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Recent Advances in Gel-Free Scalp EEG Electrodes: Materials, Structures, and Fixation Strategies

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ABSTRACT

In this review, the development of gel-free scalp electroencephalography (EEG) electrodes is summarized from three aspects: material selection, structural design, and head fixation strategies. The focus is on analyzing the role of these factors in balancing the impedance at the electrode–scalp interface with wearing comfort. Electrode materials have evolved from rigid metals to flexible polymers, textiles, and hydrogels, thus mitigating mechanical modulus mismatch with the scalp. Structural design includes optimizing the comb-like configurations of the electrode contact interfaces—such as claw, brush, and arch shapes—coupled with integrated external cushioning structures, such as springs or foams, to increase the contact stability and suppress motion artifacts. Fixation methods have evolved from rigid caps to flexible elastic caps with self-adhesive interfaces, thereby achieving conformable contact, simplified installation, and improved comfort. Collectively, these technological advances provide effective strategies for increasing electrode-scalp coupling stability and signal quality in gel-free EEG systems.

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1. Introduction

Electroencephalography (EEG) is a noninvasive technique that involves recording electrical activity in the brain using electrodes placed on the scalp [1,2]. EEG electrodes are generally classified into wet, dry, and semi-dry types on the basis of the characteristics of the electrode–skin interface, particularly the nature of the interfacial conductive medium involved in signal transmission [3–5], as shown in Fig. 1.

The use of wet electrodes, particularly Ag/AgCl-based systems combined with conductive gels, represents the most established approach for EEG acquisition [6,7]. As shown in Fig. 1(b), conductive gels serve as the interfacial medium between the electrode and the skin. Wet electrodes provide low interfacial impedance, strong resistance to environmental and motion-related noise, and high signal fidelity, which has led to their widespread use in neuroscience research, psychological studies, and clinical monitoring. Their performance relies on the formation of an ionic–electronic conduction pathway at the electrode–skin interface. Conductive gels play a dual role: They fill interfacial gaps at the interface, thereby reducing the contact impedance and increasing the charge transfer efficiency; moreover, their electrolyte content hydrates the stratum corneum, thus lowering the skin impedance through increased ionic conductivity [2].

Despite these advantages, the use of conductive gels imposes substantial practical limitations. Their application typically requires site-specific manual deposition or injection at each electrode position, which makes the preparation process time-consuming. After use, residual gel often remains on both the scalp and the surface of each electrode because of dehydration, which complicates cleaning. These limitations significantly restrict their use in long-duration monitoring and ambulatory EEG applications.

Dry electrodes eliminate the need for conductive gels and acquire EEG signals through direct physical contact with the skin [8], which significantly improves the ease of use and eliminates the gel-related preparation and cleaning steps. As shown in Fig. 1(c), dry electrodes rely primarily on naturally occurring moisture (e.g., sweat) at the electrode–skin interface rather than an externally supplied conductive medium for signal transmission. Thus, compared with wet electrodes, dry electrodes typically exhibit less stable interfacial impedance and are more susceptible

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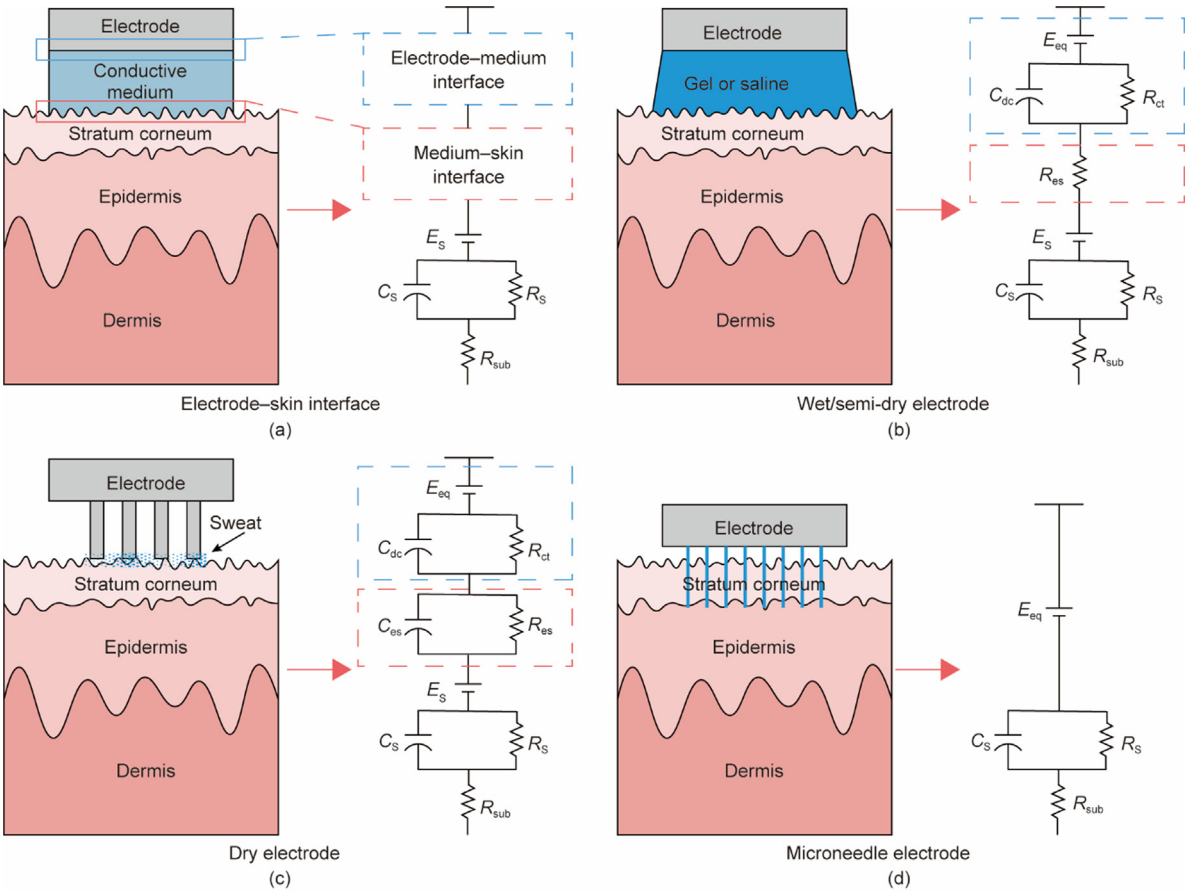


Fig. 1. Electrode–skin interface principle and equivalent circuit model. (a) Electrode–skin interface model, (b) wet/semi-dry electrode, (c) dry electrode, and (d) microneedle electrode. E_S: the skin potential arising from semipermeability and ionic gradients; C_S: the equivalent capacitance of the epidermis; R_S: the equivalent resistance of the epidermis; R_{sub}: the equivalent resistance of the dermis and subcutaneous tissue; E_{eq}: the equivalent potential difference at the electrode–electrolyte interface; C_{dc}: the double-layer capacitance between the electrode and the electrolyte; R_{ct}: the charge transfer resistance between the electrode and the electrolyte; R_{es}: the impedance of the conductive gel, saline, or sweat; C_{es}: the capacitance at the electrode–skin interface.

to motion artifacts and environmental noise, thereby leading to a reduced signal-to-noise ratio [9–15]. Among dry EEG electrodes, microneedle-array electrodes represent a distinct category designed to reduce the impedance associated with the electrode–skin interface. By penetrating the stratum corneum, microneedles bypass the conventional high-impedance electrode–skin interface to establish a lower-impedance electrical pathway to the underlying conductive tissue (Fig. 1(d)). Owing to their skin-penetrating nature and fundamentally different electrode–tissue interface, microneedle electrodes are often regarded as a minimally invasive EEG technology rather than as conventional noninvasive dry electrodes.

Semi-dry electrodes are designed to balance usability and interfacial performance. They can continuously supply a limited amount of electrolyte to the electrode–skin interface, thereby partially maintaining the conductive pathway characteristic of wet electrodes while preserving the operational simplicity of dry electrodes. As a result, they generally exhibit lower and more stable interfacial impedance than dry electrodes do [15–17]. In this review, both dry and semi-dry electrodes are collectively referred to as conductive gel-free EEG electrodes. Notably, the term "gel-free" in this review is adopted from a practical usage perspective and refers to EEG electrodes that do not require externally applied conductive gels during installation or operation. Therefore, dry and semi-dry electrodes, including hydrogel-based semi-dry systems, are discussed within a unified framework, as they all aim to simplify EEG deployment compared with conventional wet electrodes.

Although gel-free EEG electrodes improve usability, they are fundamentally constrained by a trade-off between contact impedance and wearing comfort [7,18,19]. In general, achieving a stable low-impedance interface requires either increased contact pressure or the presence of an interfacial conductive medium, both of which can reduce comfort and prolong the preparation time. This trade-off becomes even more challenging in hair-bearing regions, where hair obstructs direct electrode–scalp contact and reduces the effective contact area. Consequently, hair penetration or hair adaptability has become an additional design consideration for practical EEG systems, particularly for long-term and wearable applications.

Furthermore, the optimal balance between low interfacial impedance and wearing comfort strongly depends on the intended application scenario. For short-term high-fidelity recordings, such as recordings in laboratory EEG studies and brain–computer interface (BCI) systems, low interfacial impedance and high signal quality are often prioritized to ensure reliable signal acquisition and decoding performance. In contrast, long-term clinical monitoring places greater emphasis on user tolerance and skin compatibility during extended wear. For daily wearable applications, such as home-based health monitoring and consumer EEG devices, wearing comfort, ease of use, and long-term user compliance become particularly important. Consequently, the development of gel-free EEG electrodes that can simultaneously provide low impedance and high comfort across diverse application scenarios remains a major goal of the field.

