Title: Breath monitoring, sleep disorder detection and tracking using thin film acoustic waves and open-source electronics

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Abstract: Apnoea, a major sleep disorder, has affected many adults and caused several issues, such as fatigue, high blood pressure, liver conditions, increased risk of type II diabetes and heart problems. Therefore, advanced monitoring and diagnosing tools of apnoea disorders are needed to facilitate better treatments, with advantages such as accuracy, comfort of use, cost effectiveness and embedded computation capabilities to recognise, store, process and transmit time series data. In this work we present an adaptation of our Acousto-Pi open-source Surface Acoustic Wave (SAW) platform (Apnoea-Pi), to monitor and recognise apnoea in patients. The platform is based on thin film SAW, using bimorph ZnO and aluminium structures, including those fabricated in Al foils or plates, to achieve for breath tracking based on the humidity and temperature changes. We applied open-source electronics and provided embedded computing characteristics for signal processing, data recognition, storage, and transmission of breath signals. We show that thin film SAW devices out-perform standard and off-the-shelf capacitive electronic sensors regarding to their responses and accuracy for human breath tracking purposes. This in combination with embedded electronics makes a suitable platform for human breath monitoring and sleep disorder recognition.

Keywords: Surface acoustic waves, sleep disorder, apnoea, open-source electronics, pattern recognition, piezoelectric thin film.

Article Highlights

- An open-source system for sleep disorder monitoring
- embedded platform using SAW sensors
- proof-of-concept breath tracking system.

1. Introduction

Sleep disorders are frequently observed in adults, having a profound impact on the health and quality of life for patients. There are six major sleep disorders: insomnia, circadian rhythm disorders, sleep-disordered breathing, hypersomnia/narcolepsy, parasomnias, and restless legs syndrome/periodic limb movement disorder. Among those, apnoea sleep-disordered breathing is considered a serious medical condition due to the risk of sudden death and the reduction of life expectancy [1].

Sleep apnoea is a primary sleep disorder consisting in abnormal breathing pauses during sleep. There are three categories of sleep apnoea: obstructive sleep apnoea, central sleep apnoea, and complex sleep apnoea [2]. Obstructive sleep apnoea stops breathing for at least ten seconds, collapsing the upper respiratory system. For the central apnoea, on the contrary, the interruption of airflow happens when there is not an effort to breathe, commonly due to brain miscommunication with the muscles that control the breathing. Some patients have a combination of all the above apnoea, resulting in a complex sleep apnoea [2] In such cases, monitoring and diagnostic is challenging due to the irregular pattern, demanding advanced solutions for their diagnostic and monitoring. For this, fast response and precision sensors are needed in combination with embedded electronics for data acquisition, storage, and transmission.

Polysomnography (PSG) is considered the gold standard method for the diagnosis of sleep disorders [3]. It is a comprehensive technique where multiple sensors are attached to the patient. They monitor brain waves, blood oxygenation, heart rate, eye, and limb movements throughout the sleep study. This typically results in an extensive setup where the patient has to wear multiple sensors and monitored overnight, typically in a healthcare clinical facility [3]. Hence, such multiparametric sleep monitoring has several disadvantages including discomfort to the individuals, increased cost and being unsuitable for home settings. Complementary technologies are being implemented such as wearable sensors [3], [4], or sensors for breath pattern monitoring during sleep [5], [6]. New non-obtrusive technologies are being developed to improve patient comfort and facilitate home uses, including the uses of cameras for video-based photoplethysmography and near-infrared temperature measurements [7], [8]. Despite being useful, these techniques are based on indirect observations and do not provide direct information for gas flow associated with breathing patterns.

SAW devices have been used for respiratory monitoring applications, demonstrating fast response and good sensitivity [9], [10], including devices fabricated using a thin film piezoelectric approach [11]. Recently we developed a platform that combines SAWs with open-source electronics using Raspberry Pi hardware [13], with the potential to be used for sleep disorder diagnostics and monitoring [14]. The platform includes thermal and standard imaging that could integrate video-based photoplethysmography and near infrared monitoring. Moreover, the embedded system can be used for online or offline pattern recognition and analysis using time series [15], [16] identification and machine learning methods [17], [18]. For the thin film SAW devices we use, changes in both the relative humidity (RH) levels and temperature will cause the shifts of frequency signals as results of changes in acoustic velocity, which can be used to detect breathing. This is s combined effect of mass loading, surface sheet conductivity and elasticity changes [12].

In this paper, we investigate the ability of the SAW-Raspberry Pi platform to track and analyse breathing and the potential to detect breathing disorders. We aim to understand SAW-based breath tracking capability, and the combination of SAW temperature/humidity sensing with embedded electronics, as shown in Figure 1. Currently, as a proof-of-concept setup, we simply use the SAW-Raspberry Pi platform to interface with a vector network analyser (VNA), record the frequency spectrum and automatically track its changes. We have plans to substitute the VNA with reflectometer circuitry and radio frequency (RF) couplers to decrease the cost of the sensor. The platform offers the advantages of simplicity and potential low-cost without compromising on performance.

