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Advances in microneedle electrodes for electromyography acquisition in sports rehabilitation

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Abstract: Electromyography (EMG) is widely used in sports rehabilitation to evaluate muscle activation, coordination, fatigue, and functional recovery, yet reliable recording remains limited by the electrode-skin interface during repeated motion and prolonged wear. Microneedle electrodes offer a distinct interface strategy by penetrating the stratum corneum and forming lower-impedance, more stable electrical contact than conventional wet or dry electrodes. This review discusses microneedle electrodes for EMG acquisition in sports rehabilitation from a design-to-application perspective. We first clarify the interface requirements of EMG recording in rehabilitation settings and the technical rationale for using microneedle interfaces. We then summarize material and structural design strategies in silicon-, polymer-, and metal-based systems, focusing on how they balance penetration capability, mechanical compliance, stretchability, conductivity, and recording stability. Fabrication routes are further examined in terms of material-structure-process coupling, followed by applications in static assessment, dynamic motion monitoring, and clinical rehabilitation evaluation. This review integrates interface requirements, design, manufacturing, and validation to provide a structured overview of microneedle EMG electrodes' capabilities and limitations for sports rehabilitation.

Key words: microneedle electrode; electromyography; sports rehabilitation; flexible electronics; wearable sensor

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1. Introduction

Electromyography (EMG) records the bioelectrical activity generated by skeletal muscles and provides important information on muscle activation, coordination, fatigue, and functional recovery. It has therefore been widely used in sports rehabilitation, exercise biomechanics, and neuromuscular function assessment (Fig. 1)[1–5]. In this review, sports rehabilitation is considered in an EMG-centered sense and includes athletic training recovery, rehabilitation training after injury or functional impairment, and rehabilitation-oriented human-machine interaction, such as exoskeleton-assisted motion control. In practical rehabilitation scenarios, however, the reliability of EMG recording is often limited by the electrode-skin interface rather than by the signal source alone. Repeated movement, sweating, skin deformation, and prolonged wear can all change the contact condition between the electrode and skin, leading to impedance fluctuation, motion artifacts, and signal degradation[3–7]. Conventional Ag/AgCl gel electrodes can reduce the initial interface impedance, but they require skin preparation and conductive gel, and their performance gradually decreases as the gel dries. Dry electrodes are more convenient to use, but their surface-contact mode is still affected by the high impedance of the stratum corneum and unstable mechanical contact under dynamic conditions[6–10]. Therefore, a key issue in EMG monitoring for sports rehabilitation is how to obtain low-impedance and stable electrical contact without increasing the burden of use.

Microneedle electrodes provide a distinct solution to this interface problem. By penetrating the stratum corneum, microneedles can bypass the main high-impedance barrier of the skin and establish more direct electrical coupling with the epidermal region than conventional surface electrodes[9–12]. Recent studies have shown that the performance of microneedle EMG electrodes is not determined by penetration alone. Material stiffness, needle geometry, conductive coatings, flexible substrates, stretchable interconnects, adhesion layers, and fabrication routes all affect whether the device can maintain stable recording during motion and long-term wear[13–17]. As a result, the field has moved from simply demonstrating that microneedles can record electrophysiological signals to understanding how interface design, mechanical adaptation, and manufacturing feasibility jointly determine their practical value. This transition makes microneedle electrodes a timely topic for review, especially in sports rehabilitation, where recording stability, comfort, repeatability, and usability must be considered together[18].

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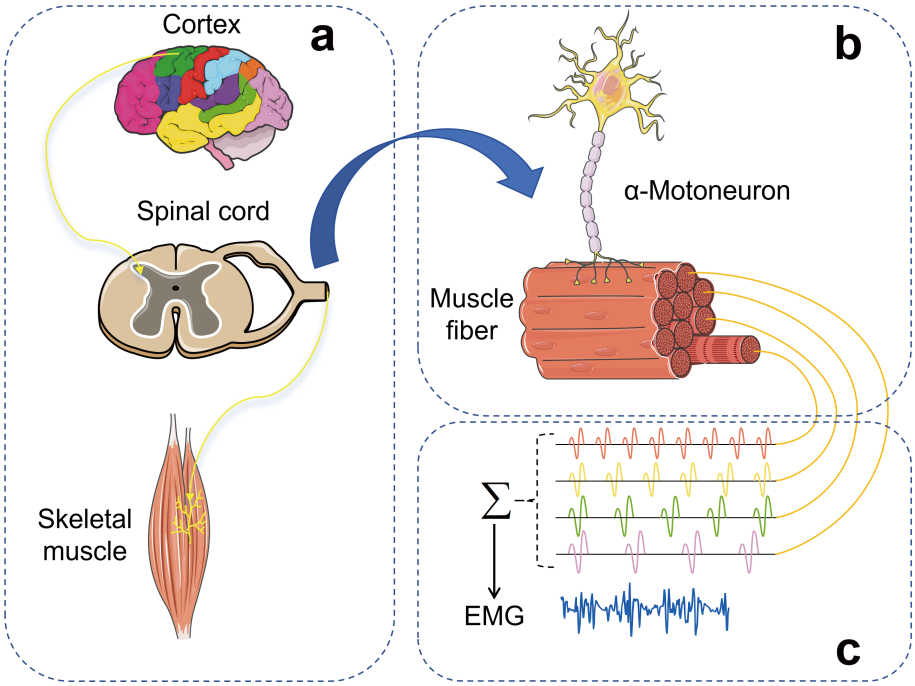

Fig. 1. (Color online) Physiological basis and acquisition principle of electromyographic signals. (a) Motor commands are transmitted from the cortex via the spinal cord to skeletal muscles. (b) α-motoneurons activate muscle fibers at the neuromuscular junction. (c) Spatial summation of asynchronous electrical activities from multiple muscle fibers (colored waveforms) forms the compound EMG signal (blue waveform). Yellow lines indicate the signal transduction pathway. Abbreviation: EMG, electromyography.

This review focuses on microneedle electrodes for EMG acquisition in sports rehabilitation and discusses the field from a design-to-application perspective. We first examine the acquisition requirements of EMG in rehabilitation settings and clarify why microneedle interfaces are technically relevant. We then summarize material and structural design strategies in silicon-, polymer-, and metal-based microneedle systems, with emphasis on how these systems balance penetration capability, mechanical compliance, stretchability, conductivity, and recording stability. Fabrication strategies are subsequently discussed in relation to material-structure-process coupling, followed by representative applications in static assessment, dynamic motion monitoring, and clinical rehabilitation evaluation. By linking interface requirements, device design, manufacturing constraints, and application validation, this review aims to provide a clear framework for understanding microneedle electrodes as practical EMG interfaces for sports rehabilitation rather than merely as alternative electrode formats.

