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Review

Wearable Applicability of Respiratory Airflow Transducers: Current Approaches and Future Directions

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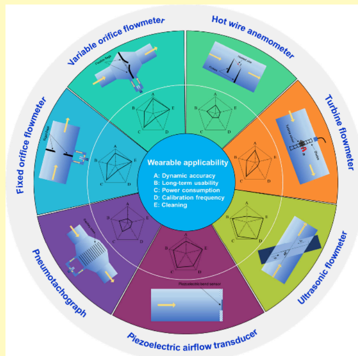
ABSTRACT: Advanced technologies employed in modern respiratory airflow transducers have exhibited powerful capabilities in accurately measuring respiratory flow under controlled and sedentary conditions, particularly in clinical settings. However, the wearable applicability of these transducers as face-mounted electronics for use in occupational and sporting activities remains unexplored. The present review addresses the critical wearability issue associated with current respiratory airflow transducers, including pneumotachographs, orifice flowmeters, turbine flowmeters, hot wire anemometers, ultrasound flowmeters, and piezoelectric airflow transducers. Furthermore, a comprehensive analysis and comparison of all factors that impact the wearable applicability of respiratory airflow transducers are conducted, considering dynamic accuracy, long-term usability, power consumption, calibration frequency, and cleaning requirements. The findings indicate that the piezoelectric airflow transducer stands out as a more viable option for wearables compared to other devices. We expect that this review will serve as a valuable engineering reference, guiding future research efforts in designing and developing wearable respiratory airflow transducers for ambulatory respiratory flow monitoring.



KEYWORDS: *respiratory airflow, wearable applicability, pneumotachographs, fixed orifice flowmeters, variable orifice flowmeters, turbine flowmeters, hot wire anemometers, ultrasound flowmeters, piezoelectric airflow transducers*

The respiratory signal is one of the most informative vital signs, providing various clinical and physiological indicators.[1] For instance, a respiratory pattern is an essential biomarker for various pulmonary diseases such as acute respiratory syndrome (ARDS), chronic obstructive pulmonary disease (COPD), and pulmonary edema.[2] Respiratory frequency is also crucial in sports science during high-intensity training.[3,4] The forced expiratory volume (FEV) and forced vital capacity (FVC) are two essential measurements in the diagnosis of obstructive and restrictive lung diseases.[5] Pulmonary ventilation is a significant metabolic rate parameter closely related to human physiological strain.[6,7] Therefore, accurate and continuous respiratory flow monitoring is indispensable in clinical settings and occupational/sports environments.[1,8]

Several types of respiratory airflow transducers, such as pneumotachographs (PTs), orifice flowmeters (OFs), turbine flowmeters (TFs), hot wire anemometers (HWAs), ultrasound flowmeters (UFs), and piezoelectric airflow transducers (PATs) have been employed to measure respiratory flow, as shown in Figure 1. These measurement techniques have been validated under stationary and controlled conditions, particularly in clinical environments.[9−11] To be suitable for use in occupational and sporting activities, respiratory airflow transducers must be securely attached to the human face using respiratory protective equipment, ensuring their wearability. However, due to their inherent characteristics, such as operating principles, components, and design structures, these transducers exhibit varying levels of wearable performance. It is, therefore, necessary and meaningful to analyze their wearable applicability to provide helpful information for developers and engineers.

Hence, we provided a detailed description of the working principles, component materials, and design structures of current respiratory airflow transducers. Based on this information, we evaluated their advantages and limitations in their wearable applicability in terms of dynamic accuracy, long-term usability, power consumption, calibration frequency, and cleaning. Our analysis indicates that the piezoelectric airflow transducer emerges as a particularly promising option for wearables, surpassing other existing flowmeters. This article aims to serve as a valuable resource for engineers and researchers seeking to design and develop the next generation of wearable airflow transducers.