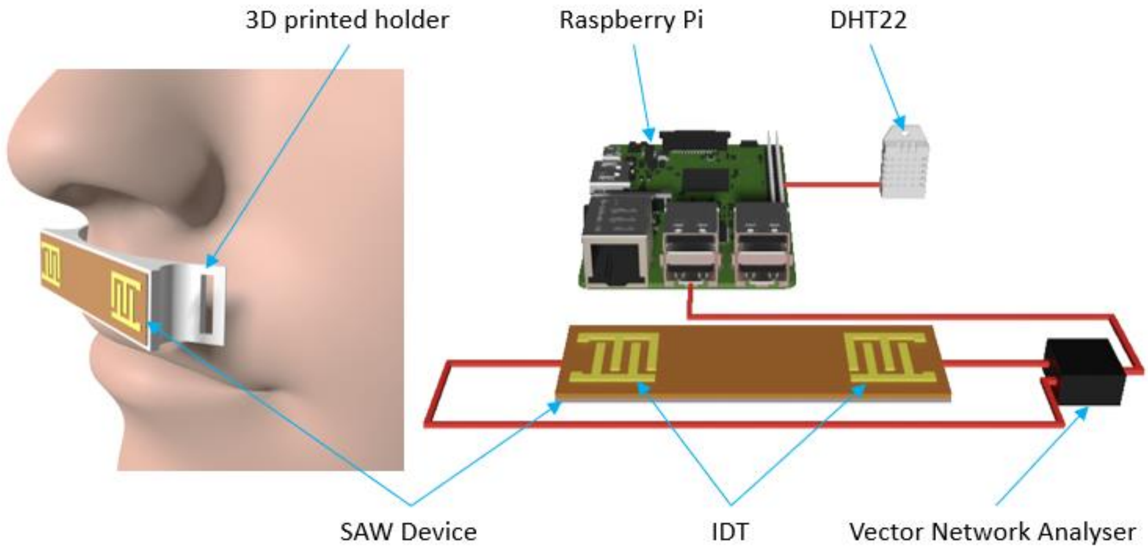

Figure 1 Proposed Apnoea-Pi Platform

2. Design methodology

2.1 Hardware description

Firstly, we explain the details for the electronic platform developed, its experimental data collections and analysis. The system is based on Raspberry Pi, Model 4, as an open-source embedded electronic processor. A similar system can also be implemented using another embedded platform, but our platform is more independent, relatively low cost, versatile, with good performance and open-source nature. We provide the setup with a combined humidity and temperature sensor (Adafuit DHT22), with capacitive humidity reading and thermistor temperature sensing functions. The humidity sensor chosen is a popular model with a compromise between cost and performance. It is easily integrated with the Raspberry Pi and can work in wide temperature and humidity ranges. The sensor is connected to the GPIO pins as shown in Figure 2. Changes in the resonant frequencies are monitored using a vector network analyser (Keysight FieldFox Network Analyzer), connected with USB ports. The embedded controller records such changes for time series representation, tracking the peak automatically. It should be noted that since SAW microfluidics are not required, the devices don't require radio frequency (RF) power inputs and can be supplied with the vector analyser, simplifying the setup.

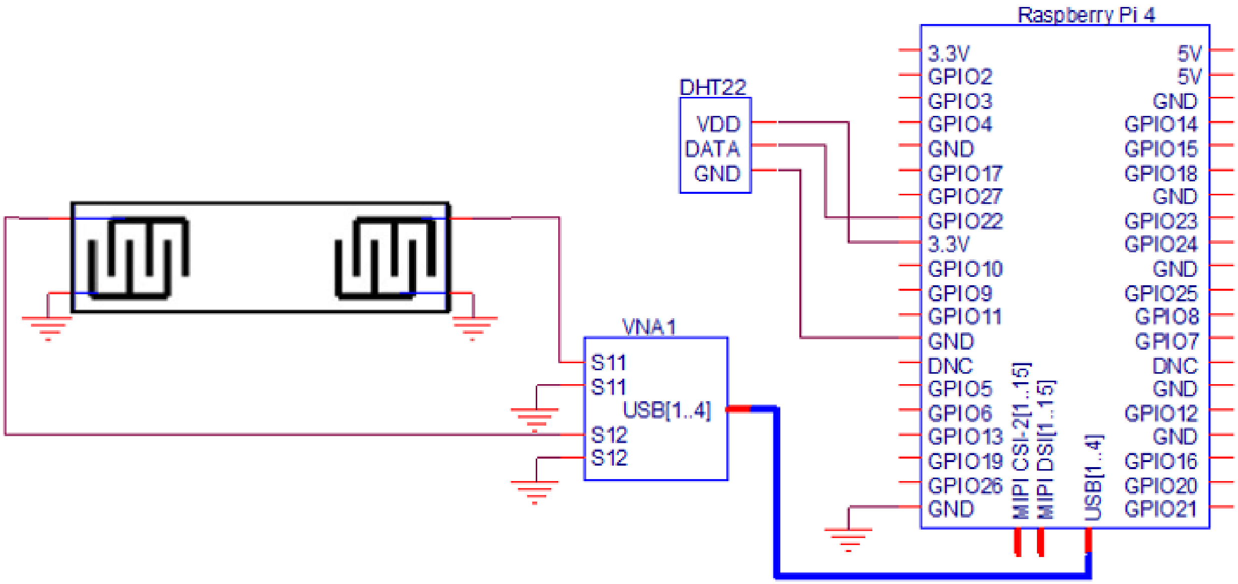

Figure 2. Electronic schematic showing Raspberry Pi General Purpose Input Output (GPIO) connections to SAW devices

2.2 Software description

Here we detail the development of the graphical user interface (GUI), implemented with Python due to the acceptance and scalability of the platform. The DHT22 sensor is connected to the Raspberry Pi and the network analyser for referencing with ambient environments, frequency measurement and drift compensation. We use the Python module Kivy [19] employed to develop a user-friendly environment at both visualisation and ergonomic levels. Using this setup, we can present the actual frequency spectrum of the SAW device and its fundamental resonant frequency along with any drifts in real time. This will be particularly useful to monitor, track and balance the frequency shift due to temperature changes. The interface also represents readings of humidity and temperature in different plots. Data are recorded in the embedded system, using Python algorithms to perform pattern identification. For instance, with off-shelf Python libraries on machine learning, it is possible to match the recorded real time data with the apnoea records, which can allow to identify the type of apnoea that the patient suffers along with its severity.

