Interdisciplinary Journal of Health, Environment and Computation
ISSN Online: 3108-1444 Vol.: 2, No.: 1 (July – September) 2026. PP. : 1-12

DESIGN AND PERFORMANCE ANALYSIS OF A PLANAR MICROSTRIP FILTER SENSOR FOR NON-INVASIVE HUMAN TENSION AND PHYSIOLOGICAL STRESS DETECTION

Nabaa S.M. Abed
College of Engineering, Cairo University, Egypt.

Email: iraqiq26@yahoo.com

https://doi.org/10.65983/ijhec.2026.01.0003

Received: July 08, 2026; Revised: August 21, 2026; Accepted : September 08, 2026

Abstract

Psychological stress is accompanied by autonomic changes that can modify sudomotor activity, skin hydration, and the ionic environment at the skin surface. Because tissue water content and conductivity contribute to microwave dielectric behavior, these physiological changes can perturb the electromagnetic loading experienced by a resonant sensor. This study presents a compact planar microstrip band-stop sensor operating near 2.45 GHz for non-invasive monitoring of stress-related physiological changes. The sensor uses a modified split-ring resonator (SRR) coupled to a 50-Ω microstrip line on Rogers RT/duroid 5880 ($\varepsilon_r = 2.2$, $\tan\delta = 0.0009$, $h = 1.57$ mm) and occupies 30 × 25 mm². HFSS simulations were performed using a three-layer skin model and the dielectric states reported in the original study. Across the modeled tissue range, ε' increased from 38.0 to 46.5 and conductivity from 1.45 to 1.95 S/m, while the resonance shifted from 2.312 to 2.185 GHz, corresponding to a 127 MHz shift and an average sensitivity of 14.94 MHz per unit change in ε'. The unloaded-to-resting transition produced an additional 138 MHz shift and should not be confused with the tissue-state sensitivity range. Experimental measurements on 12 healthy adults used a two-port VNA and GSR as a reference physiological signal. The manuscript reports phase-level values of 2.14 ± 0.35 µS at baseline, 6.85 ± 0.92 µS during the Stroop task, and 2.80 ± 0.41 µS during recovery, with corresponding measured resonances of 2.315, 2.242, and 2.301 GHz. The previously reported $R^2 = 0.984$ is retained only as a reported experimental association because individual-level raw data are not supplied in the manuscript; confidence intervals, subject-level RMSE/MAE, and repeated-measures inference require the original paired measurements. The revised interpretation therefore treats the microwave response as a stress-related physiological proxy rather than a direct or diagnostic measure of psychological stress.

Keywords: microwave sensor; microstrip filter; split-ring resonator; skin permittivity; electrodermal activity; physiological stress; non-invasive sensing; 2.45 GHz.

1. Introduction

Acute psychological stress activates coordinated autonomic and endocrine responses. In particular, sympathetic activation influences eccrine sweat-gland activity, cardiovascular function, and other peripheral physiological processes. The Trier Social Stress Test (TSST) and related laboratory stressors are widely used to evoke reproducible acute stress responses, although the magnitude of individual responses depends on the protocol and participant characteristics [1 - 3]. Importantly, physiological responses should not be treated as interchangeable measures of psychological experience; recent meta-analytic evidence emphasizes that acute stress is multisystemic and that associations among physiological and psychological measures can be modest [4].

Electrodermal activity (EDA), commonly measured as skin conductance or galvanic skin response (GSR), is closely related to sympathetic sudomotor activity and is therefore frequently used as a reference signal in stress-monitoring studies [5], [6]. Wearable EDA research has expanded substantially, but motion, electrode, temperature, humidity, and other artifacts can affect measurements, especially outside controlled laboratory settings [5], [7]. Multimodal wearable studies increasingly combine EDA with heart rate, heart-rate variability, skin temperature, and other signals because no single physiological channel completely represents the stress response [8], [9].

Microwave sensing provides a complementary measurement modality because resonant structures respond to changes in the complex permittivity and conductivity of materials placed in regions of concentrated electromagnetic fields. Planar resonators are attractive for biomedical sensing because they are compact, low-profile, relatively inexpensive, and compatible with VNA-based characterization and wearable integration [10 - 12]. The sensing principle is well established in dielectric characterization: a material perturbation changes the effective capacitance and loss of a resonator, which can modify its resonant frequency, quality factor, and S-parameters [10 - 13].

For biological tissue, however, the physiological interpretation requires particular caution. The dielectric properties of tissue are strongly influenced by water content, ionic composition, frequency, temperature, and tissue structure [14], [15]. Sympathetic activation can increase sweat secretion through neural control of eccrine glands [16]. Sweat contains water and electrolytes and changes the electrical environment at the skin surface; consequently, it is physically plausible that changes in surface hydration and conductivity can perturb a microwave near-field sensor. This mechanism provides a rationale for the present sensing concept, but it does not establish a one-to-one causal mapping between a microwave frequency shift and psychological stress. Environmental humidity, temperature, contact pressure, local anatomy, perfusion, motion, and individual sweat responses are potential confounders and must be controlled or modeled in future validation [7], [16], [17].

The objective of this revised study is therefore narrower and more defensible than the original claim: to characterize a planar SRR-loaded microstrip filter as a non-invasive electromagnetic sensor of stress-associated skin loading and to compare its response with GSR during controlled stress induction. The revision also separates simulation-

derived dielectric sensitivity from experimental stress association, corrects inconsistencies in the reported frequency shifts, expands the literature review, and explicitly identifies statistical quantities that cannot be reconstructed without the underlying subject-level data.

2. Methodology and Sensor Design

2.1 Operating Principle

The sensor consists of a 50-Ω microstrip transmission line coupled to a modified square SRR. The SRR creates a localized electric-field concentration around the split gap. When skin is placed over this active region, the local dielectric loading changes the effective capacitance and loss of the resonant network. A first-order lumped representation is

$$f_r = 1 / [2\pi\sqrt{(L(Cm + Csense))}],$$

where L is the effective resonator inductance, Cm is the intrinsic capacitance of the resonant structure, and Csense represents the additional capacitance associated with the material occupying the sensing region. This expression is a compact interpretation of the full-wave model rather than a replacement for HFSS simulation.

The complex relative permittivity can be represented as $\varepsilon^* = \varepsilon' - j\varepsilon''$, while an equivalent conductivity term contributes to the effective dielectric loss. In the electromagnetic model, the corresponding material response is therefore governed by both ε' and σ rather than by relative permittivity alone [14], [15].

2.2. Physical Configuration

Table 1 :

Parameter	Description	Value
Substrate	Rogers RT/duroid 5880	$\varepsilon r = 2.2$; $\tan\delta = 0.0009$
Substrate size	Length × width × thickness	30.0 × 25.0 × 1.57 mm³
Copper thickness	Conductor thickness	35 µm
W50	50-Ω transmission-line width	4.85 mm
LSRR	SRR outer edge length	12.20 mm
w	SRR conductor width	0.80 mm
g	SRR split gap	0.50 mm
s	Coupling gap to feedline	0.35 mm

2.3. Tissue-Loading Model

A three-layer skin model consisting of epidermis, dermis, and subcutaneous fat was used in the HFSS framework described by the manuscript. Two representative material states were used to explore the effect of increased dielectric loading. These values should be interpreted as electromagnetic simulation inputs, not as direct measurements of stress-induced permittivity in the participants.

Table 2 :

Modeled state	ε′	σ (S/m)	Interpretation
Resting / baseline	38.0	1.45	Reference tissue-loading condition
High tension / acute-stress proxy	46.5	1.95	Higher-loading simulation condition

This distinction is important because standard tissue-property databases describe dielectric behavior as a function of tissue type, frequency, temperature, and other factors; they do not by themselves provide a validated universal conversion from psychological stress level to skin ε′ and σ [14], [15]. The values above are therefore retained as the study's parametric simulation conditions, while the physiological interpretation is presented as a hypothesis supported by the sweat/EDA literature [5], [16].

3. Results

3.1. Simulated Resonance Response

The simulated unloaded structure exhibits a band-stop response centered at 2.450 GHz with a reported |S21| notch depth of −38.60 dB. With skin loading, the resonance moves downward because the dielectric loading increases the effective capacitance of the sensing region. The original manuscript contained an internal inconsistency by describing 2.450 → 2.312 GHz as the baseline-to-high-tension shift. That 138 MHz difference is actually the unloaded-air to resting-skin transition. The tissue-state sweep from ε′ = 38.0 to 46.5 corresponds to 2.312 → 2.185 GHz, i.e., 127 MHz.

Table 3 :

Physiological / loading state	ε′	σ (S/m)	f_r (GHz)	Δf_r from air (MHz)	|S11| (dB)	|S21| (dB)
Unloaded air	1.0	0.00	2.450	0	−1.12	−38.60
Resting / relaxed	38.0	1.45	2.312	138	−2.85	−34.10
Mild tension	41.0	1.60	2.268	182	−3.40	−31.80
Moderate tension	43.5	1.78	2.231	219	−4.10	−29.40
High tension / acute-stress proxy	46.5	1.95	2.185	265	−4.95	−27.10

The simulated resonance decreases monotonically as the modeled ε′ and conductivity increase. Across the resting-to-high-loading range, the frequency shift is 127 MHz over Δε′ = 8.5, giving an average sensitivity of 14.94 MHz per unit change in ε′. The

reported value of 16.2 MHz/ε′ in the original manuscript is not reproduced by the tabulated points and has therefore been removed. Likewise, the phrase 'linear frequency shift' has been replaced by 'monotonic shift' because the five tabulated points alone do not establish a statistically tested linear model.

3.2. Sensitivity

Sensitivity is defined as

$$S = |\Delta f_r| / |\Delta \varepsilon'|.$$

Using the resting and high-loading simulation points, S = |2.185 − 2.312| GHz / |46.5 − 38.0| = 0.127 GHz / 8.5 = 14.94 MHz per unit ε′. This quantity is a simulation sensitivity to the specified dielectric parameter range; it should not be reported as a direct clinical sensitivity to psychological stress.

3.3. Experimental Measurements

The prototype was fabricated on Rogers RT/duroid 5880 and measured with a Keysight N5224A two-port VNA over 1.5–3.0 GHz. The original manuscript reports 12 healthy adult participants and a protocol comprising baseline relaxation, a Stroop Color-Word task, and recovery, with GSR recorded concurrently.

Table 4 :

Phase	Mean GSR (μS)	Measured f_r (GHz)	Measured Δf_r (MHz)	Reported R^2
Baseline / resting	2.14 ± 0.35	2.315 ± 0.008	0	—
Stroop task / tension	6.85 ± 0.92	2.242 ± 0.012	73	0.984
Recovery	2.80 ± 0.41	2.301 ± 0.009	14	0.976

The experimental data show the expected direction of change at the phase-average level: GSR increased during the stress task and moved toward baseline during recovery, while the measured resonance shifted to a lower frequency during the task and partially recovered afterward. These observations are consistent with a stress-associated change in skin loading, but they do not prove that the microwave sensor uniquely measures psychological stress.

3.4. Statistical Analysis and Reporting Boundaries

The review correctly identifies that a single R^2 value is insufficient to establish statistical reliability. The manuscript does not provide the 12 participant-level paired observations needed to independently calculate a valid confidence interval for the correlation/regression coefficient, subject-level RMSE or MAE, or a repeated-measures model. It would be inappropriate to manufacture those quantities from the three phase means. Accordingly, the reported R^2 values are retained as manuscript-reported results rather than recomputed claims.

For the next experimental dataset, the recommended primary model is a repeated-measures regression in which resonance frequency shift (or normalized shift) is the dependent variable and GSR is a fixed-effect predictor, with participant included as a random intercept. A suitable model is:

$$\Delta f_{ij} = \beta_0 + \beta_1 GSR_{ij} + u_i + \varepsilon_{ij},$$

where i indexes participants, j indexes repeated observations, u_i is the participant-specific random effect, and ε_{ij} is the residual. The analysis should report β_1 with its 95% confidence interval and p-value, marginal and conditional R^2, RMSE, MAE, residual diagnostics, and within-subject versus between-subject variance. If the same subjects are measured across baseline, stress, and recovery, a repeated-measures ANOVA or linear mixed-effects model should be preferred over treating all observations as independent.

The three published phase means can support descriptive interpretation only. They are not sufficient for subject-level inferential statistics. This explicit limitation is a methodological correction to the original version and prevents overstatement of the evidence.

4. Physiological Mechanism and Discussion

The proposed mechanism can be described as a chain of physiological and electromagnetic events rather than as a direct stress-to-permittivity conversion. Acute psychosocial stress activates sympathetic pathways. Sympathetic sudomotor activity stimulates eccrine sweat secretion, and sweat changes the local water and electrolyte environment at the skin surface [16], [17]. Because water and ionic content strongly influence the dielectric and conductive properties of biological tissues at microwave frequencies [14], [15], changes in the local skin environment can alter the electromagnetic boundary condition presented to the SRR.

The sensor is particularly sensitive near the split gap because the resonator stores electric energy in a localized region. Increased effective dielectric loading increases the effective capacitance and can therefore lower the resonant frequency. Increased conductivity also increases dielectric loss, which can modify the depth and quality factor of the resonance. This interpretation is consistent with the general operating principles of planar microwave resonant sensors, in which changes in the material under test perturb the resonant response [10 - 13].

The relationship should nevertheless be described as indirect. GSR is itself an index of sympathetic sudomotor activity rather than a direct measurement of subjective psychological stress [5 - 7]. Likewise, skin dielectric properties depend on several variables besides stress, including hydration, temperature, anatomical location, tissue composition, contact conditions, and frequency [14], [15]. Consequently, a frequency shift observed during a stress task may reflect a combination of autonomic sweating and other task-related physiological changes. This is especially important for a wearable implementation, where motion and environmental effects can be substantial [7], [18].

The experimental trend is therefore best interpreted as evidence of feasibility rather than clinical validation. The agreement between the GSR trajectory and the

microwave response supports further investigation, but a larger cohort with randomized task order, repeated sessions, controlled contact pressure, skin temperature and ambient humidity, and participant-level paired data is needed to establish specificity and generalizability. Recent wearable-stress literature similarly emphasizes individual variability, multimodal sensing, and the importance of rigorous experimental protocols [8], [9], [19].

4.1. Comparison With Related Microwave Sensors

Table 5 :

Study	Application	Frequency	Platform	Reported contribution
Al-Behadili et al. [20]	Organic-fluid permittivity	1.25 GHz	Differential microstrip / FR-4	Complex-permittivity characterization; ~3% sensitivity
Morales-Lovera et al. [21]	Anisotropic permittivity	Variable	Microstrip perturbation sensor	Real/imaginary permittivity characterization
Chokenukul & Worasawate [22]	Permittivity measurement	ISM frequencies	Microstrip / lookup table	Low-cost permittivity estimation framework
Nella et al. [23]	Non-invasive glucose sensing	5.5 GHz	Microstrip band-stop / FR-4	Finger-phantom simulation and measurement verification
This work	Stress-associated skin loading	2.45 GHz	SRR-loaded microstrip / RT-duroid 5880	127 MHz simulated shift over $\varepsilon' = 38.0–46.5$; 14.94 MHz/ε'

The comparison demonstrates that planar resonators can translate dielectric perturbations into measurable frequency changes, but the sensing targets and validation standards differ substantially. In particular, liquid-permittivity and phantom studies do not validate psychological-stress detection in humans. The principal contribution of the present work is therefore the combination of an SRR-loaded planar sensor with controlled human stress-task measurements and a GSR reference, while acknowledging that physiological specificity remains to be established.

4.2. Practical Limitations and Required Validation

Several limitations should be addressed before the device is described as a clinical or diagnostic stress monitor. First, the cohort of 12 adults is small for estimating population-level performance and does not support strong claims of generalizability. Second, the manuscript does not provide participant-level raw data, preventing independent calculation of confidence intervals, RMSE/MAE, mixed-effects statistics, and subject-to-subject variability. Third, the simulated ε' and σ values are parametric states rather than experimentally measured stress-dependent skin

properties. Fourth, the measurement protocol should report sensor placement, contact pressure, stabilization time, ambient temperature and relative humidity, VNA calibration procedure, connector repeatability, and the exact timing of VNA/GSR acquisition relative to each task. Fifth, motion and temperature should be monitored because they can affect both EDA and microwave measurements [7], [18].

For a stronger Q1-level validation, the next study should include (i) at least 20–30 participants as an initial engineering-validation cohort, with a prespecified power analysis for the final study; (ii) repeated baseline, stress, and recovery measurements; (iii) randomized or counterbalanced stress-task order where feasible; (iv) synchronized raw VNA and GSR streams; (v) environmental and contact-condition logging; (vi) a sham or non-stress control condition; and (vii) participant-level statistical modeling. Where resources permit, an additional physiological reference such as heart rate/HRV or skin temperature would help distinguish general autonomic arousal from sensor-specific effects [4], [8], [9].

4.3. Technical Evidence Recommended for the Final Submission

The review requested stronger technical evidence. The revised manuscript does not fabricate field plots, equivalent circuits, or simulation-versus-measurement curves that were not present in the supplied material. The final submission should insert the actual HFSS and laboratory figures at the indicated locations:

Table 6 :

Required item	Minimum content
Sensor geometry	Top view with all dimensions and a 3D perspective or cross-section showing substrate and SRR.
Electric-field distribution	HFSS |E| map at resonance showing concentration at the SRR split gap.
Surface-current distribution	HFSS surface-current plot at resonance demonstrating current localization on the resonator.
Equivalent circuit	L–C representation linked explicitly to the SRR geometry and skin-loading capacitance.
Simulation response	|S11| and |S21| curves for unloaded and each modeled dielectric state.
Experimental setup	Photograph/schematic showing sensor, VNA, calibration plane, participant placement, and GSR reference.
Simulation–experiment comparison	Same-axis simulated and measured resonance curves with quantified frequency error.
Participant-level statistics	Scatter plot of Δf_r versus GSR with regression line, 95% CI, and subject identifiers or repeated-measure structure.

5. Conclusion

A compact planar SRR-loaded microstrip band-stop sensor operating near 2.45 GHz was evaluated as a non-invasive electromagnetic approach for monitoring stress-associated physiological changes. The revised analysis separates electromagnetic

sensitivity from psychological interpretation. In the HFSS model, the resonance decreased from 2.312 to 2.185 GHz as the specified skin ε′ increased from 38.0 to 46.5, corresponding to a 127 MHz shift and an average sensitivity of 14.94 MHz per unit ε′. The previously stated 138 MHz value corresponds instead to the unloaded-air to resting-skin transition.

The experimental measurements from 12 adults showed a lower resonance during the Stroop stress task and recovery toward baseline, while GSR increased during the task. These parallel trends are compatible with the proposed mechanism in which sympathetic sudomotor activation changes the local skin electromagnetic environment. However, the present evidence does not establish diagnostic stress detection or a unique stress-to-permittivity relationship. The reported $R^2 = 0.984$ is retained as a manuscript-reported association, while inferential statistics requiring participant-level observations are explicitly deferred until the raw dataset is available.

Future work should focus on larger and repeated-measures experiments, direct characterization of skin loading, environmental and motion control, synchronized multimodal physiological sensing, and transparent subject-level statistical analysis. These steps are necessary to determine whether the proposed microwave response can provide a robust and generalizable marker of stress-related autonomic activity.

Conflict of interest **:** Authors are declared that there is no conflict of interest regarding this study.

Ethics approval **:** Not applicable

Code availability **:** Available from the author on reasonable request.

AI-Assistance Disclosure**:** Drafting and editing of this manuscript were assisted by a general-purpose language model. The authors independently reviewed and revised all AI-assisted content, ensuring that the final text reflects the authors' own academic voice, interpretation, and responsibility for the accuracy and integrity of the work.

Availability of data and materials**:** Data sharing does not apply to this article as no new datasets were generated or analyzed during the current study.

Authors' contributions **:** The author has been written and revised the whole paper.

References

1. C. Kirschbaum, K.-M. Pirke, and D. H. Hellhammer, "The 'Trier Social Stress Test'—a tool for investigating psychobiological stress responses in a laboratory setting," Neuropsychobiology, vol. 28, no. 1–2, pp. 76–81, 1993. 10.1159/000119004.
2. N. F. Narvaez Linares, V. Charron, A. J. Ouimet, P. R. Labelle, and H. Plamondon, "A systematic review of the Trier Social Stress Test methodology: Issues in promoting study comparison and replicable research," Neurobiology of Stress, vol. 13, p. 100235, 2020. 10.1016/j.ynstr.2020.100235.
3. A. Allen, E. M. Kennedy, L. Cryan, M. Dinan, and D. Clarke, "Biological and psychological markers of stress in humans: Focus on the Trier Social Stress Test," Neuroscience & Biobehavioral Reviews, vol. 38, pp. 123–135, 2014. 10.1016/j.neubiorev.2013.11.002.
4. H. Gu, Y. Lei, Y. Yao, C. Chen, and C. Liu, "Physiological and psychological responses to acute stress: A meta-analysis of the 171 studies of Trier Social Stress Test including 8452 healthy adults," Psychoneuroendocrinology, vol. 180, p. 107566, 2025. 10.1016/j.psyneuen.2025.107566.
5. H. F. Posada-Quintero and K. H. Chon, "Innovations in electrodermal activity data collection and signal processing: A systematic review," Sensors, vol. 20, no. 2, p. 479, 2020. 10.3390/s20020479.
6. A. Klimek, I. Mannheim, G. Schouten, E. J. M. Wouters, and M. W. H. Peeters, "Wearables measuring electrodermal activity to assess perceived stress in care: A scoping review," Acta Neuropsychiatrica, vol. 37, p. e19, 2023. 10.1017/neu.2023.19.
7. D. S. Bari, H. Y. Y. Aldosky, and Ø. G. Martinsen, "Electrodermal activity measurements: A review of artifacts," Physiological Measurement, vol. 47, no. 7, 2026. 10.1088/1361-6579/ae7bae.
8. G. Vos, K. Trinh, Z. Sarnyai, and M. Rahimi Azghadi, "Generalizable machine learning for stress monitoring from wearable devices: A systematic literature review," International Journal of Medical Informatics, vol. 173, p. 105026, 2023. 10.1016/j.ijmedinf.2023.105026.
9. A. G. Peitzsch, Y. Kong, H. F. Posada-Quintero, and K. H. Chon, "Objective assessment of acute stress disorder in female subjects using wearable measures of electrodermal activity," in Proc. IEEE EMBC, 2024. 10.1109/EMBC53108.2024.10782849.
10. I. Royo, R. Fernández-García, and I. Gil, "Microwave resonators for wearable sensors design: A systematic review," Sensors, vol. 23, no. 22, p. 9103, 2023. 10.3390/s23229103.

11. M. Abdolrazzaghi, V. Nayyeri, and F. Martín, "Techniques to improve the performance of planar microwave sensors: A review and recent developments," Sensors, vol. 22, no. 18, p. 6946, 2022. 10.3390/s22186946.
12. S. Alahnomi, A. Zakaria, A. Mohd Bahar, M. A. M. Rasid, M. R. M. Hasin, and S. A. Mohd Shah, "Review of recent microwave planar resonator-based sensors: Techniques of complex permittivity extraction, applications, open challenges and future research directions," Sensors, vol. 21, no. 7, 2021. 10.3390/s21072408.
13. A. A. al-Behadili, I. A. Mocanu, T. M. Petrescu, and T. A. Elwi, "Differential microstrip sensor for complex permittivity characterization of organic fluid mixtures," Sensors, vol. 21, no. 23, p. 7865, 2021. 10.3390/s21237865.
14. C. Gabriel, S. Gabriel, and E. Corthout, "The dielectric properties of biological tissues: I. Literature survey," Physics in Medicine and Biology, vol. 41, no. 11, pp. 2231–2249, 1996. 10.1088/0031-9155/41/11/001.
15. S. Gabriel, R. W. Lau, and C. Gabriel, "The dielectric properties of biological tissues: II. Measurements in the frequency range 10 Hz to 20 GHz," Physics in Medicine and Biology, vol. 41, no. 11, pp. 2251–2269, 1996. 10.1088/0031-9155/41/11/002.
16. Y. Hu, C. Converse, M. C. Lyons, and W. H. Hsu, "Neural control of sweat secretion: A review," British Journal of Dermatology, vol. 178, no. 6, pp. 1246–1256, 2018. 10.1111/bjd.15808.
17. S. Na, H.-J. Lee, T.-H. Park, J. Kim, Y. H. Jung, M. Cha, H.-Y. Lee, H.-J. Kim, and J.-B. Lee, "Sweat glands in action: The role of biological and environmental factors, exercise, and dermatologic conditions in sudomotor function," Korean Journal of Physiology and Pharmacology, 2026.
18. S. Alotaibi et al., "Wearable devices for continuous monitoring of biosignals: Challenges and opportunities," Review of Scientific Instruments, 2022. 10.1063/5.0086935.
19. E. R. M. P. et al., "Personalized stress detection using biosignals from wearables: A scoping review," Sensors, vol. 24, no. 10, p. 3221, 2024, doi: 10.3390/s24103221.
20. H.-N. Morales-Lovera, J.-L. Olvera-Cervantes, A.-E. Perez-Ramos, A. Corona-Chavez, and C. E. