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Electronic Textile Sensors for Decoding Vital Body Signals: State-of-the-Art Review on Characterizations and Recommendations

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The digitization of textiles (textronics) has created new opportunities for integration with conformable sensors to enable unobtrusive, noninvasive, and continuous decoding of vital body signals. This article provides an in-depth review of the materials and fabrication methodologies used for textronic sensors per their form-factor in the textile manufacturing process chain—fiber, yarn, fabric, and apparel. Next, it analyzes the performance characterization techniques currently used for these sensors and highlights the needs for standardized test methods in the following aspects: biocompatibility, thermal and tactile comfort, aging, and operation of the biomedical sensing modality at standard human stretch. It also identifies the significance of pretreatment and conditioning reporting of the textile form-factors based on their impact on mechanical and electric performance of the textronic sensor. The study concludes by recommending a universal testing roadmap for textronic sensors which is expected to veritably complement the work of different standardization committees, including CEN TC-248/WG-31, IEC TC-124, ASTM D13.50, and AATCC RA111.

1. Introduction

The wearable biosensor market has seen an unprecedented growth in recent years, and is forecasted to reach over $5 billion per year by 2025.[1] Electronic textile sensors are one of the major players in this category, mostly due to their intrinsic flexible structural geometries. Supporting this, electronic industries are consolidating R&D efforts around devices offering the highest value proposition among the wearable sensor devices. Recently, the US Department of Defense (DoD) partnered with MIT to create the $300 million nonprofit institute Advanced Functional Fabrics of America (AFFOA) to develop different textile electronics, aka textronics, including physiological sensors for military soldiers and civilians conducting multidomain operations.[2,3] Tech giants like Google, Samsung, Microsoft, and Apple have already joined forces in this race to develop functional fabrics and textile sensors.[4–6] Social media and networking company Facebook has also launched Facebook Reality Labs (FRL) to develop wearable garment interfaces for interaction between the human body and virtual/augmented reality (VR/AR).[7]

Textiles, often referred to as the "second skin" of the human body, comprise an industry that can be estimated at being 27 000 years old;[8] however, the textiles manufactured in the 21st century afford a high degree of sophistication and maturity in terms of material choices, automation, and cost optimization. Scientists and researchers have leveraged this textile platform to integrate conductive elements and silicone-based electronics to fabricate a diverse range of biomedical textile sensors. Consequently, these textile sensors have been viewed as promising alternatives to expensive analytical instruments used in sports medicine or biomedical industries for the monitoring of physiological profiles. This article discusses biosensing of vital signals and emergent surface dynamics such as heart/pulse rate, respiration rate, body temperatures, blood pressure, motion, and neuromuscular response.[9–15] Monitoring of biomarkers such as glucose, hemoglobin, and blood oxygen requires a significantly different set of sensing modalities, which are often invasive and require strict medical regulation. **Figure 1** summarizes the plurality of biosensing modalities at various locations of the body and garment shape-factor (sock, headband, shirt, etc.) associated with each location. The differential drapability and pliability of textiles provide a range of good conformal fits to the body which enable effective electrical signal capture with reduced noise.[16]

The promise of the textile platform is further intensified through the diverse range of flexible form factors available (fiber, yarn, fabric) for cointegration with electronics using the traditional textile manufacturing technologies as shown in the visual summary in **Figure 2**. Leveraging the framework of the existing textile process chain and building upon current manufacturing methods and test practices for commercial textiles would greatly streamline development efforts of the multifaceted research community

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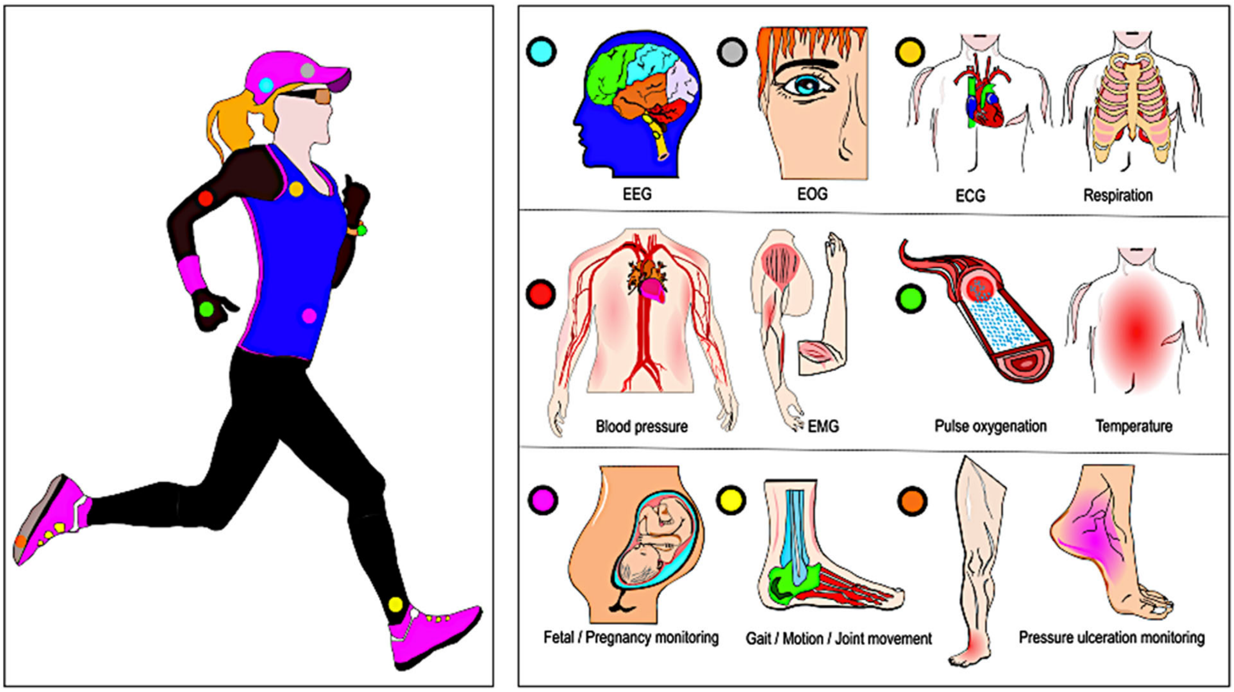

Figure 1. Locations of physiological textile biosensors for registering different vital signs of the body. As biosignals like body temperature and pulse oxygenation can be extracted from wrist (green dot), they are clustered together. Also, biosignals like electrocardiogram (ECG) and respiration rate can be extracted and detected from the chest (light orange dot). Bioelectronic measurements like surface electromyogram (sEMG) and blood pressure measurements can be performed at the arm; thus, they are clustered together at the arm (red dot). Similarly, electroencephalography (EEG) and electrooculography (EOG) biosignals are collected by mounting textile-based sensors on a headcap and a headband (preferably at the forehead location), respectively; therefore, these biosignals are placed on the head (blue dot) and forehead (gray dot). Gait (or motion analysis) and joint movements can be detected from the ankle (yellow dot). Microclimate temperature for pressure ulcer diabetic patients can be detected by integrating sensors into a sock worn over the foot (dark orange dot). A plurality of textile sensors could be integrated with maternal apparel for monitoring fetal electrical vital signals (pink dot).

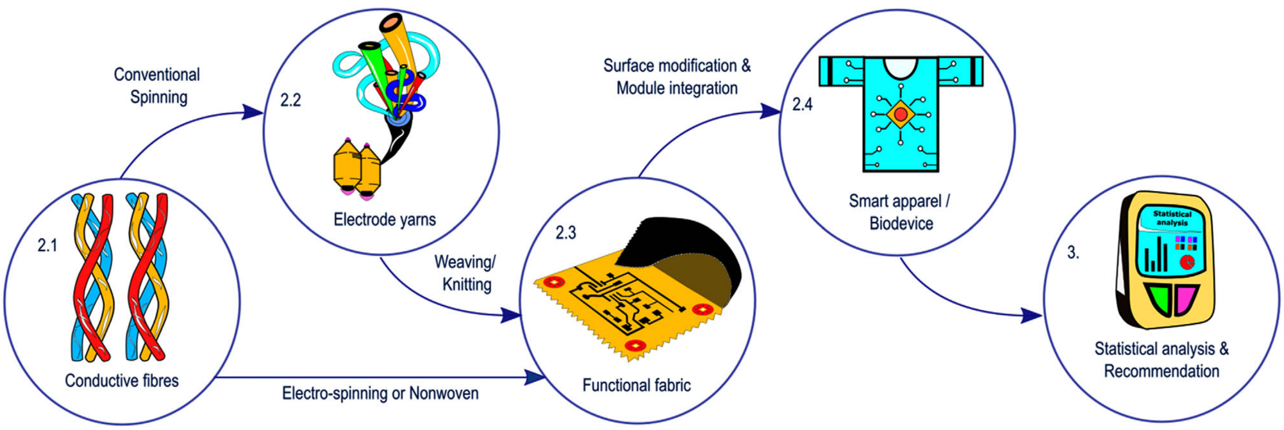

Figure 2. A graphical overview of the system flow for designing a physiological textile sensor system (from fiber to apparel). Different textile form factors with their corresponding fabrication and physiological sensing applications are discussed in Section 2.1–2.4 of this study. Section 2.1 presents various fibrous polymers and conductive elements used to produce electrode yarns and nonwoven fabrics using different spinning and other fabrication technologies. Section 2.2 discusses the biosensors designed with various industrial machinery involved in knitting and weaving operations. Optical filaments used for fiber Bragg grating sensors are also detailed in this section. Section 2.3 reports various coating and surface modification strategies to develop textile fabric-based sensors discussed in the literature. Section 2.4 introduces the process flow for biosignal acquisition using the smart electronic apparel or garment platform. Finally, Section 3 marks our statistical report on the characterization methodologies reported in the reviewed literature and our recommended benchmark for reporting the test conditions and performance parameters of any textile electronic system, including textile-based sensors.

(material scientists and textile, electrical, electronic, and mechanical engineers) associated with textronic sensor development.

A number of comprehensive reviews on textile-based health sensors have been published in recent years. The historical review by Hughes-Riley et al.[8] maps the evolution of smart textiles over the past century in terms of first-, second-, and third-generation e-textiles and presents the broad categories of their application including wearable computing, sensing, energy harvesting, and power transmission. The 2019 and 2020 reviews on flexible activity and electrophysiological sensors by Khair et al.,[17] Heo et al.,[18] and Li et al.[19] provide detailed commentary on the challenges associated with various fabrication technologies and material selection for mainly carbon-based textile sensors. The current review comes in a timely fashion to build upon the previous work with a bifocal agenda. First, it focuses exclusively on textile-based sensors for health monitoring and presents the state-of-the-art fabrication technologies based on the stage at which electrical functionality is imparted to the passive textiles, viz., fibrous polymers (Section 2.1), yarn/filament (Section 2.2), fabric (knit, woven, nonwoven) (Section 2.3), and apparel (Section 2.4). Next, our review focuses on the current sensor characterization techniques used for the textronic sensors described in the reviewed literature and identifies those being overlooked (Section 3).

A recent investigation[20] reported the need for standardized performance characterization techniques as a major barrier that smart textiles face for a successful market entry in the healthcare market. A few other studies have also identified the need for a shift in focus to the development of techniques for the practical applications and characterization of textronic sensors.[21–23] To date, however, there are no studies that have statistically shown the current state-of-the-art pertaining to the characterization needs for textile sensors in particular. Prolifically characterizing biomedical textile-based sensors requires a roadmap with a much-needed air of gravitas, something comprehensible for both textile industries and electronic sensor researchers—a convergent thinking to guide the emergence of single answer/solution, which has not been investigated in any previous studies. Our article elucidated such issue-specific barriers through an exploratory analysis and presented, for the first time, a clear and feasible roadmap to lay the groundwork for future challenge studies in this exciting area of textile electronics.

We emphasize the importance of recognizing the dual identity of these devices; both as a biomedical sensor and as a textile. From the perspective of a textile, Section 4.1 compares the characterization techniques used with the test methods and industry standards used by North American retailers for commercial apparel and provides a comprehensive breakdown of the six major categories that remain to be addressed by e-textile researchers. Section 4.2 in turn evaluates the techniques used to determine the performance of the device as a biomedical sensor and discusses the challenges that arise when characterizing transducer performance on a garment in vivo. The authors conclude by capturing the insights in the form of a bubble diagram that lays out the main considerations and interdependencies that remain to be addressed over the development life cycle of textronic sensors.

While the recommendations do not presume a prescriptive stance, the goal of the article remains to equip all participating researchers with an understanding and formal vocabulary of the textile process chain and its quality control, and a structured list of the main challenges that confront electronic-textile integration in the development of wearable health sensors. The authors believe that a holistic approach that invites the priorities of both the academic flexible-sensor research community and the textile industry will veritably complement the efforts of different standardization committees, including ASTM, AAMI, AATCC, CEN, and IEC to accelerate the development, production, and emergence of e-textiles in the wearable biosensor market.

2. Form Factors and Fabrications of Textile Sensors

2.1. Smart Fibers

The fibrous polymer can be considered the first stand-alone production unit and the fundamental building block for textile platforms. Integrating electronics at this level is desirable as it allows smart functionality to propagate upward through the textile process chain, supporting the development of increasingly complex applications at the apparel level.[24] However, the thin and long form-factor of the fiber poses a challenge from the standpoint of integrating existing electronic sensors into the fiber. This section discusses the primary raw materials and fabrication technologies for production, followed by a review of the sensing modalities that have been successfully realized for smart fibers.