2. EMG acquisition requirements and the rationale for microneedle electrodes

Reliable EMG acquisition in sports rehabilitation depends on both muscle electrical activity and the stability of the electrode-skin interface. EMG signals are weak, typically below 5 mV, and mainly distributed within 20–500 Hz; therefore, their recorded quality is highly sensitive to interface impedance, contact fluctuation, motion artifacts, and environmental noise. In repeated rehabilitation training, gait assessment, and human-machine interaction, the electrode must maintain stable electrical contact under sweating, skin deformation, and prolonged wear, while also remaining comfortable and biocompatible. Figs. 2(a) and 2(b) show that changes in skin cleanliness, sweat, and surface contamination can alter electrode contact conditions and lead to different impedance stability under comparable skin conditions. These interface-related factors make EMG acquisition in sports rehabilitation different from short-term static recording.

Conventional electrodes address the interface problem only partially. Ag/AgCl gel electrodes reduce the initial electrode-skin impedance through conductive gel, but they require cleaning, shaving, or pretreatment of the stratum corneum. During prolonged use, gel drying can increase impedance, degrade signal quality, and cause skin irritation. Dry electrodes avoid gel dependence and are more convenient for wearable monitoring, but their surface-contact mode still needs to bridge the high-impedance stratum corneum. As a result, impedance drift and motion artifacts are more likely to occur during sweating, skin deformation, and vigorous exercise[8–10]. Thus, the central limitation of conventional wet and dry electrodes is not simply insufficient conductivity, but the difficulty of maintaining a stable low-impedance interface under dynamic rehabilitation conditions.

Microneedle electrodes are technically relevant in this setting because they address the interface problem through a different contact mechanism. By penetrating the stratum corneum, microneedles form more direct electrical coupling with the viable epidermis, bypassing the main high-impedance barrier and lowering the electrode-skin interface impedance. The equivalent circuit models in Fig. 2(d) also indicate that, compared with gel electrodes and surface-type electrically conductive adhesive electrodes, flexible microneedle electrodes can simplify the impedance pathway and reduce the overall equivalent impedance. Meanwhile, microscale insertion improves contact stability. The technological evolution summarized in Fig. 2(c) shows that microneedle electrodes are shifting from rigid structures designed mainly for

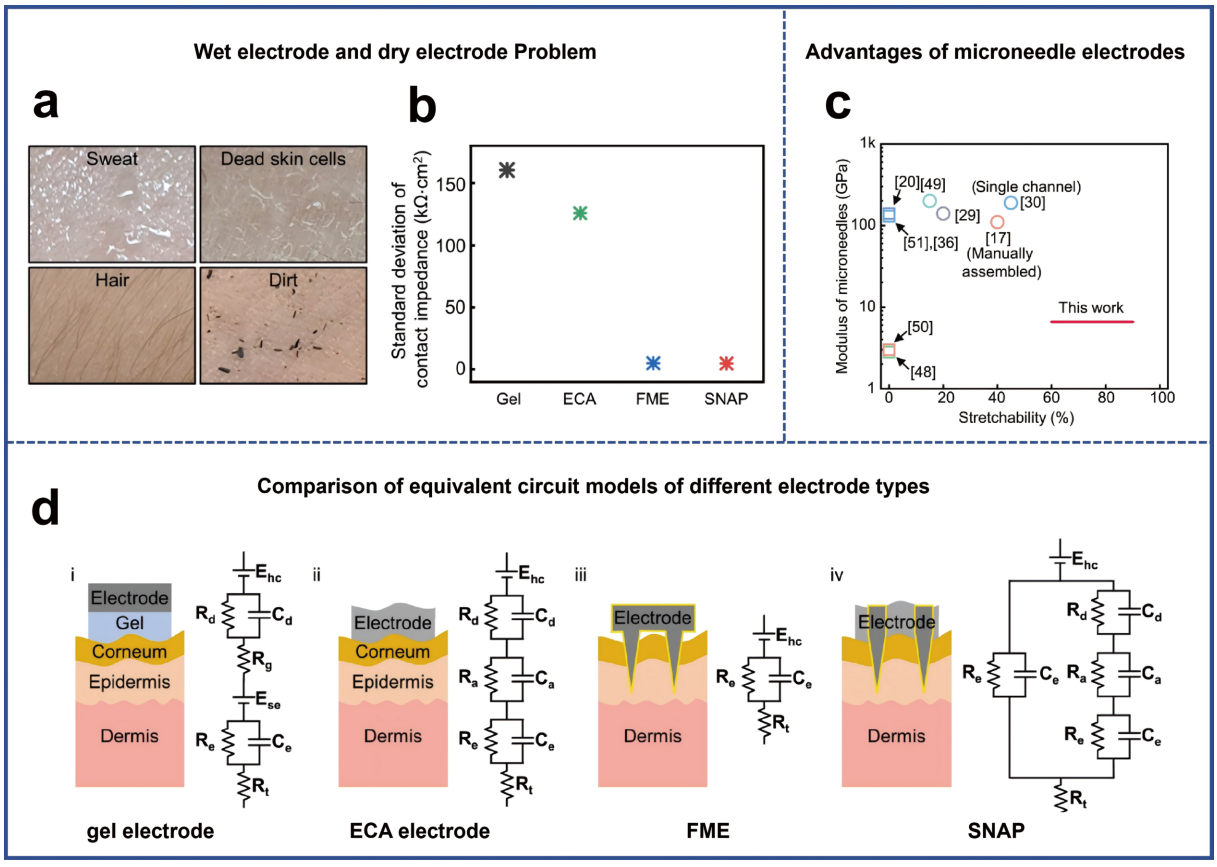

Fig. 2. (Color online) Advantages and working principles of microneedle electrodes. (a) Limitations of conventional electrodes. Optical images show different skin conditions. (b) Comparison of standard deviations of electrode impedance under clean skin conditions. (c) Evolution of microneedle electrode technologies. (d) Equivalent circuit models of different electrode-skin interfaces, (i) gel electrode, (ii) electrically conductive adhesive (ECA) electrode, (iii) flexible microneedle electrode (FME), and (iv) stretchable microneedle adhesive patch (SNAP). (a), (b), (d) Adapted with permission from Ref. [14]. Copyright 2024, The Author(s), some rights reserved; exclusive licensee American Association for the Advancement of Science. (c) Adapted with permission from Ref. [15]. Copyright 2024, The Author(s), some rights reserved; exclusive licensee American Association for the Advancement of Science.