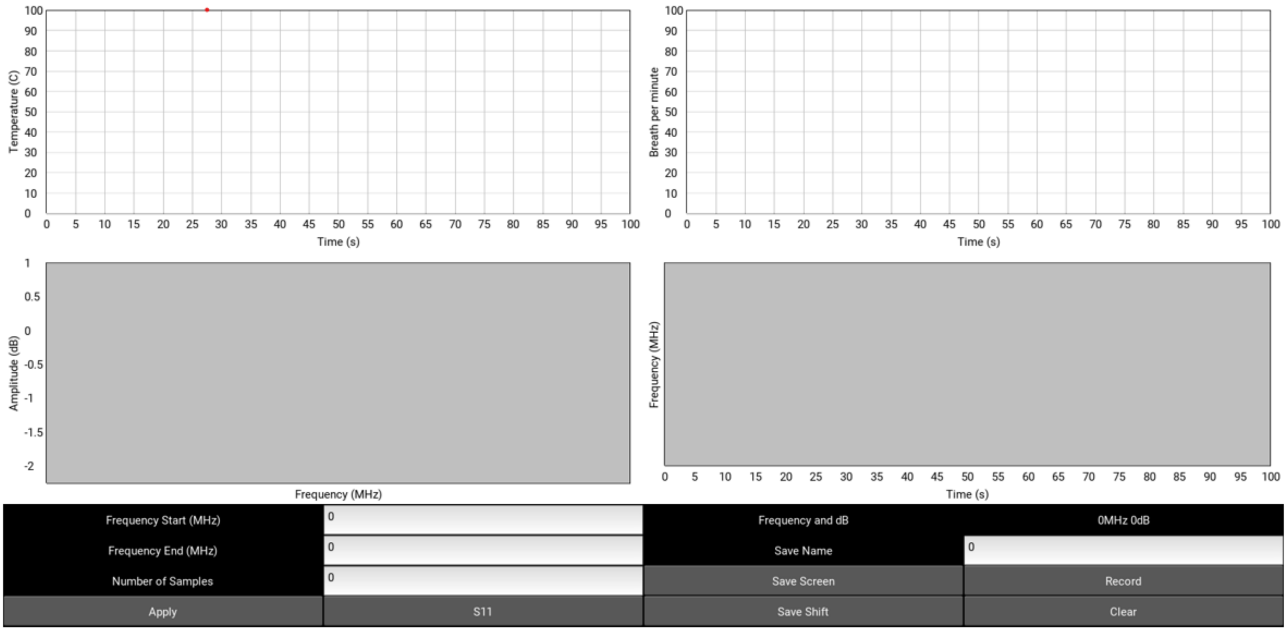

Figure 3 GUI to monitor breathing patterns using SAW and humidity sensors

3. SAW device fabrication

The concept presented in this paper is suitable for SAW device regardless of its fabrication and composition. We employ thin film SAW devices with bimorph Al/ZnO structure due to their potential to fabricate flexible SAW [20], [21]. Such SAW devices made with Al foil can be further made to be flexible and mounted on the upper lip of the patient, fitting comfortably, and allowing adequate SAW sensing during sleep. The device can also be attached on the upper lip just below the nose in a breathing mask [22] with wireless connections to the embedded controller unit, as shown in Figure 1. Further advantages of thin film SAWs include industrial scalability and ultra-low-cost fabrication potential. In addition, thin film SAWs have shown to be good humidity sensors for human breathing monitoring applications [9], [23]

Piezoelectric ZnO thin films were deposited on Al plate (1.6 mm thickness). Magnetron sputter in a direct current (DC) mode, was employed for thin film depositions using Ar/O_2 flow ration of 10/15 sccm and 99.99% pure Zn target. Direct current power was set to be 400 W, with a gas pressure of 4 $\times 10^{-4}$ mbar. A rotating sample holder was used during the deposition to realise the uniformity of the piezoelectric thin films. The deposition rate was 5.6 nm/min. IDT electrode structures were made with standard lift-off photolithography. The IDTs were metallised with sputtering Al target, achieving a thickness of 200 nm. The electrode design followed a standard finger pitching set at 200 µm, with an aperture of 5 mm and using 50 finger pairs. Resonance was measuredat ~14.3 MHz (which is corresponds to the Rayleigh wave mode).

Reflection and transmission signals were measured using the network analyser (Keysight Fieldfox vector network analyser, N9913A) for the characterisation of the fabricated devices. Figure 4 illustrates the SAW platform, with the deposited ZnO film on aluminium plate SAW device in a 3D printed device test holder. We used a DHT22 combined humidity and temperature sensor, which was placed near the sensing zone at the SAW device. Electrode connections were made with spring loaded pins.

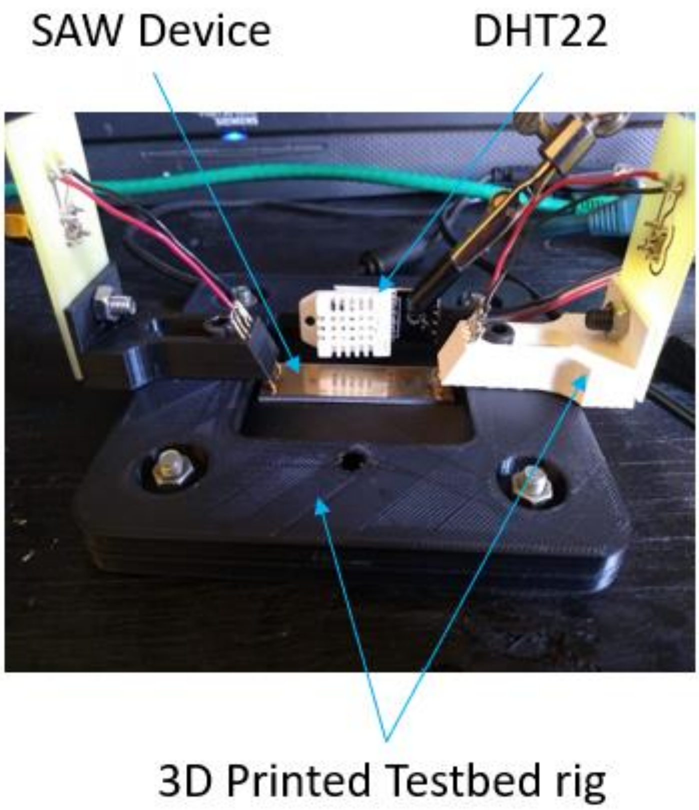

Figure 4 3D printed Testbed rig for testing humidity and temperature

4. Results and discussion

4.1 Frequency responses to humidity and temperature

In this section we report the SAW frequency changes with humidity and temperature variations in the context of human exhaling patterns. Figure 5 is an example of the recorded signal of one prolonged exhale onto a SAW device and the commercial sensor DHT22. This example is typical of what will happen when a patient is asked to take a deep breath, then exhales towards both the SAW device and the DHT22 sensor. The thin film SAW device shows a rapid change in frequency, taking 1.4 seconds to shift 50 KHz and taking 11 seconds to return back to the fundamental frequency of 14.276 MHz. The signals from the DHT22 take much longer time to respond to the same breath level. The relative humidity peaks at 77%, an increase of 25%, and the signal from the DHT22 takes ~7 seconds to respond to the exhale. The DHT22 also takes a siginficantly longer time (35 seconds) to return back to the relative humidity of the environment.