Staple fiber and filaments are two dominant form factors for producing textile-based sensors, which can be produced by employing naturally conductive metals, intrinsically conductive polymers (ICPs), and carbonaceous materials (CMs). Composites of these materials are often used to produce structures that conform with the stretchability of textiles and remain electrically active under mechanical deformation. Composite biosensors can be produced by using conductive nano- or microscale particles and elastomeric fibers.[25] Surface coating and spinning technologies are among the major techniques for producing conductive fibers for smart electronic textile applications.[26,27]

2.1.1. Coating

Textile fibers coated with particular metals, aluminum for instance,[28] are very popular for wearable electronics due to their resultant high electrical conductivity, excellent mechanical endurance, and low cost. The inherent antimicrobial properties of metallic silver (Ag) particles used in textile industries have already gained widespread attention. In 2017, for the first time, coating of microporous dielectric thin film on conductive Ag-coated fiber was demonstrated to highly sensitive fabricate capacitive pressure sensors (sensitivity: 0.278 ($\Delta C/C$) kPa^{-1}, 0–2 kPa) for noninvasive health monitoring systems.[29] The sensor was fabricated with elastomeric microporous polydimethylsiloxane (PDMS) dielectric-encapsulated silver nanoparticle (AgNP)-coated Twaron fibers. Dip coating of fibers by CMs is another fast and facile technique to produce biosensors for smart electronic textiles. For instance, by dipping cotton fibers into a graphene oxide (GO) aqueous solution followed by a reduction process, rGO-cotton pressure sensors with comparable sensitivity (0.21 ($\Delta R/R$) kPa^{-1}, 0–2 kPa) could be fabricated

(**Figure 3**A-C).[30] The sensor design process involves sandwiching the rGO-cotton between two electrodes of copper tapes, fixed by silver adhesives. Textile pressure sensors exhibit promising features for detecting pulse, muscle movements, and speech recognition. However, the dipping technology to produce wearable pressure sensors requires multilevel handling.[31] Moreover, this immersion method can be prone to producing an irregular conductive surface (as characterized through SEM imaging in the study by Li et al.[30] above), thereby degrading the overall performance of the device.

2.1.2. Electrospinning

To produce an improved homogeneous conductive surface, electrospinning technologies can be employed in designing nanostructured pressure sensors. Electrospinning is a popular technology to design nanofiber (NF)-based sensor mats, which affords a high surface-area-to-volume ratio.[32] In 2018, the first NF-based 3D membrane ultrasensitive pressure sensor (13.5 $(\Delta I/I)$ kPa^{-1}, 0–10 kPa) was introduced using electrospinning technology (Figure 3D-F).[33] Conductive core/shell NF polymers comprising poly (vinylidene fluoride-*co*-hexafluoropropene) (PVDF-HFP)/poly(3,4-ethylenedioxythiophene) (PEDOT) and poly(3,4-ethylenedioxythiophene polystyrene sulfonate (PEDOT:PSS) (an ICP) inkjet-printed polyester (PET) electrodes were used as the raw materials. Interestingly, the method simultaneously builds the 3D structure of PVDF-HFP/PEDOT spherical bumps and initiates the polymerization process. Besides large-area pressure mapping, the demonstrated technology represents a new route for fabricating polymer-based nonwoven sensors for wireless blood pressure monitoring wristbands.

In addition to pressure sensors, different kinds of NF-based acoustic, resistive, photoelectric, optical, and amperometric sensors can be fabricated using electrospinning technology.[32] Researchers have developed nylon-66 fiber and graphite nanosheet-based multifunctional sensor mats capable of detecting ECG, strain (gauge factor [GF] ≈1), temperature, and gas.[34] A recent report demonstrated the possibility of a body motion sensor (sensitivity 0.13 mV kPa^{-1}) using electrospinning δ'-phase piezoelectric nylon-11 fibers.[35] The technology also provides a platform to use CMs as well. For example, carbon nanotubes (CNTs) have been used to fabricate a multimodal stretchable textile sensor (GF 54.9, 0–1% strain) using elastomeric polyurethane (PU) fibers.[36] Additionally, the technology

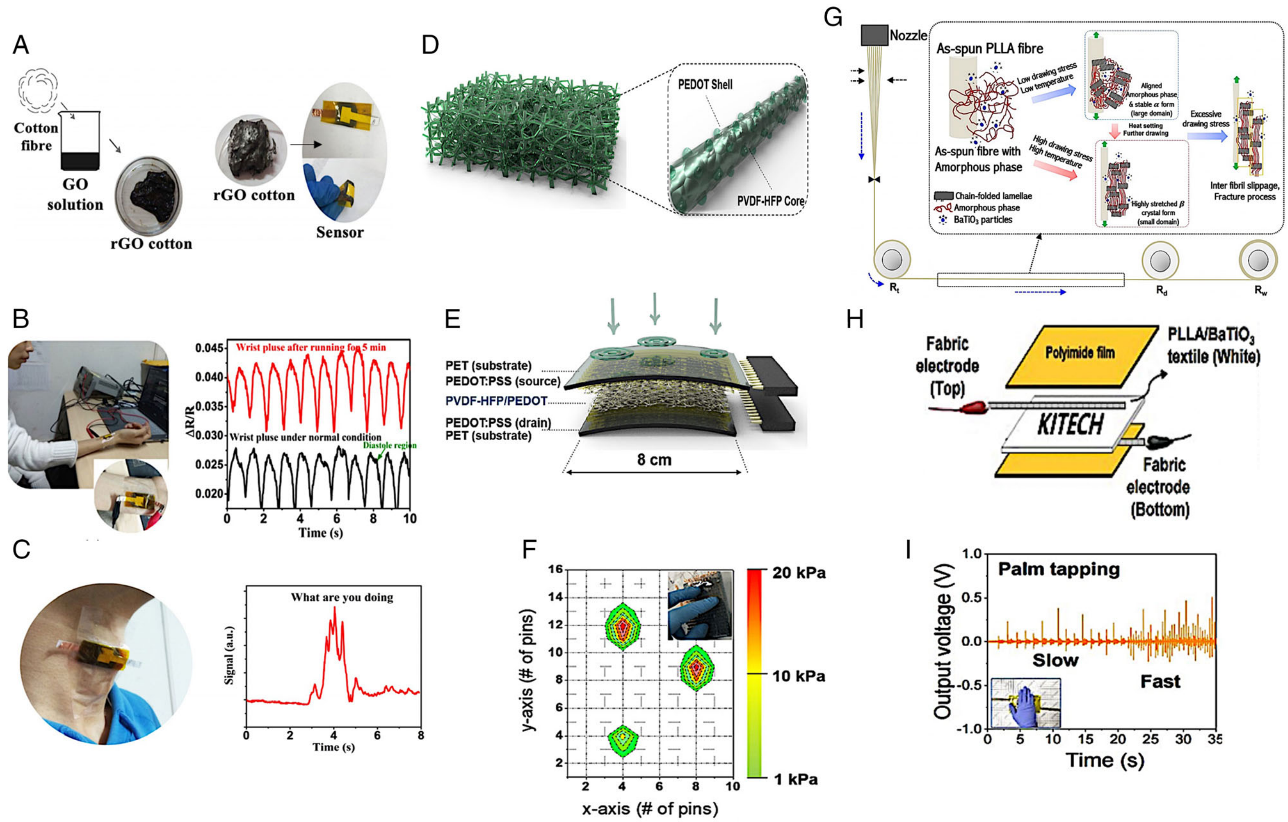

Figure 3. Textile fiber-based sensors for biosignal monitoring: A) dip coating of cotton fibers soaked in GO solution to produce rGO-cotton pressure sensor. B) Test subject's wrist pulse under normal condition and after running for 5 min (≈1.2 km); C) speech recognition response curve when the test subject spoke "I am a student"; (A-C) Reproduced with permission.[30] Copyright 2019, Springer Nature. D) electrospun 3D nanostructured PVDF-HFP/PEDOT NF mats. E) Sensor matrix for human–machine interfaces using PVDF-HFP/PEDOT 3D NF mats; F) spatial pressure mapping of a PVDF-HFP/PEDOT array sensor; (D-F) Reproduced with permission.[33] Copyright 2018, Springer Nature. G) melt spinning drawing process of forming β-crystal in PLLA/$BaTiO_3$ fibrous polymer. H) Diagram of the device electrodes jacketed between two polyimide (PI) films; I) output voltage of the piezoelectric PLLA/$BaTiO_3$ textiles upon applied force. (G-I) Reproduced with permission.[27] Copyright 2020, Springer Nature.

provides engineers with the flexibility to use a different form factor of functional fibers for designing flexible sensors, including fiber polymer blends, core–shell fibers, and hollow fibers.[37] A fibrous nonwoven mat can be further cut into small strips and twisted to produce super-stretchable electrode yarns for textile sensors. Gao et al. pursued this method to produce a CNT-based ultrastretchable helical electrode yarn for a human motion strain sensor (GF: 0.05).[38] Similarly, the 3D architecture of the electrospun nonwoven mat could be also tailored into a thin ribbon-like flexible free-standing film for sensory applications. Such an approach was introduced by Persano et al. to design piezoelectric poly(vinylidene fluoride-*co*-tri-fluoroethylene) [P(VDF-TrFE)]-based textile ribbon for accelerometers and orientation sensors.[39] Later on, Baniasadi et al. adopted this methodology to develop [P(VDF-TrFE)]-based electrospun ribbons and produced super-stretchable, up to ≈740%, piezoelectric textile yarns out of the ribbons (by twisting).[40] Another recent contribution by Hsu et al. demonstrates using such electrospun piezoelectric elastic yarns for a muscle patch sensor (0.133 pA% strain^{-1}) and a smart chest band to monitor respiration and heartbeats.[41]

2.1.3. Melt Spinning

Melt spinning (MS), another dominant textile spinning technology, is the most economical process for producing filaments or yarns from thermoplastic polymers, such as PET, nylon, or polypropylene.[42] In an MS machine, dry polymer pellets/granules are fed into the melting zone using an electric motor that guarantees a constant supply of the pellets (**Figure 4**). Next, the polymers are melted in the melting chamber and passed through a series of filters. The filtrated melt is then extruded through a spinneret in the form of filaments—a special metal die plate of small capillary holes of defined size and shape for the designed filament cross section. They are then converted back to their solid state by quenching with cool air and sent for drawing before packaging. This MS spinning technology is also the economic choice for industrial production of electroconductive fibers/yarns for CMs or conjugated polymers.[43] In 2018, University of Bolton (UK) exhibited the continuous MS process of producing functional yarns from piezoelectric polymer fibers like polyvinylidene difluoride (PVDF).[44] The cross-sectional geometry of the melt-spun filaments is, however, controlled by the dies, which can give various shapes, including circular or trilobal cross section, to the polymer melt.[45] For example, the typical cross sections of bi- or multicomponent melt-spun filaments for e-textile could include eccentric core-sheath, 50/50 side-by-side, unequal side-by-side, segmented pie or islands in the sea.[46–48] Using the MS technology, Åkerfeldt et al. were marked to be the first to use bicomponent fiber yarns to develop a textile strain sensor (sensitivity: 12 V% strain^{-1}) based on a β-crystalline PVDF fiber sheath and conductive high density polyethylene (HDPE)/carbon black (CB) core.[49] Another recent work at the Centre Européen des Textiles Innovants (CETI) in France applied MS technology to develop tricomponent piezoelectric multifilament yarns using polyethylene high-density (PEHD), polyamide (PA 12), and high β-phase content (97%) piezoelectric PVDF

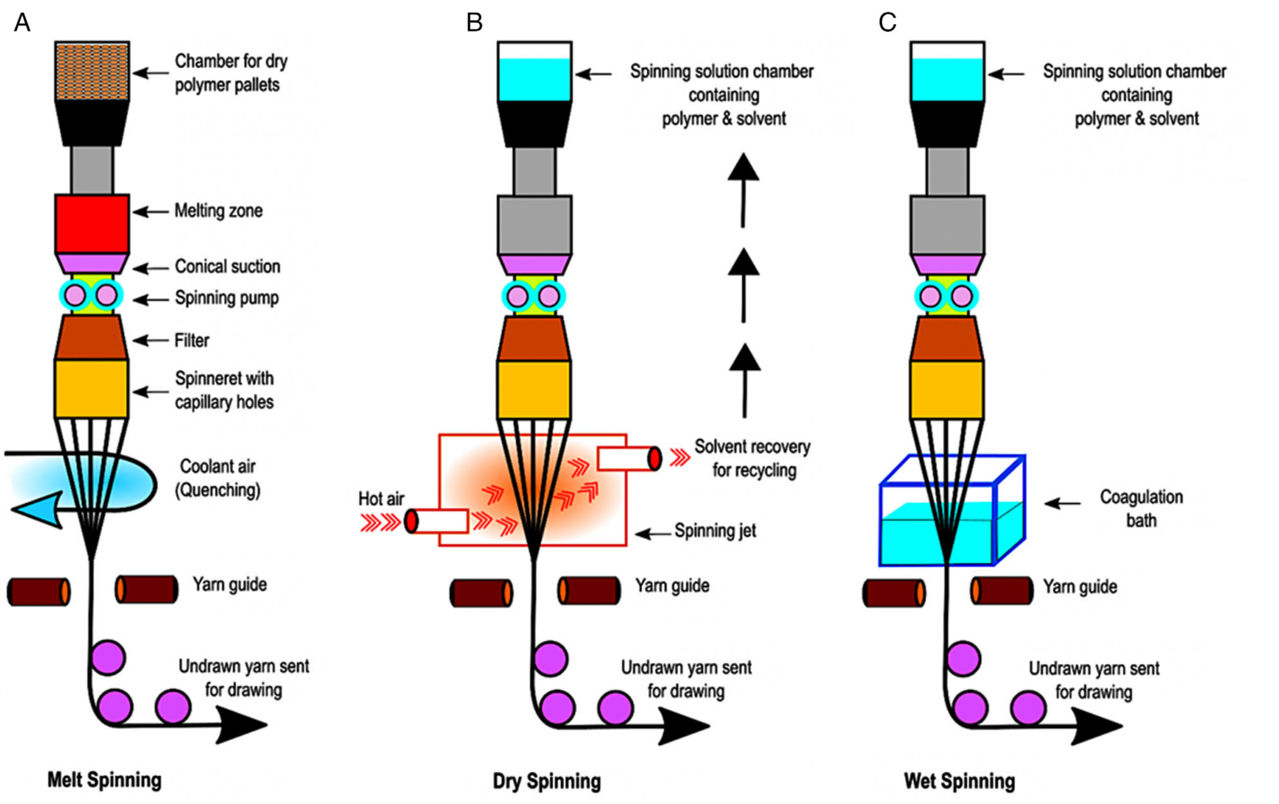

Figure 4. Schematic diagrams of A) melt, B) dry, and C) wet spinning technologies.

fibers.[50] Meanwhile, in Germany, an investigation is underway to conceptualize a capacitive multicomponent monofilament structure using MS for a novel textile actuator and sensor networks.[51] Nilsson et al. used melt-spun PVDF fibers to fabricate a textile sensor (voltage–strain ratio of $80\,\mathrm{VN}^{-1}$) to register human heartbeat and respiration.[52] Nowadays, conventional piezoelectric ceramics and polymers are being substituted by poly(L-lactic acid) (PLLA)-based piezoelectric polymers for their biodegradability, nontoxicity, low price, and nonpyroelectric nature.[27] A group of Korean scientists applied MS technology to manufacture filaments for textile sensors using PLLA/$BaTiO_3$ fibers (Figure 3G-I)[27] and also introduced a methodology to improve the piezoelectric properties by increasing the proportion of β-crystalline phase.

