

A Wearable Thermoelectric Respiratory Sensing System for Quantitative Pulmonary Function Monitoring

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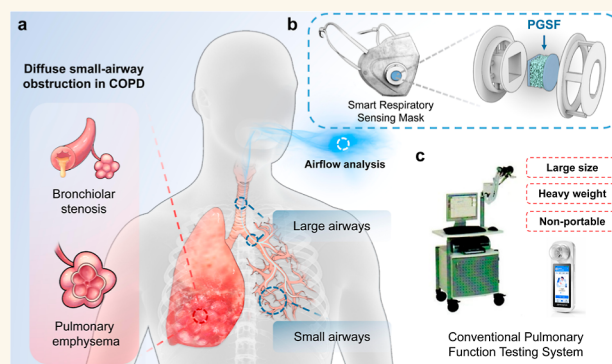
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ABSTRACT: The dynamic management of chronic obstructive pulmonary disease (COPD) and related obstructive airway diseases, such as asthma, demands respiratory monitoring technologies that combine high spatiotemporal resolution, long-term stability, and broad applicability. Yet current approaches, including bulky clinical systems, thoracoabdominal strain sensors prone to motion artifacts, and hydrogel-based airflow sensors susceptible to drift from solvent evaporation, fall short of these requirements in terms of portability, robustness against interference, and durability. Here, we report a porous graphene-based foam sensor that decouples strain–temperature to achieve simultaneous, crosstalk-free detection of deformation and airflow temperature without complex signal processing. The foam sensor maintains high mechanical and thermoelectric stability after 15,000 cyclic compression at 80% strain. Using the produced thermoelectric signal, we define a novel small airway obstruction index (SAOI) that provides individualized early warning of COPD acute exacerbation risk, enabling timely, real-time adjustment of treatment regimens.

KEYWORDS: thermoelectric respiratory sensing, mechanical–thermal signal decoupling, continuous pulmonary function monitoring, multisenario wearable detection, COPD exacerbation early warning



1. INTRODUCTION

Chronic respiratory diseases, exemplified by chronic obstructive pulmonary disease (COPD), represent a leading and escalating cause of global morbidity and mortality.^{1–5} Effective management to prevent acute exacerbations and improve outcomes depends critically on longitudinal monitoring of respiratory function. While spirometry remains the clinical gold standard for assessing lung function, its reliance on bulky, operator-dependent equipment confines measurements to episodic, in-clinic “snapshots” of a patient’s status. This approach cannot capture dynamic pathophysiological changes in daily living environments. Existing portable spirometers enable limited home monitoring but are not truly wearable or unobtrusive, precluding continuous, high-fidelity data acquisition. Consequently, a critical technological gap persists: the inability to continuously monitor lung function in real-world settings. This limitation hinders early detection of deterioration and personalized therapeutic strategies.

Efforts to overcome the limitations of traditional devices have catalyzed advances in wearable respiratory monitoring. Sensors based on various physical principles have been developed to track respiratory signals continuously. These approaches predominantly track respiratory effort through thoracic and abdominal wall kinematics, using technologies like resistive

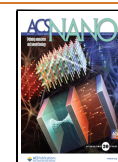
strain sensors^{6–9} or self-powered triboelectric nanogenerators that harvest energy from breathing.^{10–15} Alternative strategies detect airflow with mask-integrated piezoelectric sensors¹⁶ or leverage thermal signatures via thermistors^{17–21} and thermoelectric materials.^{22–26} Hydrogel-based airflow sensors have also been reported,^{9,15} but their susceptibility to signal drift from solvent evaporation undermines long-term reliability in continuous monitoring scenarios. However, a fundamental limitation persists: these devices primarily monitor surrogate metrics such as respiratory rate and rhythm, rather than direct physiological parameters for comprehensive pulmonary assessment. Their signal fidelity is often compromised by motion artifacts, inconsistent placement, and environmental fluctuations. More critically, they cannot provide direct, stable measurements of key spirometric parameters—airflow rate, volume, and their dynamics—which are indispensable for detecting subtle pathophysiological changes like small airway

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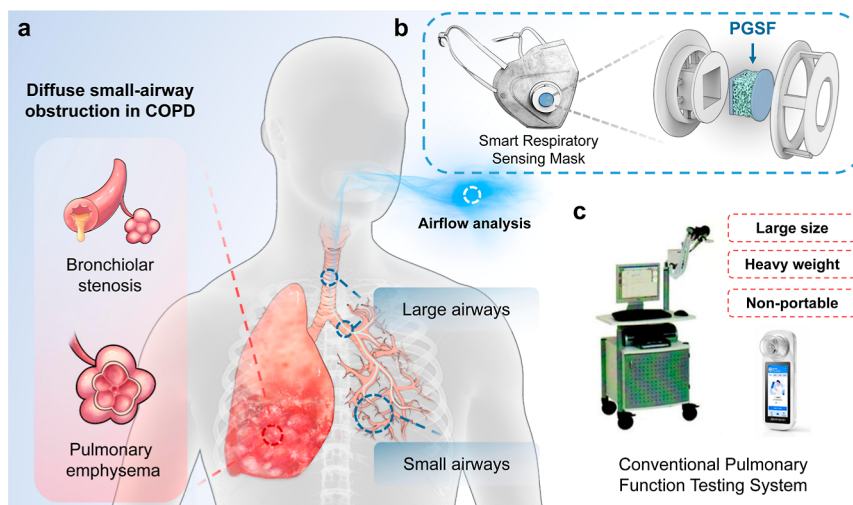


Figure 1. Smart respiratory monitoring mask for chronic obstructive pulmonary disease (COPD) monitoring. (a) Pathological features of chronic obstructive pulmonary disease (COPD) and the distribution of airways in the lung. (b) Design of the porous graphene/PDMS smart foam-integrated monitoring mask (PGSF-MM), which incorporates a porous graphene/PDMS composite smart foam. (c) Limitations of traditional pulmonary function testing systems.

dysfunction and diminished lung elastic recoil in COPD. Thus, clinical translation for longitudinal management remains severely constrained.

Lung function testing evaluates respiratory mechanics by measuring the relationship between exhaled flow rate and volume. The flow-volume loop provides a comprehensive signature of respiratory mechanics, reflecting airway patency and alveolar elastic recoil, and is the cornerstone for diagnosing obstructive diseases like COPD. Specifically, in COPD, the pathological changes are characterized by inflammation and fibrosis of the small airways (diameter < 2 mm) and destruction of alveolar structures, leading to a loss of elastic recoil.²⁷ These microanatomical alterations manifest macroscopically as characteristic changes in the flow-volume curve, most notably expiratory flow limitation (Figure 1a). Therefore, a wearable device capable of reliably capturing flow-volume information with fidelity comparable to spirometry would bridge the gap from respiratory monitoring to lung function monitoring, enabling continuous, real-world management.

