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A review of the design analysis of wearable sweat sensor circuit system

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Abstract. As a popular core device for non-invasive health monitoring, the design of the circuit system of wearable sweat sensor directly determines the monitoring accuracy, wearable adaptability and practicability. This paper focuses on the design of wearable sweat sensor circuit system, summarizes the electrochemical sensing mechanism and characteristics of various signals of signal input modules, disassembles the adaptation design logic of op-amp circuits by signals, and finally expounds the signal processing process and the synergy mechanism of different signals, and then constructs a complete circuit system design framework to provide reference for engineering design and research in this field.

Keywords: wearable sweat sensor, circuit system design, signal input, op-amp circuit, electrochemical sensors, Signal processing, signal coordination

1. Introduction

1.1. Research background and significance

With the development of health monitoring technology towards non-invasive, real-time, and wearable, wearable sweat sensors have shown broad application prospects in sports health, clinical diagnosis, chronic disease management, and other fields due to their advantages of not needing blood collection and continuously monitoring human metabolites, electrolytes, and other health indicators [1]. At present, there are rich research results in wearable sweat detection equipment and supporting electrochemical sensors, such as enome-catalyzed electrochemical sensors can detect metabolites such as sweat lactate [2], ion-selective electrodes have become the core devices for sweat electrolyte monitoring [3], and the development of flexible Organic Electrochemical Transistor (OECT) sensors has provided technical support for flexible sweat detection equipment [4]. The core performance of the above examples depends on the reasonable adaptation design of the circuit system and the electrochemical sensor. In addition, the widely developed multi-parameter sweat sensor integrates a multi-channel electrochemical sensing array (voltage type + current type + impedance type) [5], which also needs to break through the bottleneck of wearable adaptability and detection accuracy by optimizing the circuit system [6].

As the "core center" of the sweat sensor, the circuit system is responsible for the acquisition, amplification, filtering, conversion and transmission of the output signal of the electrochemical sensor, of which the signal input module is the front end of the circuit system. The stability and accuracy of the input signal directly affect the design index and overall monitoring performance of the subsequent circuit. As the connection between signal input and subsequent signal processing, op-amp circuits play important roles in signal amplification, impedance matching, and noise suppression [7]. The signal processing part is the end point of the circuit system and the key to determining whether the result is valid. Therefore, it is important to optimize the design of wearable sweat sensor circuit systems to sort out the electrochemical sensing mechanism, clarify its adaptation logic with op-amp circuits, and sort out the signal processing processes and methods [8].

1.2. Research status

At present, the signal input of wearable sweat sensors is mainly based on electrochemical sensing mechanisms, and the traditional mainstream signal types include voltage, current, and impedance, and their core mechanisms revolve around "electrochemical interaction-electrical signal conversion", which has achieved the detection of conventional indicators such as metabolites and electrolytes, and the related technologies have approached clinical-grade accuracy [8, 9], for example, flexible wearable technology based on potentiometric sensing can achieve accurate detection of electrolytes such as Na+ and K+ in sweat [10]. Lactate oxidase electrochemical sensors can continuously detect lactate concentration in sweat [2], and Electrochemical Impedance Spectroscopy (EIS) technology has been applied to the detection and analysis of blood glucose-related markers in sweat [11].

At the same time, emerging and non-mainstream signal input technologies such as FET and photoelectrochemical focus on trace marker detection, with the goal of breaking through the limitations of traditional electrochemical mechanisms and providing a new direction for high-precision health monitoring [12], for example Field-Effect Transistor (FET)-based biosensors that can achieve high-sensitivity detection of trace markers in sweat [13], and photoelectrochemical sensors for accurate monitoring of metabolites such as uric acid and glucose in sweat [14], and flexible OECT sensors have also been applied to the detection of sweat trace biomacromolecules [4]. However, most of the existing studies focus on electrochemical sensing and material design, and there is a lack of research on the adaptability of various electrochemical sensor output signal characteristics and op-amp circuits, as well as the optimization of the connection of various parts [9], resulting in signal distortion, excessive noise, and high power consumption in some circuits [15], which limits the wearability and monitoring reliability of sensors. To systematically and intuitively illustrate the research framework outlined above, the overall structure and core logic of this review are presented in Figure 1. This figure summarizes the four core sections of this review: (a) Electrochemical Signal Input Mechanisms, which systematically organizes the signal characteristics, differences, and challenges of various electrochemical sensors; (b) Front-End Op-Amp Circuit Design, which provides an in-depth analysis of the circuit optimization and component selection logic tailored to different signals; (c) Signal Processing and Collaboration, which elaborates on the complete workflow from filtering and analog-to-digital conversion to multi-signal fusion; culminating in a comprehensive summary and outlook on circuit system design for wearable sweat sensors. The arrows between the sections indicate the technical logic and information flow.

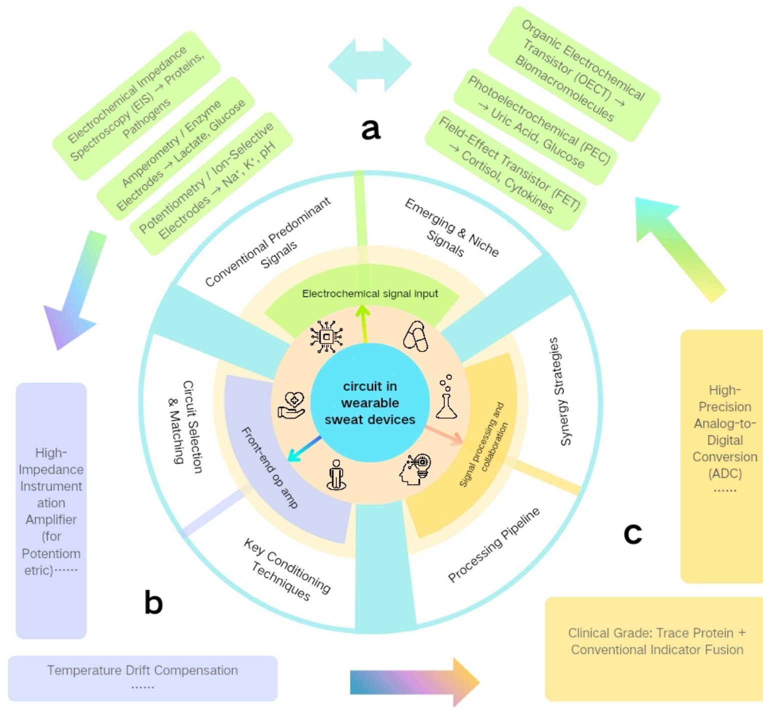

Figure 1. Framework of the circuit system design for wearable sweat sensors

Notes: (a) Signal input and sensing mechanisms: Encompasses a variety of electrochemical signal inputs, from traditional potentiometric, amperometric, and impedimetric to emerging FET-based, Photoelectrochemical (PEC), and OECT types, along with their corresponding target biomarkers. (b) Front-end op-amp and signal conditioning: Illustrates the core analog circuit modules tailored for different signal types, including key conditioning techniques such as high-impedance matching, low-noise amplification, temperature drift compensation, and filtering. (c) Signal processing and synergy strategies: Outlines the complete processing flow from Analog-to-Digital Conversion (ADC) and signal calibration to data fusion and wireless transmission, as well as multi-signal co-design strategies for both consumer-grade and clinical-grade applications.

2. Wearable sweat sensor signal input module design

The signal input module is the front end of the wearable sweat sensor circuit system, and its core function is to convert the target markers (metabolites, electrolytes, trace proteins, etc.) in sweat into identifiable electrical signals through various electrochemical sensors. The core principle of signal conversion is the specific interaction between markers and the sensing interface, which is converted into quantifiable electrical signals [6, 9]. The selection, parameter setting, and noise suppression strategy of op-amp circuits are directly determined by their signal type, amplitude, impedance, stability, and other characteristics [8].