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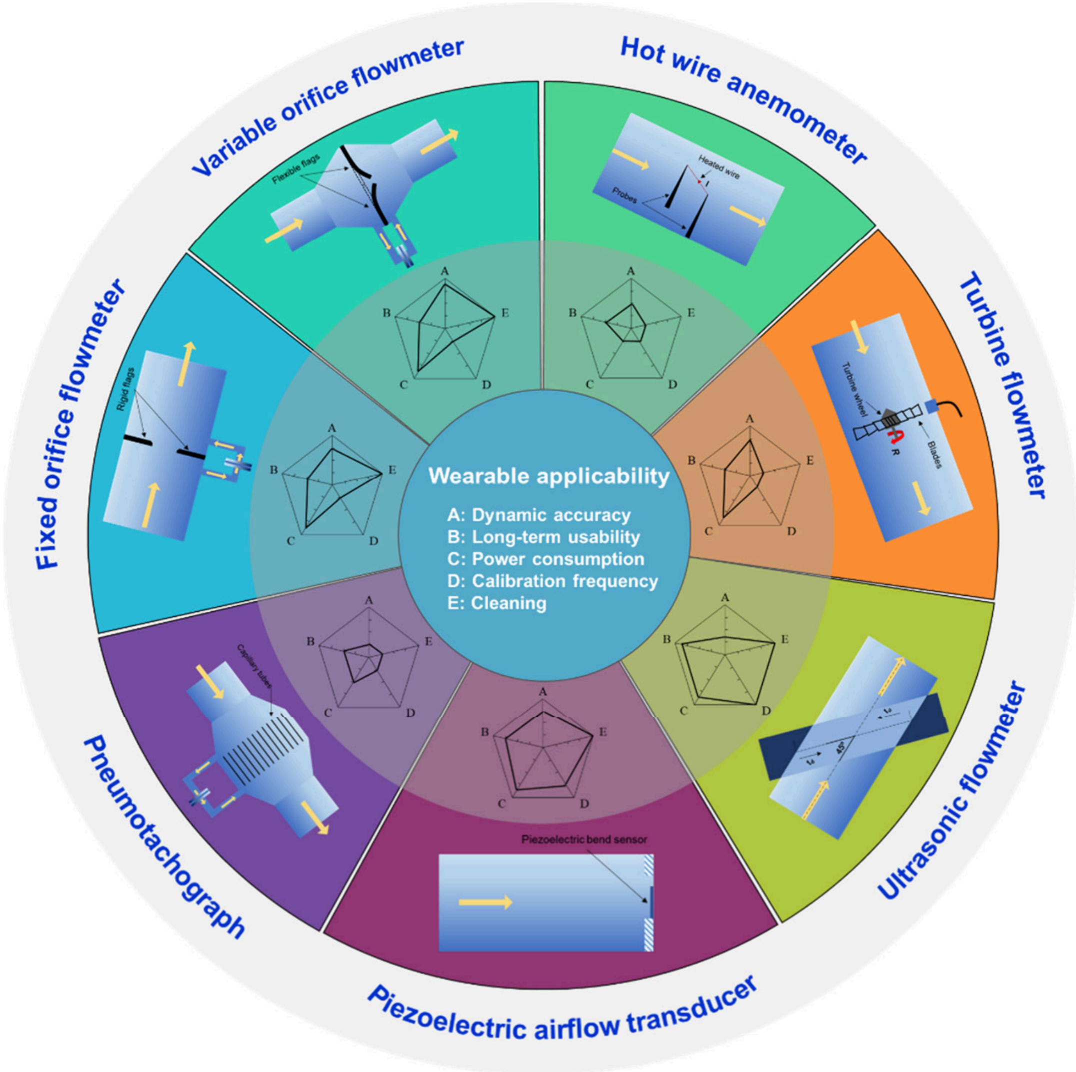

Figure 1. Current respiratory airflow transducers for respiratory flow measurement and their wearable applicability.

TECHNICAL REQUIREMENTS FOR WEARABLE RESPIRATORY AIRFLOW TRANSDUCERS

Numerous parameters, including accuracy, sensitivity, response time, measurement range, and breathing resistance, impact the performance of respiratory airflow transducers. Nonetheless, the majority of these parameters have been primarily validated for traditional respiratory flow measurement under stationary and controlled conditions.[9−11] Therefore, we try to explore the additional technical requirements for wearable respiratory airflow transducers in this section.

To evaluate the wearable applicability comprehensively, the following particular technical parameters for the respiratory airflow transducers are proposed:

(1) **Dynamic accuracy**, which is defined as the accuracy achieved under ambulatory and open environments, is a key evaluation index for wearable airflow transducers. The dynamic accuracy could be influenced by external factors such as gravity, motion artifact, noise, etc.

(2) **Long-term usability.** After long-term use, the fully saturated exhaled air quickly condenses inside the airflow transducers. The condensate can block the air pathway in some airflow transducers, reducing their long-term usability. One of the significant challenges in wearable respiratory monitoring is avoiding issues caused by water vapor condensation. Additionally, long-term usability is influenced by both the durability and degradation of components over time.

(3) **Power consumption.** To enhance mobility, wearables must be battery-powered devices. Power consumption becomes a critical aspect for battery-operated wearable respiratory airflow transducers, especially those that rely on a heating system. This review discusses and compares several factors that impact the power consumption of existing airflow transducers.

(4) **Calibration frequency.** Airflow transducers with periodic calibration or calibration-free offer great benefits in wearable applications because the calibration procedures for some airflow transducers are pretty tricky and typically require special devices.

(5) **Cleaning.** Airflow transducers are highly susceptible to contamination after use, necessitating regular cleaning and disinfecting. However, the complex structures of some airflow transducers can pose a challenge during the cleaning process.