Macroscopic preform-to-fiber thermal drawing, analogous to the operating principle of MS technology, is another promising avenue to integrate electroactive modulated components with textile fibers to produce functional multimaterials. MIT, to exemplify, developed a multimaterial piezoelectric transducer using poly(carbonate) (PC) cladding, PVDF, and carbon-loaded poly (carbonate) (CVC)/indium electrodes.[53] A contemporary progress report also marked the recent growth of multimaterials for e-textiles applications.[54]

2.1.4. Dry and Wet Spinning

Besides MS, textile industries also employ dry spinning (DS) and wet spinning (WS) techniques. Unlike the MS, DS and WS technologies are comparatively expensive as the former requires an additional evaporating chamber and the latter a solvent/coagulation bath. Through the evaporating chamber of DS, hot air—of a temperature higher than the boiling point of the solvent—is directed toward the extruded filaments. The polymeric solution solidifies, while the solvent evaporates and is collected in a recycling chamber (Figure 4). Similarly, during WS technique, the solvent is recovered from the coagulation bath while the filaments are pumped directly into the bath to coagulate and solidify (Figure 4). The DS and WS technologies are far more complex compared to the MS technology, owing on one hand to the necessity of selecting the right solvent for dissolving the polymers, and on the other to the necessity of extracting the solvent out of the filament cores prior to the drawing and packaging stages. However, researchers have developed feasible DS and WS techniques to produce physiological textile strain sensors (GF $\approx$1) using rGO-based and PEDOT:PSS-based elastomeric fibers, respectively.[55,56] A recent collaboration between USA and Chinese researchers has reported the successful continuous synthesis process of conductive, stretchable, and self-healing hydrogel fibers for strain sensors (GF = 1) by utilizing the dry–wet spinning technologies.[57] Dehydration is a major drawback of hydrogel applications in textile electronics. The researchers applied poly(methyl acrylate) (PMA) coating layer to confine water in the core fiber, so as to increase the water retention by 25%. **Table 1** and Figure 4 provide a basic outlook on the pros and cons among melt, dry, and wet spinning technologies.

In these spinning production technologies, a metering/spinning pump maintains a continuous and constant flow rate of polymer solutions to the spinneret for extruding the filaments.

Table 1. Basic comparison among textile melt, dry, and wet spinning technologies.

Parameters	Melt spinning	Dry spinning	Wet spinning
Polymer form factor	Staple and filament	Mostly filament	Staple and filament
Investment cost	Low	High	Highest
Production	High	Low	High
Solvent used	Not required	Both organic and inorganic	Volatile organic solvent
Environmental hazard	Nontoxic	Toxic	Toxic
Heat of spinning	High	Very high	Low
Typical spinning[a)]	2500–3000 ft min	2500–3000 ft min	150–300 ft min^{-1}
Spinneret slits[a)]	Many thousands	≈1000	20 000–75 000
Typical application	PET, nylon	PAN	PVA
Filament solidified by	Coolant air	Hot air/vapor	Extraction of solvent

[a)]As per data collected in 2011.

However, the extruded and solidified filaments lack adequate mechanical strength for direct industrial applications.[58] Thus, an intermediary mechanical drawing process aligns the molecules of the filaments along the yarn axis to strengthen the mechanical properties before they are spun into suitable yarn packages, e.g., bobbins or cans. However, it must be noted that the MS method is applicable only for polymers entailing a stable melt phase (i.e., having a melting temperature)[59] that can undergo a simple transformation of physical state in the coolant chamber. Conversely, polymers that do not exhibit a stable melt phase, and are vulnerable to thermal degradation at a temperature lower than their melting temperature (e.g., PAN, cellulose fibers), are dissolved in a solvent in variable concentrations to produce filaments using the solution spinning techniques (DS and WS). If the polymer solvents are of sufficiently high volatility that the filament could be formed by solvent evaporation, then DS technique is employed; whereas for polymers with low volatility, WS technique is applied to coagulate the fiber and extract the solvent using the coagulation bath. DS technologies are complex and more expensive compared to the MS processes because the production process is a function of solvent concentration and thermal energy. In the WS process, fluids of higher viscosity are used by putting high tension on the extruded filaments, and forcing the overall spinning speed to set at a low rate,[60] thus increasing the production cost as a function of operation time (as shown in Table 1). As a consequence, the DS and WS technologies are used predominantly for higher end applications to obtain superior unique properties and justify their higher production cost. Examples include the production of acrylics for high-performance carbon fiber processing, Kevlar, and Lycra.[61]

2.2. Smart Yarns and Filaments

Recent growth in wearable technologies has enabled the use of conductive yarns for biomedical textile computing, as the electrode yarns can easily be incorporated into fabrics in a noninvasive manner. Extrinsic conductive yarns or filaments for textile sensors are produced mainly by two major technologies: 1) diverse coating techniques, including electroplating, dip-and-dry, and various dyeing methods; and 2) twisting conductive filaments with nonconductive yarn/filaments. Knitting machines are very popular to embed these coated or twisted electrode yarns into the fabric to make textile sensors.[62] Weaving, stitching, adhesive coating, resin/polymer encapsulation techniques are also used to integrate optoelectronics with electrode yarns to fabricate physiological sensors. Embroidery and sewing machines offer a faster process to embed conductive filaments at a lower cost with high repeatability of geometries. It must be noted that, as 100% intrinsic conductive yarns are made from conductive fibers,[63] we have categorized them under the textile fiber section, discussed earlier.

2.2.1. Electroplating and Suffusion

Employing the electroplating and carbon suffusion technologies, Canadian researchers recently developed two different wash-durable ECG sensors of Ag- and carbon-coated nylon electrode yarn.[64] The ECG yarn electrodes demonstrated a performance comparable to the traditional gold-standard hydrogel electrodes. The electrode yarns and signal transmission lines (**Figure** 5A,B) to measure ECG biosignals were knitted onto the fabric using Colosio Magica 28-gauge circular and Stoll CMS-ADF 18-gauge flatbed knitting machines. Researchers also utilized electroplated conductive yarns to fabricate sEMG textile sensors. For instance, Lee et al. used a knitting technique to incorporate 16 channels with 32 bipolar electrodes of 99% pure silver-plated nylon yarns on a stretchable band for sEMG signal acquisition (Figure 5C-E).[65] The researchers employed an Intarsia flat-bed knitting machine to design this sEMG fabric sensor for lower limbs of the human body for controlling myoelectric hand prosthesis. Similarly, for monitoring the activation of quadriceps and hamstrings, a sEMG athletic pant was recently developed by embedding a polyester fabric patch stitched with stainless steel yarns.[66]

2.2.2. Twisting and Embroidering

Using a DirectTwist 2B multifunctional twisting machine, Babusiak et al. twisted Shieldex silver fibrous yarn around a polyamide filament to make a bicomponent electroconductive yarn for capacitive sensors to measure ECG (Figure 5F-H).[67] A single jersey circular knitting machine was used successfully to integrate the ECG electrodes into a flexible fabric that demonstrated few advantages over the conventional gel-based Ag/AgCl electrodes. For instance, the textile electrodes did not cause any

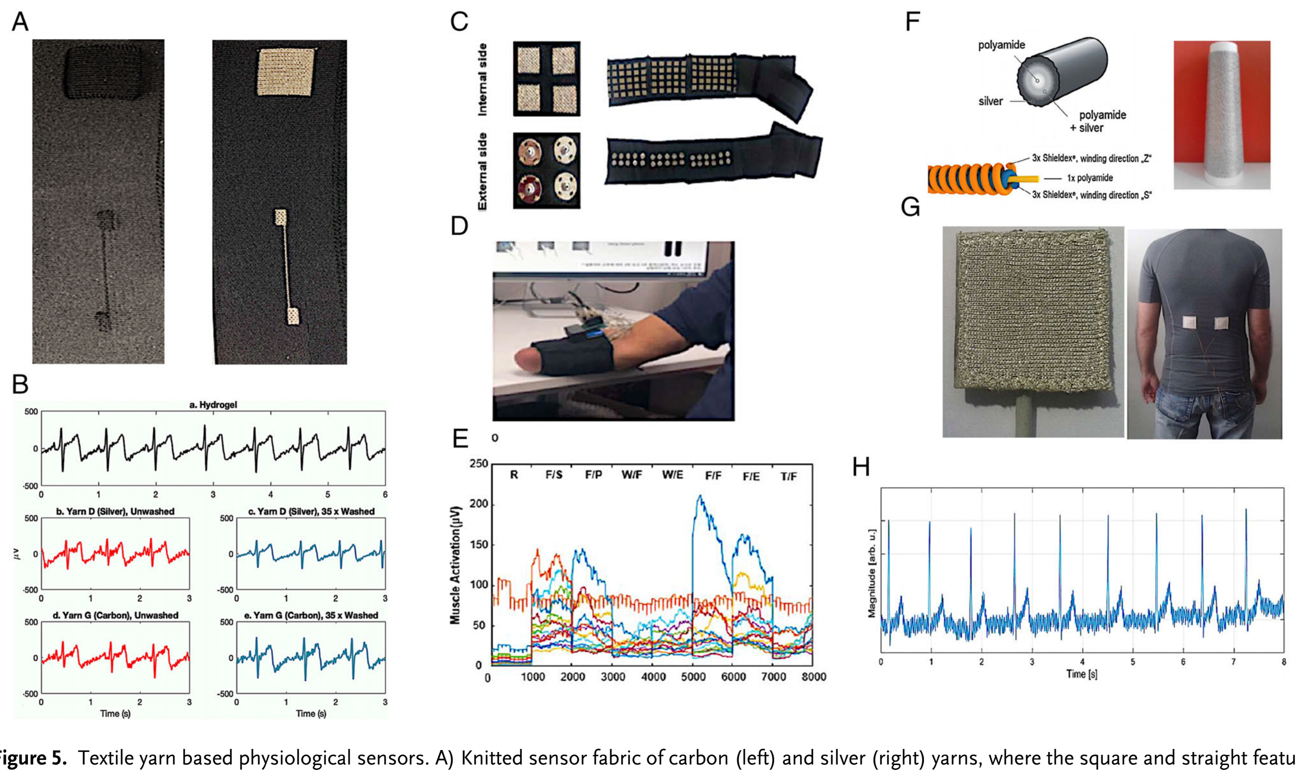

Figure 5. Textile yarn based physiological sensors. A) Knitted sensor fabric of carbon (left) and silver (right) yarns, where the square and straight feature patterns represent the ECG textile electrodes and signal transmission lines, respectively. B) The filtered ECG signals (μV vs time (s)) with silver (top panel), and carbon (bottom panel) yarn electrodes before and after 35 washing cycles; (A,B) Reproduced with permission.[64] Copyright 2020, American Chemical Society. C) wearable sEMG knitted fabric sensor made of 99% pure silver-plated nylon 66 electrode yarns—array of square knit electorodes on internal side (top panel), and 32 terminal metal snap-type connectors for electric wire connections on external side (bottom panel). D) The sensor system worn on the upper limbs to capture sEMG signals; E) root mean square values of 16-channel sEMG signals (muscle activation (μV) vs time (ms)) on the upper limb; (C-E) Reproduced with permission.[65] Copyright 2020, IOP Publishing. F) Shieldex fiber with polyamide core and thin layer of silver (top left), configuration of Shieldex bicomponent electroconductive twisted yarn (bottom left), and digital photo of the resultant yarn package (right). G) Plain jersey knitted ECG electrodes (left) attached on the back of the test subject (right); H) capacitive ECG signal (magnitude in arbitrary units (a.u.) vs time (s)) monitored on the back of the subject. (F-H) Reproduced with permission.[67] Copyright 2018, Elsevier.

allergic reaction on skin, unlike the electroconductive gel. Similar methodologies are also employed to produce stretchable electrode yarns by winding electroconductive yarns (in both S- and Z-directions), including stainless steel, copper, and silver, around elastic core yarns via hollow spindle spinning.[68] Such configuration allows up to 100% elongation with no change in resistance. Embroidery machines could be used to rapidly integrate these conductive yarns into fabrics. Zaman et al. and Ankhili et al. recently presented a novel approach of encapsulating an embroidered two-ply conductive track (Shieldex and Madeira Ag-PA66 yarn) with a nonconductive PET yarn in order to protect the electrodes against washing and drying.[69,70]

2.2.3. Other Coating Techniques

The use of coatings on textile yarns with ICPs and CMs for physiological strain sensors is making considerable progress. Examples include single-wall CNT (SWCNT)-coated stretchable cotton–PU yarn strain sensor (GF = 0.65) using vapor deposition (**Figure 6**A-C);[71] PEDOT-coated core-spun stainless steel/cotton yarn for autonomous self-powered strain sensor (GF = 0.29–0.54, 0–2.6% bending strain) via an in situ polymerization (Figure 6D-F);[72] and polypyrrole (PPy)-coated PU yarns for strains sensor (GF = 0.08–0.38, 8–60% strain) through in situ polymerization.[73] Due to the fast-paced nature and scalability of dip coating technology, researchers have applied this method to produce graphene-coated rGO-PDCY electrode yarns (DCY: double covered yarn of PU-PET fibers; PDCY: air plasma-treated DCY yarn) to monitor health or movements of robots (e.g., during "Gangnam Style" dance) (Figure 6G-I).[74] Additionally, this yarn sensor could be effectively used for monitoring pulse rate, evaluating sleep quality, recognizing accurate speech, and capturing biosignals from different human actions (such as walking, jogging, and jumping) with fast signal response (<100 ms) and high reproducibility (up to 10 000 cycles). An ultrahigh GF (873) at a large strain level (up to 200%) can also be achieved by yarn strain sensors fabricated by dip coating Dacron/spandex core filaments with MXene/Ag nanomaterials.[75] Recently, researchers from the National Graphene Institute (NGI) and University of Manchester have developed a chemical dyeing recipe to coat cellulosic yarns—using a conventional dyeing technology—with graphene flakes for a temperature sensor that could be knitted into fabrics.[76] Even though the researchers used a small-scale