For example, electrochemical sensors based on flexible materials can achieve a tight fit of the skin interface [16], printed Na+ and K+ sensors improve wearable adaptability through flexible substrate design [17], and fiber-based OECT sensors have the advantages of ultra-light and high breathability [18]. This part sorts out the signal input mechanism and characteristics of traditional, mainstream, emerging and non-mainstream electrochemical sensors, sorts out the differences between mainstream and emerging technologies, clarifies the core requirements of the connection between various signals and op-amp circuits, and provides a clear basis for the design of op-amp circuits.

2.1. Signal input mechanism of traditional mainstream electrochemical sensors

The traditional mainstream electrochemical sensor signal is based on the "marker electrochemical reaction-electrical signal conversion" as the core, and all signals are the output signals of the traditional mainstream electrochemical sensors in wearable sweat detection equipment, mainly including voltage, current, and impedance three types of signals, with mature technology, strong compatibility, and controllable cost, which is the mainstream choice of wearable sweat sensors, which is suitable for the monitoring of most conventional health indicators, and its signal type and amplitude characteristics directly determine the basic design direction and amplification requirements of op-amp circuits [8].

The core underlying mechanism of this type of signal input technology is unified, relying on the electrochemical interaction between the target marker and the electrode surface to convert the marker concentration or other information into the corresponding electrical signal, and the difference is only in the specific form of interaction and the type of signal output. For example, wearable multi-analyte electrochemical detection systems can simultaneously collect and analyze multiple electrochemical signals in sweat, and incorporate state-of-the-art AI algorithms for large-scale data analysis of complex biological and environmental systems [19], reconstructed modular systems can simultaneously monitor sweat ions and multiple physiological parameters [20], and wireless passive fabric sensors can simultaneously detect sweat multi-ions [21].

2.1.1. Current type signal input mechanism

The current signal input is based on oxidoreductase catalytic reactions, which is a common output signal for electrochemical sensors for the detection of wearable sweat metabolites. The specific principle is that the target marker undergoes a redox reaction catalyzed by specific oxidoreductase, releasing electrons or producing reduction/oxidation products, and the electrons are conducted through the electrode to form an electric current, and the amplitude of the current is linearly related to the marker concentration, which is also the core theoretical basis of this type of sensing [8, 22]. The amplitude of this type of signal is usually in the level of nA~μA, and the signal is weak and susceptible to noise interference, so it needs to be amplified by high gain through op-amp circuits, with a magnification ≥ 1,000x, while suppressing interference such as temperature and skin impedance during the reaction process [2, 7].

In practice, such signals often come from enzyme-catalyzed current electrochemical sensors in wearable sweat detection equipment, which are often used for the detection of metabolites such as glucose and lactate, and are widely used in various wearable sweat detection equipment, such as electrochemical sensors based on engineered lactate oxidase, which can realize the continuous detection of lactic acid in human sweat, and the detection performance meets the needs of health monitoring [23]. Enzyme-composite sweat sensors can achieve multiplex detection of diabetic nephropathy-related metabolites through current-based signals [24], while machine learning-assisted multimodal electrochemical analysis equipment can simultaneously detect current-based signals of metabolites such as tyrosine and uric acid in sweat [25]. This type of signal can accurately reflect the metabolic state of the human body and has important applications in scenarios such as exercise fatigue monitoring and auxiliary diagnosis of chronic diseases [9], but it has the shortcoming of insufficient accuracy in clinical-level trace detection [8].

2.1.2. Voltage type signal input mechanism

The basic principle is that "the electrode potential has a linear relationship with the logarithm of ion activity" described by the Nernst equation, and the target ion or H+ in sweat undergoes specific ion exchange with the ion carrier on the electrode surface, forming a stable potential difference, which is the core theoretical basis of this type of sensing [26]. The amplitude of this type of signal is usually at the mV level, the input impedance is high, the range is 106~109Ω, and it is easy to cause signal attenuation, and it is necessary to achieve high input impedance matching through op-amp circuits, and at the same time, the magnification is 500~1,000 times to avoid signal attenuation and distortion [7].

For example, the printed Na+, K+, and pH sensors can monitor the sweat electrolyte and acid-base state through voltage-based signal output [17], while the reconstructed modular system can simultaneously collect voltage-based signals of multiple ions in sweat [20]. Potential-type sensors based on flexible wearable technology can achieve flexible acquisition of voltage-based signals, improving wearable suitability [10], which can accurately reflect the body's electrolyte balance and acid-base metabolism, and is widely used in scenarios such as exercise dehydration monitoring and preliminary assessment of kidney function [9].

2.1.3. Impedance signal input mechanism

Based on the impedance change of the electrode-electrolyte interface, the specific recognition element modified by the electrode surface will change the charge distribution, roughness, and bilayer structure of the electrode surface after the specific combination of the target marker will change, resulting in a quantifiable change in the impedance of the electrode interface, which can be reflected by impedance analysis [27]. The frequency range of this type of signal is 10Hz~1kHz, the impedance value range is 102~108Ω, and the signal amplitude is unstable, and it is necessary to achieve impedance conversion and signal amplification through op-amp circuits, with a magnification of 1000~5000 times, and suppress power frequency interference [28].

For example, wearable EIS systems based on passive sweat such as Sankhala D can assess blood glucose changes through impedance signal analysis [11], and electrochemical impedance spectroscopy-based biosensors can achieve label-free detection of pathogens in sweat [29]. Portable configurable impedance measurement devices are used for the detection of sweat glucose [30]. The electrochemical impedance spectroscopy wearable system realizes real-time monitoring of biomarker changes in sweat [28], but this type of signal is greatly affected by factors such as sweat ion strength and temperature [17], and it needs to be coordinated with subsequent calibration circuits to improve stability.

2.2. Signal input mechanism of emerging and non-mainstream technologies

The core underlying mechanism of emerging and non-mainstream signal input technologies is to break through the limitations of traditional electrochemical interactions, combine new sensing mechanisms and nanomaterial modifications [31], to achieve a leap in signal detection performance and an expansion of scene adaptability. It forms a synergistic and complementary relationship with traditional mainstream technologies, which are adapted to conventional detection objects, and the demand for medium and low gain is mostly used in traditional electrochemical sensors in consumer-grade wearable sweat devices. Emerging technologies are suitable for trace detectables, high-gain, low-noise, and high-speed requirements, and are used in new electrochemical sensors in clinical-grade and high-end wearable testing equipment [9], such as wearable multi-mode body fluid monitoring systems, which integrate traditional and emerging signal input technologies to achieve simultaneous detection of multiple markers [32].

The core of traditional electrochemistry is the "direct electrochemical interaction between the target marker and the electrode surface", which relies on the direct conduction of electrons between the marker and the electrode, corresponding to the traditional electrochemical sensor in consumer-grade wearable devices, the detection object is mainly conventional metabolites and electrolytes, the signal type is voltage, current, and impedance, and the amplification requirement is 500~1,000 times medium and low gain [8, 9]. However, new interactions such as field effects and photocatalysis do not essentially rely on simple electrochemical redox processes, but realize signal conversion through special mechanisms such as "charge regulation" and photogenerated charge separation [33, 34], which correspond to new electrochemical sensors in clinical-grade and high-end wearable devices, which detect trace markers, and the signal types are leakage source current, photogenerated current, etc., and the signal amplitude is weaker, reaching nA or even pA level, which requires higher gain, low noise, and high-speed amplification design. The gain ≥ 10,000 times [35], suitable for clinical-grade and high-end wearable devices.

