

Wearable Biosensors on Sutures and Threads



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1 Introduction

Efforts in academia and industry have increased to explore and create smart, flexible systems for wearable applications. Several laboratory prototypes, testbeds, and industrial products have already been manufactured [1]. These adaptable/wearable devices aim to monitor the health of, and consequently improve the quality of life for those with impaired physical and cognitive abilities or suffering from various diseases. Examples range from wearable heart rate monitors for those with cardiac conditions, to fall detectors for the elderly. As a result, these systems promote independent living for the elderly, postoperative rehabilitation for patients to speed recovery, and the evaluation or enhancement of individual sporting or technical abilities [1–4]. The ideal smart flexible system could include a variety of wearable components, such as sensors, actuators, power sources, wireless communication networks, processing units, multimedia devices, user interfaces, software, and algorithms for data collection, processing, and decision support [5–7]. These systems may keep an eye on vital signals like heart rate [6], breathing rate, blood, or wound oxygenation [8], arterial blood pressure [9], body mobility and strain sensing [10], and brain activity [11]. A wireless sensor network can transmit the measurements to a central connection node, such as a personal digital assistant (PDA), or a hospital. Advances in micro-fabrication have enabled several biosensors for POC testing and wearable real-time

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monitoring [12–15]. Despite the significant advancements in wearable sensors for physiological monitoring, non-invasive chemical sensing for assessing analytes and biomarkers in bodily fluids has received very little attention.

Compared to paper as a substrate for sensing, textile is a more resilient material system that is more durable, widely accessible, and reasonably priced. It also allows capillary-based sample conveyance similar to capillary transport in paper. Additionally, textile-based electronics and sensors allow for simple integration with wearable clothing and materials, which may be utilized to create wearable sensor systems [5, 13]. The incorporation of sensors into textiles to track physiological variables like electrolytes and small molecules [16], heart rate [17–19], temperature [20], and breathing [21] has been the focus of recent research in wearable sensing. In the last decade, several research groups reported the development of electrochemical sensors on threads or smart fabrics for health monitoring. For example, Wang's group has developed electrochemical sensors on fabrics for monitoring the environment [22], wounds [23], and health [24]. A wearable electrochemical sensor for sweat analysis based on textiles has also been shown by our group (Sonkusale's group) [16]. We have reported a thread-based multiplexed sensing patch for continuous and simultaneous on-skin sweat monitoring. The sweat patch is capable and measures crucial physiological indicators such as metabolites (lactate), electrolytes (sodium and ammonium ions), and acidity (pH) in sweat. In this work, sensor fabrication was realized on flexible threads coated with conductive inks as sensing electrodes. Diamond's group has also demonstrated wearable, textile-based electrochemical sensors for sweat analysis [25]. The use of screen-printed sensors, which are frequently structurally brittle and difficult to connect with textile-based electrical components, is what allows these devices to undertake delicate analytical measurements. Recently, it was claimed that conductive thread [7] or silk yarn [26] was woven into the fabric to fabricate electrochemical sensors that were based on textiles. This approach offers improved robustness compared with screen-printed sensors but is limited to woven electrode geometries and substrates.

This chapter explores the design, fabrication, and validation of wearable electrochemical sensors on sutures/ threads for health monitoring. This chapter aims to review the necessary background and knowledge for designing and fabricating flexible sensors on sutures/threads with several key applications in sweat sensing, smart bandage, and smart sutures while emphasizing the key qualities that make them the best possible candidates for flexible bioelectronics. Last but not least, we assess the electrochemical readout circuitry's performance, which displays the sensor's capacity to provide reliable measurements under such circumstances.

2 Threads and Sutures

2.1 *Types of Threads and Sutures*

Fibers may be broadly categorized into two groups: natural, which come from biological sources including plants and animals, and synthetic, which are created by chemical processes. Both natural and synthetic fibers' molecular structures may be used to predict how their surfaces will respond to certain treatments, like dyeing, or to evaluate other factors, such as mechanical tunability and biofriendliness.

Natural fibers are frequently derived from plants and animals. Cotton, linen (flax), and jute are all cellulose-based fibers, with 94%, 80%, and 74% cellulose content, respectively [27]. Bioproteins make up animal fibers such as silk and wool. Silk is made up of the fibroin protein, which is made up of recurring sequences of glycine, serine, and alanine amino acids [28]. Wool is also formed of keratin-type protein, but its outer surface (the cuticle) defines many of its features. The epicuticle, the cuticle's topmost layer, includes fatty acids covalently bound to the wool fiber via a thioester bond, providing a hydrophobic barrier [29]. Its complex surface structure makes it less dyeable and wettable than other natural fibers [30]. Natural fibers are an excellent choice for biological sensors that contact with tissue because of their high specific strength, biodegradability, and inexpensive production costs [7, 31]. Another benefit of using natural fiber sources is their capacity to be recycled, which is becoming more and more crucial in light of the present environmental issues. Given the current state of climate change and the demand for sustainable resources, the benefit of renewable natural fiber sources is growing over time.

Polymers used to create conventional synthetic fibers often come from petroleum [32]. These include acrylic [33], polyurethane, polyester fibers, and nylon [34]. Most frequently, they are made by a "melt-spinning" procedure in which fibers are extracted from strands that have been heated to their melting point [31]. Natural fibers tend to be less durable than synthetic fibers, which also quickly take the dye. Due to their strong elastic properties, synthetic fibers like polyurethane become excellent candidates for uses like strain sensing [7, 35].

2.2 *Processing Methods*

Traditional twisting techniques have been used to create both natural and synthetic fibers. Natural fibers like cotton or wool are drawn out while being twisted to form threads and yarns, while traditional synthetic fibers like nylon are drawn out during a thermal process to form continuous threads. Extruding precursor solutions into fibers is a technique used in more recent methods including electrospinning, wet-spinning, and direct writing [36, 37]. Due to their fine control over manufactured fiber form, sub-micrometer to nanoscale regime widths, and compatibility with several inks,

these methods are being employed more often in biomedical, tissue engineering, and better versions of conventional sensing applications.

Electrospinning is a method of drawing polymer fibers from an injecting needle onto a plate while exposed to an electric field. Electrospinning produces fibers with diameters ranging from 100 nm to a few micrometers, resulting in high fiber density in a simple procedure [38, 39]. Liu et al. used electrospinning to construct a stretchy strain sensor based on a flexible polymer nanowire, whereas earlier devices based on inorganic materials revealed lesser strain sensitivity [40]. Researchers have also shown that fibers with increased modulus and tensile strength may be electrospun into fiber composites using carbon nanotubes (CNTs).

In wet spinning, a polymer precursor solution is pumped into an insoluble coagulation bath, resulting in the continuous polymerization of long fiber [38]. Wetspinning generates fibers with lengths ranging from 30 to 600 m and may make tissue-friendly fibers by using appropriate inks such as collagen and polycaprolactone. Wetspun fibers' mechanical strength may be improved by orders of magnitude by adding CNTs or graphene oxide, just as electrospun fibers. Because the fibers are greater in size here than in electrospinning, the scaffolds produced by this process have higher porosity and larger pore sizes [38].

Insects produce silk fibers through a process known as biospinning. Silk is biodegradable and has a high tensile strength ranging from 460–972 MPa [38]. These fibers consist of a fibroin core surrounded by sericin, a glue protein. After chemical processing, silk remains non-cytotoxic and non-inflammatory. The diameter of silk fibers can vary from 25 to 70 m, depending on whether they are extracted from the insect or from the cocoon. Silk fiber finds extensive use in biomedical applications including implants, tissue scaffolds, and drug delivery systems [41–43]. These applications are particularly relevant for load-bearing tissues, where a mechanical strength of around 150 MPa is required. For instance, Karageorgiou and Kaplan's work on 3D scaffolds contributed to osteogenesis [41].

2.3 Surface Modification Techniques

Surface modification techniques play a crucial role in improving the performance of threads for sensing applications. Biorecognition components (enzymes, aptamers, antibodies, etc.) are attached to the immobilization surface using various techniques. When selecting an immobilization approach, several aspects such as the immobilization surface, sample matrix, protein characteristics, buffer components, and assay performance metrics are considered. In an ideal world, reaction partners should be able to reach active regions for antibody binding or enzymatic conversion (i.e., the binding site should face away from the immobilization surface to reduce steric hindrance and should not be sterically hindered by nearby immobilized proteins). For a high-performance, repeatable experiment, receptor conformation must remain

unaltered upon immobilization to maintain receptor binding properties, permit diffusion of substrates and products, and allow excellent electron transfer. These techniques can enhance the sensitivity, selectivity, and stability of the threads, making them more suitable for specific sensing tasks. Here are some commonly used surface modification techniques for modifying threads for sensing applications.

a. Physical Adsorption

The physisorption method is the simplest method for immobilizing a bio-recognition component on a surface (i.e., physical adsorption). In physical adsorption, the baroreceptors are immobilized by van der Waals forces, hydrophobic contacts, or hydrogen bonds, whereas enzymes are joined by salt linkages in electrostatic interaction (Fig. 1a, b). The process of non-covalent immobilization is caused by the nature of the forces involved, and it can be stopped by altering the factors that determine how strong the contact is (e.g., pH, salt concentration, solvent polarity, temperature, etc. [44]). The functioning of the biomolecule is often maintained during immobilization by adsorption, which is a simple procedure. Weak attachment and unpredictable orientation, which result in desorption and poor reproducibility, are the adsorption mechanism's drawbacks [45].

b. Covalent Immobilization

One of the most prevalent techniques for immobilizing receptors involves the utilization of covalent bonding to anchor them in place (Fig. 1c–e). In the grafting process, stable bonds are formed between the biomolecule and the substrate, effectively preventing any leakage of the bioreceptor into the surrounding solution. To ensure the optimal preservation of protein activity, careful attention must be paid to the covalent attachment chemistry to the substrate, to safeguard the biomolecule's active

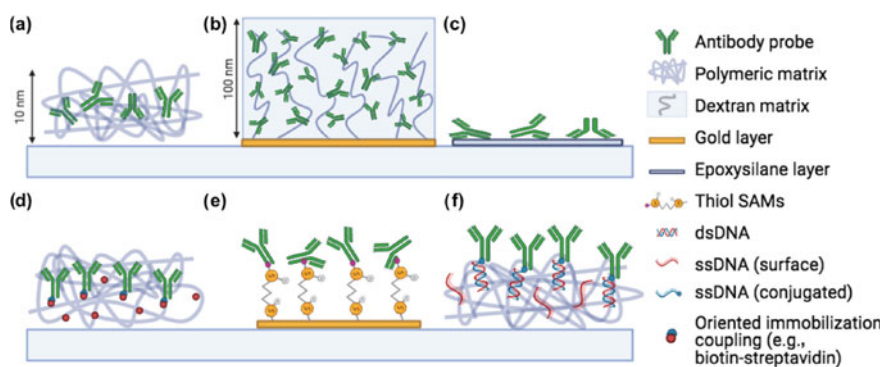


Fig. 1 Common immobilization methods for microarray applications. **a** Copoly(DMA-NAS-MAPS), **b** Carboxymethyl dextran, **c** Epoxysilane, **d** Oriented immobilization (biotin-streptavidin, click chemistry), **e** Thiol-gold coupling, **f** DNA-directed immobilization. Figure reproduced with permission from [51]

region. A critical consideration is to prevent any interference or hindrance of vital amino acid residues that govern enzyme activity or antibody recognition. However, achieving this delicate balance could prove challenging under certain circumstances [46]. Tailored to the functional groups inherent to the target molecule, several distinct reaction pathways have been developed. Despite the intricate structural complexity of biomolecules, only a limited selection of functional groups can be effectively leveraged as discriminating targets for efficient immobilization. For instance, thiol groups (-SH), integral components of cysteine residues' side chains, have demonstrated noteworthy efficacy. In the context of sugars, the Fc region represents a primary site for glycosylation. While the oxidation of diol moieties to aldehydes entails additional steps, this approach often yields antibody conjugates with heightened activity due to the specific positioning of the carbohydrate moiety. Alternatively, the reductive amination process provides a direct means to react sugars with primary amine or hydrazine-based groups. Illustrative instances of covalent immobilization are depicted in Fig. 1a–e, showcasing the versatility and application of this methodology.