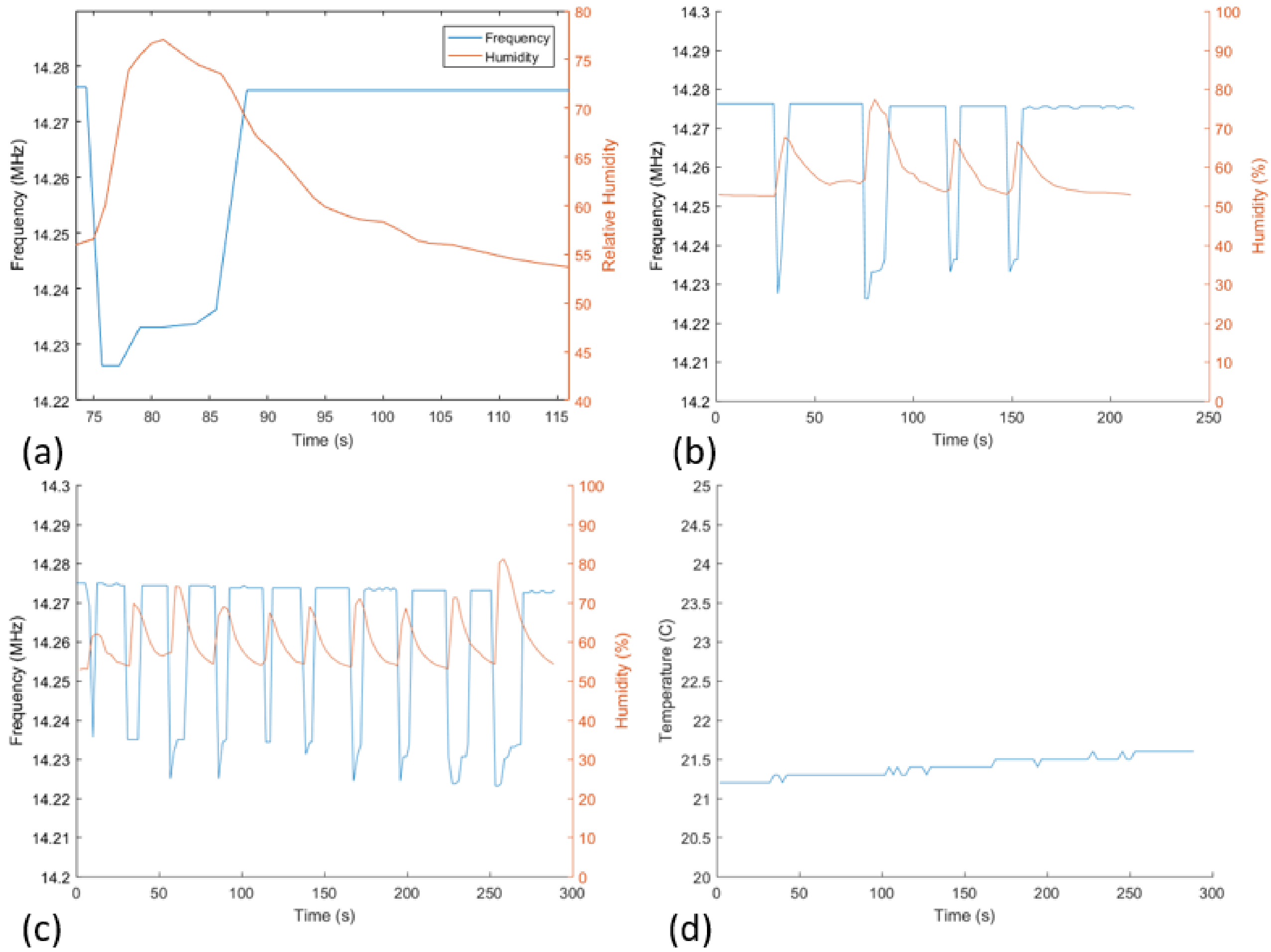

Figure 5 (a) A singular breath, the effect of humidity on frequency (b) 5 exhale breaths of approximate same duration and strength (c) 10 exhales (d) temperature over test duration

The following test consists of 10 breaths. Figure 5(b) and Figure 5(c) show breaths with the signals of humidity and temperature recorded over several minutes. Humidity would have large changes for each breath, with an increase of 10 to 20% relative humidity as shown in Figure 6(a). Temperature shown in Figure 5(d) on the other hand would have a minimal change during breath cycles, rather than gradually changing over the duration of the tests. Longer duration of breaths has affected the temperature by up to +2 ^{o}C.

There are two possible factors that could affect the temperature readings near the SAW device. One is the continuous breathing of the patient undertest, as the temperature inside a typical healthy human body is 37 °C. The temperature of an exhaled breath is around 34 °C [24]. The temperature of the SAW device was around 21 °C. The other alternative is due to environmental temperature changes. This group of results were recorded at room temperature on a warm summer day (25^{o}C). Figure 5 (d) clearly shows a steady rise in temperature over the duration of the test. This appears to be unaffected by the breathing on the device. There are no noticeable spikes when exhaling and no drops are observed when breathing is stopped. Therefore, this increase in temperature is likely caused by the environment changes due to the warming effect in the room.

In these initial tests, the device was held below the subject's nose perpendicular to the flow direction. Ideally the device should be mounted parallel to the flow directly below the nose. These tests show slower deeper breaths. When exhaling longer or harder, a greater amount of air and water vapour are expelled from the participant. These moisture or tiny droplets are deposited on the surface of the SAW device and take some time to clear, giving the signal a much longer recovery time to return to the fundamental frequency.

It is clear from these results that the SAW device responds quicker to humidity change in comparison to the standard sensor of DHT22. It is possible when breathing longer or stronger to cause the DHT22 sensor to saturate, whilst the SAW device shows a larger shift in frequency and thus has much higher sensitivity. Results show that SAW devices are a much better choice in breath monitoring in the open-source electronic domain.