2.2.1. FET type signal input mechanism

The core underlying principle is "gate potential regulates the leakage source current", and the target marker will change the charge distribution on the gate surface after the specific combination of the transistor gate modification aptamer will change the charge distribution on the gate surface, thereby regulating the gate potential, and finally causing a quantifiable change in the leakage source current, and the signal output is the change of leakage source current [36]. This type of signal is a charge-regulated signal, which does not need to rely on direct electrochemical reactions, and is suitable for trace proteins, cytokines and other detectors, with a signal amplitude of nA level and a change of 0.1~1nA, and the signal is weak and susceptible to interference.

At present, FET-based signal inputs come from FET electrochemical sensors in wearable sweat detection devices, which have been applied to the research and development of clinical-grade wearable sweat detection devices, and FET-based wearable biosensors provide technical support for the detection of trace health markers in sweat [35], while Biofield-Effect Transistors (BioFETs) have shown good application prospects in point-of-care detection of sweat trace markers [37]. For example, B.wang et al. used wearable FET for the non-invasive detection of cortisol in sweat [38]. By optimizing the design of FET-type sensors, high-sensitivity detection of trace markers in sweat can be achieved, but its gate modification process is complex and costly [35], which limits large-scale mass production applications, and is currently only used in high-end clinical monitoring scenarios.

2.2.2. Photoelectrochemical signal input mechanism

The core underlying principle is that "photoexcitation of photoelectric materials generates electron-hole pairs, and markers regulate charge separation efficiency", and the target marker undergoes a specific photoelectric reaction with photoactive materials under the excitation of specific wavelength light, regulates the separation and compounding efficiency of photogenerated electron-hole pairs, and then forms a quantifiable photogenerated current signal [34]. This type of signal belongs to the photogenerated charge separation type signal, relying on photogenerated charge separation rather than pure electrochemical action, suitable for trace metabolite detection, the signal amplitude is nA level, the range is 0.5~3nA, the detection limit is as low as nM level, the anti-power frequency interference ability is better than the traditional electrochemical signal, which can make up for the lack of trace detection ability of the traditional signal, but the signal response depends on the lighting conditions, and is easily interfered by sweat impurities, the signal amplitude is weak and fluctuating, and it is necessary to achieve low noise amplification and signal correction through op-amp circuits. Magnification ≥ 10,000x [39].

For example, the integrated wearable photoelectrochemical sensor developed by Gao et al. achieves high-sensitivity monitoring of uric acid in sweat through visible light amplification technology [12], and the ultra-high-sensitivity wearable photoelectrochemical sensor developed by Liu et al. can achieve bias-free and accurate monitoring of sweat glucose [14]. Both types of sensors realize the detection of trace metabolites through photogenerated current signal output, and the detection accuracy meets the needs of clinical-grade monitoring. Its core advantage lies in its highly specific identification of trace metabolites, which is suitable for non-invasive and accurate monitoring of chronic diseases such as diabetes and hyperuricemia [9].

2.2.3. Flexible Organic Electrochemical Transistor (OECT) type signal input mechanism

The core underlying principle is "organic semiconductor channel conductance regulation", and after the target biomolecule is combined with the specific recognition element of OECT channel surface modification, ion implantation changes the doping state of the organic semiconductor, which in turn changes the channel conductance, and finally converts the conductance change into a quantifiable leakage current signal [40]. This type of signal belongs to the conductivity regulation type of organic semiconductor, with excellent flexible adaptability, good biocompatibility, and adaptation to trace biological macromolecule detectors, the signal type is leakage source current, the amplitude is nA level, and the response speed is fast, but the signal is susceptible to interference from temperature and sweat ion intensity [41], and the stability is poor.

For example, the configurable OECT sensor realizes multiplex analysis of sweat through the leakage source current signal output [4], the flexible stretchable OECT provides the core support for the flexible design of physiological sensing equipment [42], and the vertical OECT improves the sweat pH sensing capability through structural optimization [41]. Fiber-based OECTs have both wearing comfort and detection accuracy, and are suitable for biosensing at the skin interface [18]. Its core advantage is that its flexible stretchability can conform to the curved surface and deformation of human skin surface, adapt to dynamic wearing scenarios (such as exercise monitoring and sleep monitoring), and have good biocompatibility, which can reduce skin irritation and is suitable for long-term continuous monitoring scenarios [42].

The essence of the common characteristics of emerging and non-mainstream signals is to take the demand for detection objects as the core, break through the limitations of traditional electrochemistry through new sensing mechanisms, and achieve a leap in detection performance [9], and all signals are output signals from new electrochemical sensors in wearable sweat detection equipment, which are mainly used in clinical-grade and high-end wearable sweat detection equipment, forming synergy (traditional electrochemical sensor output) [32]. First, the detection object focuses on trace markers, covering trace proteins, metabolites, biological macromolecules, pathogens, etc., to meet high-end needs such as clinical diagnosis and early warning [12, 24]. Second, the signal type breaks through the traditional voltage, current, and impedance, and is dominated by new electrical signals such as drain source current and photogenerated current, with weaker amplitude, at the nA~pA level, and more complex impedance characteristics. Third, the amplification demand shows a trend of "high gain, low noise, and high speed", with amplification covering 10,000~20,000 times, and some technologies need to achieve ultra-low noise and high-speed response, while adapting to special needs such as signal calibration, interference suppression, and temperature/deformation compensation. This also puts forward higher requirements for the design of subsequent op-amp circuits, which need to take into account the amplification needs of medium and low gain (consumer-grade equipment, traditional electrochemical sensor signals) and high gain (clinical equipment, new electrochemical sensor signals), type impedance matching, noise suppression, and synchronization processing [5], and the various signal characteristics, amplification requirements and existing pain points sorted out in this part, combined with wearable sweat detection equipment and supporting electrochemical sensor examples, are also used for the selection, parameter design, and Structural optimization provides a clear and targeted basis.

3. The difference of various input signals and the design of adapted op-amp circuits

The overall composition of the op-amp circuit of the wearable sweat sensor needs to meet the core requirements of "low power consumption, miniaturization, high precision, and flexible adaptation" [7], while taking into account the specificity and stability of signal detection to adapt to the volume and power consumption requirements of different types of wearable sweat detection equipment, and the core adapts to the characteristics of the output signals of various electrochemical sensors, such as the op-amp circuit of consumer-grade sweat detection equipment needs to give priority to controlling the volume and power consumption, and adapting to the medium and low gain requirements of traditional electrochemical sensors

[43]. The op-amp circuit of clinical-grade testing equipment needs to prioritize accuracy and low noise to meet the high gain requirements of new electrochemical sensors [44].

The core of the op-amp circuit is composed of four functional modules connected in series, and each module collaborates to realize the acquisition, processing and output of the output signal of the electrochemical sensor, laying the foundation for subsequent signal analysis and system integration. The core function of the preprocessing module is signal buffering and impedance matching, which realizes impedance conversion for high-impedance signals (e.g., voltage-type and OECT-type electrochemical sensor output signals) [45], and low-noise preamplification for weak signals (e.g., FET and photoelectrochemical electrochemical sensor output signals) [12, 35] to avoid initial signal attenuation. The gain amplification module adopts an adjustable gain design to adapt to the amplification requirements of different amplitude signals, and balances the amplification accuracy and noise suppression through a multi-stage amplification structure. The filtering module adopts the combined design of "low pass + notch + high pass" to filter out different types of noise and ensure the purity of the output signal of the electrochemical sensor. The buffer output module improves the signal driving ability and avoids the interference of the subsequent processing module on the amplified signal [45], such as the op-amp circuit of the wearable multi-parameter electrochemical detection system realizes the synchronous amplification and processing of multiple signals through this modular design [46].