Without conductive gels, the electrical and mechanical behavior of the electrode–scalp interface is governed primarily by three factors: material properties, structural design, and fixation strategy. Materials, structures, and fixation strategies are closely interconnected rather than independent design dimensions. Material properties determine the achievable electrical and mechanical characteristics of the electrode. Structural design translates these properties into specific contact morphologies and effective contact with the scalp, particularly in hair-bearing regions. Fixation strategies further influence the contact pressure and stability of the electrode–scalp interface. Together, these factors jointly govern interfacial impedance and wearing comfort.

On the basis of this framework, gel-free EEG electrodes are not simply categorized according to material type, structural configuration, or fixation method in this review. Instead, the impedance–comfort trade-off is regarded as the central challenge, and we examine how materials, structures, and fixation strategies collectively influence the electrical and mechanical behaviors of the electrode–scalp interface. By analyzing these three interconnected design dimensions, we aim to provide insights into the integrated optimization of interfacial impedance, wearing comfort, and long-term recording stability, thereby providing guidance for the development of practical high-performance EEG systems.

2. Material considerations for gel-free EEG electrodes

Materials are fundamental determinants of both signal quality and wearing comfort in gel-free EEG electrodes. They govern the intrinsic electrical and mechanical properties at the electrode–scalp interface, including the electrical conductivity, deformability, and interfacial stability.

From an electrical perspective, intrinsic properties such as electrical conductivity directly influence charge transfer at the interface, thereby affecting interfacial impedance and stability. In addition, electrochemical characteristics (e.g., polarization behavior) are closely associated with baseline drift during signal acquisition [20]. From a mechanical perspective, the Young's modulus of the material plays a key role in determining the effective contact area between the electrode and the scalp. A larger modulus mismatch with the soft tissue of the scalp reduces conformal contact, thereby decreasing the effective contact area and increasing the interfacial impedance. Moreover, such a mismatch can lead to stress concentration at the interface, thereby resulting in localized pressure discomfort. Surface properties, including hardness, friction coefficient, and roughness, further influence the microscale contact mechanics and the interfacial pressure distribution. In addition, permeability-related properties, such as breathability and moisture transport, play important roles in long-term wearing comfort.

Accordingly, the central challenge in material design for gel-free EEG electrodes lies in balancing electrical performance with mechanical compliance. In response to this challenge, electrode materials have evolved from rigid metals and ceramics to flexible polymers and further to textiles and hydrogels. This evolution reflects a general trend toward a progressively lower elastic modulus and improved wearing comfort while maintaining sufficient interfacial conductivity.

2.1. Rigid materials

Early gel-free EEG electrodes were based predominantly on rigid conductive materials such as silver, gold, stainless steel, titanium, and copper [21–27]. These materials provide excellent electrical conductivity, electrochemical stability, and mechanical robustness, and can be repeatedly used without significant degradation; hence, they were widely adopted in early-stage EEG systems. However, their high stiffness and poor mechanical compliance with soft scalp tissue inevitably lead to limited wearing comfort and suboptimal long-term usability. A representative example is the g.SAHARA gold alloy electrode developed by g.tec (Fig. 2(a)) [28,29].

Rigid electrodes benefit from relatively mature and versatile manufacturing routes. Traditional processes typically rely on molding and precision machining to define the geometry of the electrodes. For instance, Fiedler et al. [25] fabricated structured titanium electrodes by mechanical machining, followed by the deposition of a TiN coating using direct current (DC) magnetron sputtering to form a dry Ti/TiN composite electrode (Fig. 2(b)). More recently, additive manufacturing has been introduced as an alternative approach that enables the fabrication of more complex geometries and faster prototyping. Tong et al. [27] demonstrated direct three-dimensional (3D) printing of EEG electrodes using copper-based conductive filaments, which could be used directly after printing without additional postprocessing, thereby simplifying the fabrication workflow (Fig. 2(c)).

To improve interfacial electrochemical behavior, Ag/AgCl coatings are typically introduced onto metallic electrodes. Owing to their low polarization potential and relatively stable electrode–electrolyte interface, such coatings are widely used in both dry and semi-dry electrode systems [24,30]. However, these coatings are mechanically fragile and tend to degrade over repeated mechanical contact and long-term use, which leads to progressive deterioration in signal quality and necessitates periodic replacement or maintenance.

Beyond metals, ceramic materials such as Al_2O_3 and ZrO_2 have also been explored as rigid electrode components, particularly in semi-dry systems. In these designs, porous ceramics have typically been used as electrolyte reservoirs and capillary transport media. For example, Li et al. [31] developed a porous ceramic-based semi-dry electrode in which a microstructured ceramic network enables saline storage and gradual release, thereby sustaining a relatively stable interfacial impedance under hydrated conditions (Fig. 2(d)). However, the finite electrolyte storage capacity inherently limits the operational lifetime.

Mechanically, rigid materials exhibit a severe mismatch with soft scalp tissue. Their elastic modulus is on the order of hundreds of gigapascals, whereas the elastic modulus of the scalp is typically in the tens to hundreds of kilopascals, thus spanning several orders

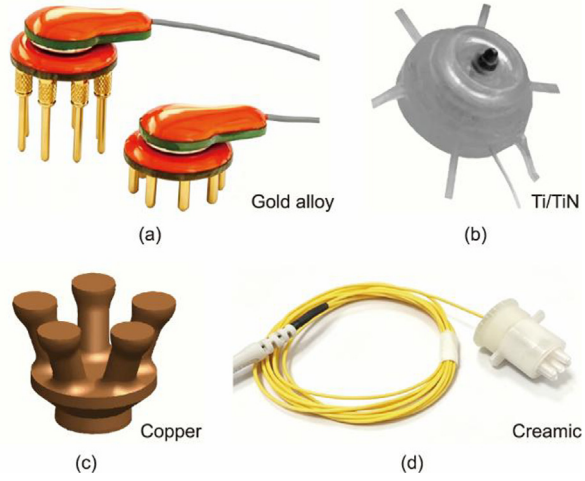


Fig. 2. Rigid materials. (a) g.tec developed g.SAHARA electrode [28,29], (b) Ti/TiN dry electrode [25], (c) 3D-printed copper-based electrode [27], and (d) porous ceramic semi-dry electrode [31].

of magnitude. This mismatch fundamentally limits conformal contact at the electrode–skin interface, thereby resulting in a reduced effective contact area and increased interfacial impedance. In addition, relative motion at the interface under dynamic conditions can introduce motion artifacts, whereas localized stress concentration may cause discomfort or even potential skin irritation during prolonged wear.

Overall, although rigid materials provide clear advantages in terms of conductivity, stability, and manufacturability, their inherent mechanical incompatibility with soft biological tissue restricts their use in long-term wearable EEG applications.

2.2. Polymeric materials

Compared with rigid materials, such as metals and ceramics, polymers generally exhibit a lower elastic modulus, which enables better conformability to the scalp. This increased conformability improves wearing comfort and reduces relative motion at the interface, thereby mitigating motion artifacts. However, most polymers are intrinsically insulating and cannot function as electrodes without additional modifications. To overcome this limitation, two main strategies have been developed: applying conductive coatings onto polymer substrates and constructing intrinsically conductive polymer composites.

The conductive coating strategy relies on the use of polymers as mechanically compliant substrates while introducing conductivity through surface coatings. In early studies, rigid plastics such as polylactic acid (PLA) and acrylonitrile butadiene styrene (ABS) were often used as substrates, with silver paste being manually applied to achieve conductivity (Fig. 3(a)) [32]. Although such designs can meet the requirements of applications with relatively low signal quality demands, their rigid substrates still limit comfort and interfacial stability. To address this issue, softer polymers such as silicone, rubber, polyurethane (PU), and polydimethylsiloxane (PDMS) have been introduced as substrates [33]. These materials better match the mechanical properties of the scalp, thereby increasing the effective contact area and reducing signal fluctuations caused by micromotion. Common conductive coatings include metals and metal compounds (e.g., Ag [34], Au [35], and TiN [12]), carbon-based materials (e.g., carbon nanotubes (CNTs), graphene, and derivatives [36]), and conductive polymers (e.g., poly(3,4-ethylenedioxythiophene):poly(styrenesulfonic acid) (PEDOT:PSS) [37]). In addition, 1D nanomaterials such as silver nanowires (AgNWs) are also frequently used because of their ability to form flexible and highly conductive percolating networks [38]. These conductive layers can be fabricated using a range of techniques depending on the material type. Metal-based coatings are typically deposited by sputtering, evaporation, or electroless plating. AgNWs can be assembled through spray coating, spin coating, or electrospinning, whereas conductive polymers such as PEDOT:PSS are typically processed via solution-based techniques, such as spin coating, spray coating, screen printing, and inkjet printing [39]. For example, Grozea et al. [40] fabricated soft electrodes by coating conductive silver ink onto flexible bristle structures, whereas Fiedler et al. [41] deposited TiN on flexible PU substrates by multiphase sputtering to create multipin (MP) electrode arrays (Fig. 3(b)). Compared with earlier rigid designs, these flexible electrodes significantly improve comfort while preserving acceptable signal quality.