impedance reduction toward integrated interfaces that combine penetration, adhesion, flexible interconnection, and skin conformity. Nevertheless, the use of microneedles also introduces concerns that are less critical for conventional surface electrodes. Although properly designed microneedles can reduce pain by limiting penetration depth and avoiding deep tissue injury, skin irritation, local discomfort, repeated insertion, and long-term safety still need to be considered in rehabilitation monitoring[9, 11, 12]. For long-term use, adhesion durability, skin compatibility, sweat-induced changes in impedance and adhesion, and repeated wearing or insertion are particularly important because they directly affect interface stability and user acceptance[8, 9, 13, 14]. This evolution also clarifies the key design trade-off introduced by microneedle electrodes. The needle body must be sufficiently stiff to penetrate the stratum corneum without buckling, whereas the overall device must remain compliant enough to accommodate skin deformation without causing discomfort or unstable contact. Therefore, the main question is not only whether microneedle electrodes can record EMG signals, but whether they can simultaneously provide reliable penetration, low impedance, reduced motion artifacts, long-term comfort, and recording stability in dynamic rehabilitation environments. These requirements provide the basis for evaluating the material selection, structural design, and fabrication strategies discussed in the following sections[9, 13, 14].

3. Material and structural design of microneedle electrodes for EMG

The preceding section shows that EMG acquisition in sports rehabilitation depends not only on electrode conductivity, but also on whether the electrode can maintain reliable penetration, low interface impedance, stable contact, reduced motion artifacts, and wearing comfort at a dynamic skin interface. For microneedle electrodes, these requirements create a basic design trade-off: the needle body must be stiff enough to penetrate the stratum corneum without buckling, whereas the whole device must remain compliant enough to accommodate muscle contraction, skin stretching, and joint movement. In addition, the penetration depth should be controlled to avoid pain, bleeding, or unnecessary tissue damage. The high stiffness required for insertion may also lead to modulus mismatch with soft skin tissue, resulting in stress concentration, discomfort, or unstable contact during motion[19]. Therefore, compliant substrates, stretchable interconnects, and strain-isolation structures are needed to accommodate skin stretching, compression, and shear deformation. The bonding between the microneedles and the substrate, the anti-detachment capability at the skin interface, and the mechanical stability during repeated insertion or long-term wearing are also important for rehabilitation applications. Therefore, material selection and structural design

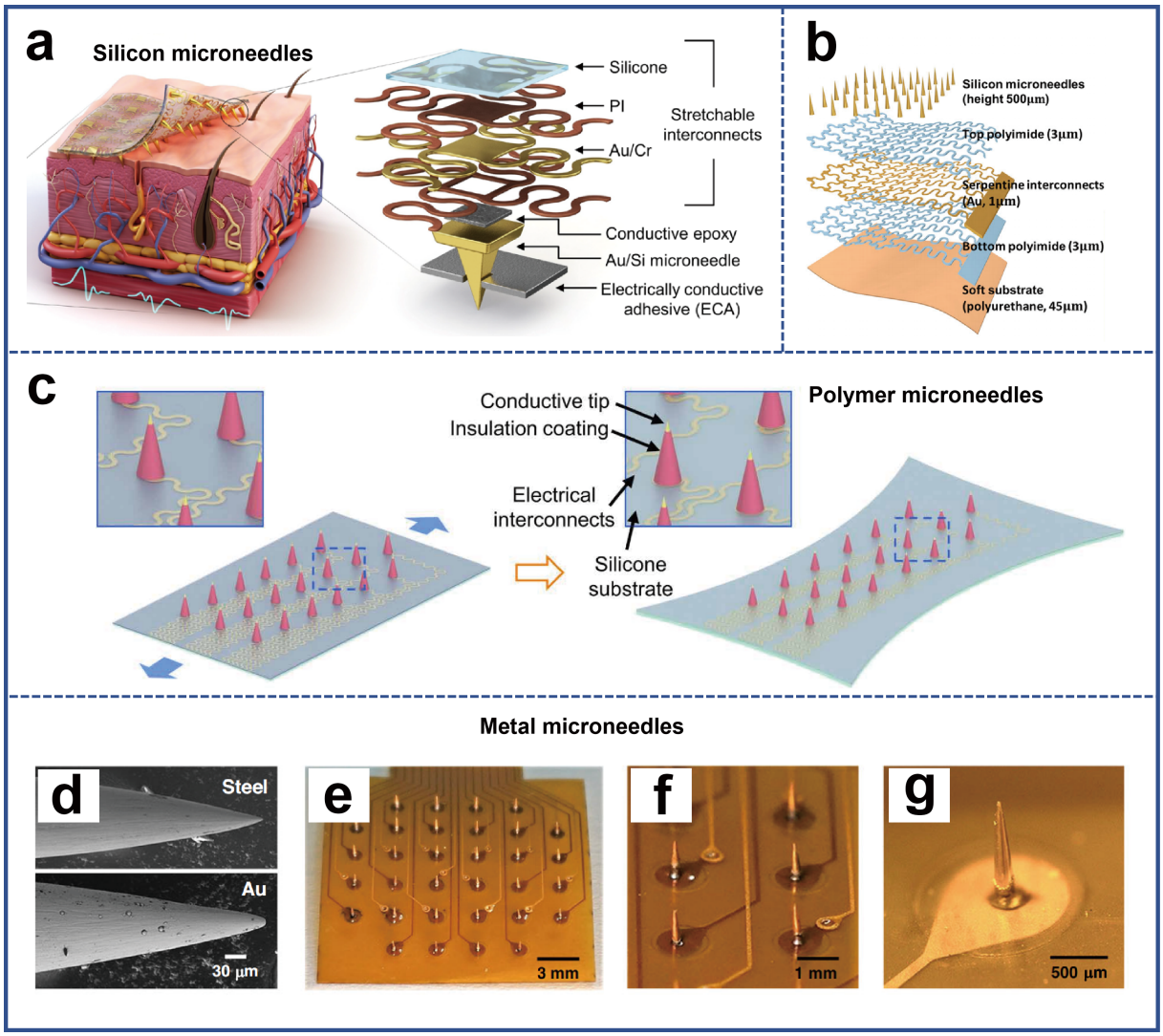

Fig. 3. (Color online) Structure and fabrication of microneedle electrodes with different materials. (a) and (b) Silicon-based microneedle electrodes, (a) skin integration schematic and exploded multilayer structure; (b) three-dimensional layered architecture with material and thickness annotations. (c) Polymer microneedle electrode fabrication process and magnified structural details. (d)–(g) Metal microneedle electrodes, (d) SEM comparison before and after Au plating; (e)–(g) optical images from different views. Abbreviations: PI, polyimide; SEM, scanning electron microscopy. (a) Adapted with permission from Ref. [14]. Copyright 2024, The Author(s), some rights reserved; exclusive licensee American Association for the Advancement of Science. (b) Adapted from Ref. [13] under the terms of the Creative Commons Attribution 4.0 International License. Copyright 2023, The Author(s), published by Springer Nature. (c) Adapted with permission from Ref. [15]. Copyright 2024, The Author(s), some rights reserved; exclusive licensee American Association for the Advancement of Science. (d)–(g) Adapted from Ref. [16] under the terms of the Creative Commons Attribution 4.0 International License. Copyright 2024, The Author(s), published by Springer Nature.

should be considered together. The needle material mainly determines penetration capability and electrical properties, while the substrate, interconnect, adhesion, and anchoring structures determine whether the interface remains stable during motion.