The underlying logic of amplification and filtering of different signals of op-amp circuit needs to be combined with the signal characteristics of wearable sweat sensors (all of which are electrochemical sensor output signals). The amplification underlying logic is based on the three-dimensional adaptation and amplification scheme of signal amplitude, impedance characteristics, and signal type, and is optimized by combining the application scenarios of wearable devices (consumer-grade, clinical-grade, flexible wearable) and supporting electrochemical sensor types. For signals with strong amplitude (voltage and current signals output by traditional electrochemical sensors), single-stage or two-stage gain amplification is used to control power consumption and cost first, and there is no need for complex low-noise design to adapt to consumer-grade devices [47]. For signals with weak amplitude (FET and photoelectrochemical signals output by new electrochemical sensors), the co-design of "low-noise preamplification + multi-level gain amplification" was adopted [35, 44], and the preamplification suppression electrochemical sensor type was optimized. For signals with strong amplitude (voltage type and standby output of traditional electrochemical sensors; For high-impedance signals (output signals from voltage-type and OECT-type electrochemical sensors), the high-impedance is converted to low-impedance by the impedance conversion module [10, 41], and then amplified to avoid signal attenuation and distortion. (current, voltage, and FET electrochemical sensor output signals), using DC amplification structure to suppress temperature drift and offset voltage [48]; for AC signals (impedance electrochemical sensor output signals), an AC amplification structure is used to match the signal frequency range and avoid DC interference [11, 28].

The underlying logic of filtering adapts the filtering method and parameters according to the noise type and signal frequency, and at the same time optimizes the use environment of wearable devices to strengthen the suppression of power frequency interference and temperature interference to ensure the accuracy of signal detection. The core noise in the sweat sensor includes three categories, the high-frequency noise (frequency ≥ 10kHz) mainly comes from circuit wiring interference and external electromagnetic radiation, using low-pass filtering, the filter cut-off frequency matches the highest frequency of the signal, filtering out the high-frequency interference while retaining the effective signal output by the electrochemical sensor; Power frequency interference (50 Hz and harmonics) mainly comes from power grid radiation, and notch filtering is used to accurately filter out power frequency and harmonic interference and improve signal purity [49].

Temperature interference mainly comes from ambient temperature changes and human body temperature fluctuations, and thermistor-assisted temperature compensation filtering design [17] is used to suppress signal drift caused by temperature. In addition, the signal frequency determines the filtering bandwidth, and high-speed response signals (such as FET and photoelectrochemical sensor output signals) need to be designed with wide-bandwidth filtering [12, 35] to avoid signal distortion and adapt to clinical-grade high-speed detection equipment. Low-frequency signals (such as traditional voltage and current-based electrochemical sensor output signals) can be designed with narrow-bandwidth filtering [8] to improve noise suppression and adapt to consumer-grade conventional detection equipment. The filtering design in wearable scenarios needs to take into account both noise suppression and signal response speed, adapt to the needs of dynamic monitoring, and control the volume and power consumption of the filtering circuit to meet the design concept of miniaturization and low power consumption [7].

3.1. Traditional mainstream signal adaptation op-amp circuit design

The core differences are concentrated in the signal output form, amplitude range, impedance characteristics, noise source, and adaptability of the detected object, which directly determine the adaptation direction of the op-amp circuit, without the need for complex low-noise, high-speed design, and focus on gain matching, impedance adaptation, and basic noise suppression [7]. It is suitable for consumer-grade wearable sweat detection equipment and supporting traditional electrochemical sensors [2, 3, 11].

The core differences of the three types of signals are summarized as follows: the current-type signal (enzyme-catalyzed electrochemical sensor output) is DC current, amplitude 0.5~5μA, and low impedance [2]; The voltage-type signal (ion-selective electrode output) is DC voltage, amplitude 10~100mV, and high impedance [3, 10]. Based on these differences, combined with the low power consumption and low cost requirements of consumer devices [7], the selection, parameter design, and structural design of op-amp circuits need to be optimized.

3.1.1. Current type signal adaptation op-amp circuit

The signal of the current type signal is the electron conduction current generated by the redox reaction, which is the exclusive output signal of the wearable sweat metabolite detection enzyme-like catalytic electrochemical sensor, the output form is continuous DC current, the amplitude range is 0.5~5μA, the input impedance is low, 10^2~10^4Ω, the noise is mainly derived from temperature fluctuations, skin impedance interference and electrode electron conduction noise (electrochemical sensor itself noise) [17], suitable for conventional metabolite detection, the signal response speed is moderate, The signal stability is good, and there is no special external auxiliary requirement [2], and it is suitable for consumer-grade sweat metabolite detection equipment and supporting enzyme-catalytic electrochemical sensors.

Its op amp selection prioritizes general-purpose operational amplifiers with low noise, high gain, and high slew rate [7], taking into account the requirements of low cost and low power consumption, such as LMP7721 and OPA2277, among which the LMP7721 has low cost and low power consumption, and is suitable for low-cost mass production scenarios. OPA2277 lower noise and suitable for consumer-grade devices with high accuracy requirements [50]. The magnification is set at 1,000~2,000x, and the inverse-phase proportional amplification circuit structure is used [51], which has the advantages of low input impedance, stable magnification, and simple circuit structure, and at the same time, it is necessary to add a temperature compensation resistor and feedback resistor in series [17] to offset the interference of temperature fluctuations, connect a 100Q current limiting resistor in series at the input end to prevent excessive current damage to the op-amp, and connect a small capacitor in parallel at the output to filter out power frequency interference [48] to ensure the stability of the output signal. This circuit design has been applied to op-amp systems of metabolite detection equipment such as sweat lactate and glucose [2], achieving efficient amplification and noise suppression of current-type signals.

Based on the above, to suppress crosstalk in multi-channel voltage signals for sweat ion detection and compensate for temperature drift, an amplification circuit with high common-mode rejection ratio and online compensation capability is required. Accordingly, Figure 2 presents the designed high common-mode rejection ratio instrumentation amplifier circuit.

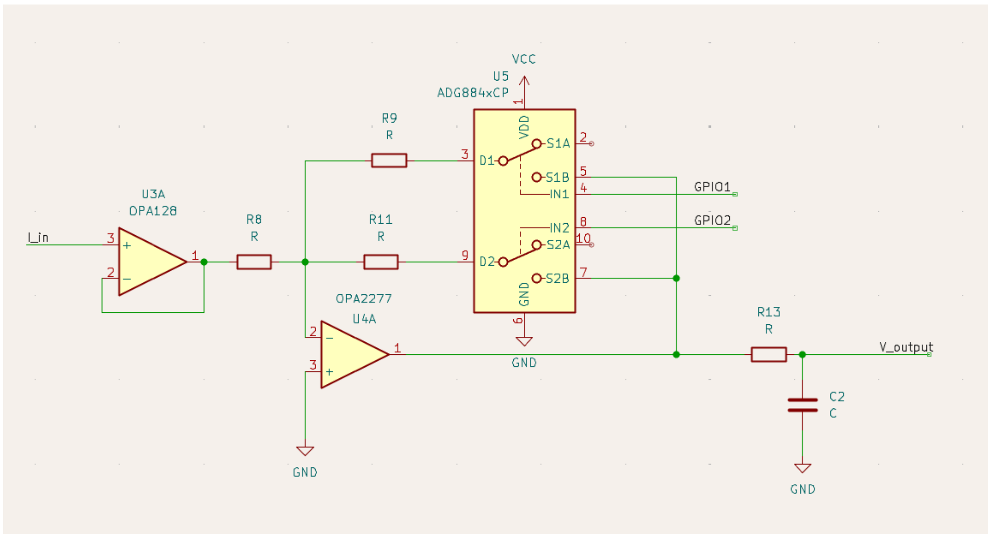

Figure 2. Low-noise programmable-gain transimpedance amplifier circuit for weak amperometric signals

Notes: The core of the circuit is a classic transimpedance amplifier structure, which is dedicated to amplifying and converting weak current signals (0.1 nA – 5 nA) from FETs, photoelectrochemistry, and other sensors. Operational amplifiers provide ultra-low input noise current. The feedback network uses a multi-level precision resistor switched by an analog switch to achieve programmable gain from 1 MΩ to 100 MΩ (corresponding to a gain of 106 – 108 V/A) to accommodate different amplitude signals. Feedback capacitors connected in parallel are used to stabilize the circuit and limit bandwidth. The GUARD drive circuit keeps the GUARD potential equal to the sensitive input node through a unity gain buffer, greatly reducing the impact of PCB leakage current and parasitic capacitance, which is the key to achieving high-precision measurements at the pA level. The circuit provides a highly sensitive front-end solution for the detection of trace markers in wearable sweat sensors.