The technical parameters such as dynamic accuracy, long-term usability, power consumption, calibration frequency, and

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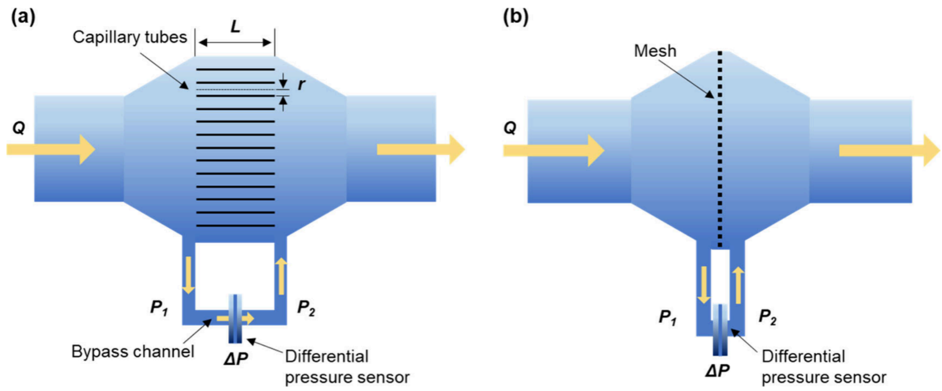

Figure 2. Schematics of (a) Fleisch and (b) Lilly pneumotachographs.

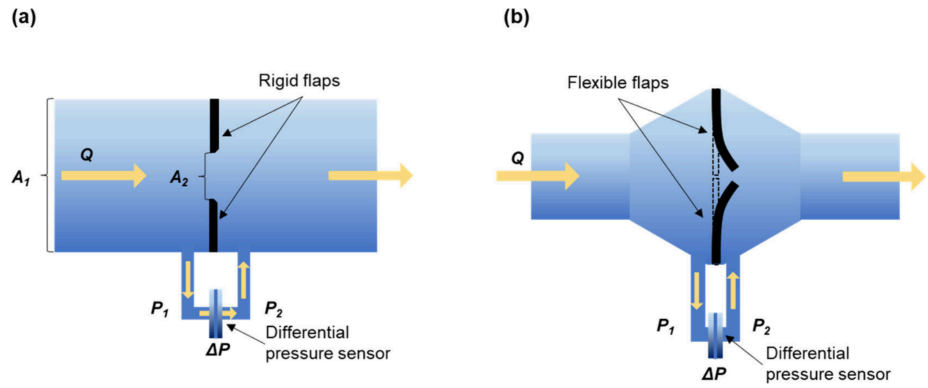

Figure 3. Schematics of (a) a fixed orifice flowmeter and (b) a variable orifice flowmeter.

cleaning difficulty could be of minor importance for traditional applications. In contrast, they could become primary factors when designing and developing wearables. Therefore, we focus on these five selection criteria to evaluate the wearable applicability of current airflow transducers.

■ WEARABLE APPLICABILITY EVALUATION

In this section, we present an overview of the operating principles, components, and design structures of airflow transducers commonly used in respiratory flow measurement. From this foundation, we deduce their technical strengths and weaknesses in areas such as dynamic accuracy, long-term usability, power consumption, calibration frequency, and cleaning, with the aim of assessing their suitability for use in wearable technology.

Pneumotachographs. A pneumotachograph (PT) generally consists of a conduit, a flow resistor, and a differential pressure sensor, as illustrated in Figure 2. There are two types of PTs, a Fleisch PT (Figure 2a) comprising a bundle of capillaries as a flow resistor and a Lilly PT (Figure 2b) with a flow resistor of a fine wire mesh.[12,13] The roles of flow resistors, i.e., capillary tubes and fine wire mesh, are to create measurable differential pressure between the upstream and downstream sides of the resistors and rectify the turbulent flow of the airstream into the laminar flow.[14] The differential pressure sensor then detects the resulting pressure through a bypass channel connecting the conduit with the differential pressure sensor (Figure 2a). Most differential pressure flowmeters adopt such a bypass configuration design. If the air fluid runs entirely in the conduit, the pressure drop across the flow resistor is proportional to the input flow rate under the laminar regime based on the Hagen–Poiseuille law. The linear relationship between the pressure drop and the flow rate for the Fleisch PT is expressed as below:[14]

$$\Delta P = \frac{8 \times \mu \times L}{n \times \pi \times r^4} Q \qquad (1)$$

