Flexible Wearable Smart Sensors for Continuous Glucose Monitoring: Original 24-Hour Data Acquisition and Comparative Analysis of Glucose Fluctuations Between Healthy Young Adults and Type 1 Diabetic Patients

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Abstract. This paper focuses on flexible wearable smart sensors for continuous glucose monitoring (CGM). It sorts out the development history of glucose collection and elaborates on the principle, structure, and performance evaluation indicators of minimally invasive CGM. The core original contribution of this work lies in the independent collection and in-depth analysis of original 24-hour CGM data from two distinct cohorts: healthy young adults (22 years old) and patients with Type 1 diabetes. Using MATLAB for data integration and visualization, we revealed novel characteristics of glucose fluctuation. (1) For healthy young individuals, the blood glucose curve is smooth. Postprandial peaks (8.3 mmol/L) occur at 1 hour after meals, and blood glucose returns to normal levels within 2 hours. Notably, there is no significant morning peak (6:00–10:00), which may be related to dietary and work-rest habits. (2) For patients with Type 1 diabetes, an obvious dawn phenomenon was observed between 00:00 and 02:00 (blood glucose increased from 10.2 mmol/L to 16.0 mmol/L). Postprandial blood glucose peaks are extremely high (>18 mmol/L) and decline slowly, taking 4–6 hours. Additionally, late-night near-hypoglycemic episodes were recorded, with the minimum blood glucose reaching 4.8 mmol/L. Furthermore, we compared our data with existing studies on middle-aged healthy populations (45 years old) and supplemented the glucose reference range for young adults. Existing glucose reference ranges are mainly based on middle-aged populations (e.g., average blood glucose: 5.3±0.5 mmol/L). In contrast, the average blood glucose of healthy young adults in this study is slightly lower (4.9±0.4 mmol/L), with no obvious morning peak. This difference provides a targeted baseline for glucose monitoring in young people, which is clinically significant for early identification of abnormal glucose metabolism and personalized health management in this cohort. Meanwhile, our results verify the consistency of CGM data accuracy. Additionally, we explore the theoretical basis, technical pathways, and existing challenges of non-invasive glucose monitoring. Our original CGM data can serve as a reliable reference for calibrating non-invasive sensors and establishing accurate glucose concentration conversion models, bridging the gap between minimally invasive monitoring results and the development of non-invasive technologies.

1. Introduction

The core step of blood glucose measurement lies in blood sample collection, and its basic operational procedure is relatively simple—ordinary people can master it with brief guidance. However, for specific groups in need of long-term and regular blood glucose monitoring (e.g., diabetic patients), traditional blood sampling methods have obvious limitations. Such groups usually need to obtain blood samples via fingertip pricking multiple times a day; repeated skin damage not only causes persistent discomfort like stinging and soreness but may also lead to thickened fingertip skin and reduced sensitivity due to long-term irritation, which impairs both the sampling experience and operational convenience. More notably, during long-term and frequent blood sampling, insufficient disinfection of sampling tools, improper care of puncture sites, or weak immunity of patients may all increase the risk of local skin infections and inflammatory reactions, imposing an additional health burden on patients [1].

Therefore, the development of minimally invasive or even non-invasive blood glucose detection technologies has become a key direction to meet clinical needs and improve patients' quality of life. Electrochemical sensors have been widely and maturely applied in the field of medical detection, thanks to their prominent technical advantages such as high sensitivity, fast response speed, and low detection limit. This type of sensor can achieve accurate blood glucose detection by specifically recognizing glucose molecules in bodily fluids like blood,

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sweat, and saliva. Among these, the solutions using sweat or saliva as detection media enable sample collection without skin puncture, completely avoiding the trauma and infection risks of traditional blood sampling. Meanwhile, they are easy to operate and can be used repeatedly, making them better suited for the long-term monitoring needs of groups like diabetic patients and providing a safer, more humanized technical approach for blood glucose detection.

Notably, existing CGM-related studies mostly focus on technical optimization of sensors or data analysis for middle-aged and elderly populations, while there is a lack of original, detailed 24-hour glucose fluctuation data for young healthy adults and comparative analysis with Type 1 diabetes patients. To fill this gap, this study independently designed a CGM monitoring scheme, collected original data from typical cohorts, and conducted in-depth characteristic analysis, aiming to provide empirical support for the refinement of CGM technology and personalized glucose management.