3.1.2. Voltage type signal adaptation op-amp circuit

The signal of the voltage type signal is essentially the electrode potential difference generated by ion exchange, which is the exclusive output signal of the wearable sweat electrolyte and pH detection ion-selective electrode (voltage electrochemical sensor) [3], the output form is DC voltage, the amplitude range is 10~100mV, the input impedance is extremely high, it is 10^6~10^{10}Ω, the noise mainly comes from electrode drift (electrochemical sensor core noise), ion interference and signal attenuation noise [52], suitable for conventional electrolyte and pH detection. The signal response speed is slow, signal drift is easy to occur in motion scenarios, and crosstalk is easy to occur when multi-channel integration [20] is suitable for consumer-grade sweat electrolyte detection equipment and supporting ion-selective electrodes.

OPA128 and AD8601 are recommended for OPA128, OPA128, OPA128 input impedance up to 10^{12}Ω, offset voltage ≤ 10 μV, and temperature drift ≤ 0.1 μV/°C, which can effectively suppress signal attenuation and drift [53,50], and are suitable for electrolyte detection equipment with high accuracy requirements [3]. The AD8601 has lower power consumption and is suitable for low-power wearable scenarios [7]. The magnification is set at 500~1,000x, and a differential amplification circuit structure is used [51], and the high common-mode rejection ratio (CMRR ≥ 80dB) of differential amplification is used to suppress multi-channel crosstalk and common-mode noise, and an anti-drift compensation circuit is designed [48], and the offset voltage is calibrated by series adjustable resistors, and an independent differential amplification unit is added to each channel when the multi-channel input is inserted, which further reduces crosstalk and ensures the accuracy of multi-parameter simultaneous detection. This design has been applied to the op-amp circuit of multi-ion synchronous detection equipment [20, 21] to achieve high-impedance matching and accurate amplification of voltage-based signals.

To achieve accurate measurement of the weak current signals from sweat sensors, the front-end circuit must possess ultra-high gain, low noise, and gain tunability. The low-noise programmable transimpedance amplifier circuit shown in Figure 3 is the solution devised to meet this objective.

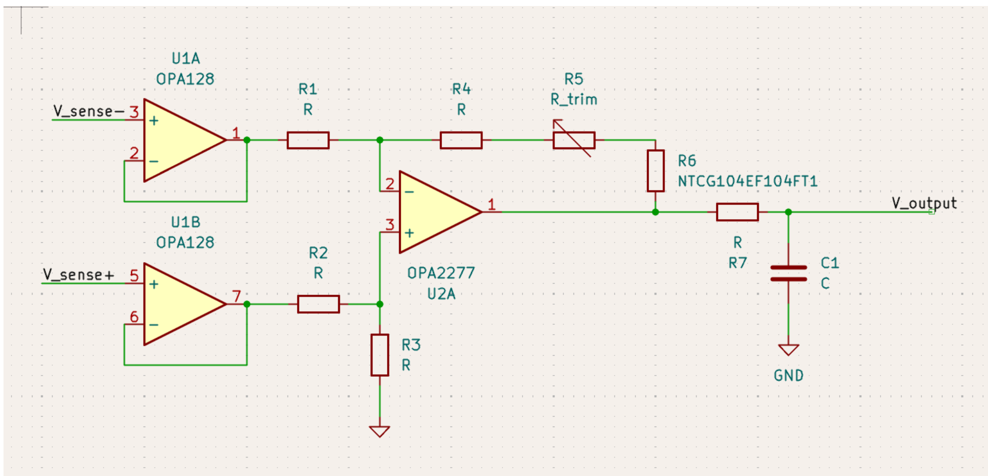

Figure 3. High Common-Mode Rejection Ratio (CMRR) instrumentation amplifier circuit for potentiometric signals

Notes: The circuit features a triple op-amp instrumentation amplifier structure specifically designed to amplify high-impedance weak voltage signals (10-100 mV) generated by ion-selective electrodes. The input buffer stages provide extremely high input impedance to avoid signal degradation. The differential amplification stage sets the gain (500-1,000x) through a resistive network and uses its high CMRR to suppress common-mode noise. A unique temperature compensation network (R_trim, R_therm) is integrated into the feedback loop to calibrate the offset voltage and dynamically compensate for temperature drift. The capacitance at the output provides low-pass filtering. This design is the core solution to the challenges of voltage-based sweat sensor signal attenuation, drift, and crosstalk.

3.1.3. Impedance signal adaptation op-amp circuit

The impedance signal is essentially the impedance change caused by the change of the electrode interface structure, and is the exclusive output signal of the wearable sweat protein, microorganism, and pathogen detection immunoimpedance electrochemical sensor [11, 29], the output form is AC impedance, the frequency range is 10 Hz ~ 1kHz, the impedance value range is $10^2 \sim 10^8$ Ω, the span is extremely large, and the noise mainly comes from electrode double layer fluctuations (electrochemical sensor noise), sweat ion intensity changes, and power frequency interference [52] For conventional protein and microbial detection, the signal amplitude is unstable, and the impedance signal needs to be converted into an amplified voltage/current signal and then amplified [11, 30], and it is suitable for consumer-grade skin health monitoring equipment and supporting immunoimpedance sensors.

OPP211 and LMV324 are recommended models, with a wide bandwidth of up to 1 MHz and an input impedance of up to 10^{11} Ω, which can adapt to the conversion and amplification of wide impedance signals

[50], and is suitable for impedance detection equipment with high accuracy [11]. LMV324 is low-cost, highly compatible, and suitable for multi-channel integration scenarios [7]. The first stage uses an operational amplifier to form an impedance follower, which converts a wide range of impedance signals (immune impedance sensor output) into voltage signals of 0~1V to achieve impedance matching and signal buffering. The second stage adopts a full-phase proportional amplification circuit [51], the magnification is set to 1,000~5,000x, and an anti-interference impedance conversion circuit and a 50Hz notch filtering circuit are designed [48], which filters out power frequency interference and electric double-layer fluctuation noise, and adds a 1μF regulated capacitor stabilization signal at the output [17] to ensure the stability and accuracy of the amplified signal. The circuit design provides core amplification support for electrochemical impedance spectroscopy-based sweat detection devices [28, 29].

3.2. Op-amp circuit design for emerging and non-mainstream signals

The core differences are focused on the signal nature, amplitude range, noise characteristics, response speed, detection object adaptation, and external auxiliary requirements, which determine the adaptation direction of op-amp circuits, focusing on ultra-high gain, ultra-low noise, and high-speed response [7], while combining signal shortcomings (The new electrochemical sensor's own noise characteristics) design a special calibration and compensation structure [17], which is suitable for clinical-grade and high-end wearable sweat detection equipment and supporting new electrochemical sensors [4, 12, 35]. Compared with traditional mainstream signals (traditional electrochemical sensor outputs), emerging signals (new electrochemical sensor outputs) have weaker amplitudes, more complex noise, and higher response speed requirements [9], so op-amp circuits need to adopt more sophisticated designs, with special suppression and calibration structures, to balance accuracy, speed, and power consumption, while meeting the high-precision requirements of clinical-grade devices and the low power consumption and miniaturization requirements of wearable devices [7, 53].