where ΔP is the pressure drop, Q is the flow rate, L is the capillary tube length, r is the capillary tube internal radius, n is the number of capillary tubes, and μ is the dynamic viscosity of the fluid. According to eq 1, the Fleisch PT's sensitivity depends on the flow resistor's geometry, especially the radius of capillary tubes. It is noteworthy that because the breathing resistance of Fleisch PT elevates with increasing sensitivity, the design of the flow resistor should be optimized for various purposes.[15] Fleisch PT's accuracy is influenced by the physical characteristics of the breathing air (i.e., dynamic viscosity, μ).[16−18] For instance, the viscosity of water vapor is approximately half that of air, and only 5% of water vapor is in the fully saturated exhaled air at 37 °C, but this causes a measurable impact on the viscosity of air, causing inaccurate measurement.[19] Similarly, changes in the content of the oxygen and carbon dioxide between inhaled and exhaled gases could affect the accuracy of Fleisch PT.[20] Because the core body heat warms the lung's air, the expired air usually is warmer than the inhaled air, resulting in air expansion to alter air viscosity as well.[21]

PTs are the established technology and are regarded as the gold standard method to measure respiratory flow and volume in various clinical settings due to the advantages of high accuracy and linear response under sedentary conditions. One decisive drawback for PTs is their dynamic accuracy because the signal of PTs is sensitive to gravity.[22] Moreover, the PTs should be operated at a condition that their measurement and calibration

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are in the same direction. Therefore, the PTs are typically used at static status, not for ambulatory conditions. As there is no moving part, the durability of PTs is not doubtful. However, water vapor can readily condense inside the flow resistor to clog the capillaries or the fine mesh, limiting its long-term usability.[23] The condensation effect was observed after five consecutive blows at an ambient temperature of 20 °C, and the error reached 7% in respiratory volume measurement.[17] The PT head is generally electrically heated to avoid such condensation, but it could significantly consume electric power. The elevated power consumption of PTs is unsuitable for use in battery-based wearable devices. PTs require frequent calibration.[24] They are difficult to be cleaned due to the thin capillaries or the fine mesh.[25]

Orifice Flowmeters. An OF is generally composed of a conduit, an orifice plate, and a differential pressure sensor. There are two configurations of OFs, which differ from each other in the orifice plate characteristics. The fixed OF (F-OF) comprises a fixed orifice plate made of rigid flaps. The variable OF (V-OF) is constituted by a variable orifice plate made of a couple of flexible flaps, as illustrated in Figure 3. The working principle of OFs is similar to that of PTs because both OFs and PTs are classified into the same category of differential pressure flowmeter. For OFs, the orifice plates act as flow resistors to produce measurable pressure drops between two sides of the orifice plate. Given that the gas passing through the conduit is one-dimensional flow and under conditions of incompressible, nonviscous, and isothermal fluid, a relationship between the pressure drop and the input volumetric flow rate can be determined by the geometry of F-OF based on the following Bernoulli's equation:[26]

$$Q = \frac{A_2}{\sqrt{1 - (A_2/A_1)^2}} \times \sqrt{\frac{2}{\rho}} \times \sqrt{\Delta P} \tag{2}$$

where ΔP is the pressure drop, Q is the flow rate, A_1 is the inlet area of the F-OF, A_2 is the passage area of the orifice place, and ρ is the fluid density. According to eq 2, the flow rate is proportional to the pressure drop's square root. The non-linearity response is the drawback of these flowmeters because the lower the flow rate, the lower the signal.[26] Conversely, the V-OF has a linear relationship between the volumetric flow rate and the pressure drop. The V-OF comprises the orifice plate made of a pair of flexible flaps so that the airflow rate varies the passage area of the orifice plate, e.g., the higher airflow, the wider passage area, and vice versa. As a result, the V-OF automatically achieves a mechanical linearization between the pressure drop and the flow rate by altering its flow resistance.[9,25] The precision of OFs is also affected by the air constituent and temperature because eq 2 is a function of the air density.[26]

F-OFs have a nonlinear (power fuction) relationship between input flow rate and pressure drop, showing limited accuracy under both dynamic and static conditions. Although V-OFs have linear responses, there is no reference regarding their dynamic accuracy. However, it can be supposed that the flexibility of the flaps plays a critical role in dynamic accuracy because flaps that are too flexible could be susceptible to vibration, whereas those that are too stiff increase breathing resistance. One significant advantage of OFs is that they are unaffected by water vapor condensation, unlike other airflow transducers.[27] As shown in Figure 4, although a large amount of water was observed in a V-OF after 4 h of the trial, its sensitivity did not change. Nevertheless, there is still a high possibility that the condensed water vapor will accumulate inside the bypass channel and eventually block the channel after long-term use, leading to incorrect measurement. Moreover, to reduce signal noise, the bypass channel diameter should be small enough,[28] accelerating the blocking. In this context, the bypass configuration design for OFs and PTs could reduce their long-term usability. OFs demand frequent calibration,[29] and they are easier to be cleaned because of their simple and robust structure compared to PTs.