Silicon-based microneedles are mainly advantageous in penetration reliability and geometric precision. The high modulus of silicon and its compatibility with microfabrication allow sharp tips, controlled needle height, and uniform array geometry, which are useful for reducing variation in stratum corneum penetration. For example, Kim *et al.* [14] developed a stretchable microneedle adhesive patch using Au-coated silicon microneedles as the penetrating units. The microneedles were approximately 200 μm in height with a tip diameter below 5 μm, allowing penetration of the stratum corneum while avoiding noticeable pain. As shown in Fig. 3(a), the device further integrated flexible encapsulation, serpentine interconnects, and an electrically conductive adhesive layer for skin integration. Ji *et al.*[13] used 500 μm silicon microneedles to construct a skin-integrated stretchable silicon microneedle electrode (SSME), in which the microneedle roots were partially wrapped by PI/metal/PI thin films. As shown in Fig. 3(b), this plant-thorn-inspired root-embedding structure improved the bonding strength between the microneedles and the flexible substrate. Silicon-based systems are suitable for EMG acquisition requiring high penetration consistency and stable geometry. However, the brittleness and high modulus of silicon also increase the mechanical mismatch with soft skin tissue; flexible substrates, root reinforcement, and strain-isolation structures are therefore necessary to improve dynamic reliability.

Polymer-based microneedles emphasize mechanical compliance and long-term wearing comfort. Low-modulus polymers favor conformity to curved skin, but excessive softness

can weaken penetration ability. Therefore, recent designs have often adopted a layered strategy consisting of a high-modulus polymer needle body, a flexible substrate, and stretchable interconnects. Zhao *et al.*[15] fabricated customizable microneedle arrays using PI-2610, with Cr/Au conductive layers, Parylene C insulation, and an Ecoflex substrate, achieving 60%–90% device stretchability for local intramuscular EMG recording. The polymer microneedle structure shown in Fig. 3(c) illustrates this design logic: the needle body provides penetration and local recording, whereas the flexible substrate and interconnects accommodate deformation. Similarly, flexible polyimide microneedle arrays, gold-nanowire-embedded PI microneedles, silk fibroin/polyurethane composite microneedles, conductive fabric-backed microneedles, and biodegradable hydrogel microneedles further extend polymer systems toward low impedance, bending durability, garment integration, biocompatibility, and long-term safety[20–24]. Overall, polymer microneedles are more suitable for low-burden and long-duration rehabilitation monitoring, but their intrinsic conductivity and penetration strength are often insufficient. Conductive coatings, metal layers, conductive polymers, or conductive nanofillers are usually required to compensate for these limitations.

Metal-based microneedles provide intrinsic conductivity, high penetration strength, and potential for multichannel recording. Compared with silicon and polymer systems, metal needles can serve as both mechanical insertion units and conductive pathways. Liu *et al.*[16] constructed a 32-channel stainless-steel/Au-modified microneedle array that can record weak electrophysiological signals, including EEG, ECG, and EMG, within a small area. The tip morphology before and after Au coating and the array configuration shown in Figs. 3(d)–3(g) indicate that metal needles can provide both mechanical support and a low-impedance conductive interface. Other metal-based approaches, such as low-melting-point Bi–In–Sn alloy microneedles and ultrathin flexible metal microneedles, attempt to reduce forming temperature, simplify fabrication, or reduce excessive rigidity[25, 26]. These systems are attractive for multichannel spatial acquisition and high-resolution recording, especially when channel density and conductivity are important. Their main limitation is mechanical mismatch with soft tissue, which may cause discomfort, unstable contact, or tissue irritation during prolonged wear. Thin substrates, flexible circuits, local insulation, and optimized array geometry are therefore important for improving their usability in sports rehabilitation.

Beyond the needle material itself, structural design directly determines whether a microneedle electrode can maintain a stable interface during motion. Serpentine interconnects are key structures for releasing tensile strain from the skin. As shown in Figs. 4(a) and 4(b), the SMNEA developed by Zhao *et al.*[15] used arc-shaped PI/Au interconnects to absorb in-plane deformation during stretching, allowing the microneedle recording sites to remain relatively stable. In the silicon-based SSME, serpentine interconnects and partially PI-encapsulated roots jointly form a rigid-island/flexible-bridge structure. As shown in Figs. 4(c) and 4(d), external strain is mainly accommodated by the interconnects, while the silicon microneedles maintain their penetration and recording functions.

Anchoring and adhesion structures further improve interface stability under shear deformation, sweat, and repeated movement. The Bioinspired root-encapsulation structure in Fig. 4(e) can significantly reduce shear stress at the microneedle roots and improve resistance to tilting and detachment[13]. Barbed microneedles enhance mechanical anchoring and reduce micromotion at the interface[27]. Between low impedance and comfort, electrically conductive adhesive layers, PEDOT:PSS coatings, and modulus-adjustable polymers offer finer control. The SNAP system uses a conductive adhesive layer to provide adhesion, conduction, and buffering simultaneously in Figs. 4(f) and 4(g)[14]. Other strategies, including Miura-ori structures for sweat ventilation and bidirectional bending, modulus-adjustable microneedles that soften after insertion, and 3D-printed arrays with tunable needle length and spacing, further indicate that interface stability requires the integration of penetration, adhesion, deformation accommodation, and tissue compatibility[28–30].