2. Development history of blood glucose collection

In the mid-19th century, the medical community discovered that the urine of diabetic patients contains sugar. Initially, judgments were made based on the physical characteristics of urine through experience, which was a crude method with no specificity. In 1908, Bayer launched Clinitest reagent strips, which produce color through a chemical reaction with Benedict's reagent, marking the first commercialized urine glucose detection tool. In 1945, Ames Company optimized the manufacturing process; however, urine glucose detection had inherent flaws such as delayed results and susceptibility to interference from renal function, so it could only serve as an auxiliary reference.

As the medical field's demand for accuracy increased, direct blood glucose detection gradually replaced urine glucose detection. The early o-toluidine method had poor specificity and used corrosive reagents. In the 1960s, the glucose oxidase (GOD) method emerged, which significantly improved specificity and was recommended by the World Health Organization (WHO) as one of the clinical standard methods. Nevertheless, venous blood glucose detection at this stage relied on professional equipment and personnel, making it impossible for patients to perform self-monitoring.

In 1965, Bayer's Dextrostix test strips, when paired with a reflectometer, became the prototype of blood glucose meters. In 1971, pocket-sized blood glucose meters were launched, followed by continuous upgrades: the second generation in the 1980s simplified operations; the third generation in 1987 enabled blood-free wiping detection; the fourth generation after 1986 adopted electrochemical methods, reducing the error rate to ±5% and shortening the reaction time to within 5 seconds; the fifth generation of blood glucose meters in 2001 required only 0.3 microliters of blood and supported multi-site blood sampling.

To address the limitations of single-point monitoring, continuous glucose monitoring (CGM) technology emerged: in 1999, Medtronic MiniMed achieved 72-hour continuous monitoring; in 2004, Dexcom STS realized real-time monitoring with an alarm function; in 2007, Dexcom extended the sensor lifespan of its products to 7 days; in 2017, Abbott launched a product that required no calibration and could be worn for 14 days. In 2013, Medtronic integrated CGM with insulin pumps. Since the 2020s, domestic enterprises (in China) have risen, launching high-precision CGM products and optimizing data through artificial intelligence (AI).

Spectroscopic technology has become a key research and development (R&D) direction. The non-invasive detection accuracy of Raman scattering spectroscopic technology, co-developed by Ruijin Hospital and Jinuan Technology, has reached 99.4%. However, Apple's silicon photonics chip still faces challenges in accuracy. Some domestic enterprises have already obtained approval for non-invasive products, but most of these are auxiliary tools that have not yet met medical-grade standards. Nevertheless, such non-invasive and unobtrusive technologies remain a key focus of future R&D efforts.

3. Continuous Glucose Monitoring (CGM)

Currently, major Continuous Glucose Monitoring (CGM) technologies can be categorized based on whether they cause trauma to the human body: minimally invasive CGM, fully implantable CGM, and non-invasive CGM. They can also be classified by different detection principles, such as spectroscopic measurement CGM, bioelectrical impedance CGM, and energy metabolism CGM. As of 2024, the global sales volume of minimally invasive CGM has reached 71.8 billion yuan, making it the most common and widely used device. During the same period, JING BAI et al. proposed an optimization method for existing CGM technologies, aiming to reduce the risks and pain experienced by patients when wearing the devices.

3.1 Basic principle of minimally invasive CGM

Continuous Glucose Monitoring (CGM) refers to a detection technology that continuously monitors the glucose concentration in subcutaneous interstitial fluid through a glucose sensor to indirectly reflect blood glucose levels. This technology can provide continuous, comprehensive, and reliable blood glucose information, enabling real-time observation of blood glucose fluctuations. It can detect occult hyperglycemia and hypoglycemia that are difficult to identify with traditional detection methods. Particularly in monitoring postprandial hyperglycemia and asymptomatic nocturnal hypoglycemia, CGM systems demonstrate significant detection advantages [3].

A continuous glucose monitor monitors interstitial fluid glucose concentration by implanting a glucose sensor under the skin. This glucose sensor, also referred

to as a "probe," is only approximately 5 mm in length and does not cause large wounds to the human body.