The commercial DHT22 is useful for referencing and data logging environmental changes surrounding the test. However due to its very slow response, it is unsuitable for rapid changes caused by breathing. The sensor took average 7 seconds to return back to the original relative humidity. Whilst there may be a small temperature change on the surface of the SAW device, the commercial sensor is unable to detect any changes in air surrounding the device. We only observe in one test that the recorded signal changes as a result of deep breathing closer to the sensor. In comparison the SAW device can react 3 to 4 seconds quicker to the fluctuations in humidity, making this a much more suitable solution for monitoring breathing.

4.2 Simultaneously monitoring humidity and temperature

In order to observe the effect of humidity, hot air flow from a hair dryer was applied to the surface in order to reduce any moisture on the device. Figure 6 shows the frequency shift, temperature, and humidity from the DHT22 sensor. Using the previous method with temperature coefficient of frequency (TCF) calculation [25], the large temperature fluctuations can be removed. It is worthwhile highlighting that we can do all these calculation steps in real time at the embedded processor. As temperature causes the device to expand, the speed of the waves is decreased, and the fundamental frequency of the device is decreased. Reducing humidity, on the other hand, increases the frequency by enhancing the wave propagation with less water mass loading effect on the device's surface. After the filtering, the resultant signals clearly show the increases of the frequency.

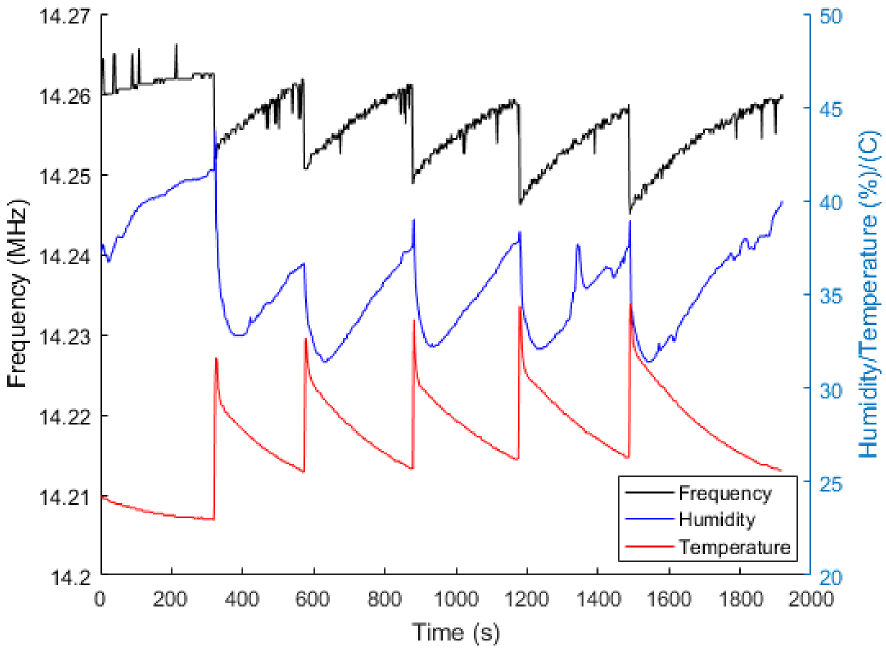

Figure 6 Effects of hot air to the frequency shifts of the SAW device

Figure 7 shows frequency and humidity extracted from the deep breathing and heating results. Above 50% relative humidity the data were recorded from the deep breathing, and this half of the data show a trend of rapid increases in frequency shift as the humidity approaches 80%. This trend is similar to what have been reported in literature [26], [27], [28]. As these data were acquired in the field during breath tests, there are other factors affecting the frequency shift such as temperature. Temperature shows a much stronger influence in the hot air tests (all results below 45% relative humidity) as the humidity which lowers the change in frequency appears to increase. Calculations to filter the effects of temperature were performed.

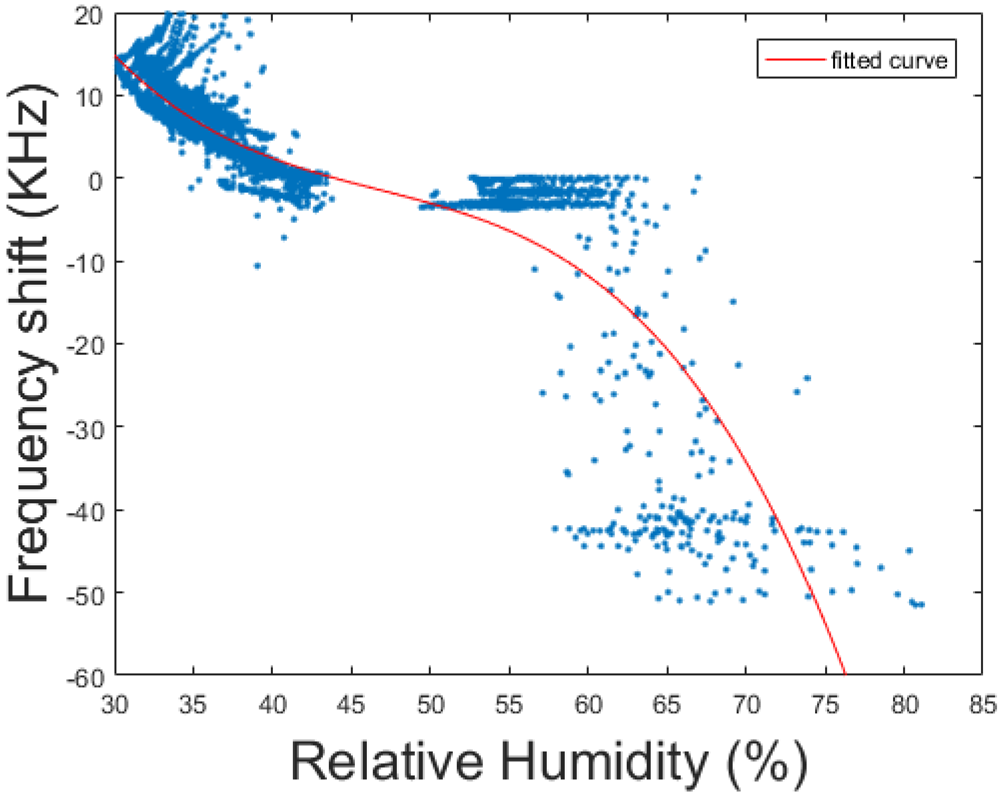

Figure 7 Combined heating and breathing tests to calculate humidity effect on frequency

4.4 Temperature compensation

We obtain the actual frequency shift caused by humidity using the TCF of the device which is measured to be -277.2 ppm. This enables the frequency changes from temperature to be calculated and deducted from the original signal data. This can also be applied in real time to the monitoring system.

