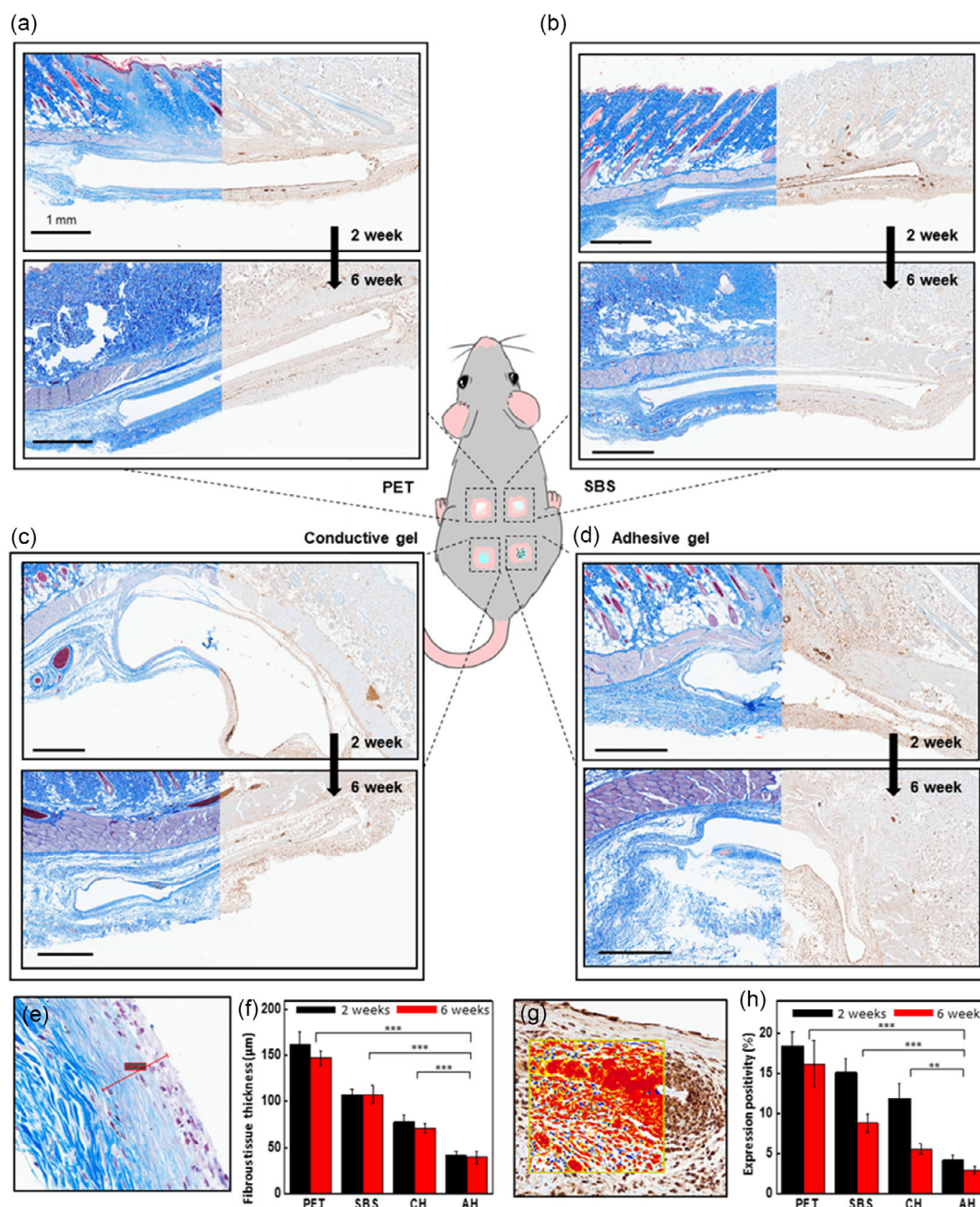


Figure 6. Biocompatibility test results of cardiac patch components. a–d) Masson's trichrome staining (left) and anti-CD 68 staining (right) of the subcutaneously implanted samples, which was 2 weeks (top) and 6 weeks (bottom) after implantation; PET (a), SBS (b), adhesive gel (c), and conductive gel (d), respectively. e) The microscope image explaining fibrous tissue measurements. f) Fibrous tissue thickness of each sample, 2 weeks (black) and 6 weeks (red) postimplantation ($n = 8$; $***p \leq 0.001$ for variables of significance). g) Microscope image explaining the positivity of the CD 68 expression. h) CD 68 expression positivity of each sample, 2 weeks (black) and 6 weeks (red) postimplantation ($n = 8$; $**p \leq 0.01$, $***p \leq 0.001$ for variables of significance).