To bridge this gap, we introduce a portable and real-time monitoring system centered on a thermoelectric sensor fabricated from a porous graphene/PDMS smart foam (PGSF) (Figure 1b). Central to our approach is an orthogonal signal pathway design that decouples thermoelectric temperature sensing from mechanical strain. This enables direct, stable acquisition of pure respiratory temperature signals even under motion artifacts, without complex computational algorithms. By integrating the PGSF sensor into a monitoring mask (PGSF-MM), we continuously track respiratory airflow dynamics in real-time, and derive lung function parameters, including flow-volume loops. Furthermore, we propose and validate a novel lung function biomarker, the expiratory flow acceleration-to-flow ratio $\dot{V}_{VVR@HPV/MVVR} = \left(\frac{dV(t)}{dt} \right)_{HPV} / \left(\frac{dV(t)}{dt} \right)_{Max}$, which demonstrates high sensitivity of the PGSF-MM in detecting the degree of small airway obstruction. Clinical validation confirms that this system enables continuous monitoring and early risk stratification for exacerbation. The PGSF-MM overcomes the environmental and portability constraints of traditional spirometry, paving the way for multisenario, continuous lung

function assessment (Figure 1c). This work presents a new paradigm for interference-resilient thermoelectric sensing and a tangible step toward precise management of obstructive lung diseases via wearable technology.

2. RESULTS AND DISCUSSION

2.1. PGSF Performance Characterization

To address the need for daily portable respiratory monitoring in patients with obstructive lung diseases, we developed a porous graphene/PDMS smart foam (PGSF) for respiratory monitoring and lung function detection. We comprehensively evaluated its sensing performance and environmental adaptability^{28,29} (Figures S1 and S2). We fabricated a high-precision testing platform to evaluate the electrical response characteristics of the PGSF under coupled mechanical–thermal fields (Figure S3). First, based on the thermoelectric sensing mechanism, a temperature difference (ΔT) between the two ends of the PGSF causes charge carriers to diffuse, generating a potential difference (Figure S4a). Measurements under different temperature gradients ($\Delta T = 1\text{--}18\text{ K}$) show that the thermoelectric voltage (V) generated by the PGSF increases linearly with ΔT (Figures 2a and S4b), and the slope of the curve corresponds to a Seebeck coefficient (S) of $58.7\text{ }\mu\text{V/K}$ (Figure 2b). The sensor is equally sensitive to opposite temperature gradients (Figure S4c). Furthermore, by applying sequentially increasing and alternating currents to a Peltier element to simulate continuous alternating hot/cold stimuli (analogous to nasal breath variations during breathing), the PGSF exhibited extremely high response sensitivity (Figure S4d). Experimentally, even under minute temperature gradients ($\Delta T \approx 0.01\text{ K}$), the PGSF generates a significant thermoelectric voltage ($\sim 0.5\text{ }\mu\text{V}$), confirming its capability for monitoring weak respiratory temperature changes (Figure 2c). It is noteworthy that the Seebeck coefficient of the PGSF remains essentially constant despite deformation-induced changes in carrier transport, which confers strong anti-interference capability for reliable respiratory monitoring (Figure 2d). A systematic comparison with recently reported sensing materials demonstrates the PGSF's superior overall performance in temperature sensing (Table S1).