Utilizing a blocking solution, such as bovine serum albumin (BSA) in PBS, can deactivate functional groups that are not bound to the receptors [47]. The inherent randomness in receptor orientation, a common outcome of covalent immobilization, ensures the irreversible attachment of receptors to surfaces. However, this randomness might influence their biological activity and their capacity to effectively bind analytes from the solution. Covalent immobilization can also be time-consuming, particularly when dealing with low-concentration receptor solutions during the attachment phase, often due to their cost considerations. In situations where protein immobilization is necessary, the combination of non-covalent and covalent immobilization can be achieved by introducing a spacer between the surface and the receptors [48]. An alternative approach termed “random immobilization” involves placing receptors on a surface without specific consideration for their orientation. In contrast, “direct immobilization” refers to attaching receptors with controlled orientation. This method frequently positions the receptor's active site away from the surface, making it more accessible to species in solution due to reduced steric hindrance. This orientation yields surfaces with a higher receptor density, which can significantly impact binding capacity and overall assay sensitivity.

c. Bioaffinity

Bio-affinity interactions, also referred to as biospecific adsorption (Fig. 1d), capitalize on specific binding processes observed in nature. These interactions offer distinct advantages over physisorption. For instance, bio-affinity association enables directed, highly specific, and significantly stronger protein immobilization. This results in reduced protein leakage due to the enhanced accessibility of fixed receptor molecules to their binding partners, unlike random orientation strategies. Furthermore, bio-affinity immobilization can be reversed through chemical means, pH adjustment, or thermal treatment. The concept of natural affinity between molecules, such as lectin-sugar, antibody-antigen, and biotin-avidin pairs, has been employed for biomolecule

immobilization. However, often this requires covalent attachment of costly affinity ligands like antibodies or lectins to the support. Notably, interactions like streptavidin–biotin and the utilization of Protein A or G for antibody immobilization stand out as the most widely adopted approaches [44].

d. Sulphur-Atom Linkage or Disulfide Bond

In the case of surfaces like gold threads or textiles doped with gold nanoparticles, although a sturdy covalent connection is established between the support and the bioreceptor, it can be cleaved through a gentle reaction with a suitable chemical, such as dithiothreitol (DTT) (Fig. 1e). Furthermore, when utilizing a thiol-reactive adsorbent with heightened specificity, the activity yield of procedures involving the formation of disulfide bonds is often significantly elevated. This enhanced yield can be attributed to the precisely controllable reactivity of thiol groups, which can be finely regulated by adjusting the pH [49].

e. Metal-Link Binding/Chelation

The metal salt or hydroxide, primarily titanium and zirconium salts, undergo precipitation and coordination with nucleophilic groups on the surface, which may include carriers based on cellulose, chitin, alginate acid, and silica [50]. Some of these metal binding sites remain unoccupied due to steric hindrances, creating opportunities for interactions with biomolecular groups. While the procedure itself is straightforward, challenges arise due to the non-uniform distribution of adsorption sites, resulting in limited repeatability, alongside notable leakage of metal ions.

f. 2D Versus 3D Surface Immobilization

The surface geometry, which can be either planar (2D) or three-dimensional (3D), has a considerable impact on immobilization techniques [52]. 2D surface immobilization on threads involves the attachment of bioreceptors onto a flat, planar substrate. This approach, although relatively straightforward, comes with limitations. Microfluidic channels within the thread's structure can be constructed from various substrates, including those coated with functional materials like gold, platinum, or graphene. This method is efficient for certain applications and allows for controlled and uniform immobilization of receptors. However, the available surface area for receptor attachment is inherently constrained due to the planar nature of the substrate. Consequently, the density and quantity of immobilized receptors are limited, potentially affecting the sensitivity and performance of the biosensor. On the other hand, 3D surface immobilization introduces a three-dimensional architecture to the thread's surface. This configuration provides several distinct advantages. Threads with embedded microstructures, such as microposts or porous membranes, offer an increased surface area for receptor attachment [53]. This enables higher receptor density, thereby enhancing the biosensor's sensitivity and allowing for the detection of lower analyte concentrations [52]. Additionally, the reduced diffusion length

between reactants on a 3D surface can expedite the binding kinetics, leading to faster response times [47]. Such enhancements are particularly valuable in rapid and real-time detection scenarios. The selection between 2 and 3D surface immobilization on thread-based biosensors depends on the specific application requirements. If uniform and controlled immobilization is sufficient and high sensitivity is not a critical factor, 2D surface immobilization may be suitable. On the other hand, if enhanced sensitivity, faster response times, and the ability to detect lower analyte concentrations are essential, 3D surface immobilization provides a more advantageous solution.

In conclusion, the decision between 2 and 3D surface immobilization on thread-based biosensors involves a trade-off between simplicity and enhanced performance. While 2D immobilization is straightforward, 3D immobilization offers superior sensitivity and binding kinetics, making it a valuable choice for applications demanding rapid and highly sensitive detection.

3 Flexible-Sensing Platform

This section provides an overview of recent advancements in utilizing thread-based fabrics as electrochemical sensors, with a specific emphasis on biochemical sensing [8, 35]. Electrochemical sensors are devices that combine a recognition component with an electrochemical transducer to generate electronic signals. In electrochemical sensing, the signal readout is directly linked to analyte binding, enabling rapid detection. Due to these attributes, electrochemical sensors find extensive applications in daily life, ranging from household carbon monoxide detectors to research laboratory pH meters. In the realm of biochemical sensing, particularly in point-of-care settings, the demand for smaller and more compact sensors is imperative to minimize invasiveness. Conversely, thread-based fabrics have garnered significant attention for creating wearable electrochemical sensors, owing to their distinctive physical and chemical attributes, including flexibility, high porosity, and mechanical stability [5, 7, 16, 20, 54–56]. The diameter of a single fiber, constituting the smallest unit of fabric, is approximately a few micrometers, promising ultra-small sensor design. Simultaneously, individual threads can be incorporated into clothing using industrial textile techniques like weaving, knitting, and embroidering, offering unparalleled mass production capabilities.

These sensors can be classified into two types based on fabrication approaches:

- **Using Fabrics as Substrates for Screen-Printed Electrodes:** This approach involves utilizing fabrics as substrates for screen-printed electrodes. This method leverages the inherent properties of the fabric as a base and integrates screen-printed electrodes for sensing purposes.
- **Using Individual Functionalized Threads as Sensing Electrodes:** Alternatively, individual functionalized threads are employed as sensing electrodes and assembled into wearable sensors. This approach enables precise customization and assembly of sensors, tailoring them to specific applications.

In summary, the integration of thread-based fabrics into electrochemical sensors presents exciting possibilities, particularly in the realm of biochemical sensing. The inherent characteristics of threads and fabrics, combined with innovative fabrication approaches, are paving the way for smaller, more versatile, and wearable sensors with applications ranging from healthcare to environmental monitoring.

3.1 Principle of Sensing

Electrochemical detection hinges on the conversion of biochemical signals into electronic ones. In the realm of electrochemical sensing, a recognition layer selectively captures the analyte, while a transducer layer translates this interaction into an electronic readout. Electrochemical methods encompass potentiometric, coulometric, and voltammetric techniques. In potentiometric measurements, the potential difference between the working and counter electrodes is gauged without the passage of current. In coulometric measurements, the cumulative current during an electrolysis process is quantified. Voltammetric methods involve measuring the current elicited by either a constant or varying voltage application. Thread-based fabrics are typically rendered electrically conductive through coatings of materials such as carbon, carbon nanotubes, or graphene oxide. Within the context of thread-based electrochemical sensors, ionic sensors designed to detect pH, sodium, potassium, calcium, ammonium, chloride, and lithium primarily rely on potentiometric approaches. Meanwhile, enzymatic sensors targeting analytes like glucose, lactate, uric acid, and ascorbic acid predominantly employ voltammetric methods. Additionally, gas sensors that measure resistance changes have garnered significant attention and will be elaborated upon. In conclusion, the integration of thread-based fabrics into electrochemical sensing holds significant promise. The diverse array of methods available, coupled with the materials used for sensor enhancement, is driving advancements in various sensing applications, ranging from ions to enzymes and gases.

3.2 Biomarker Monitoring

3.2.1 Screen-Printed Electrodes on Fabric

The application of industrial ink printing technology to fabric materials has facilitated the creation of fabric-based wearable electrochemical sensors. Screen printing stands out among these technologies, especially for disposable paper-based electrodes. Following its success in producing disposable glucose test strips [57, 58], screen-printed electrodes (SPE) have been directly applied to fabrics for a variety of biochemical analytes. Wang's group pioneered screen printing onto textiles and garments for biosensing purposes [22, 24, 59]. In their initial paper, they developed an amperometric sensor by screen printing a carbon working electrode onto the elastic

waistband of underwear. They used a platinum wire and Ag/AgCl as the counter and reference electrodes, respectively [24]. These printed sensors demonstrated high tolerance to mechanical stress caused by cloth deformation. In a subsequent study, all three electrodes were screen-printed onto textiles to detect nitroaromatic explosives [22]. The authors revealed that differences in the hydrophobicity of the textile substrate had a significant impact on the electrochemical behavior of the SPE. A waterproof fabric, GORE-TEX, was used due to its hydrophobic nature, which enabled minimal sample absorption, ensured good printing quality, and maintained the reproducible electrochemical activity of the SPE. In a subsequent investigation, the group introduced a wearable screen-printed sensor on synthetic rubber neoprene to detect pollutants and security threats in marine environments [59]. Once again, the superhydrophobic nature of the selected fabric contributed to excellent electrochemical performance. Importantly, this study also presented the first enzyme-based screen-printed sensor on fabric. Overall, these initial works demonstrated the feasibility of applying SPE to fabrics for biochemical sensing and examined how fabric properties influenced SPE performance.

In another study by Wang's group, they developed a smart bandage for sensing uric acid in a wound environment. They accomplished this by screen printing a Prussian, blue-mediated carbon electrode onto a commercial bandage, followed by chemical modification of the working electrode with uricase (Fig. 2a–c) [23]. Beyond hydrophobic, water-repellent fabrics, more common cotton- and nylon-based textiles were used as substrates for SPE. For instance, Corcoles and coworkers reported an electrochemical lactate sensor by printing electrodes onto cotton fabric [60]. They employed the wax-patterning technique to create a hydrophobic barrier serving as the sample placement/reaction zone on the hydrophilic cotton substrate. This procedure is similar to the approach widely adopted in electrochemical paper-based assays [61]. Lactate oxidase (LOx) is immobilized on the working electrode to enable amperometric measurement of lactate. The authors also explored the potential of incorporating sample collection functionality using a hydrophilic cotton thread. While most of the aforementioned studies predominantly used amperometric methods for sensing, Prasad and coworkers developed an affinity-based sensor by screen printing silver conductive electrodes and a graphene oxide transduction film onto utility textiles. The electrodes were then chemically modified with an antibody to sense influenza protein (Fig. 2d–f) [62]. The authors validated the reproducibility and stability of the fabrication procedure and the affinity-based assay using electrochemical impedance spectroscopy (EIS). Notably, EIS offers the advantage of miniaturizing the signal readout component, as the output of EIS has negligible memory consumption. However, the study did not demonstrate the sensor's performance in biological samples with possible interference from non-specific binding.