electrode at the left ventricular apex and one ground electrode) was fabricated and attached to the rat ventricle (Figure 5b). Each channel recorded electrophysiological signals across a range of regions, such as the left ventricular apex, the right ventricular base, the right ventricular apex, and the left ventricular base. The adhesive hydrogel interface ensured a secure contact between the epicardial patch and the heart, even during dynamic relaxation and contraction of the heart.

First, the diagnostic capabilities of the cardiac patch were examined. The three-channel electrode array detected spatial differences in the propagation delay of epicardial tissue and morphological variations in the electrogram (Figure 5c). Moreover, electrophysiological differences between a normal heart and a pathological heart with myocardial infarction were observed. Following the ligation of the coronary artery, notable discrepancies in the activation duration and propagation delay were discerned.^[43] In particular, elongation of the QRS duration was detected across all three electrodes (Figure 5d). Additionally, left-to-right ventricular propagation delay and trans-ventricular propagation delay were monitored, revealing myocardial damage in the right ventricular tissue (Figure 5e).

To evaluate the therapeutic potential of the patch, ventricular bradycardia was induced in rats via diltiazem injection (20 mg kg^{-1}). This resulted in a significant increase in the R-R interval from 176.2 ± 0.1 to 755.4 ± 4.4 ms. Ventricular pacing (360 beats per minute (bpm), 1.5 V, 5 ms) was successfully administered with the bifacial cardiac patch to restore the heartbeat to a normal range (Figure 5f). After 20 min of continuous ventricular pacing, the natural R-R interval decreased to 443.5 ± 4.9 ms, indicating gradual recovery from but persistent bradycardia without the help of pacing from the cardiac patch. The therapeutic potential for resynchronization was also demonstrated. After inducing ventricular fibrillation (Figure S5, Supporting Information), over-drive pacing (600 bpm, 2 V, 5 ms) was applied to the rats experiencing sustained ventricular fibrillation. After 10 s of pacing, normal sinus rhythm was restored, successfully terminating the ventricular fibrillation. These results imply the potential of the epicardial patch as a therapeutic option for the monitoring and treatment of diverse cardiac rhythm disorders.

2.4. Biocompatibility of the Cardiac Patch

The long-term biocompatibility of the heterogeneous hydrogel patch was evaluated to confirm its potential for chronic application (Figure 6). Four types of circular patches (5 mm radius, 300 μm thickness) were fabricated by using different materials: PET (Figure 6a), SBS (Figure 6b), conductive hydrogel (Figure 6c), and adhesive hydrogel (Figure 6d). PET was employed as the negative control, whereas other materials were utilized to show the change of the interface layer. After sterilization, these patches were subcutaneously implanted in the dorsal skin of rats. The tissues with the implanted patches were harvested 2 and 6 weeks postimplantation.

The harvested tissues were stained with hematoxylin and eosin (H&E), Masson's trichrome (MT), and anti-CD68 antibody, with subsequent analysis of the fibrous tissue thickness (Figures 6e,f) and inflammation positivity (Figure 6g,h). The fibrous tissue thickness, evaluated with MT staining, was highest

in the tissues with PET patches and lowest in those with adhesive hydrogel, likely due to differences in mechanical modulus and compatibility of composing materials.^[44] It is notable that the conductive hydrogel showed a higher fibrous tissue thickness than the adhesive hydrogel, but still less than that observed for SBS. This may be attributed to the higher modulus of the conductive hydrogel, which is due to the presence of metallic nanomaterials. No significant differences in fibrous tissue thickness were observed between the 2 and 6 week time points across all samples (Figure 6f).