Overall, silicon, polymer, and metal microneedles offer different design advantages. From the perspective of comfort, penetration ability, and recording stability, their trade-offs are distinct. Silicon microneedles provide penetration consistency and geometric accuracy, making them suitable for dynamic motion recording and rehabilitation training scenarios that require reliable penetration and stable needle geometry, especially when combined with soft substrates, serpentine interconnects, or adhesive layers. However, their high stiffness may reduce comfort and increase tissue mismatch if mechanical buffering is insufficient. Polymer microneedles exhibit enhanced compliance, comfort, and long-term wearability, rendering them particularly well-suited for prolonged wearable monitoring and repeated rehabilitation assessments. However, a key limitation lies in their often insufficient intrinsic conductivity and insertion strength, which typically necessitates the incorporation of conductive coatings, conductive fillers, or high-modulus polymer matrices for the needle body. Metal microneedles support high conductivity, mechanical strength, and multichannel recording, which makes them advantageous for high-density or spatially selective EMG acquisition. Nevertheless, their relatively high stiffness requires careful integration with flexible circuits or compliant substrates to reduce discomfort during prolonged use.

4. Fabrication strategies and material-structure-process coupling

Material and structural designs can be transformed into stable EMG acquisition performance only through suitable fabrication strategies. For microneedle electrodes, processing is not simply the final step of device fabrication; it determines needle geometry, array consistency, flexible integration, channel expansion, and application reliability from the outset. In other words, the penetration, low impedance, stretchability, and long-term stability discussed in Section 3 must all be realized during manufacturing. To help readers compare different fabrication routes more directly, Table 1 summarizes the main processing strategies, material types, advantages, limitations, and representative references for microneedle electrodes.

Silicon-based microneedles mainly rely on MEMS and semiconductor microfabrication routes. KOH wet etching, deep reactive ion etching (DRIE), and subsequent metallization can

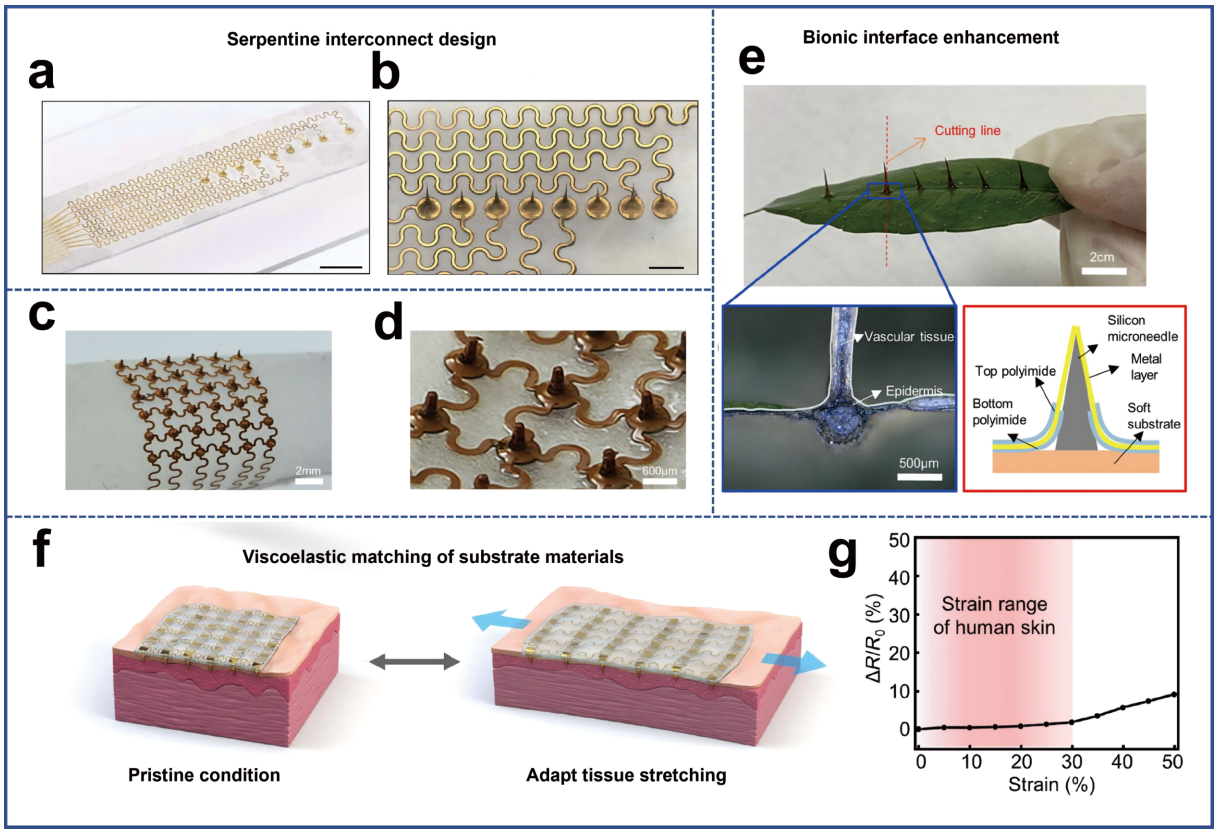

Fig. 4. (Color online) Stretchable design and biomechanical interfacing of microneedle electrodes. (a)–(d) Optical images of serpentine interconnect designs for polymer-based (a) and (b) and silicon-based (c) and (d) microneedle electrodes. (e) Bioinspired interface design and the corresponding cross-sectional view of tissue embedding. (f) Schematic illustration of viscoelastic matching between the substrate and skin. (g) Relative resistance changes as a function of tensile strain; the red shaded region indicates the physiological strain range of human skin. (a) and (b) Adapted with permission from Ref. [15]. Copyright 2024, The Author(s), some rights reserved; exclusive licensee American Association for the Advancement of Science. (c)–(e) Adapted from Ref. [13] under the terms of the Creative Commons Attribution 4.0 International License. Copyright 2023, The Author(s), published by Springer Nature. (f) and (g) Adapted with permission from Ref. [14]. Copyright 2024, The Author(s), some rights reserved; exclusive licensee American Association for the Advancement of Science.

Table 1. Comparison of representative fabrication strategies for microneedle electrodes.