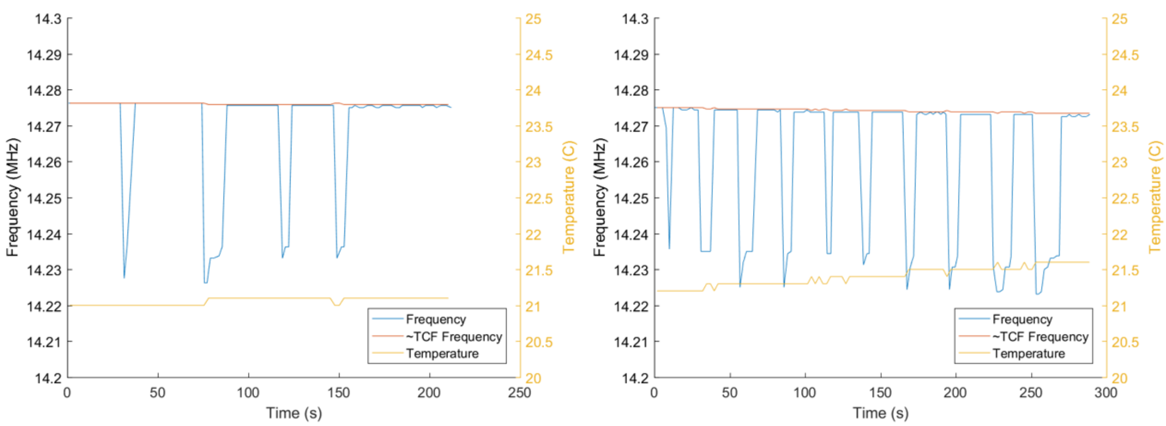

Figure 8 Frequency, temperature and TCF (a) 5 breaths test (b) 10 breahts test

Figure 8 shows the frequency, temperature, and the estimated temperature compensated frequency as the test is underway the recorded temperature gradually increases by approximately 0.5 degrees over 5 minutes. Using the calculated TCF for the device it is possibly to use this to compensate for the effects of temperature on the frequency shift reducing the chance of false peaks being detected.

4.5 Continuous breath detection and monitoring of apnoea cases

In breath tracking, real time breathing identification is of utmost importance. Figure 9 shows the points that are detected in real time. The breaths per minute (bpm) rate is calculated based on the time recorded between peaks and averages out over 1 minute. A simple breath detection method can be used to check the falling edge of each breath running live as the breathing is happening. The time until the next breath is recorded and the estimated breaths per minute are calculated, and this reading is then averaged out as there are variations between breaths withing one minute.