Its workflow is illustrated in Figure 1. Within the glucose sensor of a CGM system, glucose oxidase (or glucose dehydrogenase) serves as the core functional substance. It catalyzes the oxidation of glucose in interstitial fluid through a specific catalytic reaction, generating electroactive substances such as hydrogen peroxide. This chemical reaction is accompanied by an electron transfer process, which is captured by the working electrode built into the sensor, forming a current or potential difference signal that has a linear correlation with glucose concentration [4]. After this electrical signal is amplified and converted by the circuit module, a transmitter wirelessly transmits it to terminal devices such as smartphones or dedicated blood glucose meters. Finally, the information is presented in the form of real-time blood glucose values, fluctuation curves, etc. Meanwhile, the system supports high and low blood glucose alarm functions, enabling continuous and dynamic monitoring of blood glucose.

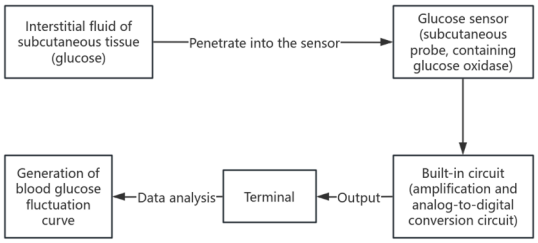

Figure 1 Workflow of CGM

3.2 Basic structure of the sensor

In minimally invasive Continuous Glucose Monitoring (CGM) systems, electrochemical sensors are utilized. Their structure typically comprises a substrate unit, a sensing unit, a decision-making unit, and an energy supply unit.

The substrate unit is a crucial component that endows electrochemical sensors with flexibility and mechanical stability. Substrate materials must not only possess processability essential for constructing sensors on them but also exhibit a series of mechanical properties required for wearable devices. These include flexibility, stretchability, and appropriate modulus [5]. Substrate unit materials for CGM need to adapt to different application scenarios like implantation or skin adhesion. They must also meet core requirements including biocompatibility, flexibility, and stability. Mainstream types include flexible polymers, rigid polymers, and specialized composite/elastomer materials. There are also various auxiliary substrate materials tailored to the needs of different components. Among these, flexible polymers are the primary choice for the electrode substrate of flexible CGM sensors. Polyimide (PI) boasts excellent high-temperature resistance and mechanical stability. It also exhibits strong compatibility with metal electrodes [6]. It is often used as the core substrate for implanted sensor electrodes. It has also been applied in manufacturing micro flexible glucose biosensors implantable in the lacrimal duct. These can meet the long-term monitoring needs of CGM (7–14 days). Polyurethane (PU) combines flexibility and biocompatibility [7]. It can serve both as an intermediate adapter layer for the sensor electrode substrate and as an outer protective film material. When paired with other layer structures, it enables the electrode to withstand a certain degree of tensile strain. In addition, polymer materials such as polyvinyl chloride (PVC) and polyethylene (PE) can be used as auxiliary substrates. They are applied in manufacturing sensing units and insulating electrode substrates respectively. Non-woven fabric acts as the substrate material for CGM fixing tapes. When combined with medical pressure-sensitive adhesives, it achieves stable fixation of the sensor. It causes no significant irritation to the skin during long-term contact.

The sensing unit is the core component of flexible electrochemical sensors. Its function is to extract and sample biological fluids. It also converts the interaction between target analytes and recognition materials into electrical signals based on a suitable electrochemical testing mechanism. The sensing unit mainly consists of an extraction and sampling module, electrodes, and wires. Within the substrate unit, the probe (i.e., micro-enzyme sensor) that directly contacts human interstitial fluid must meet miniaturization design requirements. Typically, its width is less than 1 mm and length ranges from 5 to 10 mm. Meanwhile, the biological activity of the enzyme electrode must be ensured to achieve reusable sensors. Furthermore, special attention must be paid to addressing the interference from non-glucose substances. These interfering substances tend to react with oxygen in the blood. Their reaction products adsorb onto the surface of the enzyme electrode. This hinders the diffusion of glucose to the active region of the electrode, thereby reducing the detection sensitivity of the sensor [8].

The core of the decision-making unit lies in converting electrical signals into digital signals. This is achieved through components such as signal transduction, amplification units, and analog-to-digital converters. To improve data accuracy, the data processing algorithms need to undergo multiple tests and adjustments.