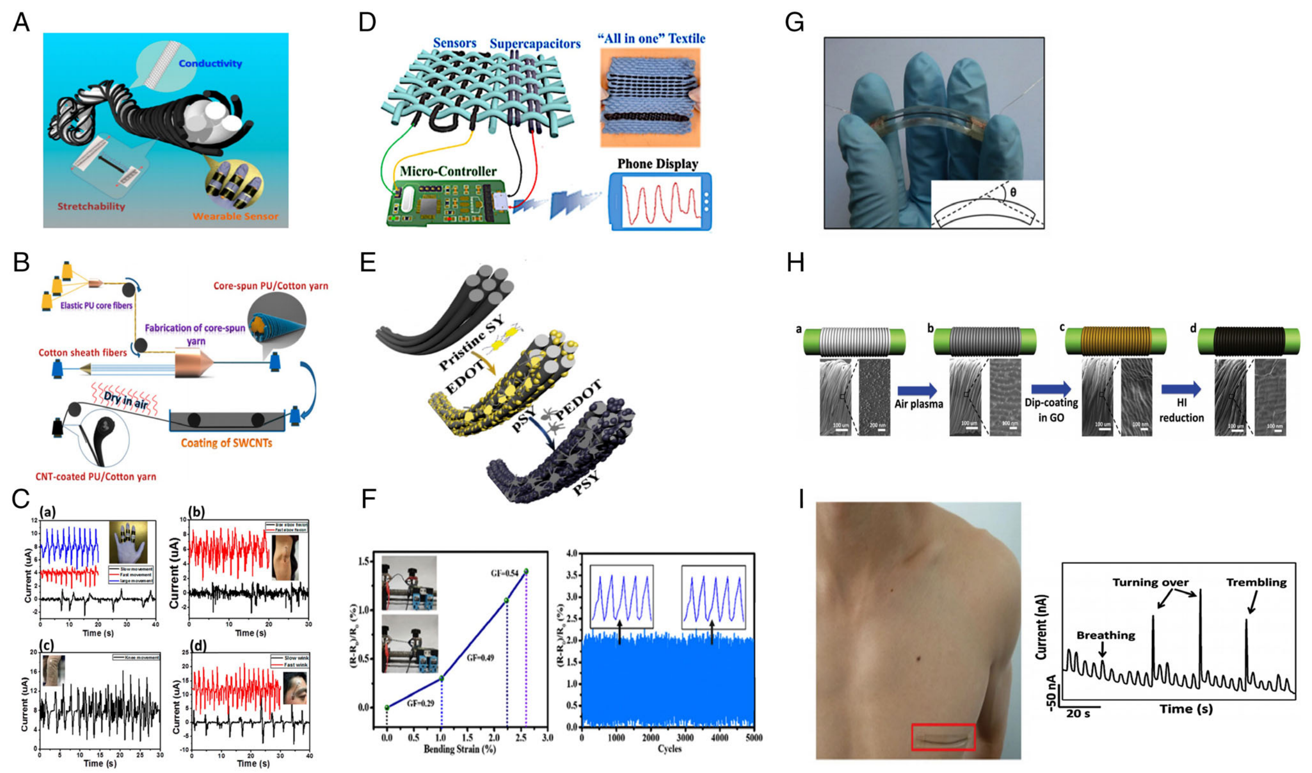

Figure 6. Textile yarn-based sensors for biosignal monitoring. A) Schematic of a flexible wearable strain sensor based on PU/cotton/CNT textile yarn. B) Coating the core-spun PU/cotton yarn with SWCNT suspension to produce electrode yarns for strain sensor applications; C) sensory signals for monitoring finger mobility (a), elbow movements (b), walking (c), and winkling (d); (A-C) Reproduced with permission.[71] Copyright 2016, American Chemical Society. D) PEDOT-based self-sensing and self-powered energy textile system with schematic showing connection to microcontroller with wireless Bluetooth module and readout on phone display. E) Production process of PEDOT (P)-coated pristine core-spun stainless steel/cotton yarn (SY); F) relative resistance change of the autonomous self-powered strain sensor (integrated with super-capacitor yarns) during mechanical strain (left) and cyclic bending stability (right) tests; (D-F) Reproduced with permission.[72] Copyright 2020, American Chemical Society. G) photograph of a flexible sensor made from rGO-PDCY yarns. H) Production steps of rGO-PDCY yarns (a: pristine yarn; b: air plasma-treated yarn; c: GO-coated yarn; d: rGO-yarns); I) attached sensors on chest (marked in red box) and the corresponding responsive curves (current (nA) vs time (s)) for simulating deep sleep (steady breathing) and light sleep (turning over and trembling), respectively. (G-I) Reproduced with permission.[74] Copyright 2015, John Wiley & Sons.

LABOMAT dyeing machine (Werner Mathis AG, Switzerland), they demonstrated that the formulated recipe could be applicable to industrial scale yarn dyeing machinery. The temperature sensor displayed excellent repeatability with a temperature sensitivity between 25 and 55 °C.

2.2.4. FBG Integration

Textile optoelectronic filament yarns, such as fiber Bragg grating (FBG) filaments, can be integrated as sensors using weaving, direct-stitching, or adhesive coating techniques. In 2015, researchers from Italy developed a respiration sensing textile by gluing FBG filaments into the textile using adhesive silicone rubber.[77] The resultant smart textile has opened a new door for respiratory textile sensors during magnetic resonance (MR) examinations, as FBG is immune to electromagnetic interferences. In 2019, another group of researchers from Italy produced flexible FBG sensors using an elastomeric Dragon Skin vulcanization method.[78] The sensors were stitched directly into a wearable elastic chest band to monitor respiratory and cardiac rates (heartbeats). Nedoma et al. produced a similar flexible PDMS-encapsulated FBG sensor pad for clinical bedsheets for the magnetic resonance imaging (MRI) chamber and for classical hospital environments to monitor ballistocardiography (BCG) signals (ballistic forces on the heart), respiration, and heart rate.[79] For body rehabilitation-related investigations—such as various static and kinematic postures or body postures arising from stroke or bone fracture—Abro et al. designed a smart sensing garment using a silica-gel-encapsulated FBG sensor stitched to a belt (**Figure** 7A-C).[80] In 2019 and 2020, Munster et al. introduced the concept design and measurement blocks for temperature and strain sensing intelligent textiles by integrating FBG filaments into nonwoven textiles using epoxy adhesives.[81,82] To simultaneously monitor plantar temperature, pressure, and joint angles for patients with risk of foot ulcers or diabetic peripheral neuropathy (DPN), Najafi et al. designed smart socks by stitching five FBG sensor filaments into knitted socks.[83] Through weaving and embroidery technologies, Rothmaier et al.[84] incorporated plastic optical fiber yarns (made from polymethyl methacrylate) into a glove designed as a wearable glove oximeter. The cotton glove was equipped with a textile-based light emitter and photodetector yarns for transmitting light through the forefinger tip and studying motion artifacts during SpO_2 measurement. The oximeter sensor utilizes the photoplethysmography (PPG) technique for measuring the blood SpO_2 and heart rate.[85,86] Recently, a group of Indian scientists introduced an optical strain sensor based on FBG finger plethysmography technique, an alternative technique to ECG for reading the pulse rate/signal (PR/PS) or heart rate (HR) at rest, to acquire arterial pulse pressure waveform from the ulnar artery.[87] The wearable FBG-based pulse plethysmograph recorder (FBGPPR) was fabricated using Germania-doped single-mode optical fiber, with a diameter of 125 mm, mounted on a thin fabric sheet of silicone diaphragm.

2.2.5. Resin Encapsulation and Braiding

Resin encapsulation is another technique to fabricate optoelectronics involving textile yarns. LEDs or different kinds of electronic devices could be attached to textile yarns for functional

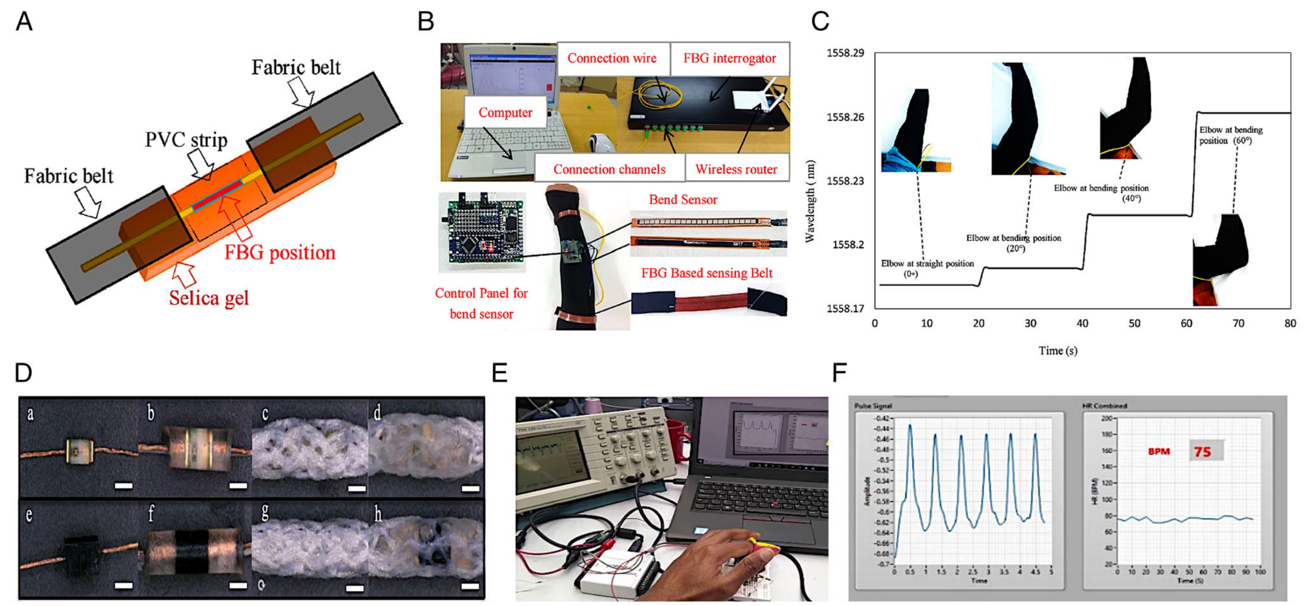

Figure 7. Optoelectronic textile based physiological sensors. A) Schematic diagram of the encapsulated FBG filament yarn inside an arm band. B) A combined monitoring system (top) based on FBG and strain sensors, where the FBG sensor is integrated into an armband attached to the elbow (bottom); (A-C) Reproduced with permission.[80] Copyright 2018, Elsevier. C) change of FBG wavelength during posture movements of the elbow (straight position corresponds to 0° and bending position corresponds to elbow positions at 20°, 40°, 60°); D) wearable textile-based heart rate sensor made of and LED (a–d) and PD (e–h) embedded yarns and encapsulated with resin micropod (a/e: LED (top)/PD (bottom) soldered onto the copper wire, b/f: encapsulated by resin micropod, and c/g: covered by the sheath of two-sets warp-knit braided polyester yarns, d/h: encapsulated by resin). E) Finger cuff with the wearable sensor; F) computing interface during the test to receive biosignal to realize the HR into beats per minute (BPM). (D-F) Reproduced with permission.[89] Copyright 2019, Elsevier.

applications.[88] Also, the resin encapsulation technique is used to attach thermistors (with a dimension of 0.5 mm × 0.8 mm × 0.8 mm) with textile yarns to produce body temperature sensors. This technique allows integration of more than one optoelectronic device together within the yarns, using a warp-knit braiding machine; for example, embedding the LED and photodiodes (PD) together with polyester yarns for fabricating wearable PPG heart rate monitoring devices (Figure 7D-F).[89] Another recent work demonstrated a novel process involving thermal drawing of polycarbonate filaments with embedded light-emitting and photo-detecting p–i–n diodes for application in textile-based heart rate monitoring PPG system.[90]

2.3. Smart Fabrics

Textile fabric configurations are generally classified into three main categories, viz., woven, knitted (warp-knit or weft-knit), and nonwoven. The previous sections have discussed how conductive fibers are deposited to form nonwoven sensory mats and how conductive yarns are directly woven or knitted into fabrics to create electronic sensors. The current section focuses on methods to convert nonfunctional solid fabrics into functional fabrics for biosensor applications. Coating is the dominant technique for fabricating fabric-based sensors as it does not considerably change the fabric flexibility, handle (tactile texture), or structural density and incorporates good conductivity.[91] In Section 2.1 and 2.2, we discussed different coating techniques for fibers and yarns, which are also applicable for textile fabrics. However, there is one particular simple coating technique that is applicable only for the fabric form factor, i.e., printing. A few research groups have applied complex coating techniques like electroless plating (or polymer grafting-assisted electroless deposition), chemical vapor deposition, and drop-casting methods to incorporate conductive electrodes onto the solid fabric surface.[63,92]

2.3.1. Printing

Printing techniques, unlike dyeing, dipping, or other coating methods, have the ability to precisely deposit localized electrode inks at only one surface of a fabric in any given time. As a result, fewer conductive materials are required, and less waste is generated. Stencil and inkjet techniques are the most widely used printing methods for wearable textile sensors. However, stencil-printing is much simpler in operation for fabricating printed electrodes[93] or conductive interconnects with PCBs on fabrics and flexible substrates. In fact, the first active electrode structure on a woven fabric for ECG measurement was produced using a stencil-printing technique.[94] Recently, researchers from the Institute of Systems and Robotics (Portugal) used this technique to develop a plug-and-play conformable headband for forehead electroencephalography (EEG) signal acquisition; the headband replaced the need of conventional individual wiring and electrode placements (**Figure 8**A-C).[95] The EEG headband was developed by cutting vias (to create electrical connection between layers) on a latex impregnated lycra-mesh fabric using a CO_2 laser followed by a screen printing treatment with Ag-based pigments. Conductive electrodes and interconnects/tracks were printed on either side of the fabric, respectively. The printed electrodes displayed a signal-to-noise ratio (SNR) ratio (−32.711 dB) comparable to that of gold standard Ag/AgCl electrodes (SNR: −32.187 dB). As stencil-printing technology is cheap, researchers often fabricate different kinds of sensors using this technology: for example, ECG electrodes by screen printing PEDOT:PSS ICPs onto a knit fabric;[96] and flexible tactile sensors by piezoresistive multiwalled CNTs (MWCNTs) and PVDF-trifluoroethylene.[97] Although stencil technique is the simplest printing method, a faster and more precise process for producing fabric sensors could be obtained by implementing inkjet-printing technique. The advantage of inkjet over other conventional printing and coating techniques is its stable ink-droplet formulation, printed circuits with higher resolution, and controllable printed patterns at precise locations with nearly zero ink waste.[98] The system also provides the flexibility of operating with composite inks in a single platform. Researchers at the National Graphene Institute (UK) recently formulated and employed a composite ink of graphene and silver on cotton fabrics to produce resistive (0.08–4.74 $\Omega\,sq^{-1}$) strain sensors for wearable electronics.[99]