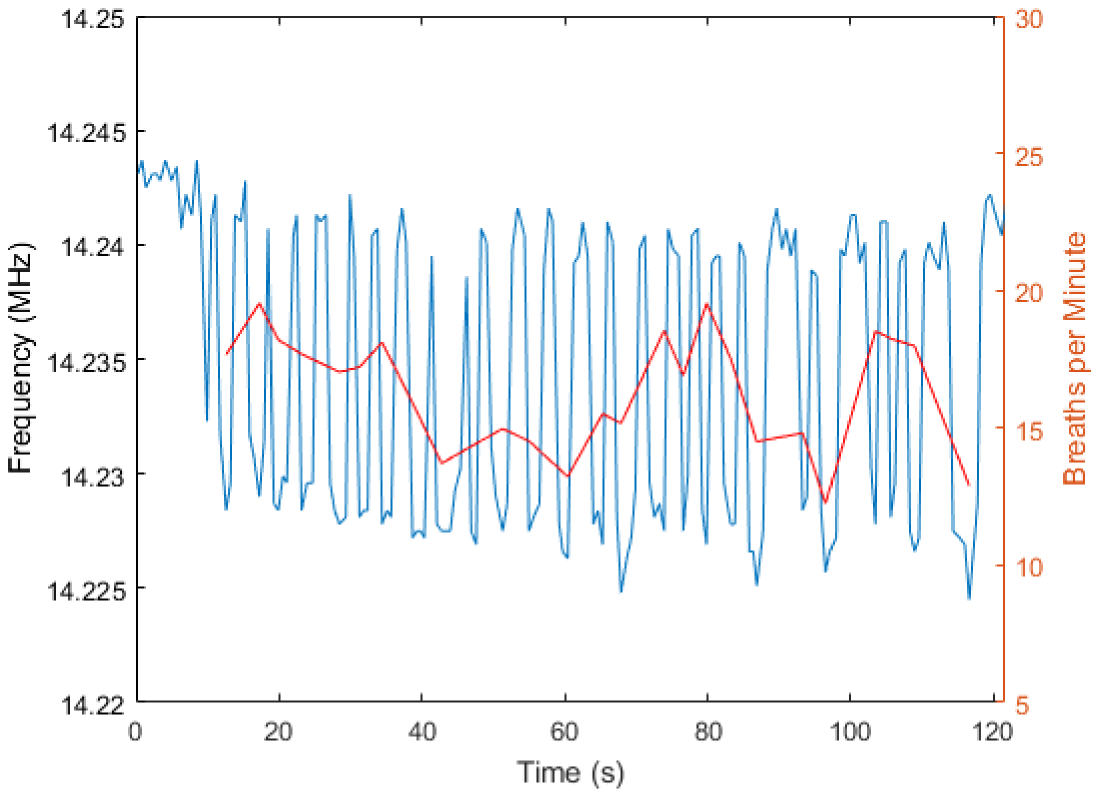

Figure 9 Continuous normal breathing and calculated breaths per minute

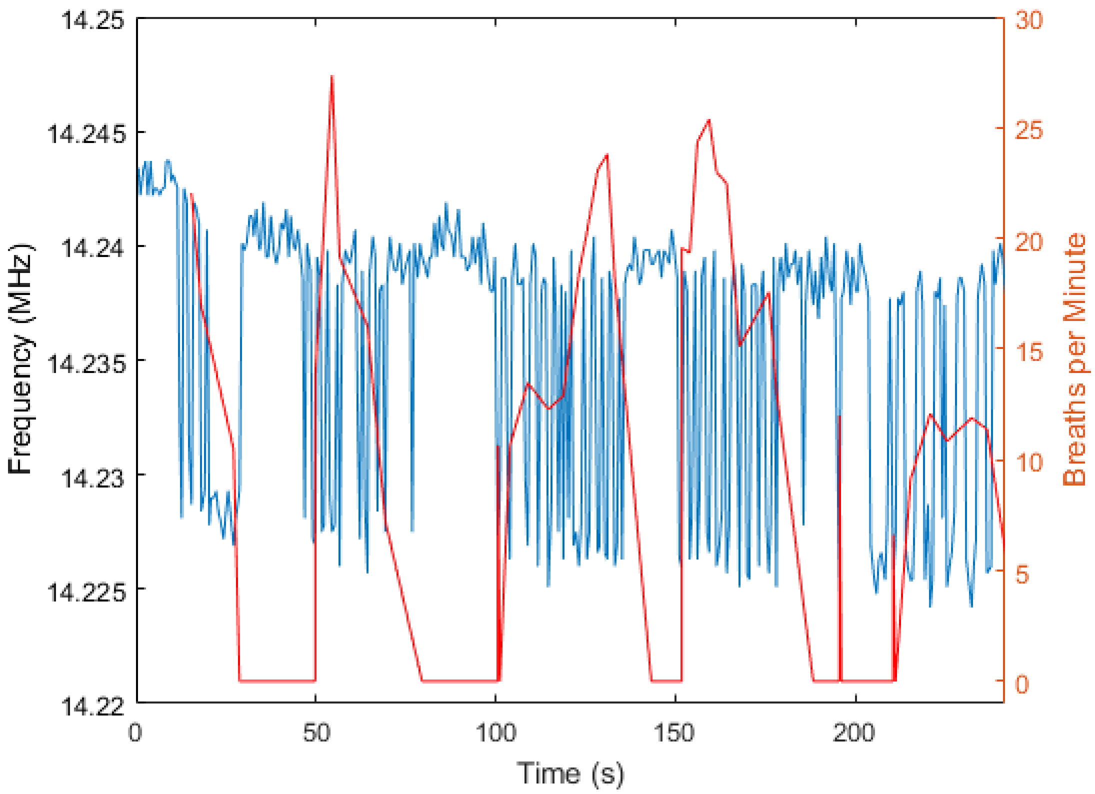

Figure 10. Demonstration of a breathing disorder and calculated bpm

Figure 10 represents a breathing disorder such as apnoea. As shown in Figure 10, the breathing is not normal with period of 30 to 50 seconds where no breathing has occurred. This is the raw frequency data captured from the SAW device. The sleeping disorder is visually identified when the user has stopped breathing and when they resume. The detection algorithm also shows that the breaths per minute momentarily drops below the average human breathing rate or completely to zero. This could be linked to a carer monitoring a person's health getting alerts when the breathing is beyond the user's normal range. There is the potential for machine learning methods to be implemented and learn the normal behaviour of a patient, further increasing the accuracy and aid in detecting when the patient might require medical attention.

4. Conclusions

In this work we explain Apnoea-Pi, an open-source system for sleep disorder monitoring and identification using SAW sensors. The platform is adequate for all embedded computational platforms and SAW sensors. We show all the different aspects that the method needs to realise for sleep disorder tracking with proof-of-concept data. The technique is suitable for a wide range of SAW devices, including flexible substrates. The concept provides embedded computing capabilities for data storage, signal processing, and transmission. We show that the SAW humidity sensor interfaced by the embedded system is faster than standard capacitive electronic sensors. In future work we will implement machine learning protocols using open-source libraries. The idea is to complement SAW sensing with pattern identification and recognition capabilities, especially over long time series recording.

Acknowledgements

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Declaration of competing interest

None.

References

[1] M. Thorpy, “International Classification of Sleep Disorders,” *Sleep Disorders Medicine: Basic Science, Technical Considerations and Clinical Aspects: Fourth Edition*, pp. 475–484, May 2017, doi: 10.1007/978-1-4939-6578-6_27.

[2] M. K. Pavlova and V. Latreille, “Sleep Disorders,” *The American Journal of Medicine*, vol. 132, no. 3, pp. 292–299, Mar. 2019, doi: 10.1016/J.AMJMED.2018.09.021.

[3] H. Zhang *et al.*, “Waist-wearable wireless respiration sensor based on triboelectric effect,” *Nano Energy*, vol. 59, pp. 75–83, May 2019, doi: 10.1016/J.NANOEN.2019.01.063.

[4] S. Kwon, H. Kim, and W. H. Yeo, “Recent advances in wearable sensors and portable electronics for sleep monitoring,” *iScience*, vol. 24, no. 5, p. 102461, May 2021, doi: 10.1016/J.ISCI.2021.102461.

[5] S. Ahmadzadeh, J. Luo, and R. Wiffen, “Review on Biomedical Sensors, Technologies and Algorithms for Diagnosis of Sleep Disordered Breathing: Comprehensive Survey,” *IEEE Reviews in Biomedical Engineering*, vol. 15, pp. 4–22, 2022, doi: 10.1109/RBME.2020.3033930.

[6] B. Feng, K. Zhao, Q. Su, Z. Yu, and H. Jin, “Passive wireless respiratory sensor,” *Electronics Letters*, vol. 57, no. 19, pp. 730–731, Sep. 2021, doi: 10.1049/ELL2.12248.

[7] M. van Gastel, S. Stuijk, S. Overeem, J. P. van Dijk, M. M. van Gilst, and G. de Haan, “Camera-Based Vital Signs Monitoring during Sleep - A Proof of Concept Study,” *IEEE Journal of Biomedical and Health Informatics*, vol. 25, no. 5, pp. 1409–1418, May 2021, doi: 10.1109/JBHI.2020.3045859.

[8] J. Allen, “Photoplethysmography and its application in clinical physiological measurement,” *Physiological Measurement*, vol. 28, no. 3, p. R1, Feb. 2007, doi: 10.1088/0967-3334/28/3/R01.