Nomura and coworkers reported a blood leakage sensor using a screen-offset printing technique on cotton textiles (Fig. 2g) [63]. Applying conventional screen printing on textile surfaces would result in ink blurring due to the highly porous textile structure unless the substrate was hydrophobic [22, 59]. To address this challenge, the authors employed screen-offset printing, adapted from screen printing. In this method, the ink was initially printed onto a silicon blanket that absorbed the

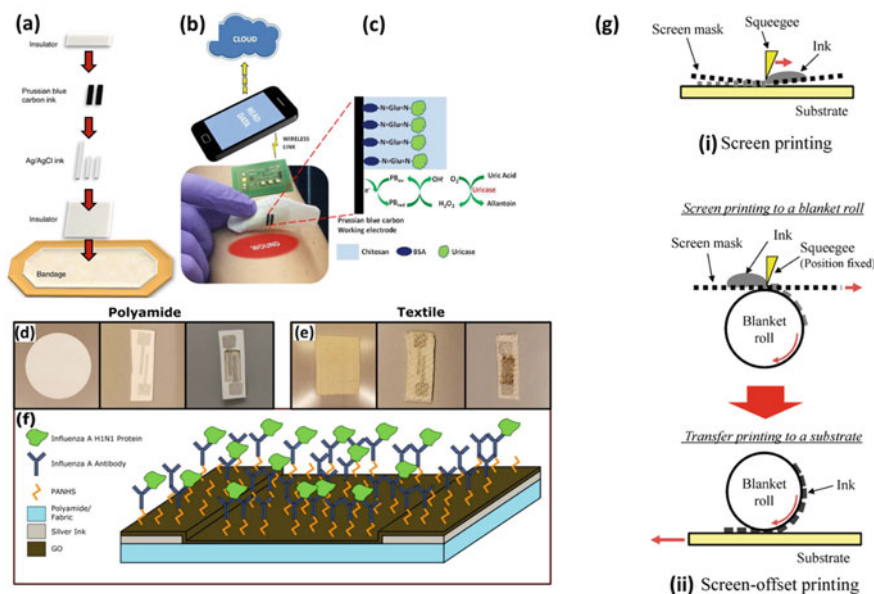


Fig. 2 **a** Screen printing the smart bandage. **b** Wearable potentiostat determines UA concentration and wirelessly communicates with a computer or Smartphone. **c** Schematic showing amperometric detection of UA with uricase immobilized on PB working electrode [23]. **d** Fabrication process flow for the flexible screen-printed electrodes on polyamide. From left to right, raw polyamide sheets are screen-printed with a $\sim 100\ \mu\text{m}$ thick pattern of conductive silver ink to form two electrodes. After, a layer of graphene oxide is screen-printed in a complimentary manner completing the fabrication. Shown are individual sensors, while fabrication occurs in sheets of 14 electrodes. **e** Fabrication process flow for the flexible screen-printed electrodes on textile. From left to right, textile sheets are screen-printed with a $\sim 100\ \mu\text{m}$ thick pattern of conductive silver ink to form two electrodes. After, a layer of graphene oxide is screen-printed in a complimentary manner completing the fabrication. **f** Cross-section schematic visually detailing the affinity assay for influenza where PANHS crosslinker binds to the graphene oxide allowing for subsequent binding of influenza protein antibody and influenza protein for detection [62]. **g** Schematic images of (i) screen printing and (ii) screen-offset printing [63]. Figure reproduced with permissions

organic solvent in the ink, making it more viscous. The pattern was then transferred onto textile substrates with improved resolution. The authors detected blood leakage volumes as low as $15\ \mu\text{L}$ by measuring the frequency dependence of blood and other liquid parameters.

3.2.2 Functionalized Thread as Individual Electrodes

As the fundamental unit of fabrics, threads can be pre-functionalized as sensing electrodes and incorporated into textiles through weaving, knitting, and embroidering methods. This “bottom-up” approach facilitates multiplex sensing with various thread electrode types, offering greater stability compared to SPE. This stability

arises from the natural embedding of sensing elements within the fabric. The mass production of such sensors is cost-effective and has high throughput, thanks to the well-developed textile industry. In this section, we will review recent advancements in crafting single thread-based electrochemical sensors, with an emphasis on different functionalization methods and integration techniques into clothing.

Methods for Individual Thread Functionalization

The functionalization of thread for electrochemical sensing has a history dating back to the 1990s when Reynolds et al. and Collins et al. experimented with coating textile threads with conductive polymers for gas sensing [64, 65]. The initial step in functionalization typically involves coating the thread with a conductive layer, as conventional thread materials are non-conductive (such as cotton, wool, and nylon). In some studies, conductive threads like carbon fiber or conductive carbon yarn have been used due to their strong electrical conductivity [66, 67]. Shim and colleagues were pioneers in using carbon nanotube (CNT)-coated cotton thread for immunosensing [68]. They employed a simple dipping method to absorb CNT and polyelectrolytes onto the cotton thread. With each dipping, the resistivity decreased, reaching as low as 20 Ω/cm . The absorbed CNT exhibited stability against washing and heating, likely due to interactions between polyelectrolytes and cotton-based materials. Anti-albumin antibodies were further coated onto the CNT layer to detect albumin by measuring changes in thread resistance. Similarly, Andrade and coworkers coated cotton yarns with CNT and ion-selective membranes for pH, K^+ , and NH_4^+ measurements in body fluids [69]. Whitesides' group demonstrated an ion sensor for Cl^- , K^+ , Na^+ , and Ca^{2+} using cotton thread [56]. They developed a surfactant-free ink consisting of carbon black as the conductive material and an organic polymer as the ink binder. This modification addressed the poor stability of the previous CNT-coated thread. Carbon black's high specific capacitance established a stable interfacial potential between the transducer and the ion-selective layer, minimizing signal drift. A detailed sensor fabrication process is illustrated in Fig. 3a–c, where the thread tip is functionalized through dip-coating and sealed with heat-shrink tubing or nail polish.

Another approach was developed by Sonkusale's group, who created a continuous coating device to sequentially deposit various inks onto a single thread fiber with drying capabilities [7]. This method enabled the fabrication of meters of functionalized thread, collected on rotating spools (Fig. 3d) [7]. Besides CNT and carbon black ink, functionalized graphene oxide (FGO) was also explored as a conductive and recognition layer for gas sensing [70]. Graphene oxide was modified with organic molecules to distinguish specific gas species. The functionalized FGO was dip-coated onto a single cotton yarn as the sensing electrode. FGO, with available sites for gas molecule absorption, showed exceptional mechanical and chemical stability. Tang et al. [71] reported the conversion of non-conductive silk fibers into conductive substrates through carbonization under 950 $^\circ\text{C}$. They further enhanced the fabric electrode's electrochemical performance by dip-coating CNT and modifying the fabric

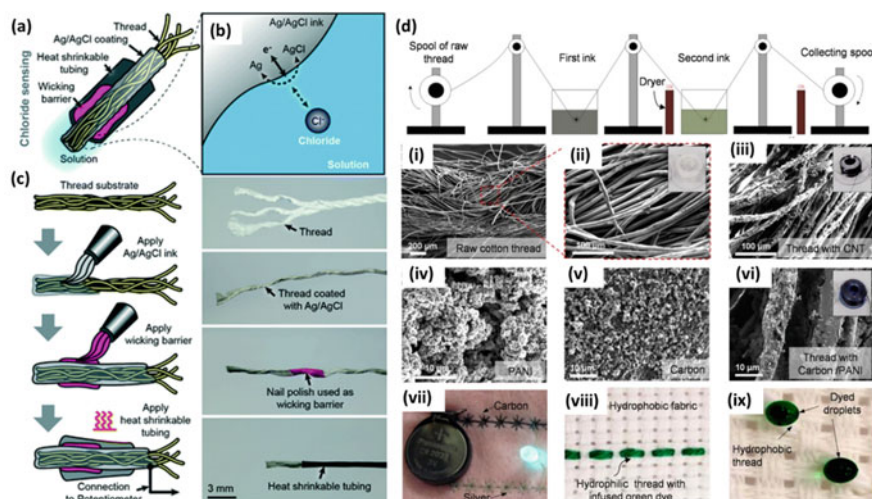


Fig. 3 Design (a), working principle (b), and fabrication process (c) for the thread-based Cl^- ISE [56]. **d** Fabrication of functional threads. (i) Schematic representation of the coating system for preparation of conductive threads. (ii) SEM image of the cotton raw thread. (iii) SEM image of a nano-infused thread coated with CNTs. (iv–vi) SEM images of the surface of PANI-, carbon-, and carbon/PANI-coated threads. (vii) Patterned conductive threads interconnect on a woolen fabric to illuminate an LED. (viii) Embroidered hydrophilic threads on a hydrophobic fabric after the green dye was infused. (ix) Hydrophobic threads repelling water. Green food dye was used to improve visualization. LED, light-emitting diode most [7]. Figure reproduced with permission

surface with platinum microspheres and glucose oxidase (GOx) for glucose sensing [72]. This combination offered good electrical conductivity and flexibility, with platinum microspheres significantly enhancing sensitivity. Sekar et al., used conductive carbon yarn (CCY) to fabricate an immunosensor for sweat cortisol detection [66]. They directly grew Fe_2O_3 nanoparticles on CCY using a hydrothermal method. These “binder-free” electrodes eliminated the need for binder materials added to conductive ink, enhancing substrate adhesion [73]. The directly grown Fe_2O_3 nanoparticles, shaped like ellipsoids, shortened the ion and electron transport path due to their high aspect ratio. These nanoparticles, decorated with anti-cortisol antibodies, enabled the detection of fg-level cortisol using high-sensitivity differential pulse voltammetry.

Methods for Integrating Individual Thread into Fabric and Clothing

Threads, as the fundamental building blocks of textiles, can be utilized and processed using various techniques to create the final clothing product. Established textile technologies are readily available for integrating individual thread electrodes into wearable sensors. Dendukuri’s group explored textile weaving to mass-produce thread-based electrochemical sensors [26, 73]. Using a handloom, enzyme-functionalized threads were woven into fabric patches, allowing for the creation of a sensing zone

with defined hydrophilic and hydrophobic silk yarns. This design confined the sample flow to hydrophilic yarns only, avoiding wicking into adjacent hydrophobic yarns. The one-layer structure offered advantages in resource-poor settings and minimized ink wastage. In contrast to top-down patterning technologies like screen printing, which often result in multi-layer structures, this weaving approach maintained a constant electrode surface area during measurements. Lillehoj's group investigated the use of embroidery to craft thread-based electrochemical sensors into fabrics like cotton gauze and t-shirts [74, 75]. Embroidery allowed for custom geometries, providing flexibility for electrochemical instrumentation. Computerized embroidery machines were used with predefined patterns, and parameters like stitch density and length were optimized for uniformity and reduced thread resistance. The embroidered dual sensor demonstrated a linear response to glucose and lactate due to pre-coated enzymes, enduring 100 bending cycles and proving highly durable for in vitro testing. Similarly, Sonkusale's group [7] reported the development of a pH sensor array and a glucose sensor by embroidering functionalized hydrophilic threads onto highly hydrophobic woven fabric (Fig. 4). In this work, a thread-based toolkit was developed to measure chemical (pH and glucose) markers and physical (strain and temperature) in an integrated thread-based diagnostic device (TDD) platform. The hydrophobic woven fabric was adorned with hydrophilic threads, ingeniously fashioned into microfluidic channels designed to precisely transport bodily fluids to specific sensing zones. To measure key physiological parameters such as glucose levels, pH, temperature, and strain, conductive threads enhanced with nanomaterials like carbon nanotubes (CNTs), carbon nanopowders, polyaniline (PANI), and various combinations thereof, were employed as the foundation for thread-based electrodes, both in vitro and in vivo. These sensor outputs were seamlessly integrated with a separate layer housing signal processing electronics and wireless communication capabilities, facilitated by conductive threads serving as interconnections, thus enabling seamless data transmission to a smartphone or computer.

3.3 *Physical Sensors*

Biophysical sensors, which monitor parameters like temperature, strain, and pressure, find another application in thread sensors. Utilizing the dip-and-dry coating technique depicted in Fig. 5, ordinary threads crafted from materials like cotton, polyester, or polyurethane can be endowed with functional conductive and dielectric layers. This conductive layer imparts strain-sensitive conductance to the thread, while the dielectric layer ensures insulation from other conductors and safeguards against physical harm. We will discuss them in this section.