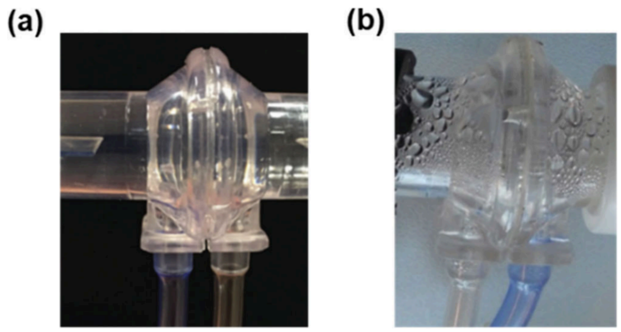

Figure 4. Pictures of a V-OF (SpiroQuant P by EnviteC, Honeywell) showing (a) before and (b) after water vapor condensation. Reproduced with permission of ref 27. Copyright 2015 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim.

Hot Wire Anemometers. An HWA consists of two probes with a heated wire stretched between them, as shown in Figure 5.

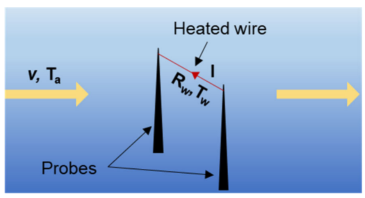

Figure 5. Schematic of a typical hot wire anemometer.

It is designed following King's law, based on which heat (Joule effect) dissipated from a heated wire by an airstream is proportional to the square root of the airstream's velocity as in the equation below.[30]

$$I^2R_w = (\alpha + \beta\sqrt{v})(\mathrm{T}_w - \mathrm{T}_a) \tag{3}$$

where I is the current supplied by the circuit, α and β are two empirical constants related to the geometric properties of wire and the air features, respectively, v is the airstream velocity, T_w is the wire temperature, R_w is the wire resistance at T_w and T_a is the airstream temperature. Since the wire resistance depends on the wire temperature, eq 3 is related to three variables: the airstream velocity, the current, and the wire temperature (or the wire resistance). If the current or the wire temperature is maintained constant, the airstream velocity will only be related to a single variable.[31] Therefore, there are two types of HWAs: a constant temperature HWA (CT-HWA) and a constant current HWA (CC-HWA). The CT-HWA is based on holding the wire temperature at a constant value so that the current depends on the airstream speed. In the CC-HWA, the current of hot wire maintains at a value; thereby, the temperature changes with the airstream speed. Compared with the CC-HWA, the CT-HWA has more merits in terms of its usability and durability. For instance, the CT-HWA operates at a constant temperature, preventing itself from burning abruptly when the airstream speed decreases. The CT-HWA has a linear relationship

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between the output voltage and the air velocity by modifying the electrical circuit.[31] Several factors influence the accuracy of HWAs. To be specific, eq 3 is a function of the airstream temperature, implying the accuracy is affected by breathing air temperature.[32,33] Moreover, the constant, α, in eq 3 is related to free convection that relies on the direction of the heated wire (horizontal or vertical); thereby, the calibration and measurement should be carried out in the same orientation.[31] The presence of water vapor in the air increases the thermal conductivity, influencing the heat exchange in the airstream and the heated wire, causing measurement errors. A change in the relative humidity from 25 to 70% at 25 °C leads to an additional 2% heat loss per degree rise of the wire temperature.[34]

HWAs take the merits of high sensitivity at low airflow and short response time.[11,35,36] As mentioned above, the calibration and operation of HWAs should be performed in the same direction,[31] but it is not always ensured during activity, limiting the dynamic accuracy. Water vapor condensation is another inherent issue for applying HWAs in wearable respiratory monitoring. As long as the condensed water vapor in exhaled air contacts the heated wire, which rapidly increases heat transfer, it will lead to inaccurate airflow measurement.[34,37] Another shortcoming of HWAs is their durability due to the fragile heated wire[1,37] that is usually made of tungsten or platinum and could be easily destroyed. In this context, a hot film-based anemometer may be more appropriate for respiratory monitoring, which is more rugged[38] and has lower signal noise[39] than those of HWAs. HWAs probably require much more power to heat the wire than another airflow transducer, limiting them in applications of battery-based wearable devices. Another concern for applying HWAs in respiratory monitoring is related to the discrimination of the respiratory airflow direction because an HWA with a single hot wire is unable to determine the airflow direction. This issue can be solved by employing more complex configurations using at least two hot wires,[9,40] as illustrated in Figure 6, but it further elevates the power consumption. HWAs require frequent calibration and are fairly difficult to calibrate.[37] Cleaning difficulty is another drawback for HWAs because of the fragile wire.[37]

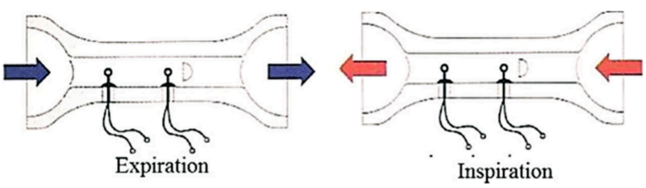

Figure 6. Schematic of an HWA comprising two heated wires to discriminate flow direction. Reproduced with permission of ref 9.