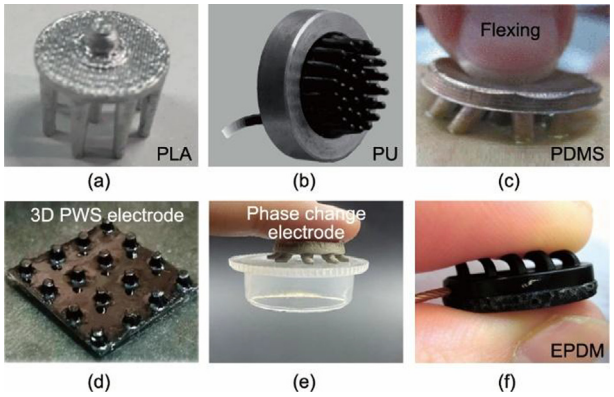


Fig. 3. Polymeric materials. (a) 3D-printed PLA electrode [32], (b) MP electrode [41], (c) soft cylindrical flexible dry electrode [44], (d) PEDOT:PSS composite film dry electrode [45], (e) body-temperature-triggered phase-change dry electrode [51], and (f) EPDM-based polymer dry electrode [18]. PWS: a blend of PEDOT:PSS, waterborne polyurethane (WPU), and *D*-sorbitol.

Despite these advantages, coating-based electrodes remain susceptible to interfacial degradation during repeated use. Mechanical friction and cyclic deformation can gradually damage or delaminate the conductive layer, thereby leading to increased impedance and deteriorated signal quality [42]. To mitigate this issue, adhesion layers or interfacial modifiers are often introduced between the substrate and the conductive layer. Common choices include Ti [43], parylene C, polydopamine (PDA), Al_2O_3, and silicone-based bonding layers. For instance, Wang et al. [44] deposited parylene C onto PDMS as an adhesion layer before gold was sputtered, which significantly improved the coating stability of soft pillar electrodes (Fig. 3(c)). Similarly, surface modification with PDA has been used to improve both the wettability and adhesion of PEDOT:PSS films on PDMS substrates (Fig. 3(d)) [45].

Although these strategies improve coating robustness to some extent, long-term reliability issues remain difficult to eliminate. To address this limitation, intrinsically conductive composites have been developed in which conductive fillers are directly incorporated into the polymer matrix to form materials that are both flexible and conductive [46]. In such composites, the polymer matrix provides mechanical support, with common materials including PDMS [47], ethylene propylene diene monomer (EPDM) [18], thermoplastic polyurethane (TPU) [48], and poly(styrene–ethylene butylene–styrene)triblock copolymer (SEBS) [49,50], all of which exhibit relatively low moduli and good compliance with skin. More recently, functional matrices with adaptive mechanical behavior have been explored. For instance, temperature-responsive phase-change materials can remain relatively stiff at room temperature to facilitate penetration through hair while softening at body temperature to form conformal, pressure-sensitive contact (Fig. 3(e)) [51]. Similarly, shear-thickening systems such as polyborosiloxane (PBS) can increase structural stability under dynamic deformation [52]. The electrical conductivity of these composites is enabled by the use of conductive fillers, which can be broadly categorized into carbon-based materials (e.g., CNTs [47,53,54], carbon black [48,55], and graphene [56–58]), metal-based materials (e.g., silver flakes [51] and stainless-steel fibers [18]), and conductive polymers (e.g., PEDOT:PSS [59] and polypyrrole) [60]. The filler content critically affects the overall performance: Insufficient loading prevents the formation of conductive networks, whereas excessive loading leads to aggregation and reduced flexibility [47]. For example, Chen et al. [18] reported that a carbon filler content of ~45% in EPDM results in an optimal balance between conductivity and mechanical properties (Fig. 3(f)), whereas Oh et al. [47] reported that conductivity in PDMS/CNT systems increases with increasing CNT content until a percolation threshold is reached. Liquid metals have also emerged as promising conductive fillers because of their high conductivity and fluidic deformability, thus enabling the establishment of adaptive conductive pathways under deformation [61]. However, conventional liquid metals such as eutectic gallium–indium (EGaIn) suffer from excessive fluidity, encapsulation challenges, and leakage risks. To

overcome these limitations, alternative approaches such as non-Newtonian liquid metal systems (e.g., $Ga_{50}In_{50}$) [62] and semisolid composites that incorporate PBS have been proposed to improve morphological stability and expand practical applicability [52].

Despite substantial progress, several fundamental challenges remain. First, achieving low interfacial impedance often requires a certain electrode thickness to ensure a sufficient contact area, which can increase the bending stiffness and reduce the conformability. Second, although polymer substrates are intrinsically soft, coatings or fillers can locally stiffen the material, thereby partially offsetting their mechanical advantages. More importantly, the modulus of most polymers (typically tens of mega pascals or higher) still exceeds that of scalp tissue (0.1–1 MPa) by one to two orders of magnitude [63–66]. Therefore, achieving high conductivity while maintaining a low modulus and structural stability remains a central challenge for polymer-based gel-free EEG electrodes.

2.3. Textile-based materials

In textile-based electrodes, fibrous substrates are used to construct flexible conductive interfaces, in which conductivity is introduced either onto fiber surfaces or within the fiber network through techniques such as chemical plating, printing, coating, and impregnation [67–69]. Common substrate materials include natural fibers (e.g., cotton), synthetic fibers (e.g., polyester and nylon), and elastomeric textiles (e.g., PU and TPU) [36]. Conductive components typically consist of metals (e.g., silver, copper wires, silver nanowires, and silver-coated nylon fibers) [70–72], carbon-based materials (e.g., CNTs, graphene, and carbon black) [73], and conductive polymers (e.g., PEDOT:PSS, polypyrrole, and polyaniline) [74–77]. Two main strategies that differ in terms of how conductivity is introduced are used to fabricate textile electrodes. The first strategy is a "fabric-first, functionalize-later" approach, where conductive layers are formed on prefabricated textiles by coating, plating, or impregnation. Examples include silver-plated cotton fabrics [69] and graphene-coated textiles prepared by dip-coating and drying processes [78,79]. For instance, Tseghai et al. [75] fabricated textile electrodes by depositing conductive pastes onto knitted fabrics using flatbed screen printing (Fig. 4(a)). The second strategy is "functionalize-first, fabricate-later," where conductive fibers are prepared in advance and subsequently woven or knitted into textile structures, such as electrodes that are based on silver-coated nylon fibers [80] or Ag/PA yarn-based systems (Fig. 4(b)) [81].

Textile electrodes have excellent flexibility and conformability, which enables them to adapt to complex scalp geometries and maintain structural integrity under dynamic deformation, thereby preserving relatively stable conductive pathways. In addition, the porous nature of textiles enables efficient air and moisture transport, which significantly improves thermal comfort during long-term wear. Their washability and reusability further contribute to mechanical durability in practical applications. However, their performance in hair-bearing regions remains limited. Dense hair can obstruct direct electrode–scalp contact, thereby reducing the effective contact area. To address this issue, structural modifications have been explored to improve hair penetration ability. For example, hook-like silver-coated nylon fibers have been used in combination with fully textile-based supporting networks to facilitate contact through hair (Fig. 4(c)) [70]. Nevertheless, such designs often introduce new challenges, including increased structural complexity, sensitivity to hair density and morphology, and limited signal stability, which restrict their practical applicability. In addition, textile electrodes that rely on surface coatings are susceptible to performance degradation during repeated use. Mechanical friction and washing processes can gradually damage or remove conductive layers, thereby leading to increased interfacial impedance and decreased signal quality over time and ultimately limiting the device lifespan.

Overall, textile-based electrodes provide a compelling combination of flexibility, breathability, and wearing comfort, which makes them particularly attractive for EEG acquisition in hairless or low-hair-density regions. However, their inability to reliably establish stable, low-impedance contact in hair-bearing areas remains a critical limitation. Addressing this challenge requires strategies that can effectively overcome hair interference while preserving the intrinsic mechanical and physiological advantages of textile structures.

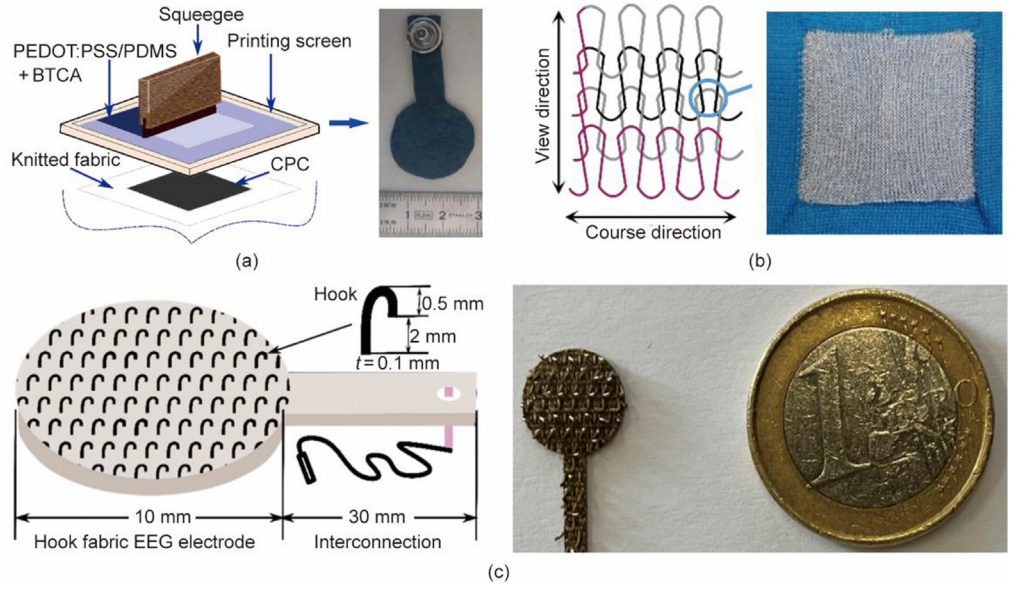


Fig. 4. Textile-based materials. (a) Screen-printed fabric electrode [75], (b) silver-plated polyamide yarn fabric electrode [81], and (c) Velcro fabric electrodes [70]. BTCA: 1,2,3,4-butanetetracarboxylic acid; CPC: conductive polymer composite (PEDOT:PSS/PDMS); *t*: base layer thickness.