3.2.1. FET type signal adaptation op-amp circuit

The signal essence of the FET type signal is the drain source current change generated by the gate charge regulation [35], which is the exclusive output signal of the wearable sweat trace protein and cytokine detection FET electrochemical sensor, the output form is the DC current change, the amplitude is extremely weak, the amplitude is 0.1~1nA, the input impedance is high, the noise is mainly due to charge fluctuations (FET sensor noise) and sweat ion intensity interference [13], suitable for trace protein and cytokine detection, the response speed is extremely fast (The response time $\leq$ 1s), poor signal stability, and susceptibility to temperature and humidity [37], are suitable for clinical-grade trace detection equipment and supporting FET electrochemical sensors.

Op amp selection prioritizes ultra-low noise, ultra-high gain, and high-speed operational amplifiers [53], and the AD8099 is recommended as the model to meet the weak signal amplification requirements of FET sensors [13]. The AD8099 has low power consumption and adjustable gain, making it suitable for low-power and high-precision scenarios [7]. The magnification is set to 10,000~15,000x, and the two-stage circuit structure of "low-noise pre-amplification + high-speed gain amplification" is adopted [35], and the pre-amplification stage adopts a common source electrode amplification structure to suppress the initial charge noise (core noise of FET-type sensors). The gain amplification stage adopts an adjustable gain design [7], dynamically adjusts the magnification according to the amount of signal change, and at the same time designs a signal calibration circuit to calibrate signal drift through a reference current source [48], and the ion interference suppression circuit uses an ion-selective membrane to isolate interfering ions in sweat [17] to ensure the accuracy and stability of the amplified signal. The circuit design provides a core signal amplification solution for FET sweat detection devices [35], achieving trace detection at the pg/mL level.

3.2.2. Op-amp circuit adaptation for photoelectrochemical signals

The signal essence of the photoelectrochemical signal is the photogenerated current formed by the separation of electron-hole pairs generated by photoexcitation [12], which is the exclusive output signal of the wearable sweat trace metabolite detection photoelectrochemical sensor, the output form is DC current, the amplitude is weak, the amplitude is 0.5~3nA, the input impedance is medium, and the noise is mainly derived from light fluctuations, sweat impurity interference, and photogenerated charge compound noise (photoelectrochemical sensor itself) [14], which is suitable for trace metabolite detection and has a fast response speed (Response time 1~2s), signal stability is average, dependent on external light, fluctuations in light intensity will lead to signal deviation [12], suitable for occupational health, chronic disease monitoring wearable devices and supporting photoelectrochemical electrochemical sensors.

OPP188 and LMV821 are recommended models, with an input noise voltage as low as 8.8 nV/Hz and an offset voltage of ≤ 2 μV, which can effectively suppress the noise of weak signals [53] and adapt to high-precision optoelectronic chemical detection equipment [12]. The LMV821 has extremely low power consumption (quiescent current of only 80 μA), making it suitable for long-term wearable scenarios [7]. The magnification is set at 10,000~12,000 times, and the inverted proportional amplification circuit structure is used [51], which has high amplification accuracy and good noise suppression effect, and the impurity interference suppression circuit is designed to remove macromolecular impurities in sweat through the filter membrane to reduce the interference of impurities on photoelectric reactions [12]. The light stabilization circuit uses a constant current source to drive the light source to ensure stable light intensity [14] and avoid signal deviation caused by light fluctuations. The signal anomaly detection circuit monitors the photogenerated current signal in real time, triggers the calibration mechanism when the signal fluctuates abnormally [48], and the input terminal adopts a light-blocking shield design to reduce external light interference. This circuit design has been applied to photoelectrochemical detection equipment for sweat, uric acid, and glucose [12, 14], achieving accurate detection at the nM level.

3.2.3. OECT type signal adaptation op-amp circuit

The signal essence of the OECT signal is the leakage source current generated by the channel conductance regulation of organic semiconductors [4], which is the exclusive output signal of the wearable sweat trace biomacromolecule detection OECT flexible electrochemical sensor, the output form is DC current, the amplitude is weak, the amplitude is 1~5nA, the input impedance is high, the noise is mainly derived from temperature fluctuations, sweat ion intensity interference, and channel charge transfer noise (OECT sensor itself noise) [41], which is suitable for trace biomacromolecule detection and has a fast response speed (response time ≤ 1.5s), poor signal stability, excellent flexible adaptability, and can follow skin deformation [42], suitable for flexible wearable devices and supporting OECT-type flexible electrochemical sensors.

Op amp selection prioritizes low-noise, high-gain, and high-speed operational amplifiers [53], while taking into account the flexible adaptability of the circuit [18], and the recommended model is LMV7219, with good adaptability to flexible packaging and low power consumption, which is suitable for flexible wearable scenarios [18]. The magnification is set at 12,000~15,000x, and a two-stage amplification structure is used [51], with the first stage being low-noise pre-amplification to suppress temperature noise, ion interference, and channel charge transfer noise (OECT sensor core noise) [41]. The second stage is high-speed gain amplification to ensure fast signal response, and a high-precision temperature compensation circuit is designed [17], which uses thermistors to detect temperature in real time and dynamically calibrate signal deviations caused by temperature fluctuations. The ion interference suppression circuit uses an ion exchange membrane to isolate interfering ions [41], and the circuit adopts a flexible PCB design [42], which adapts to the flexible characteristics of OECT to ensure wearing comfort and signal stability, and the package adopts a waterproof design to avoid short circuits caused by sweat intrusion into the circuit [18]. The circuit design provides the core support for the flexible OECT sweat detection device [4, 42], taking into account both wearable adaptability and detection accuracy.

To systematically summarize and compare the key characteristics of various electrochemical sensing signals and their differentiated requirements for front-end op-amp circuit design, Table 1 consolidates the core parameters—including amplitude, impedance, and noise sources—of mainstream and emerging electrochemical signals in wearable sweat sensors. It also specifies the corresponding design essentials such as op-amp selection, gain setting, and circuit architecture. This table provides a direct and clear reference for optimizing circuit design tailored to different detection targets.

Table 1. Characteristics of various electrochemical signals of wearable sweat sensors and key points of design points of adaptive op-amp circuits

Signal Type (Sample Object)	Signal nature and detection target	Typical amplitude range	Input impedance	Core noise/interference sources	Op amp core requirements	Typical magnification	Core circuit structure
Voltage type (Na^+, K^+, pH).	Ion exchange potential difference (electrolyte, pH)	10 – 100 mV	extremely high (10^6 – 10^9 Ω).	Electrode drift, ion interference, signal attenuation	High input impedance, low offset voltage, low drift	500 – 1,000 times	Differential Amplification Circuit (High CMRR)
Current type (lactic acid, glucose).	Redox reaction current (metabolite)	0.5 – 5 μA	Low (10^2 – 10^4 Ω).	Temperature fluctuations, skin impedance, electronic noise	Low noise, high gain, high slew rate	1,000 – 2,000 times	Inversely proportional amplification circuit
Impedance type (protein, pathogen).	Change in electrode-solution interface impedance (protein, microbe)	Impedance: 10^2 – 10^8 Ω	Wide range	Electric double layer fluctuations, ionic strength changes, power frequency interference	High input impedance, high common-mode rejection ratio, and wide bandwidth	1,000 – 5,000 times (voltage)	Impedance-voltage conversion + in-phase amplification
FET type (cortisol, cytokines).	Gate charge-regulated leakage current (trace protein)	0.1 – 1 nA (ΔI)	higher	charge noise, ionic strength interference, temperature and humidity	Ultra-low noise, ultra-high gain, high speed	≥ 10,000 times	Low-noise preamplification + high-speed gain amplification
Photoelectrochemical type (uric acid, glucose).	Photogenerated charge separation current (trace metabolites)	0.5 – 3 nA	Medium	Lighting fluctuations, impurity interference, and charge compound noise	Low noise, high gain, high slew rate	10,000 – 12,000 times	Inverting Ratio Amplification Circuit (Light Stabilization)
OECT type (biological macromolecule).	Channel conductance-regulated leakage current (trace biomacromolecule)	1 – 5 nA	higher	Temperature fluctuations, ion interference, channel charge noise	Low noise, high gain, high speed, and flexible adaptation	12,000 – 15,000 times	Two-stage amplification (low-noise pre-+ high-speed gain)