Turbine Flowmeters. A TF comprises a conduit, a multibladed turbine wheel, and a pickoff sensor, as illustrated in Figure 7. The TF is designed so that air passes through a conduit, in which a turbine wheel with multiblades is perpendicularly mounted to the airflow direction to rotate the turbine. The rotational speed of the turbine wheel is then recorded by the pickoff sensor, converting each passage into an electrical impulse. The fundamental concept of TFs is based on the linear relationship between rotational speed and airflow rate through the conduit as the following expression:[41]

$$Q = k \times R \tag{4}$$

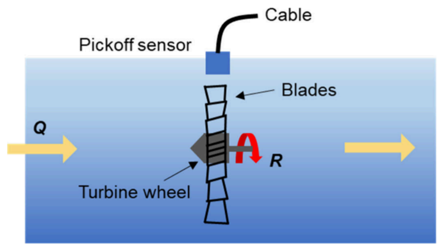

Figure 7. Schematic of a typical turbine flowmeter.

where Q is the airflow rate, R is the rotational speed of the turbine wheel, and k is the constant, which is related to the geometry of TF, but is independent of gas properties such as kinematic viscosity. This linear response is only valid when the Reynolds number is bigger than 800−1000.[1] Therefore, TFs are typically used at high flow rates and low viscosity fluid. Still, they could also be employed to measure low flow rate like respiration by adequately modifying the components of TFs.[42,43] It is noteworthy that TFs do not require any temperature compensation and are not influenced by either humidity or altitude variation.[44,45] In addition, it seems difficult to correctly measure inspired and expired gases together because of the rotational inertia of the turbine, called "turbine hysteresis".[44,46]

TFs have been used to monitor respiratory signals in various sports and exercise settings as a proven technology,[1] because these types of flowmeters are not influenced by air components and temperature, and they are inherently immune from motion artifact.[43] Two wearable metabolic trackers, such as COSMED K5 and Metamax 3B have utilized TFs as flowmeters in their equipment to measure respiratory flow under dynamic and open environments.[6] However, TFs present inferior accuracy than HWAs, PTs and UFs in low airflow measurement,[11] and the dynamic accuracy of the TFs significantly reduces at an airflow rate lower than 4 L/m.[45] Due to the degradation of moving parts like thrust bearings,[47] TFs have limited long-term usability. TFs require frequent calibration and need a special calibration syringe.[14] Cleaning difficulty is another issue for TFs owing to the complex structure of the turbine.

Ultrasonic Flowmeters. There are several operating principles for UFs, such as time-of-flight, phase shift, sing around, and so forth.[46] The only time-of-flight-based UF (TOF-UF) is reviewed in this work because it is the most frequently applied in commercial equipment for respiratory signal measurement. A TOF-UF consists of a conduit and a pair of piezoelectric transducers (PT), and two piezoelectric transducers are placed face-to-face at the extremities of the cross-sectional conduit, whose axis lies at an angle of 45° to the airflow axis, as shown in Figure 8. As a result, they can exchange ultrasonic signals alternately by rapidly altering their roles as transmitters and receivers. Since the distance (L) between two piezoelectric transducers is constant, the transit time going upstream (t_u) is decreased with the airflow while the transit time going downstream (t_d) is increased. The difference in two transit times can be used to determine the air velocity (v) and hence of flow;[48] therefore, this operating principle is called "time of flight".

The volumetric flow rate can be estimated by eq 5[48]

$$Q = \frac{\pi r}{4} \times c^2 \times \Delta t \tag{5}$$

where Q is the volumetric flow rate, r is the radius of the tube, c is the velocity of the ultrasonic signal, i.e., the acoustic velocity in

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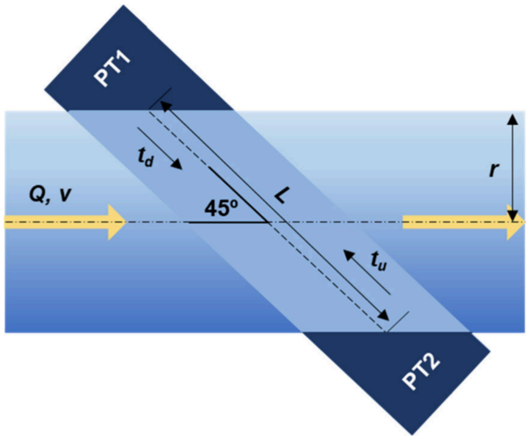