Material/system	Fabrication strategy	Advantages	Limitations	Refs.
Silicon/metal-coated Si	KOH etching or DRIE; metallization	High precision; reliable penetration	High cost; brittle; difficult transfer	[13, 14, 31]
Polymer/PLGA or magnetorheological polymer	Thermal drawing; magnetic self-assembly; MRDL	Low cost; rapid forming	Limited precision; coating required	[32–35]
Polymer/PI, epoxy, SU-8	Micromolding; replica molding; laser-assisted molding	Tunable geometry; flexible integration	Mold-dependent; multi-step process	[15, 20, 28, 37]
Functional polymer/composite microneedles	Conductive coating; composite modification	Good comfort; biocompatible	Coating uniformity; stability concern	[17, 21–24, 29]
Polymer/resin microneedles	3D printing; conductive coating	Fast prototyping; customizable	Rough surface; limited resolution	[30]
Metal/stainless steel or Au-modified metal	Laser cutting; chemical etching; FPC assembly	Conductive; strong; multichannel	Assembly variation; tissue mismatch	[16, 40–43]
Metal-coated barbed microneedles	Barbed microfabrication; metal coating	Strong anchoring; anti-detachment	Complex geometry; reproducibility issue	[27]
Metal/3D-printed stainless steel	Direct metal laser sintering	One-step forming; customizable	Rough surface; post-treatment needed	[39]
Metal/alloy or ultrathin metal	Alloy micromolding; micro/nano-electroforming	Low temperature; flexible metal	Mold dependence; uniformity issue	[25, 26]

produce microneedle structures with controllable tip dimensions and array spacing, making them suitable for high-consistency penetration and low-impedance interfaces. As shown in Fig. 5(a), silicon-based routes usually form microneedle geometry through top-down etching, followed by PI encapsulation, metal interconnection, and transfer onto flexible substrates for skin conformity. The SSME and SNAP systems both follow this logic: silicon microneedles provide penetration accuracy, while serpentine interconnects and soft substrates compensate for mechanical rigidity[13, 14]. The silicon microneedle array on a flexible substrate reported by Wang *et al.*[31] also shows the advantages of silicon processing in impedance control and long-term stability. However, these processes rely on cleanroom equipment, involve complex steps

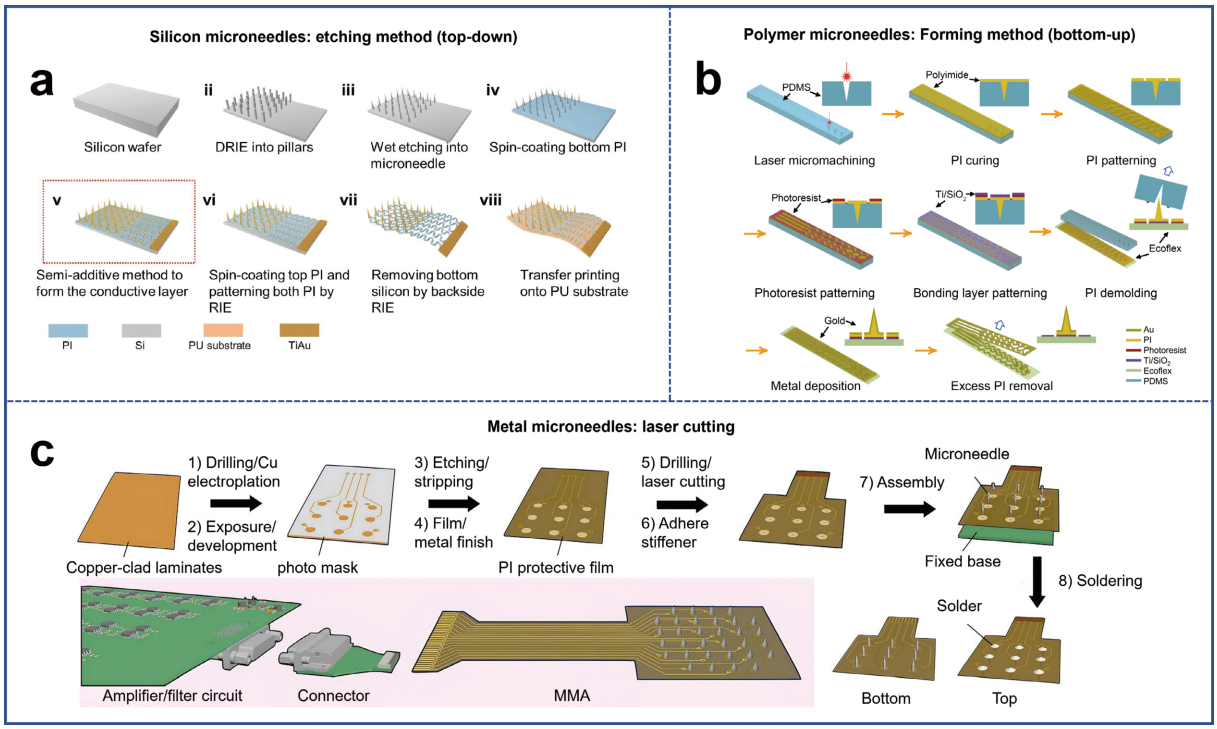

Fig. 5. (Color online) Microfabrication strategies for microneedle electrode arrays. (a) Top–down silicon microfabrication process using DRIE and anisotropic wet etching to form microneedle geometry, followed by polyimide encapsulation and metallization and transfer onto a flexible substrate. (b) Bottom–up polymer microfabrication approach based on laser micromachining, replica molding, microfabrication, and transfer printing. (c) Laser micromachining process for metal microneedles and printed circuit board processes for flexible circuit fabrication. (a) Adapted from Ref. [13] under the terms of the Creative Commons Attribution 4.0 International License. Copyright 2023, The Author(s), published by Springer Nature. (b) Adapted with permission from Ref. [15]. Copyright 2024, The Author(s), some rights reserved; exclusive licensee American Association for the Advancement of Science. (c) Adapted from Ref. [16] under the terms of the Creative Commons Attribution 4.0 International License. Copyright 2024, The Author(s), published by Springer Nature.

and high costs, and the brittleness of silicon makes subsequent transfer, packaging, and stretchable integration particularly challenging.

Polymer-based microneedle processes emphasize flexibility, low cost, and structural customization. Thermal drawing, micromolding, magnetorheological drawing lithography, 3D printing, and replica molding can all be used to form needle structures. As shown in Fig. 5(b), bottom–up forming routes are more readily integrated with flexible substrates, fabric backings, or elastomeric interconnects. PLGA thermal drawing and magnetorheological polymer forming have relatively simple equipment requirements and low cost, making them suitable for rapid fabrication and flexible attachment[32–36]. Micromolding, integrated PI molding, and laser-mold replication provide better control over array morphology, needle length, and addressable structures[15, 20, 28, 37]. More recently, PEDOT:PSS modification, silk-fibroin mesoscopic engineering, gold-nanowire-embedded PI, shape-memory polymers, and hydrogel microneedles have broadened the design space of polymer microneedles in low impedance, biocompatibility, and long-term comfort[17, 21–24, 29, 30, 38]. Their limitations are that needle strength and conductivity are usually insufficient, and the uniformity of conductive coatings, nanofiller distribution, and batch-to-batch reproducibility require careful control.