2.3.2. Vapor Deposition

Chemical vapor deposition (CVD) process could also be a suitable method to deposit a conductive thin film layer onto a fabric surface for wearable sensor application. The technique is very popular for producing high-performance solid conductor films. Peng et al. employed this method to fabricate a flexible dry fabric electrode for EOG, EEG, and ECG sensors (Figure 8D-F).[100] Due to its superior biocompatibility, flexibility, and adhesion to metallic particles, the researchers deposited a 5 μm-thick layer of Parylene C (PC) fabric onto a glass wafer using CVD process. Following the CVD process, sputtering and electroplating techniques were used to subsequently coat the fabric film with Cr/Cu and electroplated Ag layers, respectively. The electrodes demonstrated superior SNR performance (23.8 dB) as compared to similar commercial Ag/AgCl electrodes (SNR: 20.6 dB). A recent work by Korzeniewska et al. also developed a similar strategy to produce Ag-deposited nonwoven PU-fabric sensor by employing a PVD technique using a Classic 250 vacuum chamber based on the Pfeiffer Vacuum system; however, no comprehensive investigation was conducted on the resultant textile sensor.[101]

2.3.3. Electroless Plating

Electroless plating technique (EPT) refers to the deposition process of conductive materials, including metals, onto a fabric surface without applying external electricity. Such technology is often termed as an "autocatalytic technique" or a "chemical plating" technique. A cotton fabric-based capacitive sensor was recently developed by researchers from University of Salford and University of Manchester using this EPT technique (Figure 8G-I).[102] This fabric sensor has been successfully used to monitor individual respiration, speaking, blinking, and joint motions in real time during rehabilitation exercises. The capacitive sensor was developed through three successive processes: 1) polymer grafting of cotton fabric using ammonium tetrachloropalladate (II) $[(NH_4)_2PdCl_4]$, nickel sulfate hexahydrate

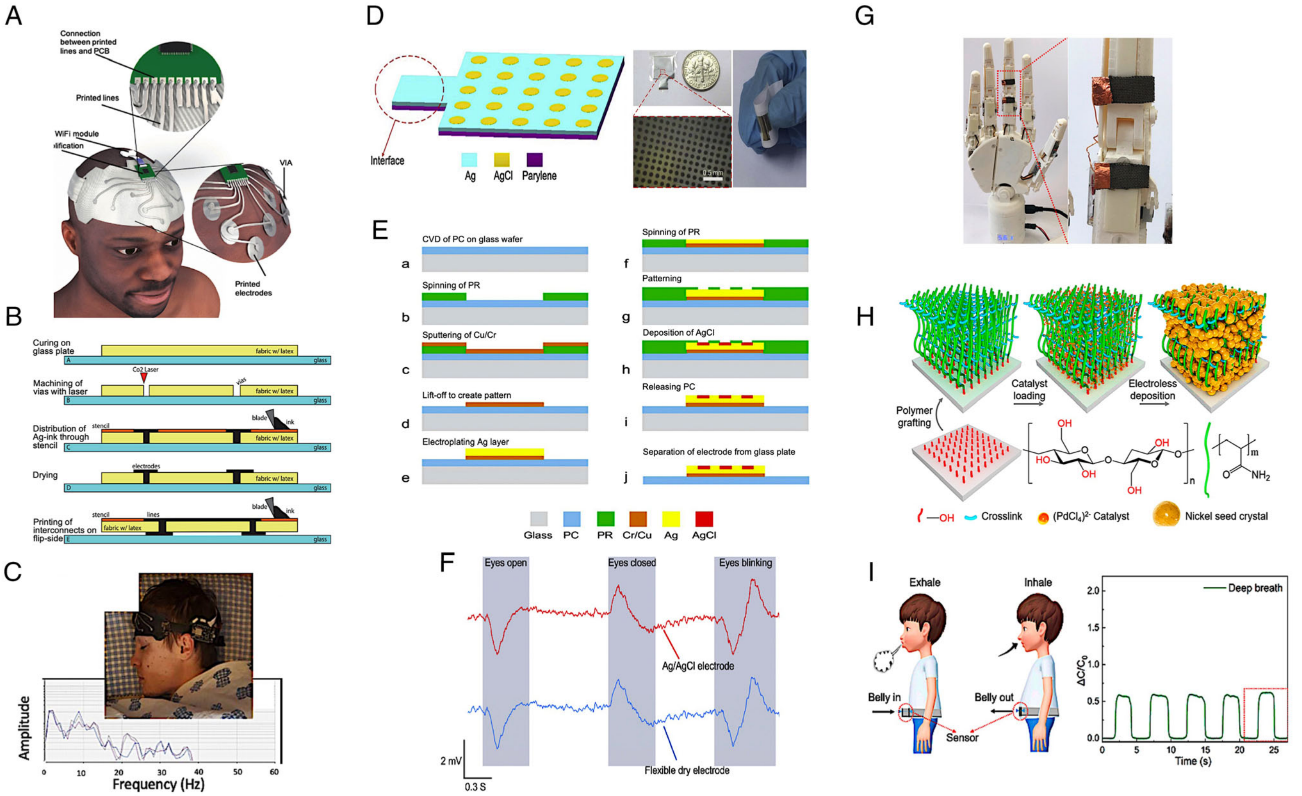

Figure 8. Physiological textile fabric-based sensors. A) EEG acquisition headband (top detail: rigid PCB and the conductive lines printed on a textile fabric is connected by an ink drop joint; right detail: contact between the forehead and printed electrodes is displayed). B) Fabrication process of the headband patch; C) sleep data acquisition using the EEG headband (top detail: normal awake EEG activity; middle detail: EEG during sleep; bottom detail: normal awake EEG); (A-C) Reproduced with permission.[95] Copyright 2020, IEEE. D) schematic of a flexible PC fabric sensor (left) with its microscopic image (right). E) Fabrication (bottom) process of the sensor using CVD method; F) EOG biosignals captured by the flexible dry and Ag/AgCl electrodes; (D-F) Reproduced with permission.[100] Copyright 2016, Elsevier. G) Cotton fabric capacitive sensor produced by polymer grafting and nanotechnology-based EPT. H) Schematic illustration of the fabrication process; I) Change of capacitance as a function of time between two conductive fabrics during deep breathing exercise, where the peaks and valleys of the curve represent inhale and exhale cycles, respectively. (G-I) Reproduced with permission[102] Copyright 2020, American Chemical Society.

($NiSO_4{\cdot}6H_2O$), acrylamide monomer (AAm); 2) palladium catalyst loading on the AAm-cotton fabric; and finally 3) the electroless deposition of Ni NPs onto the Pd-AAm-cotton fabric. The sensor was used to monitor speech, respiration, and joint movements—yielding approximate $\Delta C/C$ ratios of 0.1, 1, and 0.023, respectively.

2.3.4. Coating and Drop-Casting

Single metal and hybrid metal coated fabrics are quite popular in textiles electronics. Different ECG skin electrodes using 100% copper-coated and hybrid copper-nickel-coated polyester fabrics are an example.[69] Dip-and-dry coating technique is also popular due to its simplicity. As a result, EMG fabric sensors (SNR: 21.32 dB compared to 21.16 dB for equivalent Ag/AgCl electrodes) could easily be made from graphene-coated cotton and nylon fabrics.[103] Meng et al. used dipping technique to produce an Ag-coated woven PET wristband sensor (sensitivity: 3.88 V.kPa^{-1}) for monitoring arterial pulsation and personalized health care.[104] The flexibility of the dipping technique also allows layer-by-layer (LBL) coating on top of a predyed textile electrode. LBL technique was used to prepare ECG electrodes (SNR: 21.6 dB) and pulse rate sensors for sports bra and wrist band, respectively.[105] The LBL process involved the application of pad-dry-cure dyeing and finishing machine to dye and fix graphene onto cotton fabrics, followed by the dipping technique to coat the graphene-dyed electrodes with PEDOT:PSS. The engineers used dipping technology and developed a smart conformable headband for various EOG applications, including sleep studies, medical diagnosis, and HCI/HMI (human–computer interaction or human–machine interaction) interfaces for the disabled.[106] EOG monitors eyeblinks and eye movements by detecting the voltage difference between cornea and retina of a user.[107] Golparvar and Yapici simply dipped a knitted nylon fabric into a graphene solution to design the dry textile electrodes for integrating into the headband.[106] Completing the dipping process, thermal and chemical reduction treatments were performed to remove the intercalating oxygen groups and deposit a cladding of rGO on the fabric surface. Such EOG textile technology is gaining a widespread application in detecting drowsiness for vehicle drivers and in areas like assistive technology, cognitive neuroscience, somnolence studies, activity monitoring, mental disorder diagnosis, and so forth. Compared to the dipping method, drop-casting is a relatively slow coating technique that involves releasing controllable, large droplets of conductive ink onto a fabric surface. This distinct coating technique could be used to produce a CM-coated fabric sensor (SNR: 22 dB). Using a drop-casted PET fabric under an ultrasonic bath of edge-oxidized GO solution, Awan et al. developed a sEMG PET fabric sensor for monitoring muscular activity.[108]

2.4. Smart Garments

Thus far, this article delivers a comprehensive summary of the materials, processes and applications for biomedical sensing at the fiber, yarn and fabric level. Implementation at the garment level necessitates not only an integration of the technologies discussed above, but also a meaningful and robust system design to allow for seamless coordination between the sensors, interconnects, power supply and data storage and transfer units. This section presents selected studies that demonstrate successful approaches for autonomous operation and dynamic monitoring through textile based wearable systems.

An early prototype developed by Linz et al. in 2006 monitored electrocardiography (EKG) using embroidered electrodes and a coin-sized (27 × 27 mm) microprocessing unit printed on flexible polyimide substrate.[109] The system is implemented on a commercially available tight-fit t-shirt (**Figure 9**B). A semiprofessional embroidery machine (Bernina Artista) is used to create strain-relieving zig-zag interconnects using a silver-coated polyamide yarn. The fabric is prestretched during embroidery and encapsulation to reduce mechanical strain during wear. Multiple embroidered pathways between components and at interconnection junctions provide for good redundancy and reliable operation. EKG data are processed and transferred via low-energy Bluetooth (BLE) operating in burst-mode, which allows the suit to function autonomously for up to 24 h.

Moving toward multimodal applications, in 2018, Tao et al. presented a fully encapsulated, BLE-based activity monitoring textronic system using off-the-shelf sensors (TK InvenSense MPU-6050 for temperature and motion sensing, Texas Instruments ADS1292R for ECG and respiration)[110] (Figure 9D). The system leveraged an FPCB with five conductive pads for interconnection to the peripherals. Knitted textile electrodes of silver-plated polyamide thread were stitched in a zig-zag pattern to maintain elastic propriety of the textile. The PCB system was transferred onto a commercial stretchable sports garment via a thermoplastic polyurethane (TPU) film with a similar TPU mask applied to exposed conductive straight-stitch interconnections. The PCB was finally encapsulated in PDMS for protection during wash, dry, and wear. The study demonstrates the use of a wide variety of methods and engineered interfaces to create a robust, multimodal biosignal monitoring garment using off-the-shelf components.

In an effort to provide a wireless alternative to the manifold interconnections, Niu et al. developed a multimodal bodyNET system with stretchable on-skin sensor tags that communicated with a textile-based flexible radio-frequency identification (RFID) initiator circuit[111] (Figure 9C). The sticker-like sensor tags were fabricated out of a flexible poly-(styrene-*b*-ethylene-butadiene (poly-SEBS) substrate coupled with conductive composite inks (Ag flakes) and CNT networks that displayed excellent sensitivity and stretchability (up to 50%). The initiator circuit comprising the oscillator, amplifier, analog-to-digital convertor (ADC), microcontroller, and bluetooth transceiver was printed on an

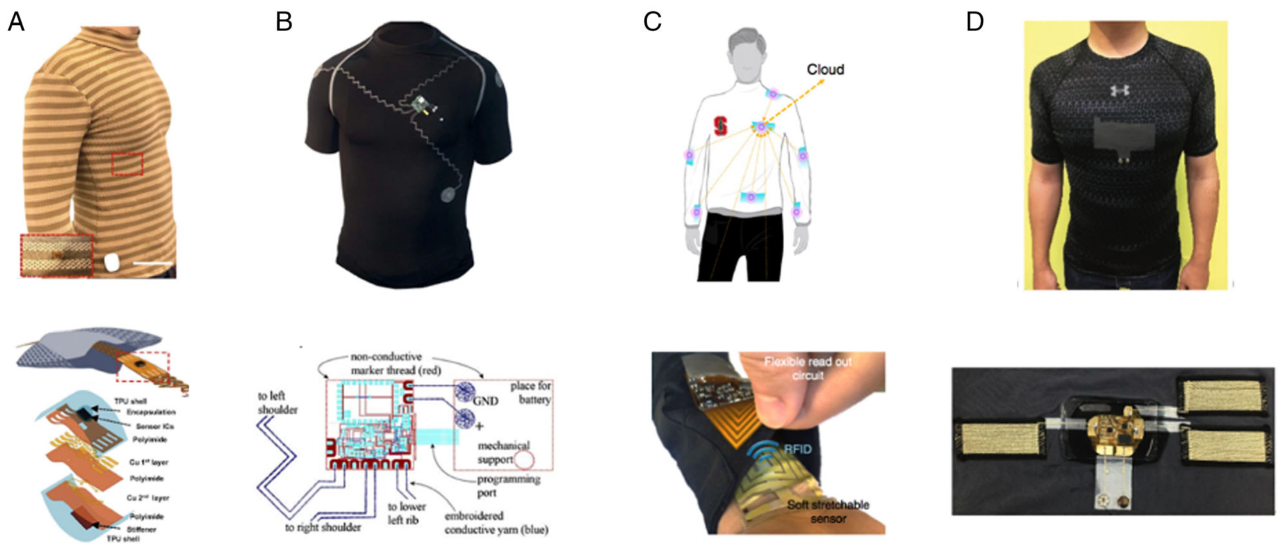

Figure 9. Garment-level implementation of multimodal physiological monitoring systems. A) Compression suit for the ETeCS with (inset) an exploded view of the PI-based sensor island. Reproduced with permission.[112] Copyright 2020, Springer Nature. B) An embroidered EKG monitoring system implemented in an encapsulate on a commercially available compression garment with (inset) a flexible PI-based PCB with embroidered interconnects. Reproduced with permission.[109] Copyright 2006, IEEE. C) A multimodal bodyNET system with (inset) on-skin sensor tags with RFID coupling to textile-mounted processing units. Reproduced with permission.[111] Copyright 2019, Springer Nature. D) A fully encapsulated, BLE-based activity monitoring textronic system with (inset) knitted textile electrodes and commercial temperature and heart rate sensors. Reproduced with permission.[110] Copyright 2018, John Wiley & Sons.