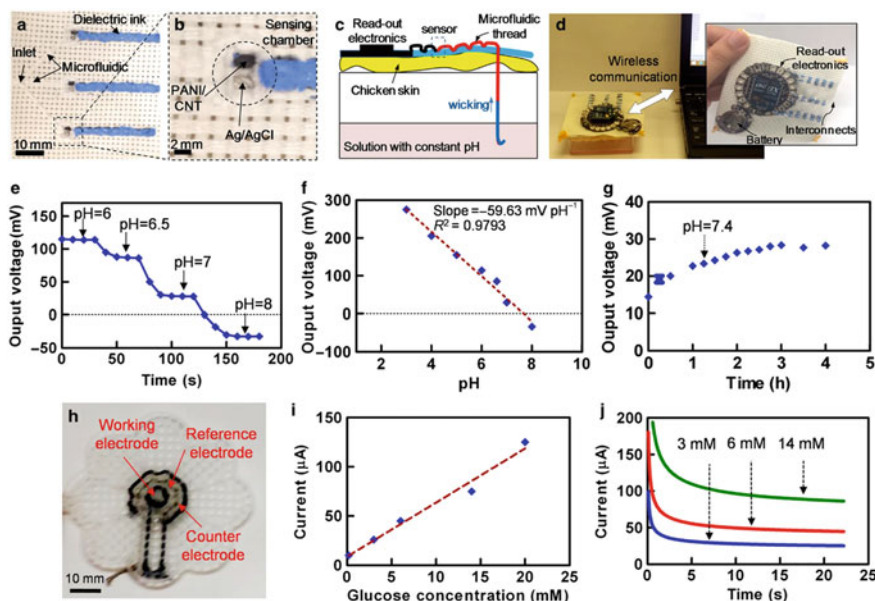


Fig. 4 Characterization of chemical sensors. **a** and **b** Optical image of a multiplexed microfluidic pH sensors assay. **c** Schematic illustration of measuring pH in an in vitro skin model. **d** Sensing system communicating with an external computer via a wireless system. **e** Transient response of the pH sensor to different pH values. **f** Calibration plot of the pH sensor. **g** Continuous pH measurement for four hours. **h** Optical image of the glucose sensor. **i** Calibration plot of the glucose sensor. **j** Transient response of the glucose sensor to different glucose concentrations in the PBS solution. PBS, phosphate-buffered saline [7]. Figure reproduced with permission

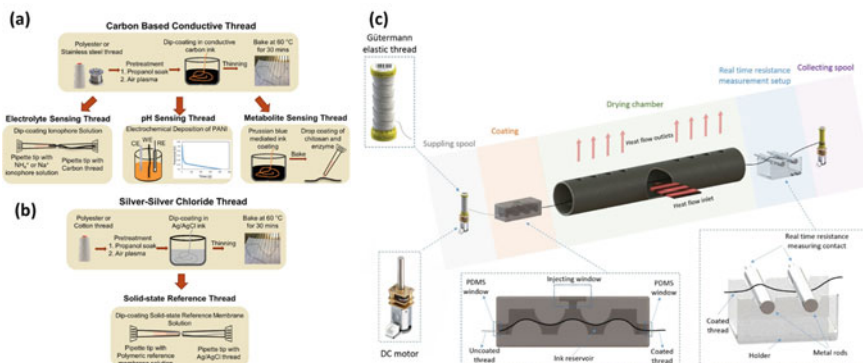


Fig. 5 Steps in preparation of **a** carbon-based conductive threads and sensing threads, **b** PVB-coated Ag/AgCl/PE thread [7], **c** Reel-to-reel automated fabrication of smart threads showing the coating process involving a specialized cartridge with ink that also enables pre-stretching, drying, real-time resistance monitoring and spool collection [76]. Figure reproduced with permission

3.3.1 Thread-Based Capacitive Strain/Pressure Sensors

Thread-based pressure and strain sensors commonly utilize a capacitive mechanism for their design. These sensors may incorporate either intrinsic or extrinsic coatings of diverse materials, yet the underlying principle remains uniform across all variations. Thread-based capacitive strain sensors operate based on the principle of changes in capacitance resulting from mechanical strain applied to the sensor. These sensors are constructed by integrating conductive threads into a flexible substrate, with a dielectric layer between them. When the sensor is subjected to mechanical deformation or strain, the distance between the conductive threads changes, leading to a change in capacitance. The fundamental equation governing the capacitance (C) of a capacitive strain sensor is similar to the general capacitance equation.

$$C = \varepsilon_0 * \varepsilon_r * A/d$$

where:

ε_0 is the vacuum permittivity

ε_r is the relative permittivity (dielectric constant) of the material between the conductive threads.

A is the area of overlap between the conductive threads.

d is the distance between the conductive threads.

As strain is applied to the sensor, the separation distance (d) between the conductive threads alters, causing a corresponding change in capacitance. This change in capacitance is directly proportional to the magnitude of the applied strain. By measuring the resulting capacitance change, the strain or deformation experienced by the sensor can be accurately determined. Thread-based capacitive strain sensors offer advantages such as flexibility, lightweight design, and suitability for wearable and conformable applications. They find use in various fields, including biomechanics, structural health monitoring, and wearable electronics, providing valuable insights into the deformation and mechanical behavior of objects and materials.

One of the initial forms of capacitive contacts was proposed by Post et al. [77], where the capacitance between two electrodes changes upon finger contact. The measurement of capacitance involves assessing the time required to charge the capacitor (formed by the electrodes and the finger) from zero to the switching threshold voltage of a CMOS logic buffer. This capacitive contact presents the advantage of time-domain measurement, offering a dynamic range of 10^4 . The electrodes within this sensor are constructed using conductive thread, while the dielectric material is composed of cloth. For ideal yarn or textile in fabric circuits, it would possess fully adjustable electrical properties, maintaining these characteristics during sewing, bending, and wear. In their study, the authors employed stainless steel yarns as conductive electrodes. Stainless steel yarn holds certain advantages, notably its inert nature, rendering it washable and impervious to sweat-related effects. These stainless-steel yarns exhibit varying compositions, ranging from 100% continuous conductive steel fibers to feltings or composites intertwined with polyester and short

steel fibers. Altering the ratio between these constituent fibers yields variations in resistivity.

Sergio et al. [78] introduced an alternative capacitive platform for pressure sensing that utilizes conductive fabric. This approach involves mapping the distribution of applied pressure across the fabric surface by detecting changes in capacitance between rows and columns of conductive fibers arranged on opposing sides of an elastic synthetic foam layer. The application of pressure to the smart fabric leads to a reduction in the dielectric layer's thickness between corresponding rows and columns, thereby increasing the coupling capacitance. To process the data, a dedicated chip is positioned at the corner of the smart fabric. The change in induced charge is transformed into a voltage value through readout circuitry integrated into the chip. The central region of the pressure sensor occupies an area of 32 mm². In a similar vein, Meyer et al. [79] employed a comparable approach in their pressure sensor design. They employed conductive textiles on both sides of a compressible spacer. Single electrodes were arrayed on one side of the spacer, while the other side featured a single common electrode. This configuration forms capacitors between each individual electrode and the common electrode. The single electrodes are skillfully embroidered with conductive yarn, while the common electrode is composed of a woven textile coated with silver. Hoffman et al. [80] devised a capacitive textile force sensor tailored for respiration monitoring. Their force sensor comprises a compressible 3D textile situated at its core, dictating its sensitivity to applied forces. Functioning as a dielectric layer, the 3D textile is sandwiched between two layers of conductive fabrics, which serve as electrodes. Notably, they opted for a fabric containing 75% silver. In a similar vein, Hollecze et al. [81] engineered a pressure insole with an extended range of pressure detection. For its electrodes, they utilized silver-coated textiles, while a compressible spacer crafted from CrosliteTM (known as Proprietary Closed Cell Resin or PCCR) served as the dielectric layer. However, the sensor's larger size rendered it inconvenient for sock integration. The employed silicone sealant exhibited quick peeling, and the connecting wires were susceptible to breakage. A significant concern was the lack of washability for the socks bearing the assembled sensors.

3.3.2 Fiber and Yarn Strain-Based Sensors

Textile sensors emerge as potent tools for monitoring bio-signals. Catrysse et al. [82] introduced a textile sensor for monitoring electrocardiogram (ECG) and respiration rate. They employed woven and knitted stainless steel electrodes named "Textrodes" for ECG signal measurement (Fig. 6a, b). The Textrodes were advantageous for their integration into garments and skin-friendly nature. However, they exhibited higher skin-electrode impedance (1–5 MW cm²) compared to commercially available gel electrodes (10 kW cm²). Similarly, Cho et al. [83] proposed various conductive fabrics as electrodes using three approaches: (1) Sputtering on Polyurethane (PU) laminated and PU dry-coated nylon fabrics (2) electrodeless Cu/Ni plated fabrics (3) Embroidering with stainless steel filament yarns over woven or knitted fabrics.

Three types of electrodes were developed using different materials by Cho's group: metal yarn embroidered on plain-woven fabric, blended knitted fabric of metal yarns and cotton, and metal yarn embroidered on blended knitted fabric. Knitted fabric (100% stainless steel yarns) and braided fabric (spandex core covered with polyester filament yarns and stainless-steel multifilament yarns) were employed as motion sensors. Zhang et al. [84] utilized carbon yarn and steel fibers to fabricate strain sensors. Carbon fiber, due to its smooth surface and intrinsic properties, outperformed stainless steel for gauge fabrication. However, steel fiber's strength made it suitable for frequency sensor application. Pacelli et al. [85] introduced textile piezoresistive sensors for post-stroke rehabilitation and cardiovascular disease monitoring. They developed knitted piezoresistive fabric (KPF) sensors using the intarsia technique and Printed Piezoresistive Fabric (PPF) sensors through a coating printing process of Conductive Elastomer (CE) material. Huang et al. [86] innovatively fashioned yarns using piezoresistive, elastic, and regular polyester fibers, which were then employed to create cloth, dresses, and sensing textiles. Various wrapping methods were employed, with double wrapping proving more linear. Different fibers, including piezoresistive, elastic, and polyester, were incorporated into the yarn to create the composite core yarn, followed by the application of the carbon-coated fiber using single and double wrapping techniques.

Recently, Sonkusale's group demonstrated the preliminary finding and fabrication of a sensing pillow to monitor head movement [87]. In this paper, we incorporate strain-sensing threads comprised of 64% polyester, 36% polyurethane, and a polydimethylsiloxane (PDMS) layer. A memory foam pillow is then created using them. The threads' covering gives them resilience, enabling the smart threads to withstand exposure to moisture from washing or perspiration. The gauge factor of the smart threads is approximately 2.5 for 50% strain. With a thread resistance range of 0.5–20.0 M, it displays linearity for applied weights of 50.0–5000.0 g. The resistance readout circuitry and wireless module are linked to the smart threads to wirelessly record and communicate strain data to a computer. Initial findings show that the smart pillow is functional and can measure a person's head motion in real time.

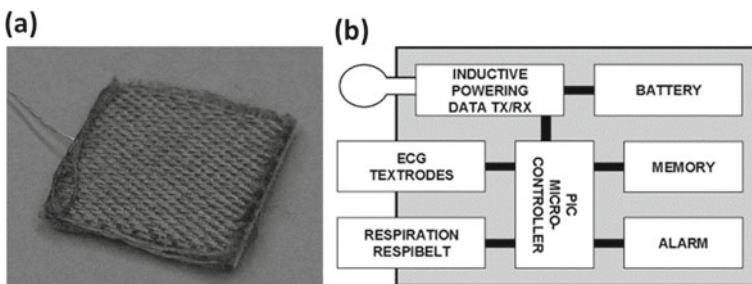


Fig. 6 a Schematic overview of the wearable electronics. b Photograph of a knitted 3×3 cm Textrode. Figure reproduced with permission [82]

4 Thread-Based Transistors, Antennas, and Circuits for Thread-Based Sensors

4.1 *E-Textile and Thread-Based Antennas*

Over the past couple of decades, there has been a growing interest in wireless technologies which enable many electronic devices to connect via WiFi, WLAN, or Bluetooth. As wireless devices and body-centric communication systems gained importance owing to the advancements in semiconductor research and nanotechnology, the future of electronics is moving toward wearable devices. Even though the design and verification of such systems are challenging due to cost, integration, and space requirements, it is now possible to shrink the size of the most sophisticated electronic components. Thus, the same idea applies to high-frequency circuits as well. One of the most crucial components in wearable devices is the antenna. This section is thereby devoted to the overview of textile-based antennas, and the design, fabrication, and discussion of thread-based antennas.

Being seen virtually everywhere, the antennas have lots of applications in automobiles, spacecraft, held mobile devices, the military, radar, and so on. The design of antennas for wearable devices has necessitated a change toward flexible materials that comprise textiles, fabrics, and threads instead of utilizing rigid and flexible antenna elements and structures. It must be noted that wearable technology and devices are not the only application area of textile- and thread-based antennas despite being the most used ones. As presented in the literature, wearable devices can be used in many applications including imaging, sensing, and healthcare [88–93], and wireless communications [94–97]. and radio-frequency identification (RFID) [98, 99]. The antennas designed for wearable applications in which wireless connectivity is required are mostly textile-based. For body-centric communications, antennas are designed to be wearable such that they can be integrated into clothes or woven fabrics. These materials should be suitable for sustaining an optimal level of RF antenna performance. Wearable and textile-based antennas occasionally comprise conductive textiles (e-textiles) in which a polymer is coated with a conductive material [100–102]. Antennas fabricated with electro-textiles offer good RF performance, flexibility, compactness, and mechanical durability [103–105].

E-textile materials and e-textile-based antennas have been developed owing to the improvements in the Internet of Things (IoT) and the demand for wearable devices in the field of sensing and electronics. Nowadays, it is possible to find many off-the-shelf e-textile materials that have various conductivity, flexibility, and mechanical durability. Being mainly used as EM shielding materials [106], e-textiles have found an important place in wearable antennas [103, 106–109]. In general, textile materials and fabrics are inherently nonconductive which necessitates techniques for obtaining conductive areas or surfaces to realize antennas. Briefly, there are two methods to make such materials conductive [110]. The first method is coating the nonconductive surface after the weaving process to form a cloth or fabric. The other technique is using ready-made threads or conductive fibers and integrating them into the cloth

or fabric by interweaving or embroidery process. Textile structures and fabrics can be made electrically conductive via knitting, weaving, or embroidery [110, 111]. To be able to fabricate antennas with optimal electromagnetic performance, surface resistivity must be kept below $1 \Omega/\text{sq}$ [112].