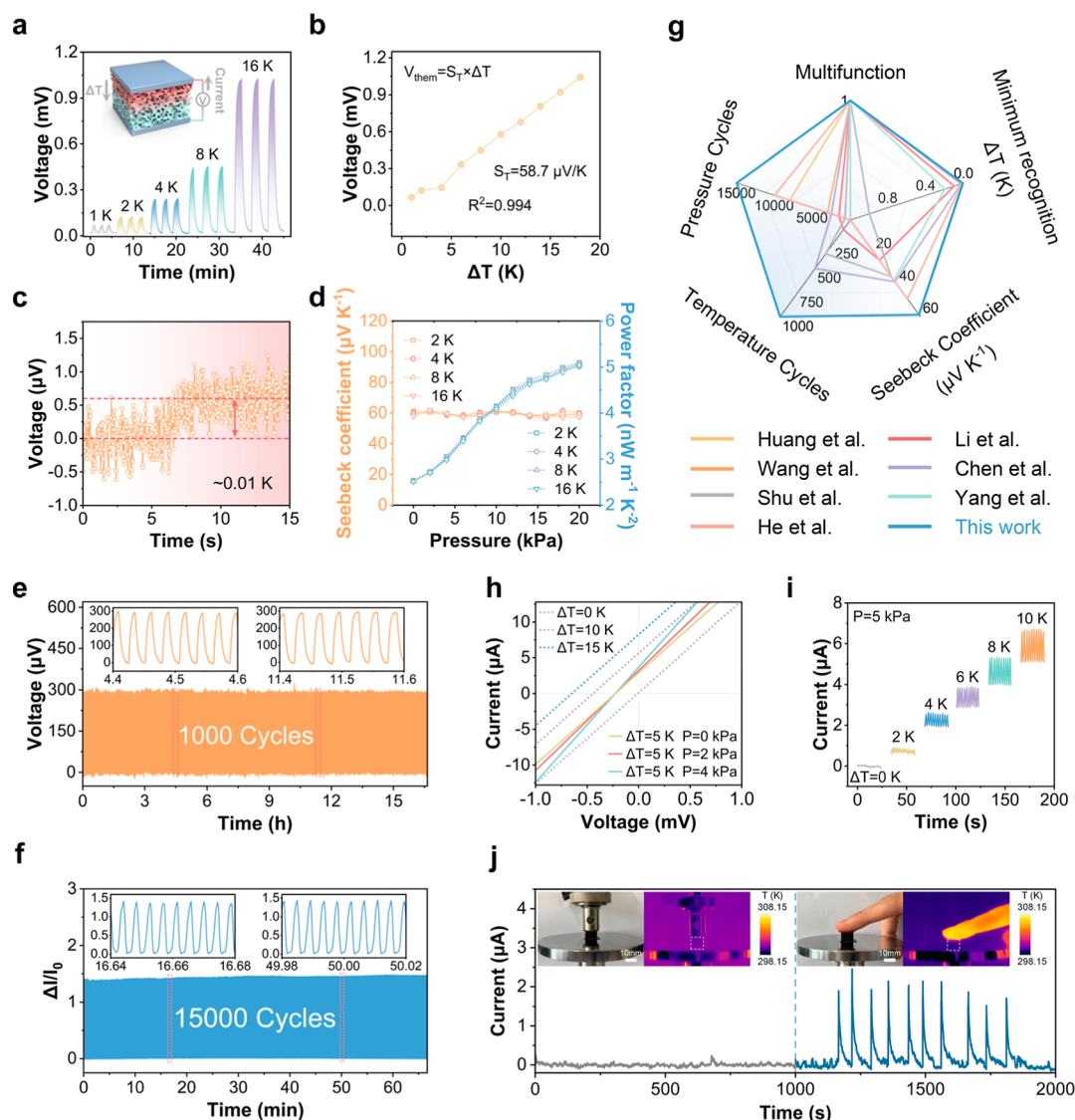


Figure 2. Electrical characteristics of the porous graphene/PDMS smart foam (PGSF). (a) Cyclic response of PGSF to various temperature differences (ΔT). Inset: schematic illustration of the thermoelectric effect, which converts ΔT into an electrical signal, enabling self-powered temperature sensing. (b) Measured output voltage as a function of temperature gradient. (c) Minimum discernible ΔT of PGSF. (d) Seebeck coefficient and power factor of PGSF as a function of pressure level under various temperature gradients (2, 4, 8, and 16 K). (e,f) Durability tests of PGSF under cyclic temperature gradients ($\Delta T = 5$ K) and cyclic compression (50 kPa), respectively. (g) Performance benchmark comparing this study with previously reported dual-mode thermoelectric materials. (h) Measured current–voltage (I – V) curves under various temperature gradients (0, 5, 10, and 15 K) and different strain levels (2 and 4 kPa) corresponding to $\Delta T = 5$ K. (i) Thermoelectric current response characteristics with an applied pressure of 5 kPa under various temperature gradients (0, 2, 4, 6, 8, and 10 K). (j) Current response of PGSF under continuous pressure from a universal testing machine and a finger. Insets show the optical and thermal images corresponding to the pressure applied by the universal testing machine, and the optical and thermal images corresponding to the pressure applied by a finger, respectively.

To further assess environmental adaptability, we systematically evaluated the humidity resistance of the PGSF using a self-built chamber (Figure S5a). The PGSF exhibits inherent hydrophobicity with a water contact angle of $>120^\circ$, repelling various liquids (Figure S5a,b). When relative humidity (RH) increases from 50% to 99%, the Seebeck coefficient and resistance vary by less than 1%, and the thermoelectric voltage under a 5 K gradient remains nearly constant (Figure S5c–e). Stepwise RH changes further confirm that both resistance and thermovoltage are insensitive to continuous humidity variations (Figure S5f,g). These results demonstrate that the PGSF maintains stable thermoelectric output under humid breath environments. Furthermore, device stability and durability are crucial. The PGSF underwent 1000 cycles of temperature

sensing tests under a 5 K gradient and 15,000 cycles of pressure sensing tests under 80% compressive strain. The sensor maintains stable temperature and pressure sensing performance after repeated loading, confirming high durability (Figure 2e,f, Table S2). The pressure-sensing capability originates from the piezoresistive effect, where compression alters the conductive network within the porous structure. This mechanism was verified by real-time LED brightness modulation (Figure S6). Systematic characterization reveals outstanding pressure sensing performance, including high sensitivity (1–50 kPa range), good linearity, fast response/recovery, and exceptional stability with minimal drift after 15,000 cycles, attributed to its robust 3D porous structure (Figure S7). Furthermore, a comparison with recent wearable multifunctional thermoelectric devices shows