2.4. Hydrogel-based materials

The hydrogel materials discussed in this section serve primarily as integrated conductive or adhesive components within EEG electrodes, particularly in semi-dry systems, rather than as externally applied conductive gels used in conventional wet EEG electrodes. Hydrogel electrodes rely on water–swollen polymer networks as their conductive medium, where ionic transport occurs through the continuous aqueous phase [82–84]. The polymer matrix typically includes synthetic polymers (e.g., polyacrylamide (PAAm), polyvinyl alcohol (PVA), and polyacrylic acid (PAA)) and natural polymers (e.g., gelatin and sodium alginate). Conductive ions (e.g., Na^+, Cl^-, and K^+) are introduced into the 3D network by solution soaking, *in situ* polymerization, or the incorporation of conductive fillers, and electrical signals are transmitted through ionic migration under an applied electric field. On the basis of the conduction mechanism, hydrogel electrodes can be broadly classified into two categories. Purely ionic conductive hydrogels rely on ion transport and typically have a low modulus and excellent biocompatibility. In contrast, ion–electron hybrid conductive hydrogels (e.g., PEDOT:PSS-based hydrogels) incorporate conductive polymers into the network, which enables improved electrical performance while retaining mechanical softness.

Hydrogel electrodes generally have elastic moduli comparable to those of skin (approximately 0.1–2 MPa), which results in excellent interfacial conformability and wearing comfort. Their high deformability enables intimate contact with the scalp and accommodates small deformations, thereby reducing motion artifacts induced by interfacial micromotion and improving signal quality. Furthermore, functional additives such as AgNWs [85] and 2D early transition metal carbides and carbonitrides (MXenes) [86], as well as bioactive components such as antimicrobial peptides [87], can be incorporated to increase conductivity and provide antibacterial or anti-inflammatory properties, thus reducing the risk of skin irritation or infection during long-term use.

A key challenge for hydrogel electrodes lies in water retention. Because ionic conductivity strongly depends on internal water content, evaporation over time reduces ion mobility, thereby leading to decreased conductivity and signal degradation. To mitigate this issue, Li et al. [88] proposed a superporous semi-dry hydrogel electrode (SPH) using PAAm/PVA, which enables gradual electrolyte replenishment through structural design (Fig. 5(a)). However, the limited electrolyte reservoir still constrains long-term operation. Conventional strategies for reducing water loss rely mainly on passive retention and include incorporating hygroscopic agents (e.g., glycerol) to slow evaporation [87,89]. For example, Shen et al. [89] improved moisture retention and long-term stability by introducing glycerol and KCl into N-acryloyl glycinamide (NAGA)-based hydrogels (Fig. 5(b)). Building on this approach, active rehydration strategies have also been explored, in which dehydrated hydrogels are reimmersed in electrolyte solutions to restore water content and ionic conductivity by swelling, thereby enabling repeated use. Beyond water retention, stable electrode–scalp contact in hair-bearing regions remains another key limitation [90]. Solid or semisolid hydrogels may not fully infiltrate the gaps between hair and skin or conform to skin microstructures, thereby leading to interfacial voids and increased impedance (Figs. 5(c) and (d)) [87,91]. To overcome this issue, adaptive hydrogel systems have been proposed [90]. One strategy involves the use of temperature-responsive phase-change hydrogels that exhibit low viscosity at elevated temperatures, which enables them to infiltrate hair-covered regions and achieve effective wetting, followed by rapid gelation at body temperature to form a stable interface. For instance, Wang et al. [92] designed hydrogels with tunable crosslinking density that undergo a liquid-to-gel transition at body temperature, thus enabling *in situ* conformal contact (Fig. 5(e)). Another approach involves the use of ionically crosslinked rapid-gelation systems, where multivalent ions trigger fast network formation, thereby achieving gelation within seconds. For example, Luo et al. [86] reported a poly(sodium acrylate)–MXene hydrogel (PAAS–MXene-based hydrogel) that gels within ~5 s (Fig. 5(f)).

Overall, hydrogel electrodes, with their skin-like mechanical properties and efficient ionic conduction, represent a promising direction for gel-free EEG electrodes. However, key challenges remain in long-term water retention, reliable interfacial contact in hair-bearing regions, and the controllability and reproducibility of adaptive fabrication processes.

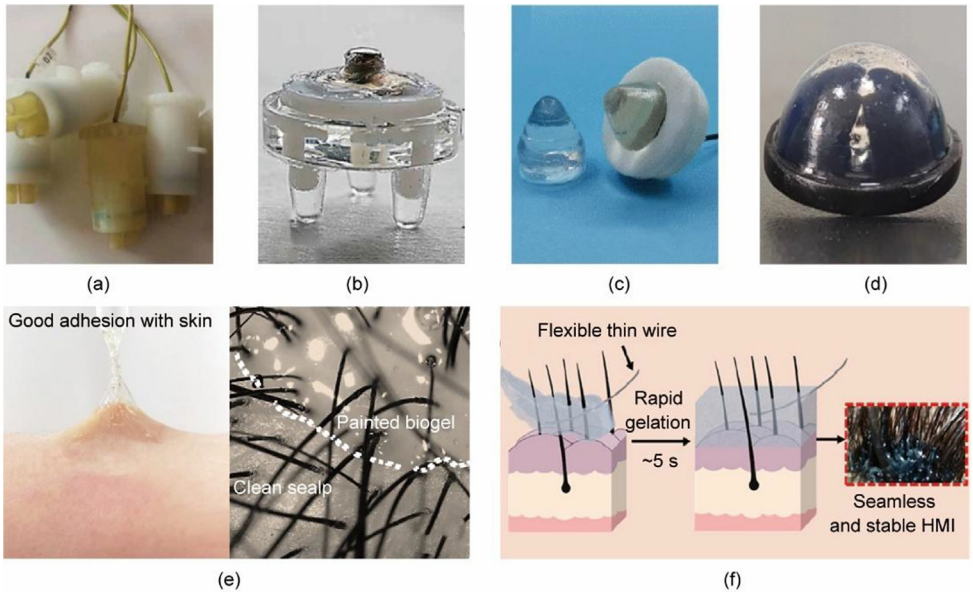


Fig. 5. Hydrogel-based materials. (a) Semi-dry hydrogel electrode (SPH) [88], (b) novel flexible hydrogel electrode (NAGA) [89], (c) solid-state pre-gelled (PreG) electrode [87], (d) semisolid hydrogel electrode [91], (e) liquid/gel phase-change biohydrogel [92], and (f) adaptive conductive hydrogel (PAAS–MXene) [86]. HMI: human–machine interface.

From rigid metals and ceramics to soft polymers, textiles, and hydrogels, electrode materials have evolved toward progressively lower stiffness and improved wearing comfort. A systematic comparison of representative material systems on the basis of mechanical properties, electrical conductivity, electrode–scalp interfacial performance, fabrication processes, and reusability is summarized in Table 1 [25,27–29,31,32,41,44,45,51,70,75,86–89,91,92], which provides insights into the advantages and limitations of various materials for gel-free EEG electrodes. However, achieving stable electrical performance while reducing material stiffness and establishing reliable electrode–scalp contact in hair-bearing regions remain key challenges in the development of gel-free EEG electrodes.

3. Structural design of gel-free EEG electrodes

In gel-free EEG electrodes, structural design governs how the electrode interfaces with the scalp by defining its geometric form and contact locations, thereby determining the effective contact area and pressure distribution. Owing to hair obstruction, planar electrodes [93] are generally limited to hairless regions for EEG recording. In hair-bearing regions, compared with planar designs, hair-penetrating structures such as pin- or comb-like geometries [94] can more effectively increase the macroscopic contact area, thereby reducing the interfacial impedance. However, at the microscale, these structures typically establish contact through multiple discrete protrusions, each with a limited local contact area. As a result, interfacial impedance becomes highly sensitive to contact pressure and susceptible to motion-induced disturbances. Moreover, the stress concentration at local contact sites increases the localized pressure, thereby leading to wearing discomfort. This effect reflects the intrinsic trade-off between low impedance and high comfort in electrode structural design [42]. Accordingly, structural optimization is fundamentally aimed at maintaining hair penetration ability while increasing the effective contact area to reduce the impedance, redistributing the contact pressure to improve static comfort, and mitigating dynamic disturbances to suppress motion artifacts and improve dynamic comfort. On the basis of this design rationale, recent research has focused mainly on optimizing the geometry of comb-like structures while introducing compliant or cushioning structures at their base.

3.1. Morphological optimization of comb-like structures

The morphological optimization of comb-like structures focuses on the geometric design of the comb elements. By adjusting their inclination and curvature, as well as the size and density of the contact features, such designs aim to improve conformability with the scalp and promote a more uniform distribution of contact pressure. From a mechanical perspective, the core objective is to regulate the interfacial pressure distribution by transforming concentrated point loading into multipoint or line contact, thereby reducing the localized stress and improving the wearing comfort. On the basis of this principle, current designs can be broadly classified into claw-like, brush-like, and arched structures.

Claw-like structures transform conventional perpendicular contact into inclined, compliant contacts, in which the prongs bend under load and conform to the curvature of the head, thereby increasing the effective contact area [95,96]. Mechanically, they behave in a cantilever-like manner: External loading induces elastic bending of the prongs, thereby redistributing concentrated forces into a more uniform interfacial pressure profile. A representative example is the Ag/AgCl electrode developed by Cognionics (Fig. 6(a)) [97]. In addition, Xing et al. [98] fabricated claw-like electrodes by TPU molding (Fig. 6(b)), and Wang et al. [95] developed TPU-based 3D-printed dry electrodes with similar configurations. Velcescu et al. [93] further demonstrated a range of claw-inspired designs using additive manufacturing (Fig. 6(c)), all of which reflected this design principle.