4. Wearable sweat sensor signal processing and collaborative design of different signals

After amplification and filtering by op-amp circuits, various electrochemical sensor signals (all of which are exclusive output signals for wearable sweat detection equipment) need to be "signal purification-conversion-analysis-feedback" through a systematic signal processing process [5], and at the same time, relying on the synergy mechanism of different signals to break through the limitations of single signal detection and improve the comprehensiveness, accuracy, and practicability of wearable sweat detection equipment [55]. This part focuses on the core process of signal processing, combines the scenario requirements of wearable sweat detection, clarifies the synergistic logic of different signals, connects the design content of the op-amp circuit in the previous article, and forms a complete link of "signal input-op-amp amplification-signal processing-collaborative output" [5], adapts to the application requirements of different types of wearable sweat detection equipment, and summarizes the core ideas of the full text to form a closed loop of review.

The core goal of signal processing of wearable sweat sensors is to adapt to the core requirements of wearable devices of "low power consumption, miniaturization, real-time, and precision" [56], convert various electrochemical signals amplified by op-amp into quantifiable and interpretable health data, and at the same time make up for the shortcomings of single signal detection through the synergy and complementarity of different signals, and finally realize the synchronous and accurate monitoring of conventional indicators and trace markers in sweat, providing reliable data support for health assessment and disease early warning [9]. Unlike the signal processing of traditional electrochemical detection equipment, the processing in wearable scenarios needs to take into account both accuracy and power consumption, adapt to the characteristics of flexible and dynamic wearability, and all processes are deeply bound to the supporting electrochemical sensors [57].

4.1. Core process of wearable sweat sensor signal processing

The signal processing process is based on "adapting to wearable scenarios and fitting the signal characteristics of electrochemical sensors" [5], connecting the output of the third part of the op-amp circuit, and is divided into four core links: Analog-to-Digital Conversion (ADC), signal calibration, signal analysis and feature extraction, and data transmission [58], and each link collaborates to achieve a complete conversion from "original electrical signal" to "health data", of which signal calibration is the key to improving detection accuracy, and all processing algorithms are adapted to the signal characteristics of different types of electrochemical sensors [5].

4.1.1. Analog-to-Digital Conversion (ADC): digital adaptation of electrical signals

Analog-to-digital conversion is the basic link of signal processing, and its core function is to convert the analog electrical signal output by the op-amp circuit into a digital signal that can be recognized and processed by the microcontroller [58], and its conversion accuracy and sampling rate directly determine the accuracy of subsequent signal analysis.

Due to the significant difference in signal type and amplitude, ADCs need to be selected in a targeted manner [57]: traditional mainstream signals use 12~16-bit low-power ADCs with a sampling rate of 10~100Hz to meet the needs of conventional detection [56], of which the impedance ADC signal bandwidth is 1kHz, and the sampling rate may be ≥ 2kHz. Emerging and non-mainstream weak signals need to use 16~24-bit high-precision ADCs, and the sampling rate should be increased to 100~1,000Hz to meet the requirements of high-speed response [59, 60]. In the multi-channel integration scenario, multi-channel ADCs are used to realize the synchronous conversion of traditional and emerging signals, reducing the circuit size [61], such as the wearable multi-parameter electrochemical detection system, which realizes the synchronous digital conversion of multiple electrochemical signals in sweat through multi-channel ADCs [5], taking into account the detection efficiency and circuit miniaturization.

4.1.2. Signal calibration: a key link in improving accuracy

Signal calibration is the core link to solve the interference problem and improve the detection accuracy in wearable scenarios [6], and the core goal is to offset the detection bias caused by factors such as temperature fluctuations, changes in sweat ion strength, sensor drift, circuit noise, etc [17]., and ensure that the processed data accurately corresponds to the concentration of target markers in sweat [5].

Calibration strategies need to be differentiated according to signal types [57]: traditional mainstream signals adopt the dual strategy of "reference calibration + temperature compensation" [17], regularly calibrate circuit conversion deviations, and dynamically correct the influence of temperature on detection through thermistors, such as printed Na+ and K+ sensors achieve temperature calibration through PEDOT: PSS-based thermistors, improving the detection accuracy of voltage-based signals [17]. Emerging and non-mainstream signals need to use "multi-parameter joint calibration" [32], which solves the fluctuation problem of photogenerated current signals by calibrating exclusive interference factors such as light and ion intensity based on temperature and reference calibration, such as constant current source light calibration and impurity filtration calibration [12]. At the same time, a user-personalized calibration model can be established through the initial detection [62], which adapts to individual differences in sweat composition and skin condition, further improving the detection accuracy, such as the blood glucose estimation model based on sweat detection, which greatly reduces the detection error through personalized calibration [62].

4.1.3. Signal analysis and feature extraction: the core of health data interpretation

The core function is to extract signal features related to the concentration of target markers through lightweight algorithms, eliminate redundant noise, and establish a quantitative relationship between signal features and marker concentrations [8], while adapting to the low-power requirements of wearable devices to avoid complex calculations occupying excessive hardware resources [56].

Analysis algorithms for different signal types have their own emphasis [5]: Traditional mainstream signals have their own emphasis on extracting current amplitude, potential difference, impedance amplitude, and phase characteristics [5]: Traditional mainstream signals extract current amplitude, potential difference, impedance amplitude, and phase characteristics respectively [8], and realize data interpretation through basic algorithms such as linear fitting, Nernster equation transformation, and impedance spectrum analysis, such as enzyme-catalyzed current-type signals to quantify lactate concentration through linear fitting [2]. The voltage-type signal completes the conversion of ion concentration through the Nernst equation [10]. For example, FET signals can detect trace proteins through differential analysis of leakage current changes [13], and photoelectrochemical signals can accurately quantify uric acid and glucose through peak photogenerated current extraction [12]. In the multi-signal collaboration scenario, a multi-feature fusion algorithm [63] is used to integrate various signal parameters to achieve simultaneous interpretation of multiple indicators and improve the comprehensiveness of health assessment [55].

4.1.4. Data transmission: the core guarantee of wearable practicality

The core requirement of data transmission is to adapt to the characteristics of "miniaturization, low power consumption, and real-time" of wearable devices [8], to stably transmit healthy data after signal analysis to terminal devices, while ensuring the stability, safety, and integrity of data transmission, and adapting to the detection frequency and signal type of supporting electrochemical sensors [5].

The transmission method is selected according to the application scenario of the device [64]: the data volume of consumer-grade devices is small, and low-power short-distance transmission technologies such as

Bluetooth BLE and NFC are preferred [65], such as sports bracelet sweat detection equipment that realizes real-time transmission of electrolyte and lactate data through Bluetooth BLE [56]. Clinical-grade equipment needs to transmit a large amount of trace indicators and continuous monitoring data [9], and uses Bluetooth BLE+LoRa dual-mode transmission [66] to take into account both close-range real-time viewing and long-distance clinical monitoring. In addition, encryption algorithms are added during transmission to ensure data security [67], and caching functions are designed to avoid data loss caused by network interruptions [64], adapting to the mobile usage scenarios of wearable devices.