It has been investigated the relationship between woven densities on a fabric and antenna performance [110]. It has been shown that the e-textile antennas with woven patterns whose grid length is less than 2 mm have antenna performance in terms of radiation efficiency and pattern. Once the distance between two conductive fibers exceeds 2 mm, an offset in antenna resonant frequency is observed. Thus, if the grid length is small enough, it is shown that the conductive e-textile antenna can be regarded as a homogeneous metal plate. In addition to the woven density, the type of the e-textile material is also effective on antenna performance. Four e-textiles are utilized in testing various types of materials that have different resistivities within $0.03 \Omega/\text{m}^2$. Having fabricated e-textiles on FR-4 substrate, antennas made of different materials show that some of the conductive e-textiles are only suitable for decent antenna performance.

Textile-based antennas that comprise electro-textiles and e-threads are primarily based on macro-level functionalization of the radiating structures using fabrication techniques such as embroidery and screen- or inkjet-printing. In general, off-the-shelf threads, yarns, or fibers are utilized in the production of antennas as described in the previous section. However, the important factor that separates the thread-based approach from the textile-based antennas is the functionalization of each fiber in the thread structure. There have been numerous studies that focus on the device-level fabrication and analysis of thread-based antennas, especially using carbon nanotubes (CNTs) even though the number of published works in this field is still relatively low. The key advantage of using CNT fibers is the possibility of attaining higher fiber conductivity which can result in a huge antenna radiation performance [113].

Current research is primarily focused on the utilization of CNT bundles and CNT threads for the implementation of antennas. Owing to their mechanical, electrical, and thermal properties, CNTs have become potential materials for achieving mechanically stable and lightweight antennas for wireless communications. Aside from the thread-based antennas, a few studies have investigated microstrip antennas in which CNTs are utilized as basic RF signal radiators [114–116]. The performance of nanotubes and nanowires has been intensively investigated in terms of input impedance and antenna efficiency in which mathematical models are developed for near- and far-field radiation patterns [117]. The analysis showed that individual CNTs possess weak radiation performance which can be attributed to the value of quantum capacitance and poor resistive part of the input impedance. Since the analyzed antenna is of a dipole type, self-impedance (0.02Ω) and mutual impedance (1.2Ω) have been calculated to determine the radiation region in which the antenna has the highest efficiency. In other words, it is observed that there is a remarkable amount of coupling between the wires which concludes that antennas are better radiators in the near field. The physical measurements that the high-frequency impedance of carbon nanotube bundles improve by the scaling of inductance with the number of each CNTs [118].

There are two basic techniques to fabricate carbon nanotube threads: dry spinning and wet spinning. The first step of the dry spinning method is the synthesis of vertically aligned CNT arrays (spinnable arrays) that comprise double or multiwall nanotubes. The density and length of CNTs inside the array are two important factors as there should be a sufficient number of contact points to ensure thread assembly by van der Waals interaction. In other words, multiwall Nanotubes are gathered and twisted into a thread structure in which every single nanotube adheres via van der Waals forces [119]. Following the synthesis of a vertically aligned CNT array, a spinnable array can be obtained using chemical vapor deposition (CVD) in which CNTs can grow under the mixture of Ar, C₂H₄, H₂, and H₂O [119, 120]. The width of the spinnable array is effective on the diameter of the final version of the thread which is also known as 1-ply (single-ply) thread. The dry spinning method and the usage of a spinnable array provide controllability to the fabrication process such that the desired density, length, and CNT thread diameter can be attained. Further weaving processes can also be used to obtain 3-ply CNT threads and ropes. Additionally, it was also proposed and validated in a postprocessing method called densification to increase the contact area between the neighboring nanotubes using dimethyl sulfoxide (DMSO) to improve electrical conductivity and mechanical strength [120]. The results of Keller's investigation of the thread spacing on the E-plane radiation pattern are related to the spacing and geometry.

The other CNT thread fabrication method, wet-spinning, can also be used to fabricate highly efficient thread-based monopole antennas [113]. It is reported that the dry-spun CNT fibers possess low radiation efficiency which is highly attributed to the density of the CNT thread. However, threads that possess very high density and conductivity can be utilized in obtaining antennas with better radiation performance. In contrast to dry spinning, much denser and highly ordered CNT fibers can be obtained via the wet-spinning method. CNT fibers from chlorosulfonic acid using single-walled carbon nanotubes (SWCNTs) are fabricated before the purification of the fibers that can create the antenna structure [113]. The CNT fibers are purified with H₂O₂ and HCl to obtain high conductivity and stabilize the overall fiber structure. Various 1/4-long monopole antennas have been fabricated to investigate the effect of different design parameters such as the number of CNT fibers and the twist of the thread. The major outcomes of this work are worth mentioning as the overall study is very important in terms of pioneering the thread-based approach to wireless connectivity. The twist of the thread which can be classified as low (~1 turn/mm) and high twist (~2 turns/mm) has no major influence on the DC resistance or antenna radiation performance. DC resistance of the CNT-thread-based antenna is inversely proportional to the cross-sectional area. The conductivity of the single fiber is almost an order of magnitude smaller than multifilament high-conductivity fibers. Concordantly, thread-based antennas made of CNT fibers have an improved S₁₁ performance over 30 AWG copper wire at least an order of magnitude. The radiation efficiency of the antenna is validated through 2-port VNA measurements inside a reverberation chamber with a standard horn antenna. The radiation efficiency of the CNT thread antenna significantly improves with increased cross-sectional area and lower operational frequencies as the antennas are tested at 1 GHz and 2.4 GHz.

It has also been reported that the frequency dependence of the radiation efficiency can be attributed to an intrinsic RF property of the CNT thread antenna because there is no impedance matching between the VNA and the antenna. Therefore, a modified version of radiation efficiency is proposed to quantitatively compare the threaded antenna with the copper wire antenna.

4.2 Readout Circuits for Thread-Based Sensors

There has been a growing interest in the utilization of thread-based sensors in various platforms such as health monitoring and wearable devices in the form of minimally invasive or non-invasive fashion. It is often desired to obtain information related to biophysical markers such as respiration rate, electrocardiogram (ECG), electrooculogram (EOG), core body temperature, or metabolism of an individual from biological fluids, sweat, and saliva which might be utilized to provide an assessment of individual's physical performance [16, 54, 121, 122]. In addition to health monitoring, flexible sensor systems are broadly used in scientific and industrial applications in which monitoring pressure, acceleration, force, gas concentration, and humidity is of primary importance [87, 122–127]. The block diagram of a flexible sensing system is shown in Fig. 7. A physical quantity (or measurand) that is being measured is sensed by the flexible sensor before feeding the signal into the signal conditioning circuit. Usually, sensors convert the measurand into a signal that could be in the form of voltage, current, capacitance, or resistance which could further be processed by the signal conditioning circuit. Different sensors could generate different outputs in which the signal amplitude could be either too weak or moderately weak which necessitates amplification. One major function of the signal conditioning circuit is to be able to amplify the signal at the output of the sensor. Furthermore, the signal generated by the sensor could possess unwanted harmonics and noise which could be suppressed and filtered out by the signal conditioning circuits [128]. Hence, the sensor front-end system can preprocess the sensed signal which can now be accepted by the digital signal processor for further digitization, visualization, and eventually data analysis.

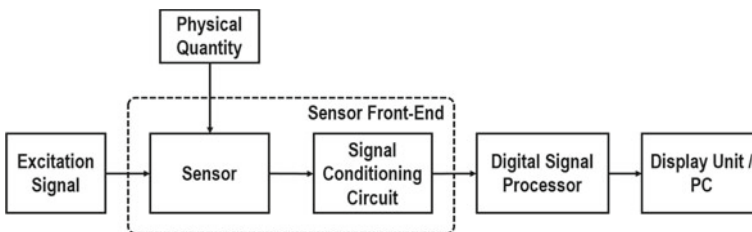


Fig. 7 Block diagram of a sensor readout system

5 Conclusions and Perspectives

The next generation of personal healthcare devices and medical therapies requires seamless integration of biosensors and power generation/storage technologies into flexible and wearable formats. In this context, threads and sutures emerge as highly promising solutions. Threads offer a plethora of advantages, including their universal availability, cost-effectiveness, material versatility, and straightforward textile-based processing. Moreover, they inherently lend themselves to wearability and seamless integration with the human body. Additionally, the thread can function as a smart flexible substrate for sensing, computing, communication, and drug delivery. Leveraging the process of surface modification by physical coating or reel-to-reel fabrication techniques, involving sequential dip-coating, and drying, enables the production of functional smart threads tailored for various sensing, therapeutic, and electronic applications. For example, smart threads have been specifically engineered to sense chemical and physical biomarkers to monitor health and well-being. Furthermore, these smart threads can be implanted in tissue as sutures or incorporated into wearable kits, such as band-like patches, to monitor multiple electrolytes and metabolites in sweat. The above-mentioned applications marked the unique ability of threads to facilitate three-dimensional, tissue-embedded bioelectronic interfaces, a feat unattainable with other flexible bioelectronic platforms.

Nevertheless, there is substantial room for further exploration and innovation in the development of smart threads capable of sensing physio-chemical activity. The flexible nature of threads or sutures provides a distinctive substrate that can be effortlessly sewn or affixed onto any item of wearable clothing or serve as a standalone device. In the near future, we can envision clothing in which threads interconnect to create an intelligent system capable of continuously monitoring an individual's health and real-time environmental changes. In the realm of fiber or thread substrates, novel fabrication, and modification techniques for creating hybrid or nanocomposite thread materials hold significant potential, providing control over material composition and functionality.

For the future, smart threads hold promise in applications such as neural implants, drug delivery, and the development of cardiac implants or patches. Ongoing research and recent findings suggest that threads can be harnessed for on-demand drug delivery and serve as scaffold substrates for cell growth, thus opening up exciting avenues for treatment. Looking ahead, it may become feasible to seamlessly integrate sensing and therapeutic functions within the same wearable platform, where threads play a dual role as a sensor and as a drug delivery device to close the loop. Lastly, there is still a need for improvement in terms of developing tighter integration and more sophisticated packaging for thread-based platforms with flexible electronics for remote health and environmental monitoring.