Brush-like structures reduce the diameter of individual contact elements to lower their bending stiffness, thus enabling each filament to deform independently under compression. This independent deformation converts concentrated loading into a distributed multipoint contact, thereby improving interfacial compliance and increasing the effective contact area. Because the bending stiffness of a single fiber scales with the fourth power of its diameter, even small reductions in size can significantly alter the mechanical response. This mechanism of independent filament deformation enables finer adaptation to microscale surface irregularities of the scalp. A representative example is the polymer bristle electrode proposed by Grozea et al. [40]; however, in hair-bearing regions, it is difficult to ensure that every filament can reliably penetrate through the hair and establish contact with the scalp (Fig. 6(d)). To improve robustness, Gao et al. [99] combined a PDMS–CNT substrate with PU–CNT microneedles, followed by tip polishing and gold electroplating to form brush-like structures (Fig. 6(e)), which significantly improved the contact stability in hair-bearing regions.

Arched structures, in contrast, do not rely on local deformation but instead achieve conformity through a preshaped curvature that matches that of the head. Examples include inverse-curvature electrodes fabricated via 3D-printed silver alloys (Fig. 6(f)) [100] and TPU-based electrodes with Ag/AgCl coatings (Fig. 6(g)) [101], both of which maintain stable contact across different hair conditions.

Overall, these three structural strategies improve the conformability in hair-bearing regions through mechanisms that operate at different scales. Claw-like structures operate at the macroscale via global curvature adaptation. Brush-like structures function as

Table 1
Comparison of representative material systems for gel-free scalp EEG electrodes.

Material system	Representative materials	Elastic modulus	Hair adaptability	Electrical conductivity	Interfacial performance	Wearing comfort	Fabrication process	Reusability	References
Metals	Au, Cu, Ag/AgCl, and stainless steel	70–210 GPa	Excellent	Very high	Moderate	Poor to moderate	Lathe, sputtering, and 3D printing	Reusable	[25,27–29,31]
Polymers	PDMS, EPDM, TPU, and ecoflex	0.1 MPa–3 GPa	Moderate to good	Moderate	Moderate to high	Moderate to good	Sputtering, mold casting, 3D printing, and drop casting	Generally reusable	[18,32,41,44,45,51]
Textiles	Conductive fabrics, CNT fabrics, and Ag-coated textiles	0.8 MPa–3 GPa	Poor	Moderate	Moderate	Excellent	Weaving, coating, dip coating, and screen printing	Generally reusable	[70,75]
Hydrogels	PAM, PVA, alginate, and PAAm	<0.5 MPa	Moderate to good	Low	High	Excellent	3D printing, mold casting, polymerization, solution mixing	Generally nonreusable	[86–89,91,92]

PAM: polyacrylamide.

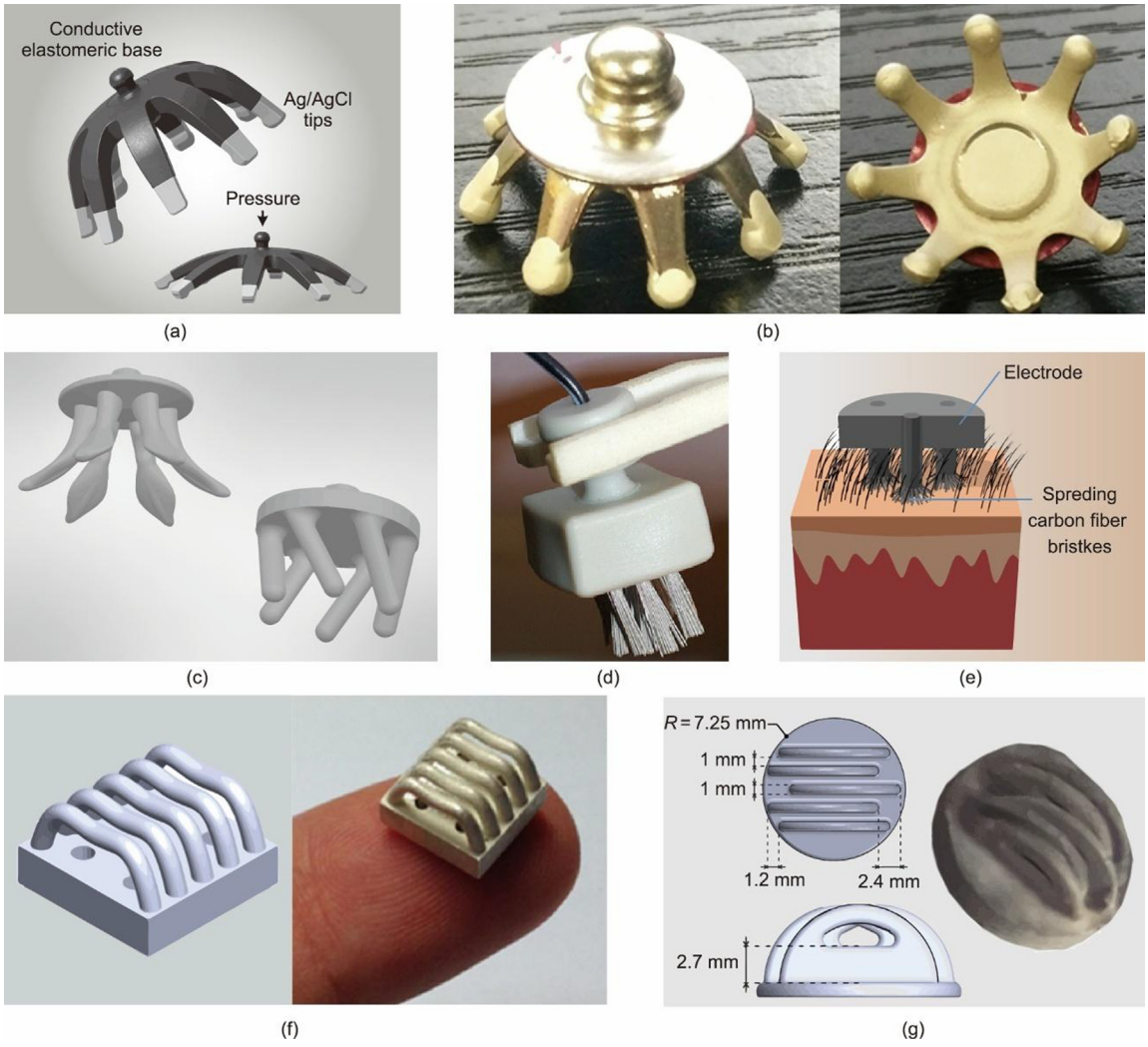


Fig. 6. Optimization of the comb-like structure at the electrode tip. (a) Cognionics claw-shaped electrode [97], (b) TPU injection-molded claw-shaped electrode [98], (c) 3D-printed TPU electrode [93], (d) polymer bristle electrode [40], (e) carbon fiber bristle electrode [99], (f) reverse-curved arch electrode [100], and (g) arched electrode [101]. *R*: outer radius.

distributed microcontact systems with increased contact density. Arched structures achieve conformity through predefined global geometry that matches the curvature of the head, thereby minimizing deformation requirements. This framework provides flexibility in selecting designs for different application scenarios and wearing conditions.

3.2. Integration of cushioning structures

The morphological optimization of comb-like structures addresses primarily the conformability and pressure distribution under static conditions. However, in practical use, electrode–scalp interfaces are inevitably subject to dynamic disturbances induced by speech, head movement, or facial expressions. These processes generate transient relative motion and fluctuations in contact pressure, which cannot be effectively accommodated by geometric design alone. Therefore, the introduction of dedicated cushioning mechanisms at the structural level is necessary to improve interface stability under dynamic conditions.

The integration of cushioning structures introduces independent elastic elements between the comb-like structure and the fixation system to provide mechanical buffering. Mechanically, this low-stiffness response stabilizes contact pressure and accommodates scalp micromotion. Two main implementation strategies have been developed: spring-based support and foam/sponge-based support.

Spring-based structures integrate individual springs at the base of each comb tooth and provide localized compliance through linear elastic deformation, thereby enabling each contact unit to independently adapt to variations in scalp geometry [102]. Typical spring-loaded probe electrodes consist of metallic contact pins (e.g., gold- or platinum-coated needles) coupled with independent spring elements, which enable individual vertical compliance to accommodate irregular scalp surfaces (Figs. 7(a) and (b)) [103,104]. Kimura et al. [105] further determined optimal spring constants using a contact mechanics model and realized an integrated electrode–spring architecture by 3D printing of resin-based materials (Blu; Siraya Tech, USA) (Fig. 7(c)). The primary advantage of spring-based designs lies in mechanical decoupling between contact units, which minimizes the cross-interference of local pressure and is particularly suitable for applications that involve large variations in scalp curvature. These systems exhibit approximately linear force–displacement behavior and fast mechanical responses. However, owing to limited intrinsic damping, their ability to dissipate high-frequency microvibrations is weak, which may still lead to motion-induced noise. In addition,