FPCB and powered by ultrathin lithium polymer (LiPO) batteries. A unique RFID near-field communication (NFC) protocol allowed for secure wireless transmission of sensor signal over 20–25 mm. This approach facilitates a meaningful separation of the conflicting form factors required for the two elements—intimate, ultraconformable sensors for optimal on-skin signal pickup and a mechanically stable textile platform for the relatively rigid high-performance silicon electronics.

However, NFC-based solutions require additional power, and there is still a need for a sustainable textile platform that allows multimodal, large-scale physiological sensing—both in terms of autonomous operation and rapid manufacturability. The electronic textiles conformable suit (ETeCS) developed by Wicaksono et al. presents a unique approach to front-end customization and integration of electronic ribbons in a tailored high-flex polyester knit garment[112] (Figure 9A). The harmonious system design provides horizontal routing of flexible PI sensor ribbons through striped textile channels throughout the suit, with vertical connections in the seams using conductive copper thread. The microprocessing unit, battery, and BLE module are housed in a mini-detachable pod that is connected to the suit using snap fasteners. The electronic ribbons are fashioned on PI-based FPCB with embedded off-the-shelf temperature sensors (Maxim Integrated MAX30205) and accelerometers (TK InvenSense MPU6050). The I2C architecture allows for data registration from up to 32 unique temperature sensors and 2 accelerometers. The compression garment is modeled to provide adequate pressure (2–20 mm Hg) for good on-skin sensor contact while maintaining a comfortable fit for the user. The ETeCS addresses the manifold challenges of developing robust, reliable wearable systems—from sensor signal integrity and calibration to garment comfort and user experience.

3. Discussion

As distinct from other works in its field, this article presents a comprehensive review of textronics for vital body signal monitoring from the unique perspective of sensor integration at the various levels of the textile manufacturing process chain—fiber, yarn, fabric, and garment. While a large number of methods and research prototypes have been developed, there is little commentary on the holistic quality and suitability of each approach for ubiquitous, personalized health-monitoring systems. Several research reviews on wearable textile platforms provide recommendations for the 1) mechanical and 2) chemical performance of the textile.[10,113] The authors further recommend analysis and clear reporting of the biocompatibility, safety, comfort, experimental conditioning, and pretreatment features for the textile sensor system. On the other hand, studies on body sensor networks (BSNs) identify and define critical performance parameters for successful BSNs such as interoperability, reliability, security, validation, and accuracy of the sensor signal.[114] A third dimension is added to the analysis when the two (body sensing electronic systems and textiles) are integrated for biomedical applications. A high-quality study by Leenen et al. systematically attempts to define metrics for success of commercial wearable vitals monitoring systems in the medical context.[115] This section aims to consolidate the critical performance requirements of the three subdomains to provide a holistic framework through which to evaluate the suitability of the textronic sensors discussed above for biomedical applications.

At present, there are no test standards for physiological textile sensors either in Europe or North America. The absence of proper standardization and quality control test methods poses the highest barrier for the biomedical and healthcare industries.[116] The first workshop in North America focusing on the need for standardization for wearable smart textiles, including textile sensors, was organized by American Society for Testing and Materials (ASTM) not more than a decade ago, and the efforts gained momentum when American Association of Textile Chemists and Colorists (AATCC) joined forces with ASTM 5 years back, resulting in a few test standards for textile electronics (developed by the ASTM D13.50 and AATCC RA111 subcommittees). More recently, Canadian institutions, such as University of Alberta and CTT Group (Quebec), have begun to collaborate with the US standardization committees. Outside North America, technical bodies of European Committee of Standardization (CEN TC-248/WG-31) and the International Electrotechnical Commission (IEC TC-124) have joined forces to develop standardized test methods for textronics. These efforts must continue for successful commercialization of textile electronics, including wearable physiological textile sensors. To support the ongoing standardization efforts, our current study of textronics has surveyed the prevailing state of the art and has identified the need for characterization test methods for physiological textile sensors, as illustrated in **Figure 10**.

We have selected sensor papers from the research articles listed over the last 10 years (with two article exceptions) in the field through a random sampling approach in order to represent the whole gamut of form-factors possible with textile-based electronics. A matrix table was developed to assign all the applicable characterization techniques as attribute properties (0, 1, N/A classification) for each study (see Table S1, Supporting Information, for detailed information). The characterization approaches are listed with a breakdown of the methods/tests included under the umbrella of each in Figure 10B-E. A total of 918 possible tests resulted, of which only 26% of tests have been attempted by the researchers of the reviewed work (Figure 10B).

Among sensor technologies and solutions we reviewed, strain and motion sensors represent 37% of the physiological textile sensors and products for monitoring different vital body signals, including acceleration, orientation, vibration, joint movement, muscle contraction, speech recognition, smiling, winking, walking, jumping, squatting, sign language, and pressure/impact (Figure 10A). In this review, ECG, respiration, and temperature sensors accounted for 27%, 11%, and 10%, respectively. Surface EMG, pulse oximetry, blood pressure, EEG, and EOG sensors make up the rest of the proportions. Figure 10C–E represents the percentages of the identified test method categories with their associated subcategories for wearable physiological sensors.

Figure 10C (textile-platform durability tests) exhibits that most of the works focus primarily on characterizing the tensile and bending performance of the textile sensor. Section 4.1 gives a detailed breakdown of the tests that remain to be addressed, and the importance of each. Figure 10D (sensor efficiency tests) indicates that a very small fraction of studies attempted a

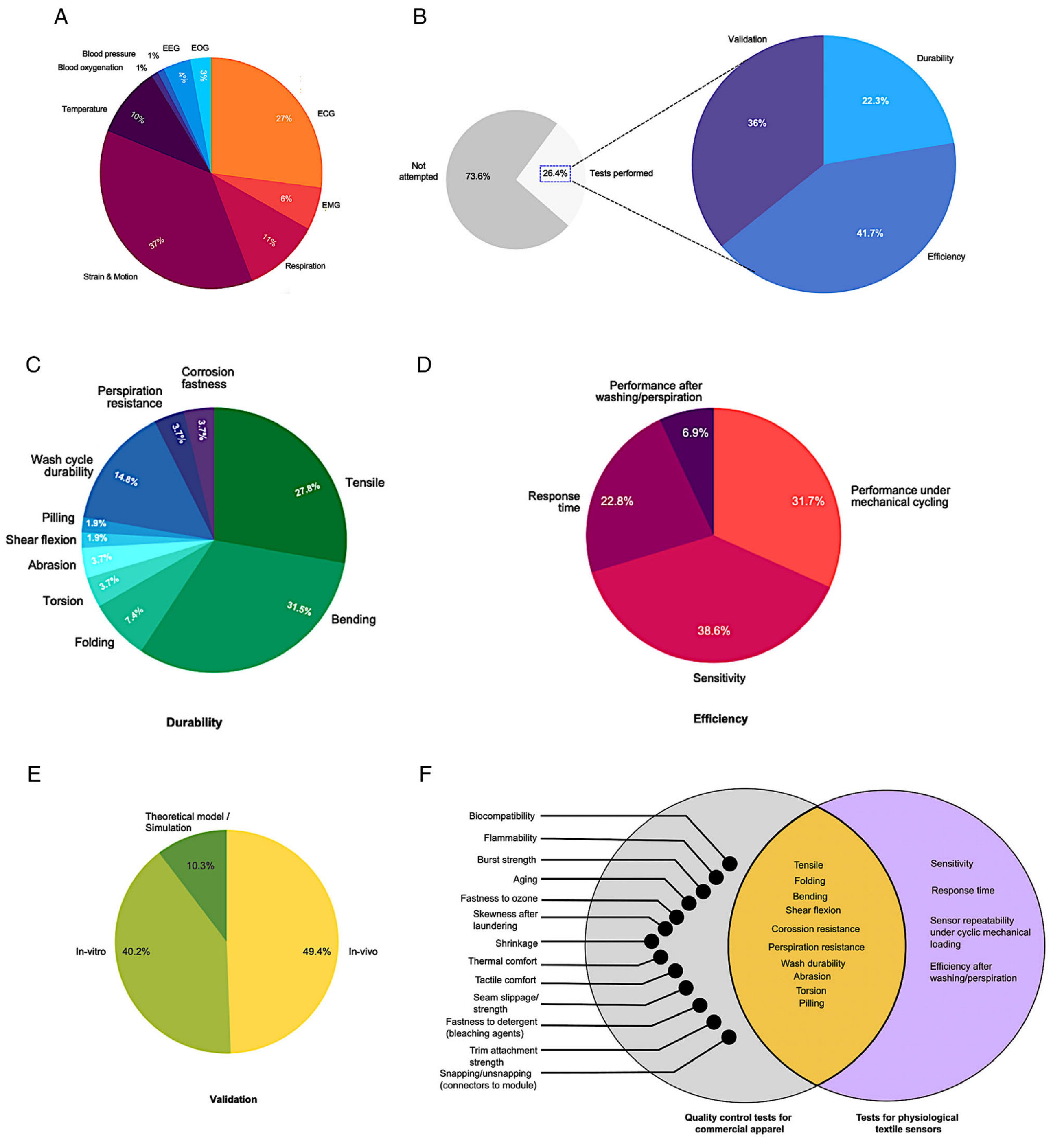

Figure 10. Proportionality diagrams for physiological sensors included in the current study with their associated testing in the context of international quality control standards for commercial apparel; A) relative proportion of textile sensors by biomedical modality (ECG, EMG, respiration, strain and motion, temperature, blood oxygenation, blood pressure, EEG, and EOG); B) relative proportion of the three main characterization tests performed for textronic sensing systems; C–E) breakdown of various test methods included under each characterization approach—sensor durability, efficiency, and validation; F) needs identified (in the red zone) for future performance characterization (comfort, safety, biocompatibility, fastness, dimensional stability, and mechanical aspects) of physiological textile sensors and textronic system.

characterization of the transducer after washing/perspiration. This is further discussed in Section 4.2.1. In Figure 10E (validation tests), the higher relative proportion of the in vivo tests (as compared to in vitro tests) exposes the fact that a number of studies may have attempted sensor system characterization in a highly variable, dynamic in vivo scenario before documenting characterization of the same under controlled, static conditions in vitro. Finally, Figure 10F is the culmination of our analysis and aims to provide a clear visual representation of the quality control tests that are currently excluded from the purview of characterization testing attempted by researchers.

To ensure stable electrical performance and durability, textronic systems, including textile biosensors, must be robust against mechanical wear such as abrasion. Surface fuzz formation is the primary effect of abrasion and deteriorates the product serviceability.[117] Due to their structural geometry, knitted materials are more susceptible to abrasion and pilling compared to woven materials;[118] hence, fibers with high elongation, elastic recovery, and work of rupture are recommended for a good degree of abrasion resistance[117] for the textronic systems. Otherwise, yarns with poor abrasion resistance will rapidly wear out because of progressive reduction of the local yarn linear density,[119] causing the biosignals registered by the electrode yarns to be interrupted. Furthermore, a poor abrasion resistance markedly affects the weaveability of yarns to be used for producing functional fabrics for biosensors and hampers the scalability of an industrial production process. Hence, reporting the test results of abrasion resistance could simulate the real-life wear behavior and help to understand performance dependence on mechanical fatigue.

Like abrasion, pilling—the entanglement of fibers—is another undesirable surface flaw of a textile product, resulting from abrasion with external surfaces or fabric-to-fabric abrasion.[120,121] Surface pilling is one of the major drawbacks of knit fabrics.[122] The seriousness of this phenomenon could be exacerbated by the integration of synthetic polymers into a sensor or wearable fabric, both by themselves or in blends with natural polymers[117] as synthetic polymers such as PET, nylon, or rayon, would create significant surface pilling.[123,124] The pill-formation process is often initiated by the slow and gradual cyclic torsional deformation of the fibers. As researchers use different kinds of polymers for sensory yarns or fabrics, it is advised to report the material robustness against mechanical surface phenomena like abrasion, pilling, torsion, folding, or shear flexion to enable a comparative performance analysis among biosensors developed with different kinds of natural, synthetic, or blended polymers. Similarly, researchers should also characterize the bursting strength, especially for knit geometries, to demonstrate the sensor strength when subjected to pressure on all possible directions at the same time (e.g., bias, course, and waleswise)—a phenomenon often experienced during squatting or fitness exercises.

Surface decay under environmental conditions is another critical issue for textile sensors, especially for electroplated or conductive metal-coated textile electrodes. For example, silver-coated textile electrodes suffer from atmospheric corrosion, which eventually degrades performance.[125] Humid air is the primary reason for this metallic corrosion; however, material species and atmosphere are also reported to be responsible for the corrosion.[126] During fabric washing, conductive metal coatings on the textile sensors become oxidized;[127] silver-plated textiles may suffer from a secondary level of surface degradation due to the sulfur releasing bacteria present in washing machines.[128] As a result of this sulfidation, a dark layer of Ag_2S (silver sulphide) is deposited onto the silver-coated textiles and decays their electrical performance.[129] Similarly, salts present in the perspiration corrode conductive metallic elements.[130,131] Therefore, it is recommended to report the performance of textile sensors against corrosion, washing, and body perspiration.