[9] X. Tao *et al.*, “Three-Dimensional Tetrapodal ZnO Microstructured Network Based Flexible Surface Acoustic Wave Device for Ultraviolet and Respiration Monitoring Applications,” *ACS Applied Nano Materials*, vol. 3, no. 2, pp. 1468–1478, Feb. 2020, doi: 10.1021/ACSANM.9B02300/SUPPL_FILE/AN9B02300_SI_001.PDF.

[10] T. Dinh, T. Nguyen, H. P. Phan, N. T. Nguyen, D. V. Dao, and J. Bell, "Stretchable respiration sensors: Advanced designs and multifunctional platforms for wearable physiological monitoring," *Biosensors and Bioelectronics*, vol. 166, p. 112460, Oct. 2020, doi: 10.1016/J.BIOS.2020.112460.

[11] Y. Q. Fu *et al.*, "Advances in piezoelectric thin films for acoustic biosensors, acoustofluidics and lab-on-chip applications," *Progress in Materials Science*, vol. 89, pp. 31–91, 2017, doi: 10.1016/j.pmatsci.2017.04.006.

[12] D. Ciplys *et al.*, "Visible–blind photoresponse of GaN-based surface acoustic wave oscillator," *Applied Physics Letters*, vol. 80, no. 11, pp. 2020–2022, Mar. 2002, doi: 10.1063/1.1459485.

[13] J. Vernon *et al.*, "Acousto-Pi: An Opto-Acoustofluidic System using Surface Acoustic Waves controlled with Open Source Electronics for Integrated In-Field Diagnostics," *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, pp. 1–1, 2021, doi: 10.1109/TUFFC.2021.3113173.

[14] J. Vernon, P. Canyelles-Pericas, H. Torun, R. Binns, W. P. Ng, and Y. Q. Fu, "Apnoea-Pi: Sleep disorder monitoring with open-source electronics and acoustics," 2021. doi: 10.23919/ICAC50006.2021.9594073.

[15] L. A. Aguirre, V. C. Barros, and Á. V. P. Souza, "Nonlinear multivariable modeling and analysis of sleep apnea time series," *Computers in Biology and Medicine*, vol. 29, no. 3, pp. 207–228, May 1999, doi: 10.1016/S0010-4825(99)00006-2.

[16] C. Cheng, C. Kan, and H. Yang, "Heterogeneous recurrence analysis of heartbeat dynamics for the identification of sleep apnea events," *Computers in Biology and Medicine*, vol. 75, pp. 10–18, Aug. 2016, doi: 10.1016/J.COMPBIOMED.2016.05.006.

[17] G. C. Gutiérrez-Tobal, D. Álvarez, L. Kheirandish-Gozal, F. del Campo, D. Gozal, and R. Hornero, "Reliability of machine learning to diagnose pediatric obstructive sleep apnea: Systematic review and meta-analysis," *Pediatric Pulmonology*, 2021, doi: 10.1002/PPUL.25423.

[18] A. Ramachandran and A. Karuppiah, "A Survey on Recent Advances in Machine Learning Based Sleep Apnea Detection Systems," *Healthcare 2021, Vol. 9, Page 914*, vol. 9, no. 7, p. 914, Jul. 2021, doi: 10.3390/HEALTHCARE9070914.

[19] Www.kivy.org, "Kivy Documentation," 2012. https://kivy.org/doc/stable/

[20] Y. Wang *et al.*, "Flexible/Bendable Acoustofluidics Based on Thin-Film Surface Acoustic Waves on Thin Aluminum Sheets," *ACS Applied Materials and Interfaces*, vol. 13, no. 14, pp. 16978–16986, Apr. 2021, doi: 10.1021/ACSAMI.0C22576/SUPPL_FILE/AM0C22576_SI_009.PDF.

[21] R. Tao *et al.*, "Thin film flexible/bendable acoustic wave devices: Evolution, hybridization and decoupling of multiple acoustic wave modes," *Surface and Coatings Technology*, vol. 357, pp. 587–594, Jan. 2019, doi: 10.1016/J.SURFCOAT.2018.10.042.

[22] R. Tao *et al.*, "Hierarchical nanotexturing enables acoustofluidics on slippery yet sticky, flexible surfaces," *Nano Letters*, vol. 20, no. 5, pp. 3263–3270, May 2020, doi: 10.1021/acs.nanolett.0c00005.

[23] D. Li *et al.*, "A flexible virtual sensor array based on laser-induced graphene and MXene for detecting volatile organic compounds in human breath," *Analyst*, vol. 146, no. 18, pp. 5704–5713, Sep. 2021, doi: 10.1039/D1AN01059J.

[24] J. M. Cowan, J. M. Burris, J. R. Hughes, and M. P. Cunningham, “The relationship of normal body temperature, end-expired breath temperature, and BAC/BrAC ratio in 98 physically fit human test subjects,” *J Anal Toxicol*, vol. 34, no. 5, pp. 238–242, 2010, doi: 10.1093/JAT/34.5.238.

[25] R. Tao *et al.*, “Bimorph material/structure designs for high sensitivity flexible surface acoustic wave temperature sensors,” *Scientific Reports*, vol. 8, no. 1, pp. 1–9, 2018, doi: 10.1038/s41598-018-27324-1.

[26] W. Xuan *et al.*, “High sensitivity flexible Lamb-wave humidity sensors with a graphene oxide sensing layer,” *Nanoscale*, vol. 7, no. 16, pp. 7430–7436, Apr. 2015, doi: 10.1039/C5NR00040H.

[27] J. Wu *et al.*, “Ultrathin Glass-Based Flexible, Transparent, and Ultrasensitive Surface Acoustic Wave Humidity Sensor with ZnO Nanowires and Graphene Quantum Dots,” *ACS Applied Materials and Interfaces*, vol. 12, no. 35, pp. 39817–39825, Sep. 2020, doi: 10.1021/ACSAMI.0C09962/SUPPL_FILE/AM0C09962_SI_001.PDF.

[28] Y. C. Han *et al.*, “Environment-friendly surface acoustic wave humidity sensor with sodium alginate sensing layer,” *Micro and Nano Engineering*, vol. 15, p. 100127, Jun. 2022, doi: 10.1016/J.MNE.2022.100127.

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