The assessment of inflammation positivity with anti-CD68 immunostaining revealed a similar trend. The highest levels of inflammatory cell infiltration were observed in tissues with PET patches, followed by those treated with SBS, conductive hydrogel, and adhesive hydrogel (Figure 6h). While there was a significant decrease in positivity at 6 weeks, the relative order remained consistent. Given that PET and SBS are commonly regarded as "biologically safe" materials,^[45,46] these findings indicate that the hydrogel interface layers, including conductive hydrogel and adhesive hydrogel, are also more mechanically and chemically biocompatible than control groups, supporting their suitability for use in chronically implantable materials.

3. Conclusion

We have developed a cardiac patch with a bifacial structure consisting of heterogeneous hydrogels, incorporating both conductive and adhesive hydrogels, along with elastomeric nanocomposite interconnects, which function as bioelectrodes. The layers of the cardiac patch are seamlessly integrated, and their softness, high stretchability, and excellent electrical performance enable secure attachment. These advantages allowed for the successful application of the cardiac patch for both diagnostic and therapeutic uses. While this study primarily focused on initial evaluations in preclinical models, specifically on epicardial interfacing, future research should incorporate long-term continuous interfacing across various organs. This can be achieved by integrating recent advances in electronics field, such as wireless power and data transmission, as well as AI-based data processing. These future developments will help advance the soft bioelectronic research toward clinical translation, not only for epicardial bioelectronics but also for a wide range of implantable electronic devices.

4. Experimental Section

Materials: Solvents such as ethyl alcohol, acetone, ethylene glycol, and toluene were purchased from Samchun Chemical (Korea). Poly(styrene-ethylene/butylene-styrene) (SEBS, H1041) was purchased from Asahi Kasei (Japan) and poly(styrene-butylene-styrene) (SBS, KTR 103) was obtained from Kumho Chemicals (Korea). Cysteamine was obtained from Tokyo Chemical Industry (Japan). The rest of the materials were purchased from Sigma Aldrich (USA). Ag–Au nanowires were synthesized as reported before.^[36]

Preparation of Hydrogels: *Adhesive Hydrogel:* The procedure to prepare the adhesive hydrogel involves adding 1.5 g of acrylamide, 50 μL of N,N'-methylenebis(acrylamide) solution (MBAA solution, 1.5 wt% in DI water), 50 μL of ammonium persulfate (APS) solution (10 wt% in DI water), and 5 μL of N,N,N',N'-Tetramethylethylenediamine (TEMED) sequentially into

10 mL of deionized (DI) water. Following these additions, 200 μL of a cysteamine solution (23 wt% in DI water) is introduced. The resulting mixture is then poured into a mold, covered, and cured in an oven at 60 °C for 30 min. This process is similar to that of creating a basic polyacrylamide hydrogel, except cysteamine is excluded. For the formation of a bifacial patch, a glass substrate is first coated with a poly(styrene-ethylene-butylene-styrene) (SEBS) solution (10 wt% in toluene) via spin-coating at 1000 rpm for 30 s. Subsequently, a PET adhesive (3 M, 200 μm thickness) is laser cut and applied to the coated substrate to form the mold.

Conductive Hydrogel: To prepare the conductive hydrogel, 0.4 g of acrylamide, 20 μL of MBAA solution, 20 μL of APS solution, and 2 μL of TEMED are sequentially added to 1 mL of deionized (DI) water. Lastly, 500 μL of cysteamine-modified Ag–Au nanowire solution (nanowire concentration: 50 mg mL^{-1} in DI) is added and mixed thoroughly. The resulting mixture is then poured into a mold, covered, and cured in an oven at 60 °C for 30 min.

Preparation of Nanocomposite Electrodes: To prepare the nanocomposite solution, 400 μL of Ag–Au nanowire solution (with a nanowire concentration of 40 mg mL^{-1} in a 1:3 mixture of ethyl alcohol and toluene) is combined with 400 μL of poly(styrene-butadiene-styrene) (SBS) solution (10 wt% in toluene) and 20 μL of hexylamine, followed by thorough mixing.