The main function of the energy supply unit is to provide energy for the aforementioned units to ensure their normal operation. In sensors, this unit primarily consists of batteries of various types. In recent years, in response to the demands of flexible smart wearables, researchers have designed and developed a variety of novel wearable energy supply units. Examples include new approaches for energy harvesting, conversion, and application. These are based on the thermoelectric effect, piezoelectric effect, triboelectric effect, giant magnetoresistance effect, photoelectric effect, and biofuel cells [9-10].

3.3 Performance evaluation

3.3.1 Data accuracy

The data accuracy of Continuous Glucose Monitoring (CGM) systems is often evaluated using the Mean

Absolute Relative Difference (MARD) as the core indicator. A smaller MARD value indicates higher accuracy. Current mainstream CGM products have achieved relatively low MARD values. For example, the MARD value of Dexcom G7 is as low as 8%. That of Sinocare Dynamic CGM is 8.83%, and that of Microtech Dynamic G7 is 9.08%. In contrast, the MARD values of most ordinary CGM systems generally range from 9% to 12%. For instance, in a study conducted by the Second Affiliated Hospital of Zhengzhou University, the average MARD of the FreeStyle Libre Flash Glucose Monitoring System reached 12.7%. Additionally, in inpatient settings, when Dexcom G6 CGM is used for hospitalized patients, its MARD is 15.1% based on bedside blood glucose testing. It is 11.4% based on laboratory blood glucose testing [11].

3.3.2 Heat resistance

The heat resistance of Continuous Glucose Monitoring (CGM) depends on the heat resistance of glucose oxidase. Beijing Weinuoen Biotechnology Co., Ltd. has conducted relevant experiments. The results show that glucose oxidase has excellent heat resistance, and the enzyme concentration has little impact on it [12].

3.3.3 Hysteresis

Continuous Glucose Monitoring (CGM) measurements require real-time performance. Only when the measured data reflects the blood glucose concentration around the current moment can timely actions such as treatment be taken. The hysteresis of CGM refers to the time difference between the blood glucose value and the glucose value in subcutaneous interstitial fluid. It has no fixed value. Its duration varies significantly under the influence of factors including blood glucose change rate, device technology, and sensor wear. Under normal circumstances, the general lag time between CGM and plasma glucose concentration changes is 4 – 10 minutes. This time can extend to 12 – 24 minutes when blood glucose fluctuates rapidly. The hysteresis of some ordinary sensors can even reach 40 minutes.

3.3.4 Practical data analysis

The Sinocare Dynamic CGM system was used to continuously monitor the 24-hour blood glucose data of healthy adults (22 years old) and patients with Type 1 diabetes. MATLAB was employed to integrate the data, and the following images (Figure 2, Figure 3) were obtained.

As shown in Figure 3, the blood glucose curve of healthy individuals is smooth. Slight fluctuations occur after meals, and blood glucose remains stable at night (ranging from 3.8 mmol/L [red line] to 6.1 mmol/L [yellow line]). Blood glucose reaches a peak of 8.3 mmol/L [green line] 1 hour after meals and returns to the normal range within 2 hours. No hypoglycemia or persistent hyperglycemia was observed throughout the day.