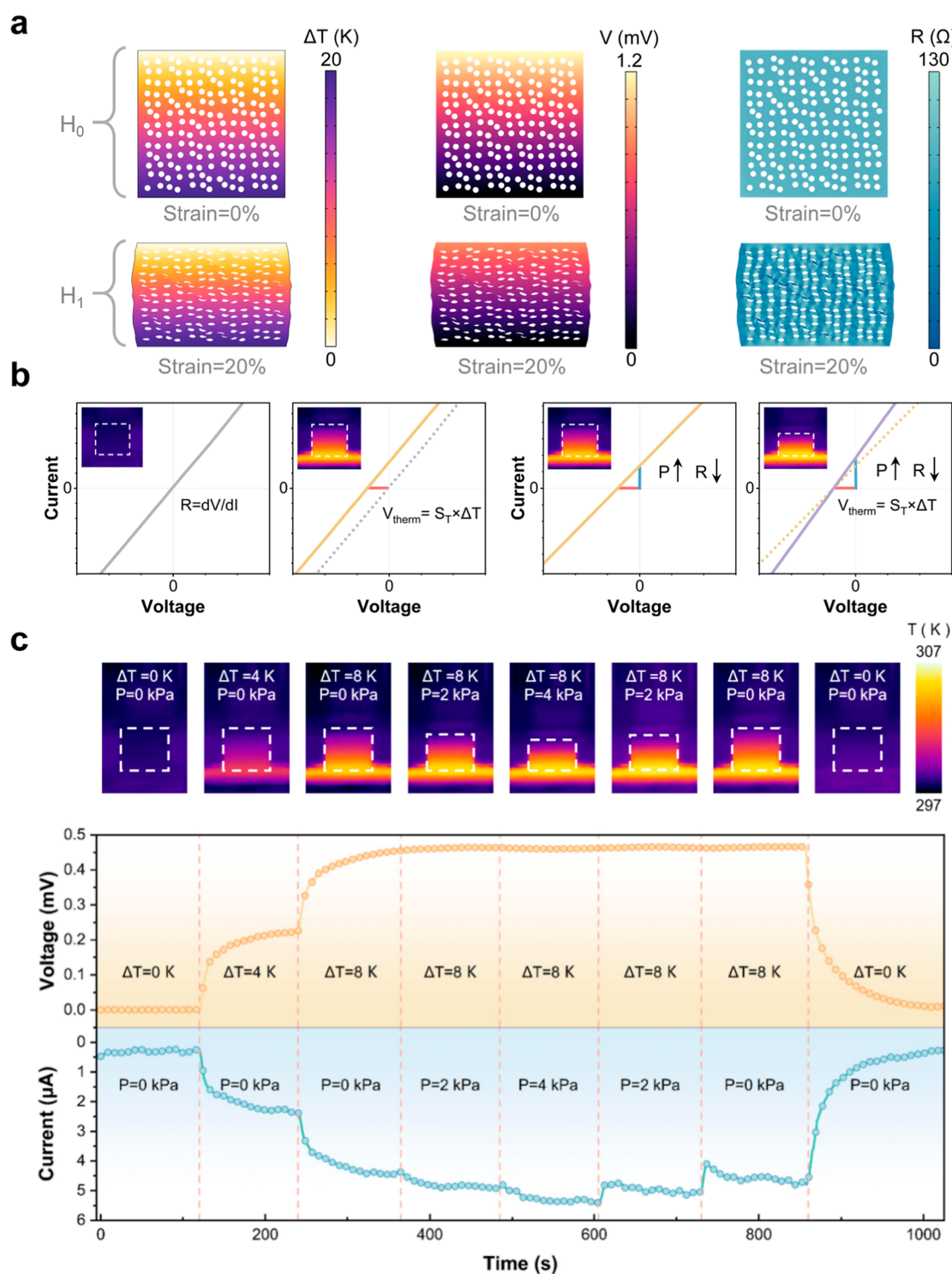


Figure 3. Decoupled sensing mechanism for temperature and strain, enabling continuous real-time monitoring. (a) Finite element simulations of PGSF's temperature distribution, output thermovoltage, and electrical conductivity under compressive strain. (b) Current–voltage (I – V) curves illustrating the mechanical and thermal decoupling mechanism of PGSF under three distinct conditions: (i) no external stimulus, (ii) thermal stimulus only, and (iii) compressive strain applied while a thermal stimulus is maintained. (c) Validation of the temperature and strain decoupling characteristics. Continuous real-time monitoring of output voltage and current responses under temperature gradients (ΔT) ranging from 0 to 8 K and strain levels from 0 to 4 kPa. Insets display infrared thermal images of PGSF under varying temperature gradients and stresses.

that the PGSF sensor exhibits significant advantages in multifunctionality, minimum detectable temperature limit, Seebeck coefficient, and long-term durability, meeting requirements for several cutting-edge performance indicators^{28,30–35} (Figure 2g).

2.2. Unique Electrical Response Characteristics of PGSF under Dual-Effect Decoupling Mechanism

In practical applications, the PGSF is often subjected to coupled fields of pressure and temperature gradients. Experiments demonstrate that its thermal voltage output remains stable

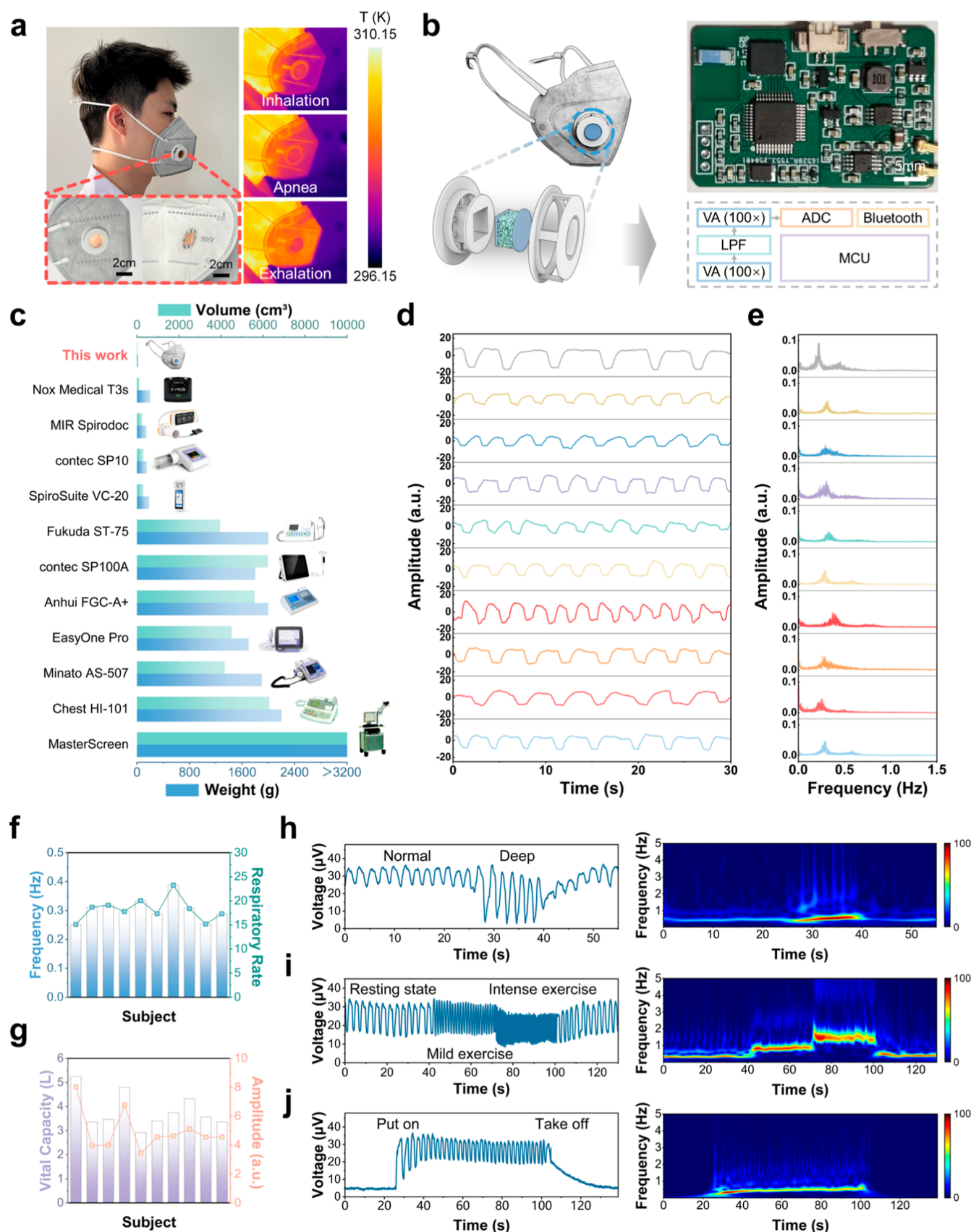


Figure 4. The porous graphene/PDMS smart foam-integrated monitoring mask (PGSF-MM) for real-time respiratory monitoring. (a) Optical image of a user wearing the PGSF-MM, alongside infrared thermal images illustrating temperature variations during different breathing patterns. (b) Schematic design of the PGSF-MM. Insets display a photograph of the circuit design and a block diagram of the system. (c) Comparison of the miniaturization and physical dimensions of the PGSF-MM developed in this study with traditional pulmonary function monitoring devices. (d) In situ respiratory monitoring of ten volunteers using the PGSF-MM, displaying respiratory waveforms during a 30 s monitoring period. (e) Frequency

Figure 4. continued

domain analysis results from 30 min of respiratory monitoring conducted on these volunteers. (f) The frequency of the respiratory signals from ten volunteers demonstrates high consistency with their actual respiratory rates. (g) The average amplitude of the time-domain respiratory signals from ten volunteers exhibits a positive correlation with their vital capacity. (h–j) Time-domain and time–frequency domain signals demonstrating different breathing patterns and actions while wearing the PGSF-MM. (h) Normal and deep breathing. (i) Resting respiration, light exercise, and strenuous exercise. (j) The actions of a volunteer donning and doffing the smart mask.

within a certain range of compressive stress. As shown in Figure 2h, under different temperature gradients (0 K, 5 K, 10 K, 15 K) and under a constant 5 K temperature difference with applied pressures of 2 and 4 kPa, the PGSF's electrical responses conform to the physical models $R = R_0(1 + \sigma\epsilon)$ and $V = S\Delta T$. Resistance (R) is primarily determined by strain (ϵ), while strain's influence on thermal voltage (V) is negligible. The temperature gradient's effect on resistance is also minimal, proving PGSF's robust electrical decoupling capability in coupled force–thermal fields. Benefiting from its thermoelectric conversion mechanism, the PGSF can achieve self-powering under small temperature differences. As shown in Figure 2i, at a 2 K temperature difference (corresponding to $\sim 116 \mu\text{V}$), the base current is $0.65 \mu\text{A}$, and it strengthens with increasing ΔT . Further experiments show that under a fixed 2 K temperature gradient, applying a 5 kPa pressure induces a current change of approximately $0.14 \mu\text{A}$. When pressure increases from 0 to 5 kPa under different ΔT (4–10 K), the current change ranges from 0.37 to $1.65 \mu\text{A}$, with an average increase of $0.16 \mu\text{A}$ per 1 K rise in temperature ($R^2 = 0.997$). This indicates that as the temperature gradient increases, the current output increases linearly and exhibits improved stability and signal-to-noise ratio. Notably, in the device's high-sensitivity pressure range (0–20 kPa), the pressure sensitivity shows no significant variation across different temperature gradients (Figure S8a,b). Furthermore, the current output is more stable and exhibits less dispersion at higher ΔT . Given that temperature gradients of ~ 2 K readily occur during respiratory monitoring, the device typically possesses self-powering capability in application scenarios. Under common environmental temperature differences (2–16 K), the PGSF's thermal voltage output remains stable, unaffected by deformation, and the conductivity and relative resistance change still meet expectations (Figure S8c,d). Analysis reveals that under different constant temperature differences, the maximum fluctuation of the Seebeck coefficient (S) and the calculated power factor for PGSF under various stress conditions (0–40 kPa) is only 3.44%, with no statistically significant difference (Figure 2d). To demonstrate self-powered and multiparameter synchronous sensing capability, as shown in Figure 2j, the PGSF response was monitored under cyclic 5 kPa pressure and finger contact ($\sim 33.6^\circ\text{C}$). Significant current is generated only upon temperature stimulus, and its amplitude is correlated with pressure, validating its concurrent self-powering and pressure–temperature dual-mode synchronous sensing ability.

To gain deeper insight into the coupling mechanism of internal resistance and output voltage in the porous foam structure during dynamic processes, a simulation model was established to systematically evaluate key PGSF parameters—including temperature difference (ΔT), thermal voltage, electrical conductivity, and stress distribution—under different compressive strains^{31,35,36} (Figures 3a and S6g). The underlying physical mechanism for temperature–strain decoupling lies in the three-dimensional interconnected porous architecture of the PGSF, which constructs two orthogonal signal pathways: the

thermoelectricity is driven solely by the temperature gradient, whereas the resistance change is governed exclusively by mechanical strain, achieving a natural physical separation of the two modalities. The electrical signal responses were observed upon independent/coupled application of thermal and mechanical stimuli: ΔT causes a shift in the I – V curve, whereas pressure reduces the resistance and alters the curve's slope by modifying the pore structure and conductive pathways, with these changes being superposed on the existing temperature difference baseline (Figure 3b). As further elucidated in Figure 3b, the I – V curves under three distinct conditions (no stimulus, thermal stimulus only, and combined thermal and compressive stimulus) demonstrate that thermal stimulation vertically shifts the curve without altering its slope, while mechanical stimulation changes the slope without affecting the vertical shift. The resulting I – V curve under dual stimuli is a linear superposition of these two independent effects, with no cross-term present. This key finding indicates that the internal resistance of the PGSF remains unaffected by the inherent temperature variation during the output of thermal voltage. To enhance experimental reliability, the electrical response of the PGSF under continuously varying temperature and stress gradients was further evaluated. As shown in Figure 3c, thermal imaging results confirm that as the temperature gradient increases from 0 to 4 K (corresponding to a thermal voltage of $222.32 \mu\text{V}$) and further to 8 K (corresponding to $465.88 \mu\text{V}$), the temperature difference across the PGSF continuously increases. Throughout this process, the resistance of the PGSF remains constant when no stress is applied (0% strain). When a 2 kPa stress is applied, the current increases to $4.87 \mu\text{A}$; increasing the stress to 4 kPa raises the current to $5.35 \mu\text{A}$; upon stress removal, the current returns to its initial value. These results demonstrate that the resistance change is sensitive only to strain, and the thermal voltage varies strictly with ΔT , with no mutual interference between the two sensing modalities (Figure S9, Movie S1). In summary, this study confirms that the PGSF possesses dual-mode independent detection and outstanding signal decoupling capabilities, which are advantageous for avoiding crosstalk issues caused by mask deformation in respiratory monitoring.

2.3. PGSF-MM for Respiratory Function Monitoring

Respiration, as one of the core human vital activities, can be monitored seamlessly by this device due to its flexibility and high integrability, allowing for easy embedding into daily wearable gear. As shown in Figure 4a, the porous graphene/PDMS smart foam-integrated monitoring mask (PGSF-MM) is visually indistinguishable from a conventional mask. Infrared thermography confirms that temperature fluctuations in the nasal region during inhalation, breath-holding, and exhalation effectively drive changes in the temperature difference (ΔT) across the device, enabling real-time respiration monitoring. The PGSF housing was fabricated via 3D printing, and its structural simplicity favors scalable production. As depicted in Figure 4b, the prefabricated PGSF is placed within the mask; the detected

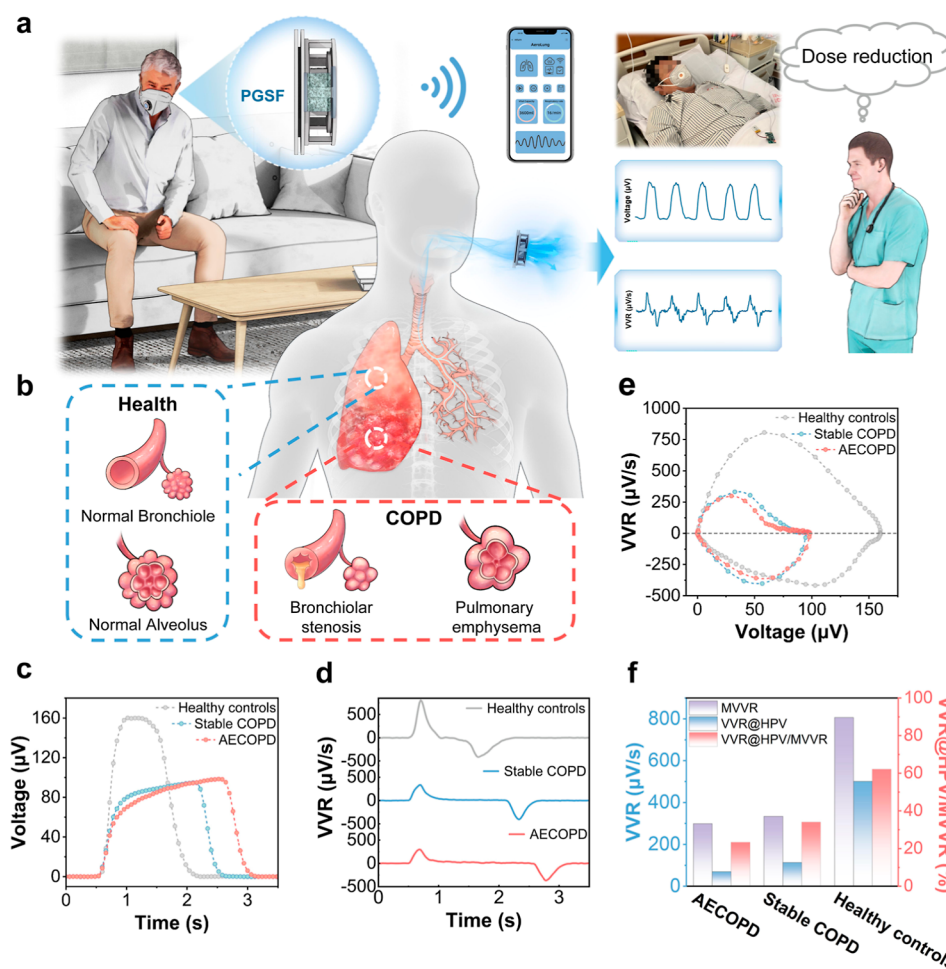


Figure 5. Diagnostic capability of the PGSF-MM for chronic obstructive pulmonary disease (COPD). (a) PGSF-MM-enabled daily pulmonary function monitoring for clinical decision-making. (b) Pathological airway structure in COPD and the functional assessment mechanism based on PGSF-MM. (c) Voltage–time curves for healthy control subjects, stable COPD patients, and patients experiencing acute exacerbations. (d) Voltage change rate (dV/dt) versus time relationship within the study cohort. (e) Signature voltage– dV/dt phase plots for distinguishing respiratory function. (f) Quantitative comparison of kinetic parameters across different subject groups. Sample sizes: healthy control ($n = 3$), stable COPD ($n = 1$), AECOPD ($n = 5$).

thermal voltage signal is conditioned by a preprocessing circuit (amplifier, low-pass filter), quantized by an ADC, and transmitted via Bluetooth to a terminal for respiratory waveform visualization (Figure S10a). As shown in Figure 4c, this integrated system demonstrates significant advantages in miniaturization compared to existing commercial respiratory monitoring and lung function testing equipment. Notably, its core measurement accuracy is comparable to that of commercial spirometers (Table S3). Specifically, the volume accuracy reaches ± 0.012 L, and the flow accuracy reaches ± 0.024 L/s. Both values are conservatively estimated based on analyses of data collected from our volunteer cohort. This high portability greatly enhances its applicability for patients in diverse daily life scenarios. Within this temperature-sensing smart mask, the exhalation process transfers heat to the inner side of the PGSF sensing unit, while inhalation causes its temperature to decrease, corresponding to the rising and falling edges of the thermal voltage signal, respectively (Figure S10b). This mechanism of respiration-driven dynamic temperature changes has been fully validated by real-time thermal imaging monitoring of the oronasal region during breathing (Figure S10c, Movie S2). Furthermore, benefiting from the foam's low thermal conductivity, temperature fluctuations at the outer copper electrode

exposed to ambient air are negligible; concurrently, its large-area design ensures temperature stability on the PGSF's outer side (a feature enhanceable by integrating heat dissipation structures).

To validate the respiratory monitoring efficacy of the smart mask, we analyzed the continuous respiratory waveforms of ten volunteers of different genders, heights, and weights during a 30 min resting state, with data subjected to detrending processing (Figure S10d). The results show that the device can effectively identify respiratory characteristics across different individuals (Figure 4d). Further amplitude and frequency analysis of the volunteer breathing data revealed that the resting respiratory rate for normal adults ranged from 12 to 20 breaths/min (0.2–0.33 Hz). Spectral analysis confirmed that all volunteers' respiratory rates fell within the normal range, while exhibiting individual variations (Figure 4e). Notably, the respiratory frequency derived from the PGSF-MM was highly synchronized with the actual breathing rate obtained by manual breath count (correlation coefficient $r^* = 1$) (Figure 4f). The respiratory signal amplitude showed a significant positive correlation with the individual vital capacity for several subjects (Spearman's rank correlation coefficient $\rho = 0.893$, $p^* < 0.01$) (Figure 4g). Given that the tidal volume is typically one-tenth of the vital capacity, this strongly suggests an inherent link between the

thermoelectric signal amplitude and respiratory volume. Moreover, under real human breathing conditions, the alternating exhalation–inhalation cycle inherently prevents thermal accumulation, ensuring that the sensor operates within its linear range without signal saturation, even during repeated forced expirations. The device's response sensitivity to different breathing patterns was further tested: during shallow and deep breathing monitoring, the waveform amplitude of deep breathing was significantly higher than that of shallow breathing, and time–frequency analysis showed obvious frequency fluctuations accompanying deep breaths (Figure 4h). The device clearly distinguished between three patterns—resting state, rapid breathing, and panting—showing a step-like increase in respiratory frequency in the time–frequency representations (Figure 4i), and effectively captured respiratory dynamics during walking, jogging, and running states (Figure S11). Moreover, the smart mask maintained effective monitoring during mask donning/doffing, indicating potential for real-time monitoring of patient mask-wearing status in healthcare settings (Figure 4j). Building on the PGSF's capability to monitor different respiratory states, its performance in identifying sleep breathing patterns was evaluated. First, a subject with simple snoring was selected for sleep monitoring. By comparing respiratory signals from nonsnoring intervals and snoring intervals in the same subject, temporal and time–frequency domain characteristics of normal breathing and snoring were obtained (Figure S12). Power density analysis revealed that during normal sleep breathing, 93.05% of the subject's respiratory frequency distribution fell within the normal range of 9–16 breaths/min (0.15–0.27 Hz). In contrast, during snoring states, the proportion within this normal frequency range significantly decreased to 83.10% (a reduction of 9.95 percentage points). Specifically, the proportion of low-frequency noise (<0.15 Hz) increased from 0.81% to 5.04%, and the proportion of high-frequency noise (>0.27 Hz) increased from 6.13% to 11.85% (Figure S13). These results confirm that the PGSF possesses high-precision quantitative analysis capabilities for sleep breathing patterns, demonstrating its clear application potential in monitoring sleep breathing disorders such as obstructive sleep apnea–hypopnea syndrome and sleep-related hypoventilation syndrome.

2.4. PGSF-MM for Lung Function Detection

Given the current inadequacy of wearable lung function monitoring technologies and their importance for real-time management and early warning of airflow limitation diseases like COPD, the PGSF-MM system developed in this study provides an effective means for daily lung function monitoring (Table S4). It reflects disease status by continuously tracking respiratory dynamics parameters, assists in the early identification of worsening trends, thereby promoting timely medication adjustment and clinical intervention to delay disease progression (Figure 5a). From a pathophysiological mechanism perspective, as illustrated in Figure 5b, healthy individuals possess intact airway structures, patent lumens, and well-supported cartilage and elastic fibers, allowing alveolar elastic recoil to maintain efficient air expulsion during exhalation. In contrast, COPD patients, due to chronic inflammation and repeated injury, exhibit airway wall thickening, mucus hypersecretion, smooth muscle hypertrophy, and destruction of alveolar attachments. This leads to loss of support structures, particularly in the small airways, making them prone to dynamic collapse during expiration, resulting in airflow limitation, gas

trapping, and persistent expiratory difficulty. To evaluate the potential of PGSF-MM in lung function monitoring, this study organized a healthy control group ($n = 1$), a stable COPD group ($n = 1$), and an acute exacerbation (AECOPD) group ($n = 5$) for forced expiratory tests. As shown in Figure 5c, the healthy control group demonstrated the fastest voltage rise rate and the shortest time to peak, indicating unobstructed expiration. Conversely, the COPD groups (especially AECOPD) showed acceptable initial flow but significant mid-to-late phase limitation with prolonged expiration, suggesting marked airflow obstruction and increased resistance. Due to the lack of a large-sample-derived calibration curve for volume-based parameters, we instead analyzed flow-related metrics by comparing the rate of voltage change (dV/dt) within this small cohort.

Analyzing the thermoelectric signal change rate (dV/dt) to characterize expiratory flow revealed that the healthy control group had a steeper voltage rise slope, and their expiratory flow was significantly higher than that of the patient groups (Figure 5d). This difference primarily stems from the dynamic airway compression phenomenon occurring in the mid-to-late expiration of COPD patients—due to small airway structural abnormalities and weakened elastic recoil, the airways collapse prematurely under intrathoracic positive pressure, leading to air trapping and a sharp drop in flow rate. The PGSF-MM system is designed based on this pathological mechanism, capturing the time–voltage and change rate–voltage relationships during expiration to generate key indicators and curve profiles analogous to traditional lung function parameters. As shown in Figure 5e, the COPD patient groups exhibited a characteristic “scallop” or concave contour in the descending limb of the curve during mid-to-late expiration, providing visual evidence of small airway dysfunction in this disease, whereas the healthy control curve descended smoothly without this feature. To further quantify the degree of small airway obstruction, this study proposes using kinetic parameters of the voltage change rate as evaluation indices, including the maximum voltage

change rate ($\dot{V}_{MVVR} = \left(\frac{dV(t)}{dt}\right)_{\text{Max}}$, corresponding to peak expiratory flow) and the voltage change rate at half peak voltage ($\dot{V}_{VVR@HPV} = \left(\frac{dV(t)}{dt}\right)_{\text{HPV}}$, corresponding to mid-to-late expiratory flow). To eliminate interindividual differences in lung volume and ventilatory capacity, a standardized parameter—the small airway obstruction index (SAOI) $\dot{V}_{VVR@HPV/MVVR} = \left(\frac{dV(t)}{dt}\right)_{\text{HPV}} / \left(\frac{dV(t)}{dt}\right)_{\text{Max}}$ —was constructed.

As shown in Figure 5f, the healthy control group had higher values for both $\dot{V}_{MVVR}$ and $\dot{V}_{VVR@HPV}$ due to superior lung function. The $\dot{V}_{VVR@HPV/MVVR}$ was above 50% in the healthy control group, while it was 34.1% and 23.4% in the stable COPD and AECOPD groups, respectively, with significant differences between groups. Since flow analysis (dV/dt) still depends on a reference database, we adopted the ratio of peak to mid-to-late expiratory flow to define a self-referential metric, which can be readily compared across individuals without relying on a database, effectively capturing differences in small airway ventilation over time. Thus, the SAOI serves as a quantitative respiratory indicator in the current proof-of-concept study; with future expansion of the cohort, this index can be readily calibrated against clinical gold standard parameters. The decrease in this index directly reflects the severity of small airway obstruction, attributable to mechanisms including airway

narrowing, mucus plugging, and loss of alveolar elastic support, leading to accelerated decay of expiratory flow in the mid-to-late phase. Therefore, $\dot{V}_{VVR@HPV}/MVVR$ can serve as a sensitive, quantitative indicator for assessing small airway dysfunction, suitable for daily monitoring and exacerbation early warning in COPD.