4. Recommendations

A textronic system must fulfill its dual purpose as a biomedical sensor as well as a garment, but the two aspects are inexorably linked. Characterization testing for textronics is thus afflicted with interdependencies. Figure 10F (gray circle) gives a summary of the different characterization techniques used for commercial apparel (data collected after consulting with a few major North American retailers) and textile-based sensors (identified from our survey) (purple circle). Only a coordinated interplay and reconciliation of the contesting needs of the textronic as (Section 4.1) a textile platform and (Section 4.2) a biomedical sensor can result in the realization of a successful textronic system. The sections below address the outstanding needs for each of these two criteria.

4.1. Performance Characterization of Textile Platform

Figure 10F (gray circle) gives a summary of the different characterization techniques used for commercial apparel (data collected after consulting with a few major North American retailers) and textile-based sensors (identified from our survey) (purple circle). As textile sensors are integrated into apparel articles, they must be capable of withstanding characterization treatments used for commercial apparel articles until new standardized test techniques specific to wearable textile electronics are introduced. After a comparative analysis between the identified test methods for physiological textile sensors and those used for commercial apparel, more than ten quality control aspects were identified that are not currently being addressed by the researchers of textronics (as shown in the grey zone of Figure 10F). Examples include test methods to check the biocompatibility, safety, flammability, aging, seam strength, and thermal comfort aspects of the textronics. These are discussed comprehensively in Section 4.1.1–4.1.6 below.

4.1.1. Biocompatibility and Safety

Ensuring biocompatibility and safety is paramount when developing any biomedical product that interfaces with the human skin. In commercial apparel, retailers typically check for the presence of formaldehyde, colorless skin-irritant compounds, and cancer-causing carcinogenic materials in the treated (scoured, desized, dyed, and finished) fabric before selling the apparel articles. Registration, Evaluation, Authorization and Restriction of Chemicals (REACH) regulations, for example, prohibit businesses from selling certain textiles and clothing articles into the European Union market that would contain carcinogenic, mutagenic, or toxic to reproduction category 1 A or 1B (CMR) substances exceeding in a concentration greater than 75, 5, and 1 $mg\,kg^{-1}$ for formaldehyde, benzene, and chlorotoluenes or extractable heavy metals (e.g., cadmium, chromium IV arsenic, lead and its compounds), respectively.[132] Researchers of wearable textile electronics also must ensure similar biocompatibility testing. For example, gallium-based liquid metal, eutectic gallium–indium (EGaIn) in particular, is the most popular liquid conductor for epidermal strain sensors and has been used for wearable gloves.[133,134] Such liquid conductor-based silicone sensors are used to track dexterous manipulation tasks or hand joint angles of patients who have lost motor control of their body's upper limbs as a result of trauma or neurological disorders. However, most wearable sensors made of EGaIn or organic solutions are not safe for skin applications.[133] Similarly, piezoelectric sensors made of $Pb(Zr,Ti)O_3$ (PZT) are toxic for skin applications because they contain lead.[135] PZT ceramic may dissolve in an aqueous environment under high temperature; hence, such toxic PZTs must be replaced with nontoxic alternatives, as the biocompatibility of such physiological sensors must be ensured against body perspiration.

Knowledge of the structural geometries of textiles would also help design a safe textronic platform for biomedical patients with conditions such as paraplegia. In medical terms, paraplegia is defined as spinal cord damage (paralysis or loss of sensory function) below thoracic 1 (T1) level, the spine in the upper back, and abdomen.[136] It may occur from an accident or chronic condition; consequently, the legs, lower limbs, trunk, and pelvic organs may become dysfunctional as well as the bowel, bladder, and sex organs. Such paraplegic patients could be vulnerable to temperature overshooting of textronic systems or thermoregulatory strain as they would be unable to detect any thermal strain due to their impaired sensation. Several reconstructive surgeries for posttraumatic paraplegic patients associated with textronic platforms were reported in Europe and the USA.[137,138] Such medical cases heighten the necessity for standardizing the safety features of textronics and investigating their structural configuration to avoid a geometrical mismatch with the needs of medical patients.

Textiles designed using cellulosic cotton materials are the dominant textile materials in the world due to their inherent softness, moisture absorption, and anti-irritant behaviors. Consequently, cotton has become a popular choice for researchers developing textile sensors; examples include a plantar pressure measuring metallic cotton sensor to monitor possible anomalies in foot-ground contact (for patients with diabetic or musculoskeletal alteration like hallux valgus, toe deformities, and flat foot)[139] or piezoresistive rGO-cotton sensor to detect body motion.[140] However, one major drawback of cotton is its high flammability, which makes them potentially dangerous species for wearable applications unless treated with specialty flame-retardant finishing chemicals (FRFCs). Common FRFCs often cannot meet the military specifications for smart electronic cotton fabrics; for such purposes, functionalization of cotton polymers with CNTs could be a solution.[141] Hence, performing the flammability tests of textile sensors, especially those designed with organic or flammable polymers, could be one of the best industrial practices to ensure consumer safety.

Like flammability, testing the fastness of a textronic system against atmospheric ozone or bleaching agents is also significant as metallic conductors like silver electrodes are susceptible to atmospheric corrosion by ozone[142] and oxidation by the bleaching agents during laundering.[143] As a consequence of corrosion and subsequent mechanical rubbing, the silver coating layer can easily tear off, rendering the textile sensors unusable to sense any biopotentials. Washing or laundering also causes shrinkage and skewness (angular displacement of yarns from a line perpendicular to the edge/selvedge), which influences key characteristics like appearance and texture of a textile platform and changes the dimensional stability. Such a change in the structural geometry of the capacitive textile sensors can negatively impact the sensing performance.

Breathability (air-permeability) also plays a dominant role in the biocompatibility of the textronic system as it regulates the dynamic interaction of the user to the external ambient environment. In a clinical setting, for instance, a lack of microclimatic breathability could introduce a high coefficient of friction between the skin of immobile patients and medical textiles due to the presence of high perspiration concentration, which would ultimately cause pressure ulcers.[144] Hence, an excellent ergonomic design paradigm of the textronic breathability is a necessary parameter to guarantee biocompatibility. Although the breathability of the textronic has long been a neglected crux of its design, modern textronic designers are adapting markedly. At present, a sensitivity up to $1.33\,\mathrm{V.kPa^{-1}}$ could be demonstrated by using a triboelectric breathable woven fabric pressure sensor made of nylon and polytetrafluoroethylene filaments.[145] Also, for a breathable tactile textronic sensor, graphene dyed woven fabrics could be another choice with a target sensitivity up to $18.65\,\mathrm{kPa^{-1}}$.[146] Techniques to improve breathability using different textile geometries and materials are discussed subsequently in Section 4.1.2.

4.1.2. Thermal and Tactile Comfort

Thermophysiological aspects of textile structures profoundly influence the adoption and use of textile platforms atop which BSNs are mounted. As textiles differ in terms of structural geometry and materials, a proper sensational evaluation is essential. Researchers often use off-the-shelf textiles for fabricating physiological sensors and the support array textronic system; however, no studies were identified that focused on the impact of electrode (yarn) linear density or conductive fabric structures on the comfort level of the users although the fabric structures are responsible for buffering and dissipating metabolic body heat. Researchers found that the integration of electronics like sensors, PCBs, conductive interconnecting paths, or power sources on the textile platform influences the thermophysiological comfort of the smart textronic system.[147] In another work, researchers recommended using woven structures for ECG sensors over nonwovens because of their superior breathability, wearability, and thermal comfort.[94] Surprisingly, no details were available on the particular weave structures despite the fact that woven fabrics differ significantly in their air-permeability based on their weaving geometries. For instance, a satin weave configuration, although inherently a loose and flexible structure, exhibits a lower breathability compared to basket weave, warp, and weft rib structures (**Figure 11**A).[148] The variance of air-permeability among the structures is a function of yarn interlacements between warp and weft. As structures like plain weave have a higher number of interlacements than other structures, the porosity and the air-permeability are the least. The thread density also influenced the

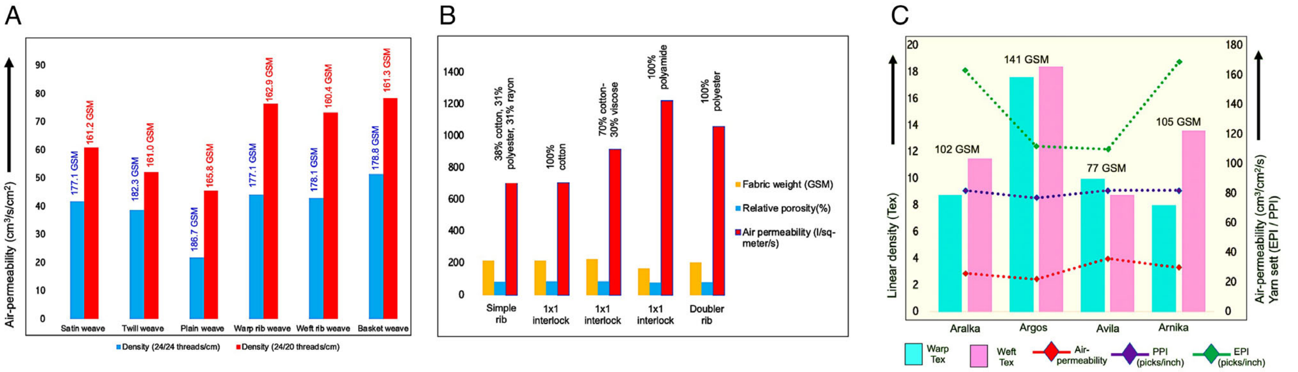

Figure 11. Effect of textile geometry on breathability and comfort. A) Influence of different weave structures and thread density on the air-permeability ($cm^3/cm^2/s$) of cellulosic fabric (based on data from [148]). B) Influence of different knit fabric structure and fiber polymers on air-permeability ($l\,m^{-2}\,s^{-1}$) (based on data from [150]). C) Impact of yarn tex and yarn settings (EPI and PPI) on air-permeability of medical-grade antiallergic PET linens, viz., Aralka, Argos, Avila, Arvinka (Based on data from [152]).

air-permeability for the mentioned structures. Air-permeability reduced with the increasing yarn density because of the reduction of porosity. Other parameters also impact the air-permeability of woven fabrics, such as yarn twist.[149] With the increase of yarn twist, the diameter of the yarn decreases and thereby reduces the air-permeability. Similarly, different knit structures affect the physiological comfort of the wearer. In fact, knit fabrics of comparable weights and porosities exhibit significant differences in air permeability as a function of structural configuration (Figure 11B).[150] Although the researchers did not provide any justification behind the permeability variance among the knit fabrics of similar structures, a possible explanation could be the slackness and tightness of the structure. After scoping a recent work, we discovered that tight structure offers a higher barrier to air-permeability than the slack structure for both single jersey and double jersey (rib and interlock) fabrics, most probably due to higher loop density.[151] Besides the structural geometries of the 2D knit/woven substrates, proper selection of mesostructures (tex) and settings (EPI: ends per inch or PPI: picks per inch) for yarn electrodes are also essential as they significantly affect the overall breathability of conductive fabrics. The influence of yarn tex and settings on the breathability of antiallergic medical linens are shown in Figure 11C.[152] One possible reason behind this variance could be the increased yarn diameters from increased tex values, which leaves fewer pores in between the interthread knitted channels. Such geometrical factors must be scoped out while designing textronic systems to offer better thermophysiological comfort levels to the users of personal protective equipment (PPE) or wearable textile sensors.

As there are no standardized test methods from ASTM, AATCC, CEN, IEC, or ISO to assess the thermal comfort level of electronic clothes, researchers can use thermal manikins[153] or different portable hardware units (based on Xtensa dual-core 32-bit and conductive textile ribbons) to test the thermal comfort level of the whole textronic system in dynamic conditions or during physical stress activities.[147] It must also be mentioned that fabric structures influence the transmission signals and SNR for textile electronics.[154] Smart textiles is a new field, and thus far, mostly simple and double jersey structures in knit configuration and plain weaves for woven configurations are being investigated. Diversified structures including sateen, twill, and honeycomb should be explored. The intersection and spacing of warp and weft conductive/nonconductive filaments impact the stretchability of the fabric, which manifests in varying the resulting resistance and SNR of the sensor.[62,154,155]

The overall comfort level of a textronic system is also associated with tactile comfort. The tactile comfort level is mainly regulated by the moisture regain of the constituent fiber polymers as polymers of higher moisture regain would provide better tactile comfort because of improved thermal effusivity (first thermal contact feeling) and cooling effect.[150] In contrast, the structural fabric geometry would dictate the porosity, thermal conductivity, and water vapor permeability. Also, gram per square meter (GSM) (thickness) and porosity should be adequately adjusted as GSM significantly impacts thermal insulation and porosity.[151]

4.1.3. Aging

Aging of textile sensors is another critical issue that is associated with laundering and exposure to environmental conditions. Although the direct effect of aging is not visible in the short term, the gradual long-term effects may prove damaging over time. The synergetic influence of solvent, temperature, chemicals, and mechanical stresses involved in laundering could cause systematic aging and gradual reduction of performance of textile sensors. Other than laundering, various environmental factors (ambient heat, UV, or humidity) may also age the textiles sensors in the long run. Despite such reasonable concerns, very few works exist concerning the effect of aging phenomena on textile sensors. Therefore, monitoring the impact of aging is significant.