To form the encapsulated nanocomposite electrode, the SBS solution is first blade-coated onto a PDMS negative mold, to create the encapsulated nanocomposite electrode. The nanocomposite solution is then blade-coated on top of the SBS layer several times. Finally, another layer of SBS solution is blade-coated over the entire surface, excluding both ends. The method for TMSPMA treatment to enable integration with conductive hydrogel was referenced from a previous publication.

Surface Characterization: FTIR spectroscopy was performed to identify the surface functional groups of the hydrogel. The analysis was conducted using an FTIR spectrometer (VERTEX80v, BRUKER) installed at the National Instrumentation Center for Environmental Management (NICEM) at Seoul National University.

Contact angle analysis was performed to measure the hydrophilicity of the surface, following ASTM D5946. This analysis was conducted using Phoenix 300 Touch (S.E.O) installed at Korea Polymer Testing & Research Institute (KOPTRI, Korea).

Mechanical Characterization: A tensile test was performed to measure stretchability and calculate the modulus of each material using a universal testing machine (UTM, 1 kN load cell, Instron-5543) at a unidirectional speed of 50 mm min^{-1} , according to modified ISO 527.

A lap shear test was performed to evaluate the adhesion strength of the hydrogel using UTM, at a unidirectional speed of 5 mm min^{-1} . The porcine skin was stored in an acetic acid solution (0.5 wt% in DI) for 30 min and successively stored in a PBS solution for 20 min before the test. The hydrogel was synthesized with an adhesion area of 25 mm² (width $\times$ length: 5 $\times$ 5 mm) according to modified ASTM F2205-05 and then transferred to pretreated porcine skin then sandwiched with another porcine skin. Weight (about 100 g) was applied to the sandwiched sample to ensure stable adhesion between the hydrogel and porcine skin.

Electrical Characterization: For analysis of the electrochemical properties of the conductive hydrogel, Cr and Au were deposited (active site length $\times$ width: 1 $\times$ 1 cm) onto a glass substrate, and a hydrogel (of 1 mm thickness) was cured on the Au-coated side. Nanocomposites were cut into rectangles (length $\times$ width: 1.2 $\times$ 1 cm), and Cu foil was attached with a 2 mm interval of nanocomposite. Except for the active site, the remaining areas were encapsulated using polyimide tape to ensure complete coverage.

These measure samples were connected to a working electrode in three-electrode system consisting of Ag/AgCl reference electrode and platinum counter electrode. Measurements were performed by using electrochemical workstation (CHI660E, CH instruments) and conducted in the PBS solution at room temperature.

Electrical impedance was measured in the 10–10 000 Hz frequency range with a potential amplitude of 5 mV. Cyclic voltammetry (CV) curves were measured with a potential ± 0.5 V and scan rate of 0.1 V s^{-1} , then charge storage capacity value was calculated by integrating the CV curve. To measure charge injection capacity, biphasic pulses of 200 ms and

± 0.5 V were applied, and the output voltage and current were used to analyze the charge injection capacity.

For the cyclic test, conductive hydrogel (length $\times$ width $\times$ height: 2 $\times$ 2 $\times$ 1 cm) was cured on the TMSPMA-treated nanocomposite (length $\times$ width $\times$ height: 4 $\times$ 0.5 $\times$ 0.02 cm). The thickness of the nanocomposite was measured by a field-emission scanning electron microscope (SEM; SIGMA, ZEISS) installed at NICEM in Seoul National University. The hydrogel and nanocomposite ends were fixed and mounted on the uniaxial tensile stage (Jaeil Optical System, Korea) and Cu film was connected at each end with liquid metal. The repeated external strain of 100% was applied at a frequency of 0.05 Hz, and the line resistance was measured using a source meter (KEITHLEY 2450, KEITHLEY).

Biocompatibility Analysis: For the biocompatibility analysis, four types of polymer films (300 μm -thick) consisting of PET, SBS, conductive gel, and adhesive gel were prepared and cut into circular shape with 5 mm radius. The samples were sterilized with 70% ethanol and ultraviolet ray irradiation.

Eight-week-old male SD rats were purchased from OrientBio Inc. (Republic of Korea). After hair removal using clippers, the dorsal skin of rats were disinfected using ethanol and iodine swab. Then the animal was anesthetized in the isoflurane chamber. The anesthesia was maintained through the facial mask. After a 5-mm skin incision with iris scissor, the samples were implanted subcutaneously through the incised skin. The incision was sutured with 4-0 Black Silk suture (AILEE, Republic of Korea). Following 2 and 6 weeks of recovery, the skin tissue containing the sample was explanted after euthanizing the animals. The explanted tissues were sectioned via microtome (Leica, Germany) and stained with three types of dyes for histological analysis; H&E, Masson's trichrome, and anti-CD68 antibody.

Animal Preparation for Epicardial Experiments: Eight-week-old male Sprague–Dawley rats were purchased from OrientBio Inc (Republic of Korea). Following intratracheal intubation with a 18 G intravenous catheter, the animal was connected to a small animal ventilator (RoVent Jr., Kent Scientific, CT, USA), which was already connected to the isoflurane machine. Under stable anesthesia, the upper abdominal tissue, right under the diaphragm was incised transversely to expose the abdominal cavity. The thoracic cavity was exposed by incision of the diaphragm through the abdominal cavity and the epicardial patch was attached on the left ventricular region of the rat's heart, after removal of the pericardium.

Pathological Monitoring and Management: After preparation of the animal, myocardial infarction, ventricular bradycardia, and ventricular tachyarrhythmias were induced as follows. To induce myocardial infarction, the coronary artery was ligated with 7-0 Black Silk Suture (AILEE, Republic of Korea). The electrogram of the heart before and after the ligation was monitored through the data acquisition system (PowerLab, AD Instruments, New Zealand), which was connected with the epicardial patch. The bradycardia model was induced through the intraperitoneal injection of diltiazem (20 mg kg^{-1}). After inducing bradycardia, the ventricular pacing was followed with the identical DAQ device. The 1.5 V-electric stimulation with 360 bpm and 5 ms duration was delivered to maintain normal heartbeat under the bradycardia. Ventricular tachyarrhythmia was induced through the intraperitoneal injection of an excessive amount of norepinephrine (0.5 mg kg^{-1}). Following sustained tachyarrhythmia (duration of more than 30 s), overdrive stimulation (600 bpm, 2 V, and 5 ms duration) was delivered through the epicardial patch. A commercialized software (Lab Chart 8, AD Instruments, New Zealand) was used for the electrogram collection and analysis.

Statistical Analysis: Paired data are analyzed by two-sample *t* test. Analysis of variation (ANOVA) is used to evaluate the correlations among data from three or more groups. All statistical indications in the graph were presented as mean with standard deviation. Sample size (*n*) = 5 for experimental data, unless otherwise noted in figure caption. Statistical details (such as *p* value) are described in each figure caption.

Supporting Information

Supporting Information is available from the Wiley Online Library or from the author.

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Conflict of Interest

The authors declare no conflict of interest.

Author Contributions

Jeeyoung Kim: conceptualization (lead); data curation (lead); formal analysis (lead); investigation (lead); validation (lead); visualization (lead); writing—original draft (equal); writing—review & editing (equal). **Gi Doo Cha:** conceptualization (equal); data curation (equal); project administration (lead); supervision (lead); validation (lead); visualization (supporting); writing—original draft (lead); writing—review & editing (lead). **Minsung Kim:** conceptualization (supporting); investigation (equal); methodology (supporting); resources (supporting). **Seung-Pyo Lee:** conceptualization (supporting); funding acquisition (lead); methodology (equal); project administration (equal); writing—review & editing (supporting). **Sung-Hyuk Sunwoo:** conceptualization (equal); data curation (equal); formal analysis (equal); methodology (supporting); project administration (equal); supervision (equal); validation (equal); visualization (equal); writing—original draft (equal); writing—review & editing (equal). **Dae-Hyeong Kim:** funding acquisition (supporting); project administration (lead); supervision (lead); writing—review & editing (supporting). **Jeeyoung Kim, Gi Doo Cha, and Minsung Kim** contributed equally to this work.

Data Availability Statement

The data that support the findings of this study are available from the corresponding author upon reasonable request.

Keywords

cardiac interfacing, elastomeric nanocomposites, hydrogel nanocomposites, implantable device, soft bioelectronics

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