References

1. Chan, M., Estève, D., Fourniols, J.Y., Escriba, C., Campo, E.: Smart wearable systems: current status and future challenges. *Artif. Intell. Med.* **56**, 137–156 (2012). <https://doi.org/10.1016/J.ARTMED.2012.09.003>
2. Becerik-Gerber, B., Lucas, G., Aryal, A., Awada, M., Bergés, M., Billington, S., Boric-Lubecke, O., Ghahramani, A., Heydarian, A., Höelscher, C., Jazizadeh, F., Khan, A., Langevin, J., Liu, R., Marks, F., Mauriello, M.L., Murnane, E., Noh, H., Pritoni, M., Roll, S., Schaumann, D., Seyedrezaei, M., Taylor, J.E., Zhao, J., Zhu, R.: The field of human building interaction for convergent research and innovation for intelligent built environments. *Sci. Reports* **12**(1), 1–19 (2022). <https://doi.org/10.1038/s41598-022-25047-y>
3. Stanberry, B.: Legal ethical and risk issues in telemedicine. *Comput. Methods Programs Biomed.* **64**, 225–233 (2001). [https://doi.org/10.1016/S0169-2607\(00\)00142-5](https://doi.org/10.1016/S0169-2607(00)00142-5)
4. Gatzoulis, L., Iakovidis, I.: Wearable and portable eHealth systems. *IEEE Eng. Med. Biol. Mag.* **26**, 51–56 (2007). <https://doi.org/10.1109/EMB.2007.901787>
5. Sonkusale, S.: Smart threads for tissue-embedded bioelectronics. In: *Proceedings of the Custom Integrated Circuits Conference. 2022-April* (2022). <https://doi.org/10.1109/CICC53496.2022.9772846>
6. Servati, A., Zou, L., Jane Wang, Z., Ko, F., Servati, P.: Novel flexible wearable sensor materials and signal processing for vital sign and human activity monitoring. *Sensors* **17**, 1622 (2017). <https://doi.org/10.3390/S17071622>
7. Mostafalu, P., Akbari, M., Alberti, K.A., Xu, Q., Khademhosseini, A., Sonkusale, S.R.: A toolkit of thread-based microfluidics, sensors, and electronics for 3D tissue embedding for medical diagnostics. *Microsyst. Nanoeng.* **2**(1), 1–10 (2016). <https://doi.org/10.1038/micronano.2016.39>
8. Mostafalu, P., Lenk, W., Dokmeci, M.R., Ziaie, B., Khademhosseini, A., Sonkusale, S.R.: Wireless flexible smart bandage for continuous monitoring of wound oxygenation. *IEEE Trans. Biomed. Circuits Syst.* **9**, 670–677 (2015). <https://doi.org/10.1109/TBCAS.2015.2488582>
9. Chen, S., Qi, J., Fan, S., Qiao, Z., Yeo, J.C., Lim, C.T.: Flexible wearable sensors for cardiovascular health monitoring. *Adv. Healthc. Mater.* **10**, 2100116 (2021). <https://doi.org/10.1002/ADHM.202100116>
10. Qin, H., Owyung, R.E., Sonkusale, S.R., Panzer, M.J.: Highly stretchable and nonvolatile gelatin-supported deep eutectic solvent gel electrolyte-based ionic skins for strain and pressure sensing. *J. Mater. Chem. C Mater.* **7**, 601–608 (2019). <https://doi.org/10.1039/C8TC05918G>
11. Sun, W., Guo, Z., Yang, Z., Wu, Y., Lan, W., Liao, Y., Wu, X., Liu, Y.: A review of recent advances in vital signals monitoring of sports and health via flexible wearable sensors. *Sensors* **22**, 7784 (2022). <https://doi.org/10.3390/S22207784>
12. Punjiya, M., Rezaei, H., Zeeshan, M.A., Sonkusale, S.: A flexible pH sensing smart bandage with wireless CMOS readout for chronic wound monitoring. In: *TRANSDUCERS 2017—19th International Conference on Solid-State Sensors, Actuators and Microsystems*, pp. 1700–1702 (2017). <https://doi.org/10.1109/TRANSDUCERS.2017.7994393>
13. Sonkusale, S.: Sutures for the wireless sensing of deep wounds. *Nat. Biomed. Eng.* **5**(10), 1113–1114 (2021). <https://doi.org/10.1038/s41551-021-00806-w>
14. Gao, W., Emaminejad, S., Nyein, H.Y.Y., Challa, S., Chen, K., Peck, A., Fahad, H.M., Ota, H., Shiraki, H., Kiriya, D., Lien, D.H., Brooks, G.A., Davis, R.W., Javey, A.: Fully integrated wearable sensor arrays for multiplexed in situ perspiration analysis. *Nature* **529**:7587, 509–514 (2016). <https://doi.org/10.1038/nature16521>
15. Mondal, S., Zehra, N., Choudhury, A., Iyer, P.K.: Wearable sensing devices for point of care diagnostics. *ACS Appl. Bio Mater.* **4**, 47–70 (2021). https://doi.org/10.1021/ACSABM.0C00798/ASSET/IMAGES/LARGE/MT0C00798_0015.JPEG
16. Terse-Thakoor, T., Punjiya, M., Matharu, Z., Lyu, B., Ahmad, M., Giles, G.E., Owyung, R., Alaïmo, F., Shojaei Baghini, M., Brunyé, T.T., Sonkusale, S.: Thread-based multiplexed

- sensor patch for real-time sweat monitoring. *npj Flexible Electron.* **4**(1), 1–10 (2020). <https://doi.org/10.1038/s41528-020-00081-w>
17. Arquilla, K., Webb, A.K., Anderson, A.P.: Textile Electrocardiogram (ECG) electrodes for wearable health monitoring. *Sensors* **20**, 1013 (2020). <https://doi.org/10.3390/S20041013>
 18. Shen, S., Xiao, X., Xiao, X., Chen, J.: Wearable triboelectric nanogenerators for heart rate monitoring. *Chem. Commun.* **57**, 5871–5879 (2021). <https://doi.org/10.1039/D1CC02091A>
 19. Coyle, S., Morris, D., Lau, K.-T., Diamond, D., Moyna, N.: Textile-based wearable sensors for assisting sports performance, pp. 307–311 (2009). <https://doi.org/10.1109/BSN.2009.57>
 20. Lee, J.W., Han, D.C., Shin, H.J., Yeom, S.H., Ju, B.K., Lee, W.: PEDOT:PSS-based temperature-detection thread for wearable devices. *Sensors* **18**, 2996 (2018). <https://doi.org/10.3390/S18092996>
 21. El Gharbi, M., Fernández-García, R., Gil, I.: Embroidered wearable Antenna-based sensor for real-time breath monitoring. *Measurement* **195**, 111080 (2022). <https://doi.org/10.1016/J.MEASUREMENT.2022.111080>
 22. Chuang, M.C., Windmiller, J.R., Santhosh, P., Ramírez, G.V., Galik, M., Chou, T.Y., Wang, J.: Textile-based electrochemical sensing: effect of fabric substrate and detection of nitroaromatic explosives. *Electroanalysis* **22**, 2511–2518 (2010). <https://doi.org/10.1002/ELAN.201000434>
 23. Kassal, P., Kim, J., Kumar, R., De Araujo, W.R., Steinberg, I.M., Steinberg, M.D., Wang, J.: Smart bandage with wireless connectivity for uric acid biosensing as an indicator of wound status. *Electrochem. Commun.* **56**, 6–10 (2015). <https://doi.org/10.1016/J.ELECOM.2015.03.018>
 24. Yang, Y.L., Chuang, M.C., Lou, S.L., Wang, J.: Thick-film textile-based amperometric sensors and biosensors. *Analyst* **135**, 1230–1234 (2010). <https://doi.org/10.1039/B926339J>
 25. Schazmann, B., Morris, D., Slater, C., Beirne, S., Fay, C., Reuveny, R., Moyna, N., Diamond, D.: A wearable electrochemical sensor for the real-time measurement of sweat sodium concentration. *Anal. Methods* **2**, 342–348 (2010). <https://doi.org/10.1039/B9AY00184K>
 26. Choudhary, T., Rajamanickam, G.P., Dendukuri, D.: Woven electrochemical fabric-based test sensors (WEFTS): a new class of multiplexed electrochemical sensors. *Lab Chip* **15**, 2064–2072 (2015). <https://doi.org/10.1039/C5LC00041F>
 27. Woven Fabrics - Google Books. https://books.google.com/books?hl=en&lr=&id=r-KdDwAAQBAJ&oi=fnd&pg=PA267&dq=B.+Gutarowska+and+A.+Michalski,+Microbial+Degradation+of+Woven+Fabrics+and+Protection+Against+Biodegradation+&ots=rqgkFves_i&sig=r_9UtVKMDWzuswPapXoF5L1Dp9Y#v=onepage&q=B.%20Gutarowska%20and%20A.%20Michalski%2C%20Microbial%20Degradation%20of%20Woven%20Fabrics%20and%20Protection%20Against%20Biodegradation&f=false
 28. Murphy, A.R., John, P.S., Kaplan, D.L.: Modification of silk fibroin using diazonium coupling chemistry and the effects on hMSC proliferation and differentiation. *Biomaterials* **29**, 2829–2838 (2008). <https://doi.org/10.1016/J.BIOMATERIALS.2008.03.039>
 29. Barani, H., Calvimontes, A.: Effects of oxygen plasma treatment on the physical and chemical properties of wool fiber surface. *Plasma Chem. Plasma Process.* **34**, 1291–1302 (2014). <https://doi.org/10.1007/S11090-014-9581-X/TABLES/3>
 30. Barani, H., Maleki, H.: Plasma and ultrasonic process in dyeing of wool fibers with madder in presence of Lecithin **32**, 1191–1199 (2011). <https://doi.org/10.1080/01932691.2010.505525>
 31. Tajuddin, M., Ahmad, Z., Ismail, H.: A review of natural fibers and processing operations for the production of binderless boards. *Bioresources* (2016). <https://doi.org/10.15376/BIORES.11.2>
 32. Jamir, M.R.M., Majid, M.S.A., Khasri, A.: Natural lightweight hybrid composites for aircraft structural applications. In: *Sustainable Composites for Aerospace Applications*, pp. 155–170 (2018). <https://doi.org/10.1016/B978-0-08-102131-6.00008-6>
 33. Delebecq, E., Pascault, J.P., Boutevin, B., Ganachaud, F.: On the versatility of urethane/urea bonds: reversibility, blocked isocyanate, and non-isocyanate polyurethane. *Chem. Rev.* **113**, 80–118 (2013). https://doi.org/10.1021/CR300195N/ASSET/IMAGES/LARGE/CR-2012-00195N_0014.JPEG

34. Du, Y., George, S.M.: Molecular layer deposition of nylon 66 films examined using in situ FTIR spectroscopy. *J. Phys. Chem. C* **111**, 8509–8517 (2007). <https://doi.org/10.1021/JP067041NE00005.GIF>
35. Sadeqi, A., Rezaei Nejad, H., Alaimo, F., Yun, H., Punjiya, M., Sonkusale, S.R.: Washable smart threads for strain sensing fabrics. *IEEE Sens. J.* **18**, 9137–9144 (2018). <https://doi.org/10.1109/JSEN.2018.2870640>
36. Li, W., Liu, J., Wei, J., Yang, Z., Ren, C., Li, B.: Recent progress of conductive hydrogel fibers for flexible electronics: fabrications, applications, and perspectives. *Adv. Funct. Mater.* **33**, 2213485 (2023). <https://doi.org/10.1002/ADFM.202213485>
37. De France, K., Zeng, Z., Wu, T., Nyström, G., De France, K., Zeng, Z., Wu, T., Nyström, G.: Functional materials from nanocellulose: utilizing structure-property relationships in bottom-up fabrication. *Adv. Mater.* **33**, 2000657 (2021). <https://doi.org/10.1002/ADMA.202000657>
38. Tamayol, A., Akbari, M., Annabi, N., Paul, A., Khademhosseini, A., Juncker, D.: Fiber-based tissue engineering: progress, challenges, and opportunities. *Biotechnol. Adv.* **31**, 669–687 (2013). <https://doi.org/10.1016/J.BIOTECHADV.2012.11.007>
39. Rutledge, G.C., Fridrikh, S.V.: Formation of fibers by electrospinning. *Adv. Drug Deliv. Rev.* **59**, 1384–1391 (2007). <https://doi.org/10.1016/J.ADDR.2007.04.020>
40. Liu, N., Fang, G., Wan, J., Zhou, H., Long, H., Zhao, X.: Electrospun PEDOT:PSS-PVA nanofiber based ultrahigh-strain sensors with controllable electrical conductivity. *J. Mater. Chem.* **21**, 18962–18966 (2011). <https://doi.org/10.1039/C1JM14491J>
41. Karageorgiou, V., Kaplan, D.: Porosity of 3D biomaterial scaffolds and osteogenesis. *Biomaterials* **26**, 5474–5491 (2005). <https://doi.org/10.1016/J.BIOMATERIALS.2005.02.002>
42. Kim, D.W., Lee, O.J., Kim, S.W., Ki, C.S., Chao, J.R., Yoo, H., Yoon, S. il, Lee, J.E., Park, Y.R., Kweon, H.Y., Lee, K.G., Kaplan, D.L., Park, C.H.: Novel fabrication of fluorescent silk utilized in biotechnological and medical applications. *Biomaterials* **70**, 48–56 (2015). <https://doi.org/10.1016/J.BIOMATERIALS.2015.08.025>
43. Murphy, A.R., Kaplan, D.L.: Biomedical applications of chemically-modified silk fibroin. *J. Mater. Chem.* **19**, 6443–6450 (2009). <https://doi.org/10.1039/B905802H>
44. Kim, D., Herr, A.E.: Protein immobilization techniques for microfluidic assays. *Biomicrofluidics* **7**, 041501 (2013). <https://doi.org/10.1063/1.4816934>
45. Nimse, S.B., Song, K., Sonawane, M.D., Sayyed, D.R., Kim, T.: Immobilization techniques for microarray: challenges and applications. *Sensors* **14**, 22208–22229 (2014). <https://doi.org/10.3390/S141222208>
46. Prieto-Simon, B., Campas, M., Marty, J.-L.: Biomolecule immobilization in biosensor development: tailored strategies based on affinity interactions. *Protein Pept. Lett.* **15**, 757–763 (2008). <https://doi.org/10.2174/092986608785203791>
47. Salva, M.L., Rocca, M., Niemeyer, C.M., Delamar, E.: Methods for immobilizing receptors in microfluidic devices: a review (2021)
48. Welch, N.G., Scoble, J.A., Muir, B.W., Pigram, P.J.: Orientation and characterization of immobilized antibodies for improved immunoassays (Review). *Biointerphases* **12**, 02D301 (2017). <https://doi.org/10.1116/1.4978435>
49. Bulaj, G.: Formation of disulfide bonds in proteins and peptides. *Biotechnol. Adv.* **23**, 87–92 (2005). <https://doi.org/10.1016/J.BIOTECHADV.2004.09.002>
50. Limo, M.J., Sola-Rabada, A., Boix, E., Thota, V., Westcott, Z.C., Puddu, V., Perry, C.C.: Interactions between metal oxides and biomolecules: from fundamental understanding to applications. *Chem. Rev.* **118**, 11118–11193 (2018). https://doi.org/10.1021/ACS.CHEMREV.7B00660/ASSET/IMAGES/LARGE/CR-2017-006605_0023.JPEG
51. Chiodi, E., Marn, A.M., Geib, M.T., Selim Ünlü, M.: The role of surface chemistry in the efficacy of protein and DNA microarrays for label-free detection: an overview. *Polymers* **13**, 1026 (2021). <https://doi.org/10.3390/POLYM13071026>
52. Jain, P., Rauer, S.B., Möller, M., Singh, S.: Mimicking the natural basement membrane for advanced tissue engineering. *Biomacromol* **23**, 3081–3103 (2022). https://doi.org/10.1021/ACS.BIOMAC.2C00402/ASSET/IMAGES/LARGE/BM2C00402_0006.JPEG