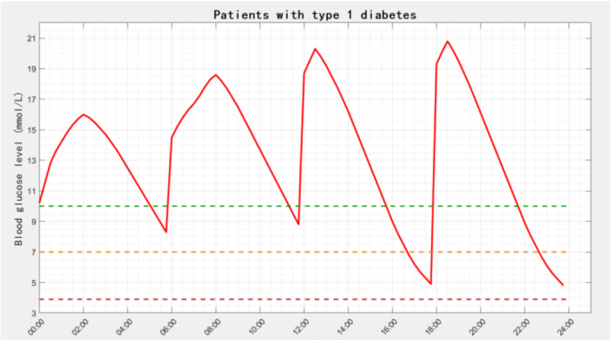

Figure 2 Blood Glucose of Patients with Type 1 Diabetes

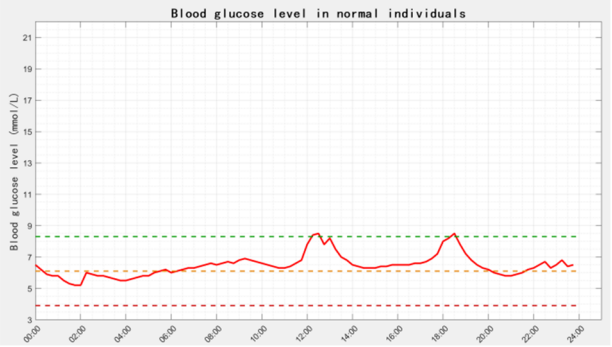

Figure 3 Blood Glucose of Normal Individuals

Figure 2 shows the 24-hour blood glucose changes in patients with Type 1 diabetes. The blood glucose curve rises and falls sharply. An obvious dawn phenomenon occurs at night: blood glucose increases from 10.2 mmol/L to 16.0 mmol/L between 00:00 and 02:00. The postprandial blood glucose peak is extremely high (>18 mmol/L), and the decline is slow, taking 4–6 hours. From 23:00 to 00:00 (late night), due to residual effects of insulin, the blood glucose is close to the hypoglycemia threshold. The minimum blood glucose level is 4.8 mmol/L.

According to the measurement results by professionals [13], the average blood glucose level of normal people (aged 45 years old) is (5.3±0.5) mmol/L. The blood glucose level is low before three meals and rises rapidly after meals until reaching the peak. The peak of postprandial blood glucose occurs at 1–2 hours after three meals. At 3 hours after three meals, the blood glucose level shows a downward trend. The maximum blood glucose value is (7.8±1.4) mmol/L, and the minimum value is (3.9±0.7) mmol/L. Compared with other measured data, the changes in blood glucose level of the individual in this paper are roughly the same, and the values are also similar. However, considering that the test subject is 22 years old, the average blood glucose level, peak value, etc., should be slightly lower. In addition, in the data of this paper, there is no obvious fluctuation near the morning period (6:00–10:00). This may be due to not having breakfast or consuming very little (the curve shows a slight rise). Based on the above data, the test subject may have problems such as irregular work and rest (staying up late and getting up late).

The CGM-measured data is consistent with the actual physiological characteristics of the subjects. This verifies

the high accuracy and fast response of the CGM system. More importantly, the original data supplements the 24-hour glucose fluctuation baseline for young healthy adults. It reveals the correlation between lifestyle (such as work-rest and diet) and glucose changes[14]. It also provides a reference for personalized CGM data interpretation and health management.

4. Non-invasive blood glucose monitoring

A key opportunity for the technological transition from minimally invasive to non-invasive blood glucose monitoring lies in the discovery that glucose molecules also exist in human peripheral body fluids (such as sweat, saliva, tears, etc.). Among these, sweat has become a core research target for non-invasive monitoring. It has the characteristics of easy collection and real-time accessibility. Although a unified conclusion has not yet been reached regarding the specific source of glucose in sweat, existing hypotheses provide possible explanations. It may be produced through the passive diffusion of skin epidermal cells, the selective permeation of blood glucose by sweat gland ducts, or the local synthesis of skin metabolism. Researchers have confirmed that there is a significant correlation between sweat glucose content and blood glucose concentration. By establishing an accurate mathematical model, the specific content of blood glucose and its dynamic change patterns can be inferred from the sweat glucose detection data. This lays a core theoretical foundation for the realization of non-invasive blood glucose monitoring.

However, the practical implementation of non-invasive blood glucose monitoring faces severe technical challenges. The glucose concentration in sweat is extremely low, only about 1/100 of that in blood [15]. The detection of such trace levels places extremely high demands on the sensitivity of the sensing system. Especially in hypoglycemic scenarios, the blood glucose concentration itself is already at a low level. The corresponding sweat glucose content is even more negligible, which is easily masked by the background noise of the sensor. Furthermore, residual glucose contaminants on the skin surface (such as glucose in skincare products and food residues) interfere with the target analytes in sweat. This further increases the difficulty of accurate detection. Therefore, the development of a sensing system with ultra-sensitive response and high-specificity recognition capabilities has become the key to breaking through the technical bottleneck of sweat-based non-invasive blood glucose monitoring. Current mainstream research directions include electrochemical sensors modified with nanomaterials and aptamer-functionalized optical sensing chips. The aim is to achieve accurate capture and quantitative analysis of trace glucose in sweat.