Figure 8. Schematic of a "time of flight" based ultrasonic flowmeter.

air medium, Δt is the difference time between transit times. As seen from eq 5, the linear relationship between the volumetric flow rate and the differential transit time is greatly influenced by the acoustic velocity. As a result, the accuracy of TOF-UF could be affected by the composition, temperature, and moisture content of breathing air because these parameters alter the velocity of sound propagation.[48] For instance, it leads to a maximum measurement error of 3% in pure oxygen medium after calibration is carried out with air.[49] A TOF-UF calibrated with dry air at 30 °C causes a 5.3% error if the measurement is performed in a humid and hot environment with 44 mmHg water vapor pressure at 40 °C.[48]

TOF-UFs provide some unique advantages over other airflow transducers, such as inherent bidirectional measurement and negligible pneumatic resistance.[50] Due to good linearity and stability, it is free from calibration as well.[46] Generally, TOF-UFs for airflow measurement operate at a specific frequency ranging from 40 to 200 kHz. This is because, at frequencies higher than 200 kHz, the signal absorbed by gas exhibits significantly large, whereas sound lower than 40 kHz becomes audible.[46] Therefore, even the stray internal sound of TOF-UFs caused by breathing could interfere with the signal baseline. As the respiratory volume is the time integration of the signal, even a slight drift in the signal baseline could lead to a significant error in respiratory volume measurement.[50] Such influence could worsen in an uncontrolled and noisy environment, limiting the dynamic accuracy of TOF-UFs. The cost of TOF-UF is another concern. It is much more expensive than other airflow transducers due to the sophisticated circuit, which requires submicron to nanosecond time resolution to discriminate the differential transit times.[51] TOF-UFs are easily cleaned because there is no obstacle in the air pathway.

Piezoelectric Airflow Transducers. A piezoelectric airflow transducer (PAT) consists of a conduit and a piezoelectric bend sensor, as illustrated in Figure 9. The bend sensor is particularly noteworthy as it is crafted from uniaxially drawn piezoelectric poly L-lactic acid (PLLA) film, which effectively eliminates sensor signal fluctuation caused by the temperature variations between inhaled and exhaled air, thanks to the piezoelectric PLLA film's lack of pyroelectricity. Furthermore, the bend sensor incorporates a unique double-layered structure design, wherein the outer electrode is grounded, providing a self-shielding capability that effectively mitigates motion artifacts, thereby enhancing the sensor's wearable performance. The airflow detecting mechanism of the PAT is as follows: An airflow

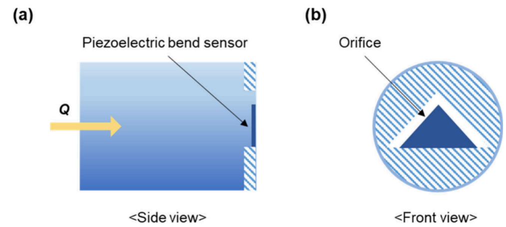

Figure 9. Schematic of a piezoelectric airflow transducer, (a) side view and (b) front view.

passes through a conduit, in which a piezoelectric PLLA bend sensor is perpendicularly installed with respect to the airflow direction, produces a differential pressure on both sides of the bend sensor. The bend sensor is deflected because of the differential pressure, generating piezoelectricity proportional to the airflow rate square. Therefore, the relationship between the airflow rate and generated piezoelectricity for the PAT is expressed as below:[52]

$$Q = k \times \sqrt{P} \tag{6}$$

where Q is the airflow rate, P is the produced piezoelectricity, and k is the constant, which is related to the geometry and property of the piezoelectric bend sensor. According to eq 6, the flow rate is proportional to the piezoelectric signal's square root, showing a nonlinear respons. Nevertheless, it has been proven that the dynamic accuracy of the PAT presented by Lu Jin et al. is comparable to a commercial flowmeter, i.e., Lily PT.[52] Unlike the above-mentioned differential pressure flowmeters, PAT has no bypass channel, so there is no channel-blocking issue, improving its long-term usability. Regarding calibration, it has been reported that the PAT's calibration equations have not changed remarkably during ten months, indicating frequent calibration is unnecessary.[52] Due to the simple structure of PAT, it is not difficult to be cleaned and sterilized. However, it should avoid high temperatures as the piezoelectricity of the bend sensor of the PAT could be changed above a critical temperature, i.e., curie temperature.[53]