3. CONCLUSIONS

In this study, we have engineered a wearable respiratory monitoring system (PGSF-MM) that provides continuous, high-fidelity tracking of pulmonary mechanics in ambulatory settings. The system's performance is rooted in a novel porous graphene/PDMS composite foam, which features a unique architecture designed to physically decouple thermoelectric and mechanical sensing pathways. This intrinsic decoupling mechanism effectively nullifies motion artifacts and signal crosstalk—a persistent challenge that has limited the reliability of previous wearable sensors. This high degree of signal fidelity enables the extraction of a new, physiologically grounded biomarker: the small airway obstruction index $\dot{V}_{VVR@HPV}/MVVR$. This index quantitatively captures the accelerated flow decay characteristic of small airway collapse in COPD, a direct manifestation of the disease's underlying pathophysiology. Our clinical validation demonstrates that the SAOI robustly stratifies healthy individuals, stable COPD patients, and those experiencing acute exacerbation. Critically, it provides early warning of impending exacerbations, offering a crucial window for pre-emptive clinical intervention. By matching clinical spirometer accuracy while offering superior wearability, PGSF-MM bridges episodic clinical assessment and continuous disease management. It represents a paradigm shift from institution-based testing to patient-centered, home-based monitoring. Ultimately, this work not only introduces a powerful tool for the proactive management of COPD but also establishes a versatile technological platform for developing next-generation wearable diagnostics.³⁷ Though limited by the current sample size, future expansion of cohort studies and integration of deep learning algorithms will further enhance monitoring. By translating a fundamental materials science innovation into a clinically validated solution, we pave the way for a future of predictive, personalized medicine for chronic conditions.

It should be acknowledged that the respiratory monitoring validation in this study involved a relatively small sample size, which was determined based on exploratory objectives without formal statistical power analysis. The limited sample may constrain the generalizability of the findings across diverse populations with varying sex, age, and body habitus. Therefore, the current observations regarding respiratory waveform characteristics and pulmonary function parameters should be interpreted as proof-of-concept. Future studies with larger, independent cohorts and formal sample size estimation are warranted to further validate the generalizability and sensitivity of the SAOI across different stages of COPD and in broader populations.

4. EXPERIMENTAL SECTION

4.1. Materials

During the fabrication of the porous graphene/PDMS composite smart foam (PGSF), graphene was procured from Kenye Technology Co., Ltd. (Deyang, Sichuan, China). Sodium chloride (NaCl) powder (particle size: 0.4 mm) was sourced from Tianjin Zhiyuan Chemical

Reagent Co., Ltd. (Tianjin, China). The PDMS precursor was obtained from Dow Silicones Co., Ltd. (Germany).

4.2. Fabrication of PGSF

The porous graphene/polydimethylsiloxane (PDMS) composite smart foam (PGSF) was fabricated using the following procedure. First, 70 mg of graphene, 9 g of sodium chloride (NaCl) powder (particle size: 0.4 mm), and 1 g of PDMS precursor were weighed. The PDMS precursor was mixed with its curing agent at a mass ratio of 10:1 to prepare the PDMS elastomer. Subsequently, the weighed graphene, NaCl powder, and PDMS elastomer were thoroughly mixed. This mixture was then placed into a cubic mold with an edge length of 10 mm and compacted to a height of 10 mm. The mold was transferred to an oven and cured at 80 °C for 2 h. Following curing, the mixture was immersed in hot water for 12 h to dissolve the NaCl particles, thereby generating the porous structure. Finally, the resulting sponge was dried in an oven at 80 °C for 2 h, yielding the final PGSF.

4.3. Characterization and Performance Evaluation

The microstructure and surface morphology of the samples, including surface features and pore distribution, were characterized using scanning electron microscopy (SEM, ZEISS Sigma 300). Compression deformation behavior and mechanical response were quantified using an electromechanical tensile testing machine (ZHIQU ZQ-990B). A comprehensive test platform was constructed to simulate coupled multiphysical field environments. This system integrated Peltier elements (TEC1-04903) with an adjustable DC stabilized power supply (MAISHENG MS155DS) to generate controlled pressure and temperature gradients. During testing, thermoelectric and other electrical signals were acquired using a digital source meter (Keithley 2611B). Temperature distribution was monitored and recorded via an infrared thermal imager (FOTRIC 618C). Triboelectric signals were measured separately using an electrometer (Keithley 6514). Additionally, a porous foam model was developed using COMSOL Multiphysics software. This enabled dual-effect simulation studies to investigate the sample's performance characteristics and underlying mechanisms under complex operating conditions.

4.4. Ethics Statement

This study involving human participants was reviewed and approved by the Ethics Committee of the First Affiliated Hospital of Xinjiang Medical University (ethics approval number: K202412-23). All experiments were performed in accordance with the approved guidelines and regulations. Written informed consent was obtained from all individual participants involved in the study.

■ ASSOCIATED CONTENT

Data Availability Statement

All data supporting the findings of this study are available within the article and its [Supporting Information 1](#), [Supporting Information 2](#), [Supporting Information 3](#), [Supporting Information 4](#) files. The underlying data for the graphs have been provided in the Source Data file. Custom code used for signal processing and analysis is available from the corresponding author upon reasonable request. The materials described in this study can be obtained from the corresponding author under a materials transfer agreement for academic research purposes.

Supporting Information

The Supporting Information is available free of charge at <https://pubs.acs.org/doi/10.1021/acsnano.6c03384>.

Informed consent form (PDF)

Robust thermoelectric voltage output of PGSF under varied mechanical strain (Movie S1) (MP4)

Demonstration of PGSF-MM for respiratory monitoring (Movie S2) (MP4)

Scanning electron microscope images of graphene material (Figure S1); microstructure of porous gra-

phene/PDMS foam and pure PDMS foam (Figure S2); test platform for applying controlled levels of ΔT and strain (Figure S3); thermoelectric output performance of PGSF (Figure S4); humidity resistance and environmental stability of the PGSF (Figure S5); piezoresistive effect and physical demonstration of PGSF (Figure S6); characterization of pressure sensing and stability performance in PGSF (Figure S7); electrical characteristics of PGSF under simultaneous application of temperature gradient and strain without external power supply (Figure S8); continuous real-time monitoring of resistance changes under temperature gradients and applied strain (Figure S9); design and monitoring principles of PGSF-MM (Figure S10); respiratory waveforms, rhythm, and time–frequency domain analysis under different exercise states (Figure S11); waveform characteristics and time–frequency domain analysis of normal breathing and snoring during sleep (Figure S12); spectral analysis of normal breathing and snoring during sleep: classification and quantitative evaluation of respiratory frequency (Figure S13); comparison of temperature sensing performance (Table S1); comparison of durability performance (Table S2); comparison of volume accuracy and flow accuracy across commercial spirometers and the present device (Table S3); PGSF-MM respiratory monitoring advantage comparison (Table S4) (PDF)

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Author Contributions

◆H.S. and H.L. contributed equally to this work. H.W. and C.W. conceived the idea and supervised the research; H.S. and H.L. fabricated the device and optimized its performance; H.F., C.H., and X.C. assisted in the synthesis of the samples and experimental investigations; S.J., X.H., and D.Z. assisted with performance testing and simulation studies; Z.L., X.Z., and Y.X. guided and optimized the clinical testing part. H.S. wrote the initial draft, and all authors read, discussed, and edited the manuscript.

Notes

The authors declare no competing financial interest.

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