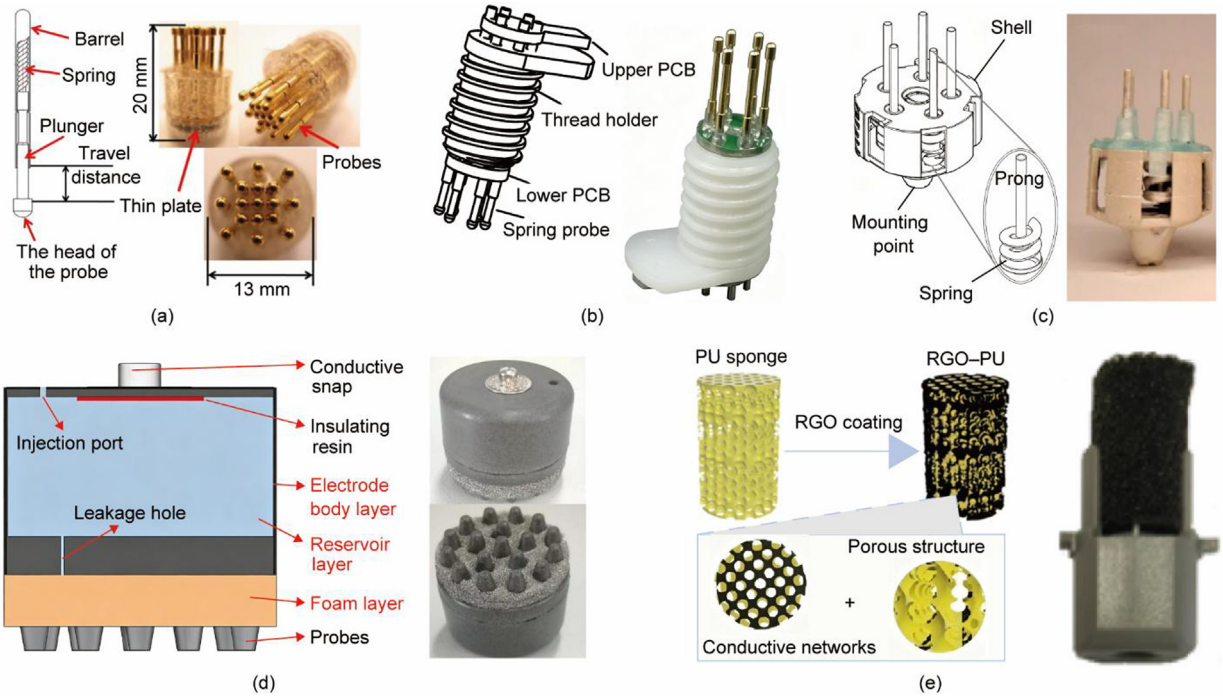


Fig. 7. Integrated rear-end buffer structures. (a) Spring-contact probe-type dry electrode [103], (b) spring probe electrode [104], (c) 3D-printed spiral spring fork dry electrode [105], (d) flexible multilayer semi-dry electrode [107], and (e) RGO–PU semi-dry electrode [56]. PCB: printed circuit board.

the relatively high stiffness and contact impact of metallic springs may reduce wearing comfort under dynamic conditions, whereas the discrete assembly of spring elements increases fabrication and integration complexity.

In contrast, in foam- or sponge-based supports, porous continuous media are used to achieve distributed compliance, which eliminates the need for individual elastic elements for each comb tooth, thereby simplifying the structural design. Representative implementations include conductive PU foam electrodes, where conductive textile layers combined with Ni/Cu coatings establish electrical pathways [106]. These materials exhibit pronounced nonlinear compressive behavior: under low strain, they present very low effective moduli and efficiently absorb minor perturbations; as compression increases, their stiffness progressively increases, which provides structural support. The related designs also include PDMS/silver nanoparticle composite multilayer structures (Fig. 7(d)) [107] and reduced graphene oxide (RGO)–PU sponge systems (Fig. 7(e)) [56]. These porous architectures leverage compressibility and energy dissipation to increase signal stability and comfort under dynamic conditions while maintaining stable electrode–scalp contact.

Overall, spring- and foam-based strategies represent two distinct cushioning paradigms: discrete linear compliance and continuous nonlinear compliance, respectively. The former emphasizes independent tuning of individual contact points, whereas the latter prioritizes global conformity and structural simplicity.

From a system-level perspective, the central challenge remains the trade-off between hair penetration ability and stable low-impedance contact. Representative structural strategies, including comb-like contact structures and cushioning architectures, are compared in Table 2 [40,59,93,97–101,103–105,107] in terms of hair adaptability, contact stability, motion stability, and wearing comfort, thus highlighting their distinct contributions to improving EEG acquisition performance. Whereas comb morphology optimization primarily improves static conformability, cushioning integration improves dynamic robustness by mitigating motion-induced disturbances. However, the suppression of motion artifacts has not yet been fully resolved, and a trade-off between structural complexity and wearability remains.

4. Fixation strategies for gel-free EEG electrodes

Owing to the absence of conductive gels, gel-free EEG electrodes rely primarily on contact pressure to determine interfacial performance. Accordingly, the fixation strategy determines both the magnitude of this pressure and its temporal stability. The fixation method involves first determining the contact pressure between the electrode and the scalp. For electrodes with the same geometry, increasing the contact pressure appropriately can improve the conformity to the scalp and increase the effective contact area, thereby reducing the interfacial impedance [1,44,108]. In addition,

Table 2
Comparison of representative structural designs for gel-free scalp EEG electrodes.

Structural design	Structure type	Hair adaptability	Contact stability	Motion stability	Wearing comfort	References
Comb-like structures	Claw-shaped	Excellent	Moderate to good	Unknown	Moderate to good	[93,97,98]
	Brush-shaped	Good	Moderate to good	Unknown	Excellent	[40,99]
	Arch-shaped	Moderate to good	Good	Unknown	Moderate to good	[100,101]
Cushioning structures	Spring-buffered	Unknown	Good	Good	Good	[103–105]
	Porous-buffered	Unknown	Moderate to good	Unknown	Good	[59,107]

the fixation strategy affects the stability of the interfacial impedance. The pressure provided by EEG caps is not constant; user movements, speech, and facial expressions may induce slight electrode motion, thereby leading to dynamic changes in interfacial impedance, polarization potential, and capacitance, which in turn generate motion artifacts or noise and degrade signal quality. In addition, the fixation strategy strongly influences wearing comfort. The magnitude of the contact pressure directly affects short-term pressure-related discomfort, whereas the mode of pressure maintenance (e.g., elastic structures or viscoelastic pads) affects mainly long-term wear stability and comfort. Early fixation strategies rely mainly on externally applied contact pressure to maintain electrode–scalp contact. However, achieving low interfacial impedance often requires relatively high pressure, which could compromise wearing comfort. To address this limitation, self-adhesive electrode designs have been developed, thus enabling improved scalp conformity and more stable electrode–scalp interfaces under lower external pressures. Accordingly, the fixation strategies for scalp EEG electrodes can be broadly categorized into EEG cap-based fixation and self-adhesive fixation.

4.1. EEG cap-based fixation

The use of an EEG cap is among the most widely used fixation strategies for gel-free EEG electrodes. Its primary function is to stabilize electrode–scalp contact by applying controlled normal pressure, thereby ensuring reliable signal acquisition. From a system perspective, EEG caps can be broadly categorized into rigid-frame support systems and textile-based compliant systems.

Rigid or semirigid frame configurations represent the most common early-stage fixation solutions in gel-free EEG systems [31,97,109]. These systems rely on externally applied normal forces (typically by mechanical locking) to maintain electrode–scalp contact. A representative example is the early semirigid head-mounted EPOC system (Emotiv, USA). Their main advantages are structural stability, strong fixation ability, and ease of standardized deployment. However, in hair-bearing regions, these systems often require relatively long setup procedures and may introduce localized pressure concentrations and discomfort during wear, with prolonged use potentially causing skin indentation. In addition, owing to their relatively rigid architecture, they exhibit limited adaptability to interindividual variations in head geometry and lack fine control over local contact states, which can result in nonuniform pressure distribution, degraded signal quality, and increased susceptibility to motion artifacts.

To address these limitations, adjustable EEG caps have been developed, with tunability introduced at two levels of the electrode–head system: local interface control of electrode–scalp contact conditions and global structural adaptation to the geometry of the head. At the local level, two main strategies are used: passive adaptation and active tuning. Passive systems typically rely on compliant elements such as springs or gimbal-like mechanisms, which enable electrodes to automatically adjust their orientation under load to better conform to the scalp. For example, Chen et al. [110] proposed an electrode fixation system in which steel springs are incorporated between the electrode and the supporting platform, thus enabling pressure adaptation through elastic compression (Fig. 8(a)). In contrast, active tuning strategies rely on mechanical actuation, for example, via knobs or other adjustment mechanisms, to enable independent control of electrode contact conditions. For instance, in Quick series EEG caps (Cognionics, USA), modular electrodes are integrated with real-time impedance feedback, which enables point-by-point tuning of electrode–scalp contact on the basis of impedance variations. At the global level, cap structures are designed to accommodate variations in head geometry, particularly head circumference and overall curvature. For example, Wu et al. [111] developed a wireless EEG cap that consists of a dual-ring structural system and electrode panels, where the adjustable ring enables both rapid head-size adaptation and independent vertical adjustment of the electrodes (Fig. 8(b)). Lin et al. [112] used fused deposition modeling (FDM) 3D printing to fabricate a customizable electrode support framework, where head fitting is achieved through adjustable connector lengths and modular assembly strategies (Fig. 8(c)). Although these designs improve adaptability to interindividual differences, they remain predominantly rigid in nature, which limits wearing comfort and compliance, and they remain susceptible to micromotion-induced disturbances under dynamic conditions.