In addition to sweat detection, the mainstream approach, non-invasive blood glucose monitoring also includes a variety of optical detection technologies. Among these, polarimetry and near-infrared (NIR) spectroscopy are two relatively mature research methods. As a typical optically active substance, glucose has chiral characteristics in its molecular structure. When polarized light passes through a biological medium containing glucose, the polarization plane rotates by a specific angle. This rotation angle is positively correlated with the glucose concentration. Polarimetry utilizes this property to achieve non-invasive determination of blood glucose content by detecting changes in the rotation angle. Near-infrared spectroscopy, on the other hand, is based on the characteristic absorption peaks of glucose molecules in the near-infrared band. It captures the characteristic spectral signals of glucose in blood using spectral analysis technology. Then, it eliminates the background interference from tissues such as skin and fat through chemometric methods. Finally, it completes the quantitative analysis of blood glucose concentration [16]. Such optical detection methods have become an important research branch in the field of non-invasive blood glucose monitoring. They do not require contact with body fluids and can achieve fully non-invasive real-time monitoring. However, they are still limited by issues such as detection accuracy and anti-interference ability. They have not yet been applied on a large scale at the clinical level.

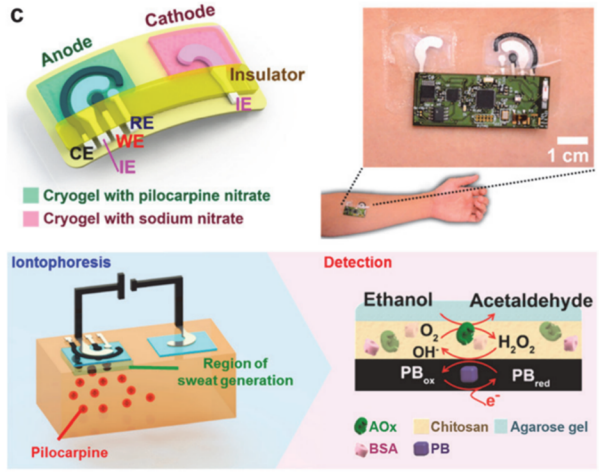

Figure 4 Used for Accurate Measurement of Electrophysiological Signals in Body Surface Sweat [5]

The Kim research team has developed a series of wearable electrochemical sensors based on flexible materials. These are used for the accurate measurement of electrophysiological signals in sweat on the body surface [17], and the sensor structure is shown in Figure 4. In their initial work, the team synthesized high-quality, large-area graphene through a chemical vapor deposition (CVD) process. This constitutes a very useful new type of transparent and stretchable electronic material. It has been reported that wrist patches developed using graphene technology can use sensors to detect the glucose content and pH value in sweat. Then, they combine with body temperature to calculate the glucose level in the patient's blood [18].

5. Conclusion and outlook

In the development history of blood glucose monitoring technology, blood sample-based detection technology has

evolved into a mature technical system after decades of iteration and optimization. Among these, minimally invasive Continuous Glucose Monitoring (CGM) systems have become the mainstream choice in current clinical applications and consumer markets due to their unique technical advantages. From a market performance perspective, minimally invasive CGM products represented by Sinocare Dynamic and Dexcom G7 have achieved large-scale commercialization. Global annual sales show a year-on-year growth trend—particularly their penetration rate among diabetic patients continues to rise. The core breakthrough of this type of technology lies in the organic integration of "continuous monitoring" and "minimally invasive sampling." By implanting a micro-sensing probe into subcutaneous tissue via a minimally invasive method, it can automatically collect interstitial fluid glucose concentration data every 5–15 minutes. After algorithm calibration, the data is converted into blood glucose values and transmitted to terminal devices in real time. This not only avoids the pain of frequent needle pricks associated with traditional fingerstick blood glucose meters but also fully captures key information such as the circadian rhythm of blood glucose fluctuations, postprandial peaks, and nocturnal hypoglycemia. It provides continuous and three-dimensional data support for the precise diagnosis and treatment of diabetes.