53. Gharib, G., Bütün, İ., Munganlı, Z., Kozalak, G., Namlı, İ., Sarraf, S.S., Ahmadi, V.E., Toyran, E., van Wijnen, A.J., Koşar, A.: Biomedical applications of microfluidic devices: a review. *Biosensors (Basel)* **12** (2022). <https://doi.org/10.3390/BIOS12111023/S1>
54. Xia, J., Khaliliazar, S., Hamed, M.M., Sonkusale, S.: Thread-based wearable devices. *MRS Bull.* **46**, 502–511 (2021). <https://doi.org/10.1557/S43577-021-00116-1/FIGURES/7>
55. Oweyung, R.E., Terse-Thakoor, T., Rezaei Nejad, H., Panzer, M.J., Sonkusale, S.R.: Highly flexible transistor threads for all-thread based integrated circuits and multiplexed diagnostics. *ACS Appl. Mater. Interf.* **11**, 31096–31104 (2019). https://doi.org/10.1021/ACSAMI.9B09522/SUPPL_FILE/AM9B09522_SI_001.PDF
56. Mousavi, M.P.S., Ainla, A., Tan, E.K.W., Abd El-Rahman, M., Yoshida, Y., Yuan, L., Sigursslid, H.H., Arkan, N., Yip, M.C., Abrahamsson, C.K., Homer-Vanniasinkam, S., Whitesides, G.M.: Ion sensing with thread-based potentiometric electrodes. *Lab Chip.* **18**, 2279–2290 (2018). <https://doi.org/10.1039/C8LC00352A>
57. Ricci, F., Palleschi, G.: Sensor and biosensor preparation, optimisation and applications of Prussian blue modified electrodes. *Biosens. Bioelectron.* **21**, 389–407 (2005). <https://doi.org/10.1016/J.BIOS.2004.12.001>
58. Renedo, O.D., Alonso-Lomillo, M.A., Martínez, M.J.A.: Recent developments in the field of screen-printed electrodes and their related applications. *Talanta* **73**, 202–219 (2007). <https://doi.org/10.1016/J.TALANTA.2007.03.050>
59. Malzahn, K., Windmiller, J.R., Valdés-Ramírez, G., Schöning, M.J., Wang, J.: Wearable electrochemical sensors for in situ analysis in marine environments. *Analyst* **136**, 2912–2917 (2011). <https://doi.org/10.1039/C1AN15193B>
60. Malon, R.S.P., Chua, K.Y., Wicaksono, D.H.B., Córcoles, E.P.: Cotton fabric-based electrochemical device for lactate measurement in saliva. *Analyst* **139**, 3009–3016 (2014). <https://doi.org/10.1039/C4AN00201F>
61. Carrilho, E., Martinez, A.W., Whitesides, G.M.: Understanding wax printing: A simple micropatterning process for paper-based microfluidics. *Anal. Chem.* **81**, 7091–7095 (2009). https://doi.org/10.1021/AC901071P/SUPPL_FILE/AC901071P_SI_001.PDF
62. Kinnamon, D.S., Krishnan, S., Brosler, S., Sun, E., Prasad, S.: Screen printed graphene oxide textile biosensor for applications in inexpensive and wearable point-of-exposure detection of influenza for at-risk populations. *J. Electrochem. Soc.* **165**, B3084–B3090 (2018). <https://doi.org/10.1149/2.0131808JES/XML>
63. Nomura, K.I., Horii, Y., Kanazawa, S., Kusaka, Y., Ushijima, H.: Fabrication of a textile-based wearable blood leakage sensor using screen-offset printing. *Sensors* **18**, 240 (2018). <https://doi.org/10.3390/S18010240>
64. Kincal, D., Kumar, A., Child, A.D., Reynolds, J.R.: Conductivity switching in polypyrrole-coated textile fabrics as gas sensors. *Synth. Met.* **92**, 53–56 (1998)
65. Collins, G.E., Buckley, L.J.: Conductive polymer-coated fabrics for chemical sensing. *Synth. Met.* **78**, 93–101 (1996). [https://doi.org/10.1016/0379-6779\(96\)80108-1](https://doi.org/10.1016/0379-6779(96)80108-1)
66. Sekar, M., Pandiaraj, M., Bhansali, S., Ponpandian, N., Viswanathan, C.: Carbon fiber based electrochemical sensor for sweat cortisol measurement. *Sci. Reports* **9**(1), 1–14 (2019). <https://doi.org/10.1038/s41598-018-37243-w>
67. Parrilla, M., Ferrø, J., Guinovart, T., Andrade, F.J., Parrilla, J M, Ferrø, J., Guinovart, T., Andrade, F.J.: Wearable potentiometric sensors based on commercial carbon fibres for monitoring sodium in sweat. *Electroanalysis* **28**, 1267–1275 (2016). <https://doi.org/10.1002/ELAN.201600070>
68. Shim, B.S., Chen, W., Doty, C., Xu, C., Kotov, N.A.: Smart electronic yarns and wearable fabrics for human biomonitoring made by carbon nanotube coating with polyelectrolytes. *Nano Lett.* **8**, 4151–4157 (2008). <https://doi.org/10.1021/NL801495P>
69. Guinovart, T., Parrilla, M., Crespo, G.A., Rius, F.X., Andrade, F.J.: Potentiometric sensors using cotton yarns, carbon nanotubes and polymeric membranes. *Analyst* **138**, 5208–5215 (2013). <https://doi.org/10.1039/C3AN00710C>
70. Kang, M.A., Ji, S., Kim, S., Park, C.Y., Myung, S., Song, W., Lee, S.S., Lim, J., An, K.S.: Highly sensitive and wearable gas sensors consisting of chemically functionalized graphene

- oxide assembled on cotton yarn. *RSC Adv.* **8**, 11991–11996 (2018). <https://doi.org/10.1039/C8RA01184B>
71. Tang, J., Wei, L., He, S., Li, J., Nan, D., Ma, L., Shen, W., Kang, F., Lv, R., Huang, Z.: A highly sensitive electrochemical glucose sensor based on room temperature exfoliated graphite-derived film decorated with dendritic copper. *Materials* **14** (2021). <https://doi.org/10.3390/MA14175067/S1>
 72. Chen, C., Ran, R., Yang, Z., Lv, R., Shen, W., Kang, F., Huang, Z.H.: An efficient flexible electrochemical glucose sensor based on carbon nanotubes/carbonized silk fabrics decorated with Pt microspheres. *Sens Actuators B Chem.* **256**, 63–70 (2018). <https://doi.org/10.1016/J.SNB.2017.10.067>
 73. Wang, Y., Yang, B., Hua, Z., al, Wang, D., Liu, Z., Li, J.: Review-Flexible and Stretchable Electrochemical Sensing Systems: Materials, Energy Sources, and Integrations You may also like Recent advancements in flexible and wearable sensors for biomedical and healthcare applications Thin-film transistor arrays for biological sensing systems (2020). <https://doi.org/10.1149/1945-7111/ab7117>
 74. Liu, X., Lillehoj, P.B.: Embroidered biosensors on gauze for rapid electrochemical measurements. In: *Proceedings of the IEEE International Conference on Micro Electro Mechanical Systems (MEMS)*, pp. 377–380 (2017). <https://doi.org/10.1109/MEMSYS.2017.7863420>
 75. Liu, X., Lillehoj, P.B.: Embroidered electrochemical sensors for biomolecular detection. *Lab Chip* **16**, 2093–2098 (2016). <https://doi.org/10.1039/C6LC00307A>
 76. Alaimo, F., Sadeqi, A., Rezaei Nejad, H., Jiang, Y., Wang, W., Demarchi, D., Sonkusale, S.: Reel-to-reel fabrication of strain sensing threads and realization of smart insole. *Sens Actuators A Phys.* **301**, 111741 (2020). <https://doi.org/10.1016/J.SNA.2019.111741>
 77. Post, E.R., Orth, M., Russo, R.R., Gershenfeld, N.: E-broüder: Design and fabrication of textile-based computing. *IBM Syst. J.* **39**, 840–860 (2000). <https://doi.org/10.1147/SJ.393.0840>
 78. Sergio, M., Manaresi, N., Campi, F., Canegallo, R., Tartagni, M., Guerrieri, R.: A dynamically reconfigurable monolithic CMOS pressure sensor for smart fabric. *IEEE J. Solid-State Circuits* **38**, 966–975 (2003). <https://doi.org/10.1109/JSSC.2003.811977>
 79. Meyer, J., Arnrich, B., Schumm, J., Troster, G.: Design and modeling of a textile pressure sensor for sitting posture classification. *IEEE Sens. J.* **10**, 1391–1398 (2010). <https://doi.org/10.1109/JSEN.2009.2037330>
 80. Hoffmann, T., Eilebrecht, B., Leonhardt, S.: Respiratory monitoring system on the basis of capacitive textile force sensors. *IEEE Sens. J.* **11**, 1112–1119 (2011). <https://doi.org/10.1109/JSEN.2010.2082524>
 81. Holleczeck, T., Rüegg, A., Harms, H., Tröster, G.: Textile pressure sensors for sports applications. In: *Proceedings of IEEE Sensors*, pp. 732–737 (2010). <https://doi.org/10.1109/ICSENS.2010.5690041>
 82. Catrysse, M., Puers, R., Hertleer, C., Van Langenhove, L., Van Egmond, H., Matthys, D.: Towards the integration of textile sensors in a wireless monitoring suit. *Sens Actuators A Phys.* **114**, 302–311 (2004). <https://doi.org/10.1016/J.SNA.2003.10.071>
 83. Cho, G., Jeong, K., Paik, M.J., Kwun, Y., Sung, M.: Performance evaluation of textile-based electrodes and motion sensors for smart clothing. *IEEE Sens. J.* **11**, 3183–3193 (2011). <https://doi.org/10.1109/JSEN.2011.2167508>
 84. Zhang, H., Tao, X., Yu, T., Wang, S.: Conductive knitted fabric as large-strain gauge under high temperature. *Sens Actuators A Phys.* **126**, 129–140 (2006). <https://doi.org/10.1016/J.SNA.2005.10.026>
 85. Pacelli, M., Caldani, L., Paradiso, R.: Textile piezoresistive sensors for biomechanical variables monitoring. In: *Annual International Conference of the IEEE Engineering in Medicine and Biology—Proceedings*, pp. 5358–5361 (2006). <https://doi.org/10.1109/IEMBS.2006.259287>
 86. Huang, C.T., Shen, C.L., Tang, C.F., Chang, S.H.: A wearable yarn-based piezo-resistive sensor. *Sens Actuators A Phys.* **141**, 396–403 (2008). <https://doi.org/10.1016/J.SNA.2007.10.069>