CHALLENGES AND DIRECTIONS OF WEARABLE RESPIRATORY AIRFLOW TRANSDUCERS

The wearable applicability of the current airflow transducer in dynamic accuracy, long-term usability, power consumption, calibration, and cleaning has been evaluated. Table 1 provides a comprehensive overview of the various factors that impact the wearable applicability of current respiratory airflow transducers, clearly indicating that none of the existing transducers can perfectly satisfy the five technical requirements of wearable applicability together. OFs and TFs show acceptable dynamic performance but still present some limitations in the long term usability and calibration frequency over other airflow transducers. OFs are much more beneficial than TFs in terms of cleaning and durability because of their simple structure. The main problem of OFs is their long-term usability due to the bypass configuration design. In this context, PAT without a bypass channel could be the more optimal device for wearable applications than other flowmeters, as compared in Figure 10. Although cost is not explicitly mentioned due to its inherent difficulty in quantification, it remains a decisive factor in determining the commercial viability of wearable products. The PAT could potentially offer a competitive price point, primarily

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Table 1. Factors Limiting the Wearable Applicability of Respiratory Airflow Transducers

Type	Dymanic accuracy	Long-term usability	Power consumption	Calibration frequency	Cleaning
PT	Gravity	Moisture condensation in bypass channel	Heating capillary	Frequent	Capillary or mesh
F-OF	Nonlinear response	Moisture condensation in bypass channel	No factor increases power consumptions	Frequent	Ease
V-OF	Flag flexibility	Moisture condensation in bypass channel	No factor increases power consumptions	Frequent	Ease
HWA	Measurement direction and moisture condensation	Fragile heated wire	Heating wire	Frequent	Fragile heated wire
TF	Limited at low flow rate	Mechanical wear out	No factor increases power consumptions	Frequent	Turbine structure
UF	Noise	No factor	No factor increases power consumptions	Free	Ease
PAT	Nonlinear response	No factor	No factor increases power consumptions	Not frequent	Ease

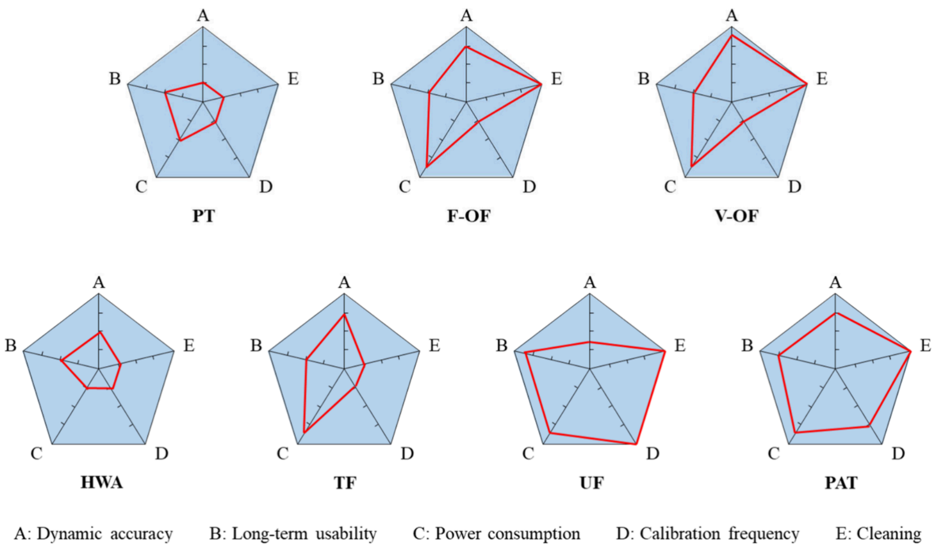

Figure 10. Estimation and comparison of wearable applicability of current respiratory airflow transducers in dynamic accuracy, long-term usability, power consumption, calibration frequency, and cleaning.

because its sensing component consists solely of low-cost piezoelectric PLLA film.

Developing a novel airflow transducer with high dynamic accuracy, long-term usability, low-power consumption, calibration-free, and easy cleaning is essential and significant for ambulatory respiratory flow monitoring, but it is challenging. Another approach is to overcome the drawbacks of current airflow transducers by innovating design and integrating new technology. For instance, allowing the airflow transducers to measure only inhaled air by separating the breathing pathway can effectively avoid the influence of moisture condensation. Frequent calibration can automatically be achieved through upgrading the electric circuit and device program. Introducing the low-power design of the airflow transducer and a high-density Lithium-ion battery, the burden from the power consumption would reduce to a certain level. The longevity of some airflow transducers would be enhanced by adopting more durable and rigid hot wires for HWAs.

Wearable airflow transducers are expected to undergo significant advancements in the future, with anticipations of increased wearability, providing better monitoring and management of respiratory health. Improvements in design will also lead to enhanced comfort and usability, with smaller and more lightweight devices that can be worn for extended periods. Integration with other wearable devices, such as smartwatches, smartglasses and fitness trackers, will offer a more comprehensive view of overall health and wellness. Furthermore, wearable airflow transducers will play a larger role in medical research and occupational health settings, expanding their use and impact. We expect future studies and advancements that will build upon this engineering reference to further improve the wearable applicability of respiratory airflow transducers.

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