4.1.4. Seam Reinforcement

Optimum seam reinforcement is essential to hold the overall structure of a textile platform during donning and doffing and prevents the rupture of stitch lines. It resists the damaging mechanisms like stances that cause maximum apparel stress at different locations, for example, during squatting, elbow, or knee bending. Seam efficiency—the ratio of seam strength to fabric strength—represents the durability of a seam, and typically ranges up to 85%. It can be further optimized by proper selection of polymers for sewing thread and seam configurations.[156] A seam efficacy of 80% is the standard used by the Canadian Forces for designing their combat clothing.[157] The efficacy of seam strength becomes even more crucial while developing a conformable electronic sensor suit that would hold a sensor island for registering physiological biosignals.[112] Any breakage of the seam line could expose the sensor island and make it vulnerable to mechanical damage or connection loss. Hence, high stitch density and poly/poly core-spun threads could be used to ensure good seam strength due to their high sewing thread strength.[158] Power supply and management systems are an important consideration for textronic development. Self-powered and energy harvesting modules are being increasingly integrated into state-of-the-art sensor design,[104,111] while several studies still utilize external batteries which are attached via fasteners as a module to the garment.[110,112] These attachments must be secured and tested during system characterization in compliance with existing standards such as ASTM D4846-96. Sufficient attachment strength is required for trim components to ensure robustness and reliability of the textronic system to stand up to the demands of mass commercialization.

4.1.5. Conditioning

Surprisingly, very few researchers mentioned the conditioning parameters of ambient temperature and relative humidity while fabricating the textile sensors or registering the biosignal measurements. Although relative humidity has a significant impact on the fiber properties, especially on moisture regain of hydrophilic polymers,[159] researchers appear to overlook this critical aspect. Because the moisture regain increases with the increase of relative humidity, and depletes with the increase of air temperature,[160] natural polymers are more susceptible to humidity variations than synthetic fibers, especially in terms of tensile strength, elasticity, fiber diameter, and friction. Tensile strength of natural hydrophilic polymers, for instance, increases with the increase of moisture regain.[117] As a result, natural hydrophilic polymers, such as cellulosic fibers, are processed at a controlled high relative humidity to prevent fiber breakage and to ensure optimum flexibility against mechanical stresses. However, exceeding the regain threshold also lowers the overall performance. For example, high moisture regain causes high transverse swelling to polymers—around 40% for cotton and 1.6% for nylon—that provides accessibility to the ambient moisture.[161,162] As a result, an exothermic reaction takes place and liberates heat, resulting from hydrogen bond formation.[160] Such unwanted exothermic reactions may influence the preciseness of biosignals captured by a temperature or heat-flux sensor.

Furthermore, the ambient relative humidity, i.e., the moisture content between air and fiber interface, also influences the electrical sensitivity and static electrification of different fiber polymers. The fiber releases moisture and will become dry if the atmospheric air is drier than the relative humidity equilibrium of the textile. As the fiber becomes drier, its electrical resistance increases. In addition, dry fibers tend to build their static charge, a scenario that could become dangerous for natural polymers at low humidity, as the accumulated static electricity could build a high voltage, possibly in the magnitude of hundreds or more, posing a health risk to people with poor heart conditions. An uncontrolled static discharge at the wrong place could easily damage the miniaturized semiconductors, microprocessors, or electronic modules of BSNs and textronics. Static discharge by polyester filaments is the ultimate result of poor moisture regain. Therefore, fabrication and measurements of the textile polymer-based physiological sensors should be conducted under polymer specific standardized atmospheric humidity and temperature, as recommended by ASTM or ISO test methods.

4.1.6. Pretreatment Conditions

Pretreatment of textile form-factors is another critical issue that is often ignored by researchers. Natural fiber polymers are always pretreated with alkaline scouring and H_2O_2 chemicals to improve wettability, eliminate impurities (oil, wax, fat, etc., added during fabrication process), and improve flexibility.[163–167] Synthetic polymers are also pretreated, although they are free from natural impurities. For example, nylon filaments are alkali-scoured with surfactants and soda ash (Na_2CO_3) at 70 °C to remove the spin finishes, lubricants and PVA residues or other synthetic size materials on woven fabrics.[160] A majority of the researchers use the pristine form or off-the-shelf commercially available textiles for fabricating sensors, which typically comes pretreated. However, scoured-bleached textiles could lower the performance of sensors as they demonstrate higher sheet resistance against conductive print patterns.[168] Again, if a fabric is woven or knitted with electrode yarns (like silver plated or stainless steel yarns), and then treated with the scouring and bleaching chemicals, the effect (increased resistance values) remains the same and irreversible, mostly because of the chemical damage on the surface of the electrode yarns.[154] Any further laundry washing is likely to exacerbate the overall transmission properties, thus lowering the SNR. More assertive application of new pretreatment technologies for textile electrodes would reduce the sheet resistance and increase the SNR. Plasma treatments or covalent polymer grafting with ICPs like polyaniline (PANI) have shown promising results in lowering the sheet resistance of conductive textiles.[168,169] Hence, it is highly recommended to report the pretreatment conditions of the textile substrates used for fabricating textile sensors.

4.2. Performance Characterization of Biomedical Sensing Modalities

The section above captures the criteria for evaluation of the impact of the textronic system on the user. An equivalent investigation is warranted vice versa in order to bridge the gap between the stand-alone operation of flexible physiological sensors and their implementation on daily-use textiles. Further, for the wearable sensor to be relevant in a medical context, the sensor data must exhibit a certain degree of accuracy, repeatability, and sensitivity throughout its life. Several studies discuss the considerations for testing and standardization of wearable sensors in the context of the medical device market, including quality control and ethics[20,170,171] The study by Leenen et al. attempts to set limits of agreement (LoA) and accuracy constraints for the various physiological sensing modalities: 10 ± 10 beats per minute for HR, 3 ± 3 breaths per minute for RR, $0.5 \pm 1.0\,°C$ for temperature, 10 ± 20 mmHg for systolic BP, and $3 \pm 5\%$ for SpO_2.[115] While this cannot be vetted as an industrial standard, it sets a reasonable guideline for researchers as they characterize sensor performance across a range of physiologically relevant temperatures as well as standard strains associated with day-to-day movement and motions such as walking and stretching. This section identifies these as two major areas that call for standardized testing and characterization of the interdependencies afflicting the electrical response of sensors intended for textile-based wearable applications.

4.2.1. Resistive Strain Sensors

The typical stretchability range of textiles for a tailored garment is 15–25%, for sportswear is 20–35%, and for a form-fit compression garment is between 30% and 40%.[172,173] For strain-motion sensors, this behavior is inherently captured through GF plots. Based on the application, a critical distinction must be drawn between the sensor's functionality as intended for large-strain or supersensitive applications. Examples of both types of sensors are comprehensively discussed in Section 2.1 and 2.2. For body-motion and respiration sensors, sensors with GFs in the range of 0.1–10 are typically anticipated as given in Refs. [71] and [75], while for the small strains associated with heart-rate capture, strain gauges with high sensitivities (GFs of 10^2–10^4) are required.[174] Based on the GF plot characteristic, the sensor can be prestrained and tuned to operate in the appropriate region of linearity corresponding to the needs of the physiological application.[174] Once the sensor has been optimized for the intended application, efforts must be made to flatten out the sensitivity curves ($\Delta R/R\%$ vs ε) in the other strain regions. This "isolation" of the sensor response can be achieved through mechanical manipulation of the sensor geometry, back-end analog signal postprocessing,[175] or through innovative strain-dissipation approaches through the fabric such as the use of gradated auxetic patches or mechanically stabilizing encapsulations surrounding the sensor.

4.2.2. Optical Sensors

FBGs are an excellent choice for textile-based physiological sensors owing to their innate yarn-like form factor, high sampling rates, low power operation, and high-sensitivity. However, it is commonly known that FBGs respond to both temperature as well as strain stimuli, and it becomes necessary to decouple the responses. The thermal sensitivity of the average FBG sensor is around $10\,\text{pm}\,°C^{-1}$, while deformation sensitivities are in the range of $1–2\,\text{pm}\,\mu\varepsilon^{-1}$. Thus, for body-mounted sensors, a change in $0.1\,°C$ in body temperature can confound the wavelength shift registered due to strain. There is a definitive need to identify a standardized approach to characterize and represent the contribution from each, as well as their interdependence. Current methods to isolate the contributions from each stimulus include 1) placement of an additional temperature probe within the FBG assembly[82] to generate an independent measurement and compensate for temperature, 2) manipulating the specific pattern of the sensor grating to generate largely different sensitivities for the thermal and strain components,[176] and 3) encapsulating the sensor to render it temperature resistant. Other optical based sensors used for SpO_2 measurement are similarly impacted by body movement, as demonstrated by Rothmaier in a study with an optical SpO_2 glove susceptible to finger motion artifacts.[84] None of the studies in our survey validate the performance of the optical sensor in a dynamic monitoring scenario, though some identify it in the scope of their future research.[78] This lack of in vivo validation presents a critical gap in the applied research of optical based sensors, one that will determine their relevance in textronic systems of the future.

4.2.3. Thermocouple/Thermistor-Based Sensors

Ink-based temperature sensors remain relatively stable with stretching cycles up to 20% strain (800 cycles), as is well characterized in a study by Jung using AgNP ink on PEDOT:PSS substrate.[177] Lugoda et al. report errors as small as $0.5\,°C$ for a commercially available thermistor embedded in a polymer resin and multithread yarn subjected to stretch, compression, and bending.[178] An rGO fiber-based resistive temperature sensor fabricated using wet spinning shows excellent performance under mechanical deformation over 10 000 cycles.[179] Hussain evaluated the performance of an RTD based chest-mounted TSF (metallic wire in polyester knit fabric) in a dynamic monitoring scenario, and observed uneven temperature readings (temperature drop up to $1.6\,°C$) during body movements.[180] As is clear from the review above, distinct efforts have been made to capture the behavior of various types of temperature sensors under mechanical deformation. The methods used, however, are largely personal to each study, and there is a lack of common terminology and parameterization techniques to characterize the stretch behavior of these sensors.

The subsections above provide a concise insight into the performance interdependencies of three types of sensors commonly used in the textile industry—resistive, optical, and thermal. A similar discussion is warranted for other sensing modalities to identify the main confounding factors for in vivo measurement—interested readers are encouraged to explore the associated complexities of in vivo testing for other sensors including capacitive, inductive, triboelectric, and piezoelectric sensors discussed in recent articles.[41,67,72,181–184]

The smart textile sensor industry is young but has experienced unprecedented growth in recent years. The underlying motivation of our work is to introduce a protocol to ensure proper quality control of this young but vibrant industry. Due to the aforementioned challenges with the textronic systems, developing a fully documented experimental protocol with precise methods of standardized testing is highly recommended. Therefore, to reduce the diversity in reporting characterization test series or performance parameters for textronic or textile-based sensors, we propose the following test-report protocol (as summarized in **Figure 12**) that could be adopted while designing and evaluating the system.

Figure 12. Bubble diagram summarizing the recommended flow for investigating and characterizing textile sensing materials and systems. As textile fiber materials are susceptible to ambient conditions, it is recommended to maintain a standardized ambient condition (1), e.g., relative humidity and temperature, before characterizing the performance of textile sensors. For this purpose, AATCC, International Organization of Standardization (ISO), or ASTM standards for textile materials could be followed. Also, polymer/material composition, biocompatibility (reactivity to skin or concentration of the chemicals), and nature of pretreatments (scoured, bleached, or other surface modifications) conducted on the textile form factors (e.g., fiber, yarn, fabric, or apparel) must be reported before the fabrication of the sensor architecture (2). During the fabrication (3) stages, it is vital to ensure sufficient structural strength of the BSN for robustness and uninterrupted vital signal registration. At this stage, it is recommended to choose the correct fabric structure, yarn tex or setting to maximize the tactile comfort level and minimize the thermophysiological strain on the user of the textronic platform. Scrutinizing the system safety in terms of flammability and electric shock resistivity is highly recommended, while the sensor efficiency and mechanical behaviors are benchmarked during postfabrication (4) in vivo characterization under physiologically relevant thermal loads and movement.

5. Conclusion

An agreement on the high-level quality control of textronics can trickle down to provide a structured framework within which R&D efforts in this field can be meaningfully translated.

In our literature review, we have made a critical selection of 58 articles to represent the whole gamut of textile form factors (fiber, yarn, fabric, and apparel) used in the fabrication of textronic sensor systems. We have then provided a comprehensive discussion around the characterization techniques used for the sensors. Although current advancements present a diverse array of novel materials and fabrication technologies, very few works apply a rounded approach to sensor characterization that aligns with the current industry standards for textile performance. While the final route to approval for wearable sensors lies through Class II regulatory pathways with the FDA, there is a definitive need for standards and testing frameworks dedicated to textronic applications. The emergence of specialized, interdisciplinary standards will greatly streamline 510(k) or premarket approval (PMA) submissions for these devices and ensure timely and well-informed decision-making by regulating bodies. Our work attempts to highlight the research gaps by combining and reconciling the perspectives of textile, mechanical, electrical, and materials engineering. The pie charts in Section 3.1 provide a detailed breakdown of all the mechanical, sensor efficiency, and validation tests performed for the reviewed sensors.

For the mechanical performance of the textronic system, we identify four main criteria given in Section 3 (burst strength, trim attachment, seam reinforcement, and snap-fastening strength) that have not been addressed by any of the reviewed works. The mechanical impact of angular displacement and shrinkage of yarn-based electrodes after washing also calls for quantitative measurement. Measures must be taken to offset the reaction of ambient chemical factors such as ozone, bleaching agents, and perspiration (alkaline and acidic) on the sensor. As textiles age silently, accelerated aging tests must also be performed at physiologically relevant temperatures and relative humidity to simulate the wear behavior of the apparel. In terms of safety control, a larger focus should be placed on addressing the biocompatibility and flammability, as is substantiated in Section 4.1.1. Finally, we recommend that researchers consider the thermal and tactile comfort of the resulting garment during system design to optimize the microclimate in the vicinity of human skin.

Our discussion is based on a comprehensive review of 58 selected works. In the future, we aim to perform a similar analysis on large datasets (>1000 articles) using tailored algorithms based on the framework established in this study to distill insights at a statistically significant level. We hope that our work will serve as a guide to both researchers in the field as well as the work of different international and national technical committees responsible for developing standardized test methods for smart electronic textiles, such as CEN TC-248/WG-31, IEC TC-124, ASTM D13.50, and AATCC RA111.

Supporting Information

Supporting Information is available from the Wiley Online Library or from the author.

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Conflict of Interest

The authors declare no conflict of interest.

Keywords

biomedical devices, conformable devices, flexible electronics, quality control, testing and characterization, textile sensors

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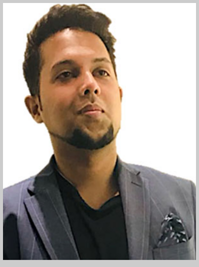

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