87. Liu, M., Del-Rio-Ruiz, R., Sharma, A., Asci, C., Oweyung, R., Sonkusale, S.: Preliminary results on sensing pillow to monitor head movement using strain sensing threads. In: Proceedings of IEEE Sensors. 2022-October (2022). <https://doi.org/10.1109/SENSORSS2175.2022.9967140>
88. Pantelopoulos, A., Bourbakis, N.G.: A survey on wearable sensor-based systems for health monitoring and prognosis. *IEEE Trans. Syst. Man Cybern. Part C Appl. Rev.* **40**, 1–12 (2010). <https://doi.org/10.1109/TSMCC.2009.2032660>
89. Coyle, S., Lau, K.T., Moyna, N., O’Gorman, D., Diamond, D., Di Francesco, F., Costanzo, D., Salvo, P., Trivella, M.G., De Rossi, D.E., Taccini, N., Paradiso, R., Porchet, J.A., Ridolfi, A., Luprano, J., Chuzel, C., Lanier, T., Revol-Cavalier, F., Schoumacker, S., Mourier, V., Chartier, I., Convert, R., De-Moncuit, H., Bini, C.: BIOTEXBiosensing textiles for personalised health-care management. *IEEE Trans. Inf. Technol. Biomed.* **14**, 364–370 (2010). <https://doi.org/10.1109/TITB.2009.2038484>
90. Sibinski, M., Jakubowska, M., Sloma, M.: Flexible temperature sensors on fibers. *Sensors* **10**, 7934–7946 (2010). <https://doi.org/10.3390/S100907934>
91. Linz, T., Gourmelon, L., Langereis, G.: Contactless EMG sensors embroidered onto textile. *IFMBE Proc.* **13**, 29–34 (2007). https://doi.org/10.1007/978-3-540-70994-7_5/COVER
92. Coosemans, J., Hermans, B., Puers, R.: Integrating wireless ECG monitoring in textiles. *Sens Actuators A Phys.* **130–131**, 48–53 (2006). <https://doi.org/10.1016/J.SNA.2005.10.052>
93. Kim, C., Kim, J.K., Kim, K.T., Yoon, Y.K.: Micromachined wearable/foldable super wide-band (SWA) monopole antenna based on a flexible liquid crystal polymer (LCP) substrate toward imaging/sensing/health monitoring systems. In: Proceedings—Electronic Components and Technology Conference, pp. 1926–1932 (2013). <https://doi.org/10.1109/ECTC.2013.6575841>
94. Tronquo, A., Rogier, H., Hertleer, C., Van Langenhove, L.: Applying textile materials for the design of antennas for wireless body area networks. European Space Agency, (Special Publication) ESA SP. 626 SP (2006). <https://doi.org/10.1109/EUCAP.2006.4584573>
95. Kiourti, A., Volakis, J.L.: Wearable antennas using electronic textiles for RF communications and medical monitoring. In: 2016 10th European Conference on Antennas and Propagation, EuCAP 2016 (2016). <https://doi.org/10.1109/EUCAP.2016.7481222>
96. Ivsic, B., Bonefacic, D., Bartolic, J.: Considerations on embroidered textile antennas for wearable applications. *IEEE Antennas Wirel. Propag. Lett.* **12**, 1708–1711 (2013). <https://doi.org/10.1109/LAWP.2013.2297698>
97. Kiourti, A., Nikita, K.S.: A review of implantable patch antennas for biomedical telemetry: challenges and solutions. *IEEE Antennas Propag. Mag.* **54**, 210–228 (2012). <https://doi.org/10.1109/MAP.2012.6293992>
98. Choi, J.H., Kim, Y., Lee, K., Chung, Y.: Various wearable embroidery RFID tag antenna using electro-thread. In: 2008 IEEE International Symposium on Antennas and Propagation and USNC/URSI National Radio Science Meeting, APSURSI (2008). <https://doi.org/10.1109/APS.2008.4619730>
99. Amendola, S., Lodato, R., Manzari, S., Occhiuzzi, C., Marrocco, G.: RFID technology for IoT-based personal healthcare in smart spaces. *IEEE Internet Things J.* **1**, 144–152 (2014). <https://doi.org/10.1109/JIOT.2014.2313981>
100. Stoppa, M., Chiolerio, A.: Wearable electronics and smart textiles: a critical review. *Sensors (Basel)* **14**, 11957 (2014). <https://doi.org/10.3390/S140711957>
101. Krifa, M.: Electrically conductive textile materials—Application in flexible sensors and antennas. *Textiles* **1**, 239–257 (2021). <https://doi.org/10.3390/TEXTILES1020012>
102. Nurul, H.M.R., Soh, P.J., Malek, M.F.A., Vandenbosch, G.A.E.: Dual-band suspended-plate wearable textile antenna. *IEEE Antennas Wirel. Propag. Lett.* **12**, 583–586 (2013). <https://doi.org/10.1109/LAWP.2013.2259211>
103. Koski, K., Sydänheimo, L., Rahmat-Samii, Y., Ukkonen, L.: Fundamental characteristics of electro-textiles in wearable UHF RFID patch antennas for body-centric sensing systems. *IEEE Trans. Antennas Propag.* **62**, 6454–6462 (2014). <https://doi.org/10.1109/TAP.2014.2364071>

104. Abbas, B., Khamas, S.K., Ismail, A., Sali, A.: Full embroidery designed electro-textile wearable tag antenna for WBAN application. *Sensors* **19**, 2470 (2019). <https://doi.org/10.3390/S19112470>
105. Bayram, Y., Zhou, Y., Shim, B.S., Xu, S., Zhu, J., Kotov, N.A., Volakis, J.L.: E-textile conductors and polymer composites for conformal lightweight antennas. *IEEE Trans. Antennas Propag.* **58**, 2732–2736 (2010). <https://doi.org/10.1109/TAP.2010.2050439>
106. Guy, A.W., Chou, C.K., McDougall, J.A., Sorensen, C.: Measurement of shielding effectiveness of microwave-protective suits. *IEEE Trans. Microw. Theory Tech.* **35**, 984–994 (1987). <https://doi.org/10.1109/TMTT.1987.1133796>
107. Liu, G., Wang, Q.-A., Jiao, G., Dang, P., Nie, G., Liu, Z., Sun, J., Liu, G., Wang, Q.-A., Jiao, G., Dang, P., Nie, G., Liu, Z., Sun, J.: Review of wireless RFID strain sensing technology in structural health monitoring. *Sensors* **23**, 6925 (2023). <https://doi.org/10.3390/S23156925>
108. Chen, X., Ukkonen, L., Bjorninen, T.: Passive E-Textile UHF RFID-based wireless strain sensors with integrated references. *IEEE Sens. J.* **16**, 7835–7836 (2016). <https://doi.org/10.1109/JSEN.2016.2608659>
109. Kennedy, T.F., Fink, P.W., Chu, A.W., Champagne, N.J., Lin, G.Y., Khayat, M.A.: Body-worn E-textile antennas: the good, the low-mass, and the conformal. *IEEE Trans. Antennas Propag.* **57**, 910–918 (2009). <https://doi.org/10.1109/TAP.2009.2014602>
110. Liu, N., Lu, Y., Qiu, S., Li, P.: Electromagnetic properties of electro-textiles for wearable antennas applications. *Front. Electr. Electron. Eng. China* **6**, 563–566 (2011). <https://doi.org/10.1007/S11460-011-0182-7/METRICS>
111. Ouyang, Y., Chappell, W.J.: High frequency properties of electro-textiles for wearable antenna applications. *IEEE Trans. Antennas Propag.* **56**, 381–389 (2008). <https://doi.org/10.1109/TAP.2007.915435>
112. Corchia, L., Monti, G., Tarricone, L.: Wearable antennas: nontextile versus fully textile solutions. *IEEE Antennas Propag. Mag.* **61**, 71–83 (2019). <https://doi.org/10.1109/MAP.2019.2895665>
113. Amram Bengio, E., Senic, D., Taylor, L.W., Tsentalovich, D.E., Chen, P., Holloway, C.L., Babakhani, A., Long, C.J., Novotny, D.R., Booth, J.C., Orloff, N.D., Pasquali, M.: High efficiency carbon nanotube thread antennas. *Appl Phys Lett.* **111** (2017). <https://doi.org/10.1063/1.4991822>
114. Elwi, T.A., Al-Rizzo, H.M., Rucker, D.G., Dervishi, E., Li, Z., Biris, A.S.: Multi-walled carbon nanotube-based RF antennas. *Nanotechnology* **21**, 045301 (2009). <https://doi.org/10.1088/0957-4484/21/4/045301>
115. Puchades, I., Rossi, J.E., Cress, C.D., Naglich, E., Landi, B.J.: Carbon nanotube thin-film antennas. *ACS Appl. Mater. Interf.* **8**, 20986–20992 (2016). <https://doi.org/10.1021/ACSAMI.6B05146>
116. Keller, S.D., Zaghoul, A.I., Shanov, V., Schulz, M.J., Mast, D.B., Alvarez, N.T.: Radiation performance of polarization selective carbon nanotube sheet patch antennas. *IEEE Trans. Antennas Propag.* **62**, 48–55 (2014). <https://doi.org/10.1109/TAP.2013.2287272>
117. Burke, P.J., Li, S., Yu, Z.: Quantitative theory of nanowire and nanotube antenna performance. *IEEE Trans. Nanotechnol.* **5**, 314–334 (2006). <https://doi.org/10.1109/TNANO.2006.877430>
118. Plombon, J.J., O'Brien, K.P., Gstrein, F., Dubin, V.M., Jiao, Y.: High-frequency electrical properties of individual and bundled carbon nanotubes. *Appl. Phys. Lett.* **90**, 8 (2007). <https://doi.org/10.1063/1.2437724/327159>
119. Keller, S.D., Zaghoul, A.I., Shanov, V., Schulz, M.J., Mast, D.B.: Design considerations for a meshed carbon nanotube thread patch antenna. *IEEE Antennas Wirel. Propag. Lett.* **12**, 1192–1195 (2013). <https://doi.org/10.1109/LAWP.2013.2282176>
120. Jayasinghe, C., Chakrabarti, S., Schulz, M.J., Shanov, V.: Spinning yarn from long carbon nanotube arrays. *J. Mater. Res.* **26**, 645–651 (2011). <https://doi.org/10.1557/JMR.2010.91>
121. Oweyung, R., Zeng, W., Sonkusale, S.: Eutectogel electrodes for long-term biosignal monitoring. In: *Proceedings of IEEE Sensors*, October (2022). <https://doi.org/10.1109/SENSOR552175.2022.9967321>

122. Jiang, Y., Sadeqi, A., Miller, E.L., Sonkusale, S.: Head motion classification using thread-based sensor and machine learning algorithm. *Sci. Reports* **11**(1), 1–11 (2021). <https://doi.org/10.1038/s41598-021-81284-7>
123. Oweyung, R.E., Panzer, M.J., Sonkusale, S.R.: Colorimetric gas sensing washable threads for smart textiles. *Sci. Reports* **9**(1), 1–8 (2019). <https://doi.org/10.1038/s41598-019-42054-8>
124. Liu, Y., Wang, H., Zhao, W., Zhang, M., Qin, H., Xie, Y.: Flexible, Stretchable sensors for wearable health monitoring: sensing mechanisms, materials, fabrication strategies and features. *Sensors* **18**, 645 (2018). <https://doi.org/10.3390/S18020645>
125. Zhu, G.J., Ren, P.G., Guo, H., Jin, Y.L., Yan, D.X., Li, Z.M.: Highly sensitive and stretchable polyurethane fiber strain sensors with embedded silver nanowires. *ACS Appl. Mater. Interf.* **11**, 23649–23658 (2019). https://doi.org/10.1021/ACSAMI.9B08611/SUPPL_FILE/AM9B08611_SI_008.MP4
126. He, Z., Wang, K., Zhao, Z., Zhang, T., Li, Y., Wang, L.: A Wearable flexible acceleration sensor for monitoring human motion. *Biosensors* **12**, 620 (2022). <https://doi.org/10.3390/BIOS12080620>
127. Alrammouz, R., Podlecki, J., Abboud, P., Sorli, B., Habchi, R.: A review on flexible gas sensors: from materials to devices. *Sens Actuators A Phys.* **284**, 209–231 (2018). <https://doi.org/10.1016/j.sna.2018.10.036>
128. Somappa, L., Malik, S., Aeron, S., Sonkusale, S., Baghini, M.S.: High resolution frequency measurement techniques for relaxation oscillator based capacitive sensors. *IEEE Sens. J.* **21**, 13394–13404 (2021). <https://doi.org/10.1109/JSEN.2021.3068351>