Metal-based microneedle processes serve high conductivity, high penetration strength, and multichannel expansion. Laser cutting, chemical etching, direct metal laser sintering (DMLS), low-melting-point alloy vacuum micromolding, micro/nano-electroforming, and FPC post-assembly can all be used to construct metal microneedle arrays. As shown in Fig. 5(c), metal microneedles are often fabricated separately from flexible circuits and then integrated, which facilitates channel routing and array expansion. The 32-channel stainless-steel/Au microneedle array developed by Liu *et al.*[16] demonstrates the advantage of FPC integration for multichannel acquisition. DMLS printing, low-melting-point alloy molding, and ultrathin electroformed microneedles respectively reduce the fabrication barrier of metal microneedles through rapid customization, low-temperature forming, and flexible metallization[25, 26, 39]. In addition, curved metal microneedles, PDMS-embedded stainless-steel arrays, Parylene-coated microneedles, and barbed microneedles provide different process routes for long-term sEMG, intramuscular EMG, and dynamic anti-detachment applications[27, 40–43]. Nevertheless, metal processing still faces issues such as surface roughness, assembly inconsistency, tissue-mechanics mismatch, and complex scalable packaging.

Overall, fabrication routes define the practical boundary of microneedle EMG electrodes. Silicon processes provide the highest geometric precision and reproducibility, but their cost and brittleness complicate flexible integration. Polymer processes are more compatible with soft, wearable, and personalized devices, but they require better control over conductivity, needle strength, and batch consistency. Metal processes offer conductivity, mechanical strength, and multichannel recording capability, but they need careful packaging and mechanical design to reduce tissue mismatch. These differ-

ences directly affect application performance: tip precision determines penetration reliability, flexible integration determines resistance to motion artifacts, channel fabrication determines spatial resolution, and process consistency determines scalability. The following section discusses how these material-structure-process choices are reflected in static assessment, dynamic motion monitoring, and clinical rehabilitation evaluation.

5. Application progress in sports rehabilitation

The value of microneedle EMG electrodes should ultimately be evaluated by whether they improve recording reliability in sports rehabilitation tasks. Compared with conventional surface electrodes, their application advantages are mainly reflected in three aspects: improved spatial selectivity for local muscle assessment, better motion robustness during dynamic monitoring, and more stable interfaces for repeatable rehabilitation evaluation.

For static rehabilitation assessment, microneedle electrodes first demonstrate improved spatial selectivity. Isometric contraction, posture maintenance, and local muscle activation analysis require electrodes to distinguish target muscle activity from neighboring muscles as much as possible and to reduce volume conduction and crosstalk. The comparison between intramuscular EMG and surface EMG in Figs. 6(a) and 6(b) indicate that customizable microneedle arrays can obtain more localized muscle electrical activity distributions through local insertion and selectively exposed recording regions[15]. Curved metal microneedle arrays and 3D-printed stainless-steel microneedle electrodes further show that high-stiffness needles and controllable geometry can improve the local selectivity and signal stability of sEMG[39, 40]. When higher channel numbers are required, FPC-integrated multichannel metal microneedle patches can realize multipoint recording within a limited area, providing richer spatial information for muscle coordination analysis, local activation mapping, and movement recognition[16, 30].

During dynamic monitoring and motion assistance, the advantages of microneedle electrodes are more strongly reflected in motion robustness and long-term stability[44, 45]. Rehabilitation training often involves skin stretching, sweat, joint movement, and microsliding of electrodes, which can easily cause impedance drift and motion artifacts in conventional surface electrodes. Figs. 6(c)–6(e) presents the application of the SNAP system in dynamic electrophysiological sensing and exoskeleton control. By combining Au-coated silicon microneedles, an electrically conductive adhesive layer, and stretchable serpentine interconnects, this system maintains stable skin coupling without skin preparation. EMG signals acquired during movement can be used for motion intention recognition and can drive an exoskeleton to perform assisted actions, demonstrating the potential of microneedle electrodes in rehabilitation training and human-machine interaction[14]. In parallel, the stretchable silicon microneedle electrode developed by Ji *et al.*[13] uses PI root encapsulation and serpentine interconnects to distribute skin deformation, thereby improving EMG recording stability during long-term motion scenarios. Stretchable intramuscular microneedle arrays, Miura-ori microneedles, fabric-backed microneedles, silk-fibroin composite microneedles, nanowire-embedded PI microneedles, ultrathin flexible metal microneedles, and modulus-adaptive microneedles improve dynamic monitoring reliability from the perspectives of stretch adaptation, sweat ventilation, garment integration, low motion noise, bending durability, flexible metallization, and tissue-mechanics matching[21–23, 26, 28, 29] cost processes such as thermal drawing, magnetization-induced self-assembly, and magnetorheological drawing lithography further provide supplementary routes for scalable wearable EMG monitoring[32, 33, 35].

In clinical rehabilitation assessment, the value of microneedle electrodes lies not only in improving signal quality, but also in reducing the burden of operation and supporting continuous, repeatable functional evaluation. The facial palsy monitoring system shown in Fig. 6(f) uses PEDOT:PSS-modified flexible microneedle arrays to record EMG from the frontalis, orbicularis oculi, orbicularis oris, and mentalis muscles. It can compare muscle activity differences between the affected and healthy sides and can be used for intraoperative nerve stimulation response and follow-up assessment[17]. In terms of rehabilitation translation, the SNAP system demonstrates the feasibility of using microneedle EMG as a closed-loop input for exoskeleton control[14]. In addition, microneedle arrays that combine EMG acquisition and sensory feedback, reconfigurable multimodal microneedle systems, and facial EMG-based silent speech interfaces expand the application boundary toward prosthetic feedback, neuromuscular injury assessment, and facial function assistance[46–48]. These studies indicate that the role of microneedle electrodes in sports rehabilitation is not limited to replacing conventional electrodes; rather, by providing a more stable interface, better structural adaptation, and more convenient system integration, they offer continuous and reliable EMG information for rehabilitation assessment, training monitoring, and functional recovery evaluation.