4.2. Collaborative design of different signals: improve the comprehensiveness and accuracy of monitoring

Different types of electrochemical sensor signals do not exist independently, but achieve performance improvement through the synergistic mechanism of scenario adaptation + complementary advantages [55], break through the limitations of single signal detection, and improve the comprehensiveness, accuracy, and practicability of wearable sweat detection equipment [63].

Consumer-grade devices are mainly based on traditional mainstream signal synergy [56], and through the combination of voltage + current + impedance signals, comprehensive monitoring of electrolytes, metabolites, and skin health-related indicators is realized. Clinical-grade devices are mainly based on emerging signals and supplemented by traditional signals [32], and through the synergy of FET/photoelectrochemical/OECT and traditional signals, the simultaneous detection of conventional indicators and trace markers is completed, realizing the dual functions of health monitoring and early clinical early warning. Flexible wearable devices use OECT-type flexible signals as the core, combined with traditional voltage and current signals [4], taking into account skin adhesion and dynamic monitoring accuracy, such as the synergy of fiber-based OECT sensors and ion-selective electrodes, to achieve simultaneous detection of biomolecules and electrolytes in flexible wearable scenarios [18].

The key to the collaborative design of different signals is to solve the three major problems of "signal interference, power consumption control, and synchronous processing" [68]: First, to reduce signal crosstalk through circuit shielding design [54], such as using shielding wires to isolate each signal channel in a multi-channel amplification circuit to avoid mutual interference between voltage and current signals [5]. The second is to optimize the selection of op-amp circuits and ADCs [59], and use adjustable gain and multi-channel design to achieve synchronous amplification and conversion to reduce circuit complexity, such as the integrated design of multi-channel op-amp and ADC of wearable multi-parameter electrochemical detection systems to achieve low-power synchronous processing of multiple signals [5]. Third, a lightweight multi-feature fusion algorithm [63] is used to realize the simultaneous analysis of multi-signal data and improve the interpretation accuracy, such as a multimodal electrochemical analysis algorithm based on machine learning, which improves the accuracy of sweat marker detection by fusing current, voltage, and impedance signal characteristics [25].

Table 2. Design strategies of wearable sweat sensor systems for different application scenarios

Design dimensions	Consumer-grade equipment (e.g., sports bands)	Clinical/high-end equipment (e.g., disease monitoring patches)	Flexible wearable devices (e.g., e-skins)
Core objectives	Low cost, real-time monitoring, and user-friendly experience	High accuracy, trace detection, clinical reliability	High comfort, dynamic deformation adaptation, long-term wear
Adapt signal type (dominant)	Traditional signal combination: voltage type (electrolyte) + current type (metabolite).	Emerging signals are the main ones, supplemented by traditional ones: FET/photoelectrochemical type (trace material) + voltage/current type (conventional index).	OECT flexible signal at the core: with traditional voltage/current type signals
Circuit design priorities	1. Low power consumption 2. Miniaturization/low cost 3. Basic precision	1. Ultra-high accuracy and low noise 2. High-speed response 3. Special calibration capabilities	1. Flexible/stretchable integration 2. Signal stability under deformation 3. Low power consumption
Collaborative design is key	• Circuit: Multi-channel basic integration • Algorithm: basic filtering and linear fitting • Strategy: Functional combination to meet routine monitoring	• Circuit: precision shielding, independent grounding, high-precision ADC • Algorithm: multi-parameter joint calibration, machine learning fusion • Strategy: complementary advantages to achieve early warning	• Circuit: Flexible interconnection, waterproof packaging, distributed sensing • Algorithm: Dynamic temperature/deformation compensation • Strategy: Fit physiological curves to achieve natural monitoring
Typical detection indicators	Sweat Na^+/K^+, lactic acid, glucose, pH	Cortisol, cytokines, trace uric acid/glucose, specific proteins	Dynamic biomolecules, local electrolytes, pH (in exercise/sleep, etc.)

4.3. Summary

The signal processing process of wearable sweat sensors revolves around "adapting to wearable scenarios and fitting electrochemical sensor signal characteristics" [57], and realizes the complete conversion from original electrical signals to interpretable health data through four links: analog-to-digital conversion, signal calibration, signal analysis and feature extraction, and data transmission, of which signal calibration is the key to improving detection accuracy [17], and lightweight algorithms and low-power transmission are the core to ensure wearable practicality [56]. The collaborative design of different signals complements the advantages of traditional mainstream and emerging and non-mainstream signals [32] to improve the comprehensiveness and accuracy of monitoring, and adapt to the needs of different scenarios such as consumer, clinical, and flexible wearables [9]. Signal processing and collaborative design need to be deeply connected with the signal input and op-amp circuit design mentioned above, forming a complete circuit system link, which provides core support for the engineering design of wearable sweat sensors [5].

To achieve optimal performance for wearable sweat sensors across different application scenarios, the system design must be strategically tailored according to the core objectives. From an application-oriented perspective, Table 2 provides a systematic comparison of the differentiated design strategies for consumer-grade, clinical-grade, and flexible wearable devices across multiple dimensions, including design goals, compatible signal types, circuit priorities, and synergy strategies. This table clearly illustrates how co-design of hardware and algorithms can transform general sensing technologies into customized solutions that meet the demands of specific application.

5. Summary and outlook

5.1. Research summary

This paper reviews the circuit system design of wearable sweat sensors, systematically sorts out the signal input mechanism, op-amp circuit adaptation design, signal processing process, and different signal synergy mechanisms of various electrochemical sensors in wearable sweat detection equipment [69], and constructs a complete circuit system design framework based on relevant research examples, and draws the following core conclusions: First, the signal input module is the front-end core of the circuit system, and all signals are the exclusive output signals of the electrochemical sensor in the wearable sweat detection equipment. It is divided into two categories: traditional mainstream and emerging and non-mainstream [8]. The traditional mainstream signals (voltage, current, and impedance) are mature and cost-controllable, and the conventional index monitoring of consumer-grade equipment requires medium and low gain amplification [56]. Emerging and non-mainstream signals (FET, photoelectrochemical, OECT) focus on trace marker detection, signal characteristics are more complex, requiring high gain, low noise, and high-speed amplification [12]. The following section provides a streamlined outlook, presenting a coherent discussion without further subdivision.

5.2. Research prospects

In the future, circuit system design will evolve in the direction of high integration, low noise, flexibility and intelligence. In response to the needs of multi-parameter synchronous detection, the integration of reconfigurable op amp arrays and multi-channel ADCs into the on-chip system will be the key [70], realizing the unified acquisition of nA~nA current, mV voltage and wide-range impedance signals through adaptive gain adjustment, and breaking through the bottleneck of discrete component volume and power consumption. For weak signals such as FET type and photoelectrochemical type, ultra-low noise front-end circuits such as capacitive feedback transimpedance amplifier and chopper stabilization technology can reduce the equivalent input noise to the fA level [71], and combine active shielding and digital filtering to suppress sweat ion interference and motion artifacts. In flexible wearable scenarios, the optimization of serpentine wire layout and liquid metal interconnection technology need to take into account both stretchability and signal integrity to ensure stable gain and channel isolation in the deformation state [72]. In addition, the edge intelligent calibration module embedded with micro temperature sensors and lightweight machine learning accelerators can realize real-time temperature compensation and personalized baseline establishment, reduce wireless transmission power consumption [73], and improve the real-time monitoring and accuracy. These technological iterations will drive the transformation of wearable sweat sensors from laboratory prototypes to clinical, consumer-grade products.

6. Epilogue

The design of wearable sweat sensor circuit system needs to continuously deepen the adaptation mechanism of signal characteristics and circuit parameters, and provide core support for the engineering application of non-invasive health monitoring through the deep coupling of electronic systems and electrochemical sensors.

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