Although minimally invasive CGM systems have demonstrated significant clinical value and market vitality, their current technical promotion still faces many practical bottlenecks. Among these, cost control is particularly prominent. From the perspective of technical cost composition, the high pricing of minimally invasive CGM systems stems from technical barriers across multiple links. First, the core sensing chip requires precious metal materials with excellent biocompatibility (e.g., gold, platinum electrodes) and specific enzyme preparations (e.g., glucose oxidase). The material preparation and modification processes are complex. Second, the implantable probe must maintain stable operation for 14 days. This imposes strict standards on the sealing and anti-interference performance of the packaging process. Third, the development of data calibration algorithms requires the accumulation of large amounts of clinical data. This results in high upfront R&D investment. These factors lead to a common price tag of around 1,000 yuan (RMB) for a single set of mainstream minimally invasive CGM products. Probes need replacement every 14 days—pushing the annual usage cost to 20,000–30,000 yuan (RMB). For groups with weak economic capacity, such as low-income populations and elderly diabetic patients, the economic burden of long-term use is heavy. This makes it difficult for this technology to achieve universal access and results in an imbalance between "technical accessibility and economic affordability."

Compared with minimally invasive technology, non-invasive blood glucose monitoring technology—using sweat or saliva as detection media—has become a research hotspot in the field of blood glucose monitoring in recent years. It completely avoids the pain and infection risks of invasive procedures. It is regarded as a crucial development direction for the next generation of blood glucose monitoring technology. By collecting glucose molecules from human peripheral body fluids for in vitro detection, this technology meets patients' core needs for "painless and convenient" monitoring. It holds broad application prospects in scenarios such as health management and chronic disease prevention. However, the industrialization process of current non-invasive monitoring technology is still limited by multiple technical bottlenecks. Complex matrix interference and insufficient detection accuracy are the two core challenges. From the perspective of matrix characteristics, the composition of peripheral body fluids such as sweat and saliva is significantly affected by the external environment. Residual skincare products on the skin surface and sugar deposits from food can directly lead to false positive results. Sweat secretion volume is regulated by factors such as temperature, humidity, and exercise status. This easily causes fluctuations in sensor response signals. Proteins and enzymes in saliva may also undergo non-specific binding with the sensing interface, interfering with the specific recognition of glucose. These complex interference factors place extremely high demands on the selectivity and anti-contamination capabilities of sensors. There is an urgent need to optimize the sensing interface performance through technical means such as nanomaterial modification and aptamer functionalization.

In terms of detection accuracy, non-invasive blood glucose monitoring technology still has much room for improvement. On one hand, there is a significant concentration difference between glucose in peripheral body fluids and blood (e.g., sweat glucose concentration is only 1/100–1/50 of blood glucose). The correlation between the two is affected by factors such as individual metabolic differences and physiological status. This necessitates the establishment of more accurate mathematical models to achieve concentration conversion. On the other hand, the core performance indicators of existing non-invasive detection devices have not yet met clinical-grade requirements. The Mean Absolute Relative Difference (MARD) of most prototypes is still higher than the clinically acceptable threshold of 15%. Detection stability is insufficient especially under extreme blood glucose conditions such as hypoglycemia and hyperglycemia. Additionally, the clinical verification system for non-invasive monitoring technology is not yet complete. It lacks multi-center and large-sample clinical data to support the reliability of its detection results. This also restricts the translation of this technology into clinical applications.

Notably, the original 24-hour glucose fluctuation data from this study—covering age-specific characteristics of healthy young adults and pathological patterns of Type 1 diabetes patients—can provide critical support for addressing the bottlenecks of non-invasive technology. For example, the accurate blood glucose concentration data from minimally invasive CGM can serve as a gold standard for calibrating non-invasive sensors. It helps optimize the mathematical models that convert peripheral body fluid glucose (e.g., sweat glucose) to blood glucose, especially for age-specific populations. This strengthens the relevance between current minimally invasive

monitoring results and the future development of non-invasive technologies.

In the future, on the basis of this study's original data, we will further expand the sample size, include more age groups and disease types, and establish a more comprehensive glucose fluctuation database. This will support the refinement of CGM algorithms and the development of non-invasive monitoring technologies. Meanwhile, addressing the technical bottlenecks of non-invasive monitoring (e.g., improving sensor sensitivity and anti-interference ability) and reducing the cost of minimally invasive CGM will be key research directions to promote the universal application of glucose monitoring technology.

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