Overall, applications of microneedle EMG electrodes in sports rehabilitation can be understood through three linked requirements. Static assessment emphasizes spatial selectivity and localized recording; dynamic monitoring emphasizes motion robustness and long-term interface stability; clinical rehabilitation evaluation emphasizes repeatability, usability, and functional relevance. These application demands correspond directly to the material, structural, and fabrication strategies discussed above. High-stiffness needles and exposed recording regions support localized acquisition; stretchable interconnects, adhesive layers, and compliant substrates reduce motion artifacts; and scalable, reproducible fabrication determines whether these devices can move from laboratory demonstrations toward practical rehabilitation monitoring. Among these directions, dynamic motion monitoring and rehabilitation training assistance appear closest to practical use, because microneedle EMG electrodes have already been demonstrated in motion-state recording and exoskeleton-assisted control[13, 14]. Static assessment and localized muscle activity mapping show clear advantages in spatial selectivity, especially with customizable or multichannel microneedle arrays, but standardized comparisons across muscle groups, electrode configurations, and rehabilitation tasks are still needed[15, 16, 30, 39, 40]. Clinical rehabilitation evaluation has strong translational relevance, as shown by facial palsy monitoring and several rehabilitation-oriented interface demonstra-

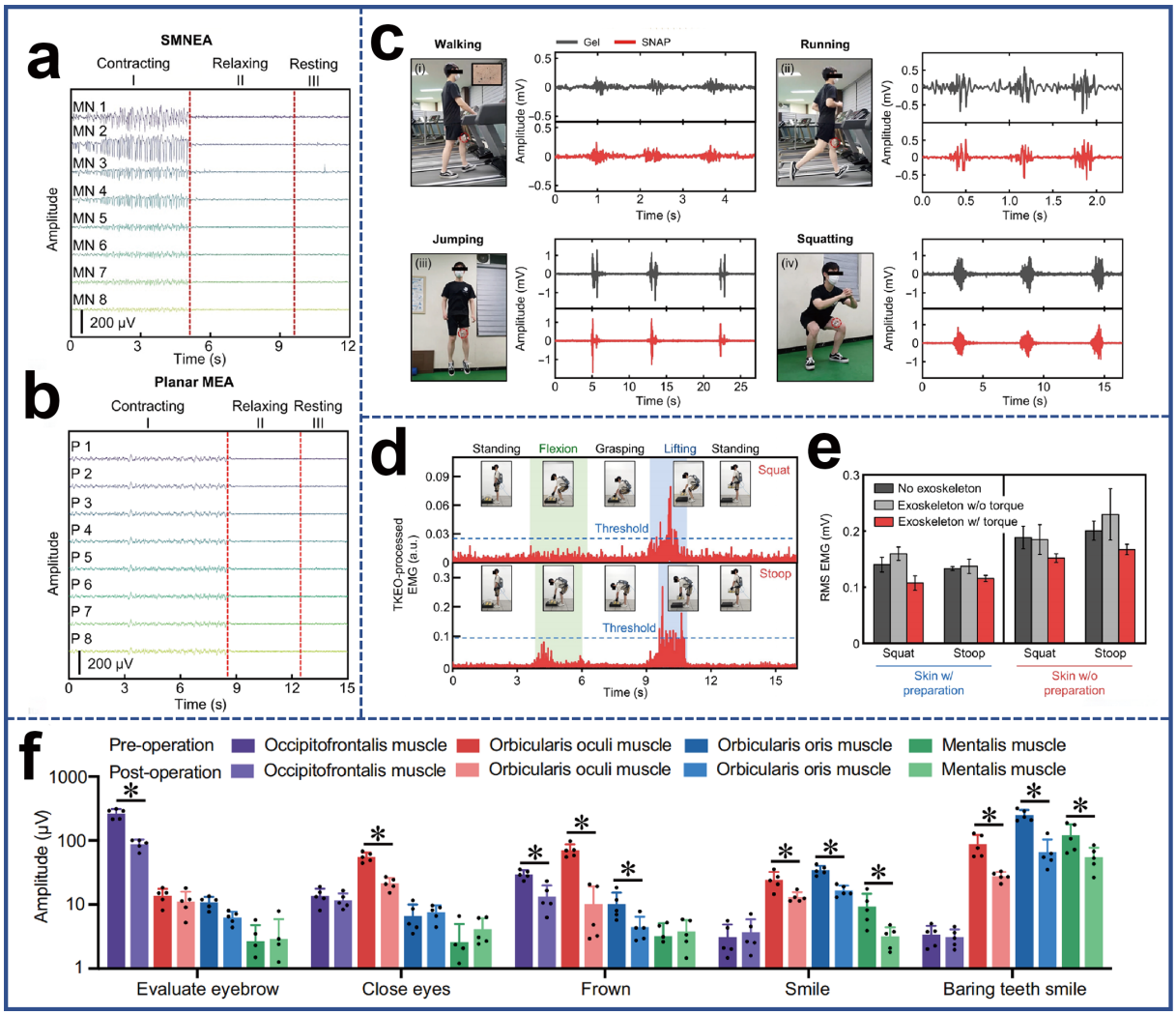

Fig. 6. (Color online) Applications of microneedle electrodes in sports rehabilitation. (a) and (b) Comparison of intramuscular EMG and surface EMG. (c) Movement patterns and corresponding EMG recordings. (d) Demonstration of closed-loop EMG control for an exoskeleton robot. (e) Comparison of root-mean-square (RMS) EMG values under robot-assisted and non-assisted conditions. (f) Facial EMG amplitude differences before and after surgery in a patient with unilateral facial palsy under different facial expressions. (a) and (b) Adapted with permission from Ref. [15]. Copyright 2024, The Author(s), some rights reserved; exclusive licensee American Association for the Advancement of Science. (c)–(e) Adapted with permission from Ref. [14]. Copyright 2024, The Author(s), some rights reserved; exclusive licensee American Association for the Advancement of Science. (f) Adapted from Ref. [17] under the terms of the Creative Commons Attribution 4.0 International License. Copyright 2024, The Author(s), published by Springer Nature.

tions, but it remains the least validated direction and will require larger patient cohorts, longer follow-up, and testing under real clinical rehabilitation workflows before broad deployment[17, 46–48].

6. Conclusions

Microneedle electrodes provide a distinct interface strategy for EMG acquisition in sports rehabilitation by bypassing the high-impedance stratum corneum and forming more stable electrical contact with the skin. Current studies show that their performance depends not on needle penetration alone, but on the coordinated design of penetration capability, conductivity, mechanical compliance, adhesion, flexible integration, and manufacturable processes. Silicon microneedles offer penetration accuracy and geometric control, polymer microneedles favor skin conformity and wearing comfort, and metal microneedles support high conductivity and multichannel recording; however, no single material system can meet all requirements for dynamic and long-term rehabilitation monitoring. The main unresolved issues include insufficient validation of long-term interface stability under realistic motion and sweating conditions, the trade-off between penetration reliability and wearing comfort, uneven fabrication scalability and device reproducibility, and the lack of standardized evaluation methods for impedance, signal quality, motion artifacts, comfort, and rehabilitation relevance. Future work should therefore move beyond demonstrating individual device advantages and focus on reliable, comparable, and rehabilitation-oriented EMG interfaces through long-term testing, stiffness-compliance optimization, reproducible multichannel fabrication, and application metrics that reflect functional assessment, training monitoring, and recovery evaluation.

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