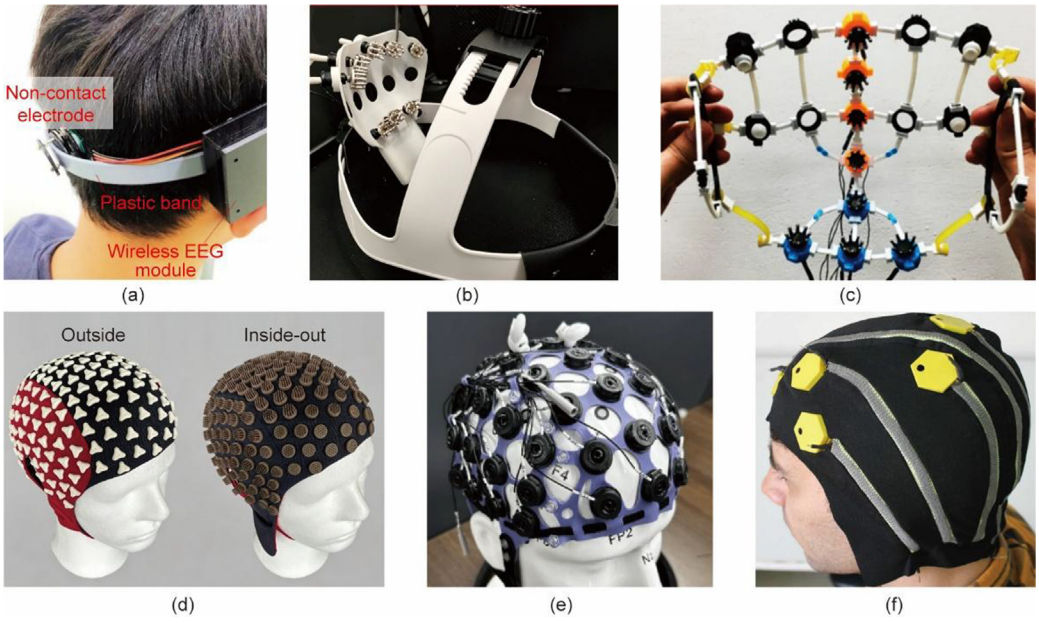


Fig. 8. EEG cap fixation. (a) Adaptive spring-structured EEG cap [110], (b) wireless EEG cap with independently adjustable electrodes [111], (c) 3D-printed EEG cap [112], (d) double-layer fabric EEG cap [114], (e) four-way adjustable silicone EEG cap [115], and (f) stretchable electronic EEG cap [116].

In contrast, flexible EEG caps achieve a more uniform pressure distribution through material compliance and represent an important recent direction in gel-free EEG systems. Headband-type designs [113] provide high portability and comfort but are limited in terms of channel count, which makes them unsuitable for high-density EEG recording. More typical flexible caps are constructed from cotton fabrics, mesh nylon, silicone, or stretchable electronic textiles, and rely on intrinsic material elasticity to achieve global conformity. Among these, cotton and mesh nylon structures provide good breathability and softness but limited mechanical support. For example, the dual-layer textile cap developed by Fiedler et al. [114] (Fig. 8(d)) demonstrates a balance between wearing comfort and structural stability. Silicone-based materials provide improved elastic stability and design flexibility, thereby enabling localized pressure regulation, as demonstrated by the 4D adjustable EEG cap proposed by Zhang et al. [115] (Fig. 8(e)). In addition, in stretchable electronic textiles, conductive pathways and electrodes are integrated within fabric architectures to form fully integrated flexible systems; for example, Motiepor et al. [116] developed a double-layer knitted electronic textile EEG cap (Fig. 8(f)).

The transition from rigid to flexible cap-based fixation represents an important step toward more user-friendly and adaptable EEG systems. The distinct characteristics of these fixation strategies in terms of placement efficiency, procedural complexity, wearing comfort, individual adaptability, and channel capacity are presented in Table 3 [111,112,114–116].

Overall, flexible EEG caps achieve a more uniform interfacial pressure distribution through material compliance, thereby reducing localized pressure discomfort and improving wearing comfort. They also improve adaptability to head geometry and hair distribution and, to some extent, improve stability under motion. However, fundamentally, these pressure-driven fixation strategies still rely on external normal forces to maintain electrode–scalp contact, which may introduce sustained mechanical loading during long-term wear, and remain susceptible to micromotion-induced disturbances in dynamic environments. These limitations become increasingly significant as EEG systems evolve toward high- and ultrahigh-density configurations. Owing to the limited scalp area available for electrode placement, current high-density EEG systems are based predominantly on 128- and 256-channel configurations [114], whereas further increases in electrode density require substantial reductions in the dimensions of the electrodes. Emerging technologies such as microneedle electrodes and ultrathin flexible-film electrodes [117] provide promising routes toward dense spatial integration by reducing the electrode footprint while maintaining acceptable interfacial performance. However, challenges associated with hair interference, electrode positioning, and stable electrode–scalp coupling remain difficult to resolve as the electrode density increases. Consequently, future ultrahigh-density EEG systems will likely require the coordinated optimization of electrode materials, structural design, and fixation strategies to simultaneously achieve low interfacial impedance, robust hair adaptability, and acceptable wearing comfort while enabling high-spatial-resolution neural recording.

4.2. Self-adhesive fixation

To eliminate reliance on continuous external pressure, self-adhesive fixation strategies have been developed. These approaches enable electrodes to actively conform to the scalp and maintain stable contact through either intrinsic material adhesion or externally triggered interfacial modulation mechanisms, thereby achieving fixation without sustained compression from EEG caps. This transition provides clear advantages in terms of reducing discomfort, improving breathability, and mitigating motion artifacts. Self-adhesive fixation strategies can be broadly categorized into three types, namely, bioinspired structural adhesion, stimulus-responsive adhesion, and intrinsic material adhesion, with the stimulus-responsive class further divided into temperature-responsive systems and pressure-responsive systems.

Bioinspired structural adhesion improves interfacial attachment by mimicking natural biological adhesion mechanisms through microstructural design. For example, Lan et al. [118] designed a skin-interfacing structure that was inspired by tree frog toe pads and features hexagonal pillar arrays and microchannel architectures, which enable stable attachment via capillary effects and localized negative pressure while maintaining breathability and durability. Inspired by the hierarchical micro- and nano-scale structures of gecko setae, Kim et al. [119] achieved dry adhesion through enhanced van der Waals interactions, thus providing strong yet reusable adhesion performance (Fig. 9(a)). A common feature of these structure-based approaches is that they avoid chemical intervention and generally exhibit low skin irritation; however, their performance is often compromised in hair-bearing regions because of interference from hair, which makes them more suitable for hairless or low-hair areas.

Stimulus-responsive adhesion enables reversible switching between adhesive and non-adhesive states through external stimuli that modulate material properties. In temperature-responsive systems, adhesion is governed primarily by temperature-induced changes in the mechanical modulus or phase state. One typical class relies on body-temperature-triggered modulus softening: Materials remain relatively stiff at room temperature for handling and positioning but soften upon skin contact at body temperature, thereby transitioning into a highly viscoelastic state that enables conformal adhesion and recovery of stiffness upon cooling for removal. Importantly, this process does not involve a true phase transition but rather thermally induced mechanical softening of the solid matrix. Representative examples include the thermal-responsive dry electrode material of Wang et al. [51], which exhibits a high shear-sensitive adhesive state at approximately 36.5 °C (Fig. 9(b)), and the bistable adhesive polymer system proposed by Lai et al. [120] (Fig. 9(c)). Another category involves externally heated liquid–solid phase transition systems, where heating induces a flowable state that enables interface filling, followed by solidification at body temperature to form a stable contact layer, with reversibility achieved by reheating. This mechanism relies primarily on the phase transitions of hydrogel materials to establish interfacial adhesion. Representative examples include the *in situ* coatable and rapidly gelling highly adhesive hydrogel

Table 3
Comparison of EEG cap-based fixation strategies for gel-free scalp EEG electrodes.

Fixation strategy	Placement time	Procedural complexity	Wearing comfort	Individual adjustability	Channels	Reference
Rigid cap	3 min	Simple	Moderate	Yes	8	[111]
	15 min	Moderate	Unknown	Unknown	8	[112]
Flexible cap	(19 ± 4) min	Simple	Moderate	Unknown	256	[114]
	(14 ± 3.2) min	Moderate	Excellent	Yes	64	[115]
	Unknown	Simple	Excellent	Unknown	8	[116]

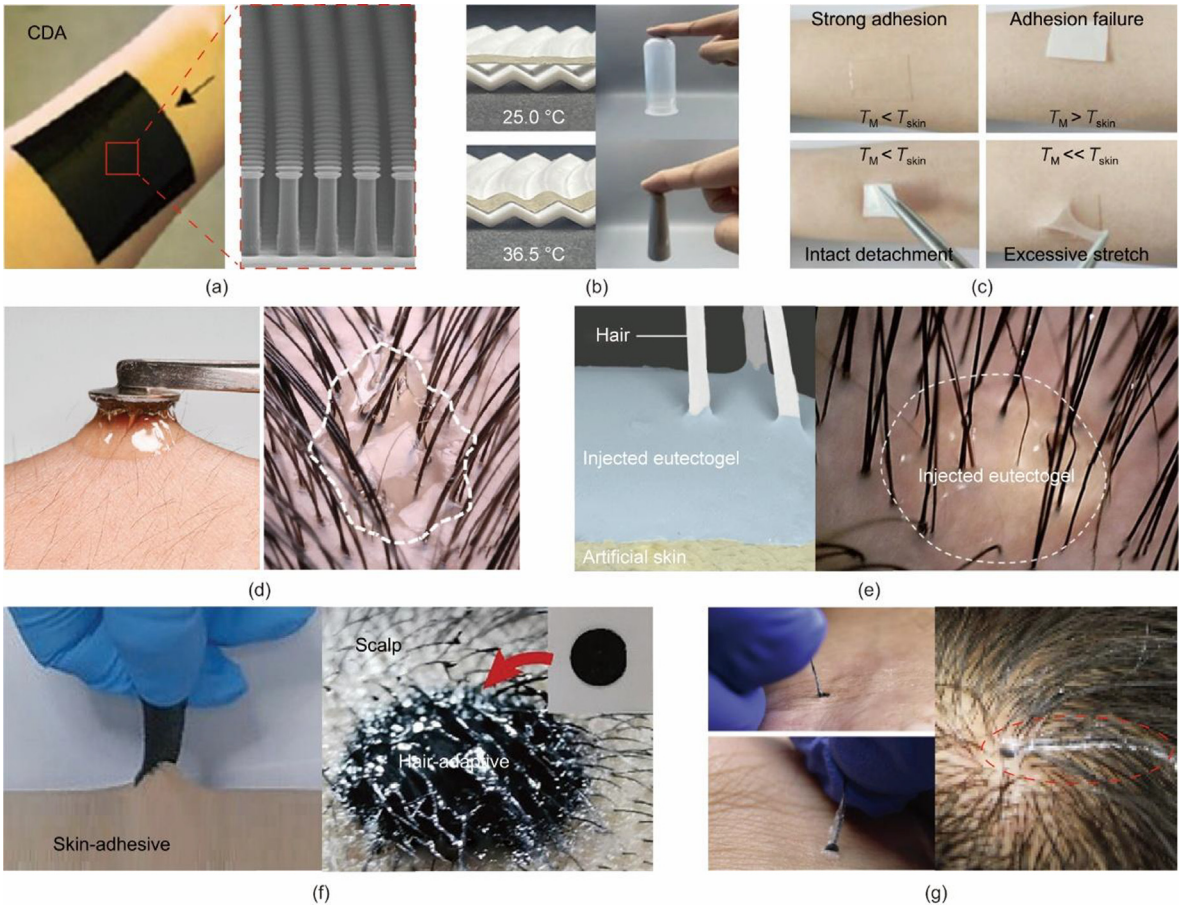


Fig. 9. Self-adhesive fixation. (a) Conductive dry adhesive (CAD) with a high-aspect-ratio fractal structure inspired by geckos [119], (b) adhesion performance of a pressure-sensitive adhesive for body-temperature-triggered phase-change dry electrodes [51], (c) changes in the adhesion force under temperature switching [120], (d) optical images of the skin adhesion and conformal scalp contact of a thermoreversible high-adhesion hydrogel (Gel-GGLiCit) [121], (e) scanning electron microscopy (SEM) and optical microscopy images of the gel–scalp interface of an *in situ* gelled thermally responsive phase-transition hydrogel (EGCDLi) [122], (f) adhesion performance of viscoelastic dry electrodes suitable for hair-bearing areas [123], and (g) adhesion performance of hair-like bioadhesive electrodes [126]. T_M: melting temperature; T_{skin}: skin temperature.

(Gel-GGLiCit) developed by Li et al. [121] (Fig. 9(d)), the injectable eutectogel electrode (EGCDLi) proposed by Wang et al. [122] (Fig. 9(e)), and the thermally controlled phase-transition hydrogel system reported by Wang et al. [92]. In all of these systems, gelatin is used as the primary matrix, and its thermally reversible phase-transition behavior imparts temperature-controllable adhesion to the hydrogel electrodes. In addition, in some systems, such as the MXene-reinforced hydrogel reported by Luo et al. [86], auxiliary components are introduced to trigger *in situ* gelation for interfacial fixation. Pressure-responsive adhesion involves leveraging the rheological behavior of materials to achieve interfacial adaptation. Under external pressure, the materials undergo shear thinning or localized flow, which enables them to infiltrate hair spaces and conform to microscale skin topography, thereby increasing the effective contact area and reducing the interfacial impedance. Upon the removal of stress, the viscosity recovers, thus enabling stable fixation. A representative example is a viscoelastic conductive electrode that is based on sorbitol-modified polydimethylsiloxane (SPBS) (Fig. 9(f)) [123]. Such systems require no external energy input and exhibit good adaptability in hair-bearing regions; however, their long-term stability can be affected by environmental humidity, temperature, and material aging.

Intrinsic adhesive electrodes rely on intrinsic interfacial adhesion for direct attachment without external triggering or auxiliary structures. Their compliant substrates improve the conformal contact with the microscale skin topography, thereby effectively reducing the interfacial impedance and mitigating motion artifacts. However, their effectiveness remains limited in dense hair regions, and they are susceptible to degradation under physiological conditions such as sweat and sebum exposure [124,125]. To address these issues, using 3D printing combined with bioadhesive inks, Ahmed et al. [126] developed hair-mimicking electrode structures in which each filament can simultaneously provide conductivity and adhesion, thereby maintaining relatively stable interfacial performance under dynamic disturbances (Fig. 9(g)).

Three self-adhesive strategies, namely, bioinspired structural adhesion, stimulus-responsive adhesion, and intrinsic adhesion, are summarized in Table 4 [51,92,118–120,122,123,126], with a focus on comparing their adhesion mechanisms, adhesion strength, adhesion durability, motion stability, and hair applicability in hair-bearing regions. Overall, self-adhesive fixation enables a shift in electrode–scalp interfacing from externally applied pressure toward material-driven interfacial coupling. This paradigm provides clear advantages in reducing discomfort and improving dynamic stability. However, challenges remain in ensuring reliable performance in hair-bearing regions and in translating these materials into practical, easy-to-use engineering systems.

5. Conclusions and perspectives

In this review, the recent progress on and development pathways of gel-free EEG electrodes are systematically summarized

Table 4
Comparison of adhesion-based fixation approaches for gel-free scalp EEG electrodes.

Adhesive strategy	Adhesion mechanism	Adhesion strength	Motion stability	Adhesion durability	Hair adaptability	Reference
Bioinspired structural	Microstructure	≈10 kP	Unknown	>4 h	Unknown	[118]
		≈1.3 N·cm^{-1}	Good	Unknown	Unknown	[119]
Stimulus-responsive	Thermally induced modulus transition	29.6 N·m^{-1} @36.5 °C	Unknown	48 h	Good	[51]
		(71 ± 4) kPa @32 °C	Good	7 d	Unknown	[120]
	Thermal sol-gel transition	(0.80 ± 0.076) N·cm^{-1}	Good	72 h	Good	[92]
		0.70 N·cm^{-1}	Good	24 h	Good	[122]
	Reversible interactions	Unknown	Good	48 d	Good	[123]
Intrinsic adhesion	Interfacial interactions	0.7 N·cm^{-1}	Unknown	>24 h	Good	[126]

from three perspectives: materials, structural design, and fixation strategies. From a materials perspective, electrode systems have evolved from rigid metals and ceramics to compliant elastomers, textiles, and hydrogel-based systems. This evolution has significantly improved the mechanical matching between electrodes and the scalp, thereby leading to improved wearing comfort. However, achieving stable electrical conductivity while reducing the modulus and establishing reliable electrode–scalp contact in hair-bearing regions remain fundamental challenges for current material systems. From a structural design perspective, comb-like morphological optimization improves static conformability by tuning geometric features to achieve macroscopic curvature matching and microscale pressure redistribution. In parallel, the integration of cushioning structures introduces elastic buffering elements that absorb dynamic disturbances and reduce motion artifacts. Together, these strategies improve the adaptability in hair-bearing regions under both static and dynamic conditions. However, the increased structural complexity is often accompanied by increased fabrication and integration challenges, and systematic co-optimization remains insufficiently explored. With respect to fixation strategies, EEG cap-based systems achieve electrode positioning and contact pressure control through mechanical frameworks and provide advantages in multichannel integration and rapid deployment. Nevertheless, they fundamentally rely on an externally applied continuous pressure, which imposes inherent limitations in terms of wearing comfort and dynamic stability. In contrast, self-adhesive fixation strategies enable active electrode–scalp attachment through material adhesion or stimulus-responsive mechanisms, thereby substantially reducing the contact pressure and improving the adaptability under motion. However, their reliability in hair-bearing regions and long-term stability still require further improvement.

Overall, the development of gel-free EEG electrodes is clearly transitioning from "rigid-passive-externally loaded" systems toward "compliant-adaptive-actively conformal" interfaces. Materials, structures, and fixation strategies are evolving toward greater flexibility, conformability, and stability. However, synergistic integration across these three dimensions remains limited, and a unified system-level design framework is still lacking. For instance, the coupling between compliant materials and structural buffering mechanisms, as well as the coordinated optimization of structural design and adhesion strategies, has yet to be fully established.

Future research may proceed in several directions. First, material systems that simultaneously exhibit high conductivity, low modulus, and long-term stability, particularly adaptive conductive composites tailored for hair-bearing regions, should be developed. Second, integrated material–structure codesign strategies that unify comb-like morphology optimization with cushioning mechanisms to achieve a balance between static conformability and dynamic stability should be established. Finally, intelligent fixation strategies for real-world applications that combine bioinspired adhesion and stimulus-responsive materials should be developed to achieve both unobtrusive wear and reliable long-term monitoring. With advances in materials, microfabrication, and responsive systems, gel-free EEG electrodes are expected to become comfortable, reliable, and truly wearable neural interfaces for use in BCIs and neural engineering.

CRediT authorship contribution statement

Dangxiao Yi: Writing – original draft, Visualization, Methodology, Investigation, Formal analysis. **Junyi Zhang:** Writing – review & editing, Conceptualization. **Songhao Liu:** Writing – review & editing, Visualization. **Yijun Wang:** Writing – review & editing, Conceptualization. **Yang Wang:** Writing – review & editing, Methodology, Formal analysis, Conceptualization. **Weihua Pei:** Writing – review & editing, Supervision, Resources, Funding acquisition.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Declaration of generative AI and AI-assisted technologies

During the preparation of this review manuscript, the author(s) used AI-assisted tools (e.g., DeepSeek) for language polishing and grammatical refinement. No AI tool was used to generate any substantive content, ideas, conclusions, or bibliographic information in this work. The author(s) take full responsibility for all the content presented in this manuscript.

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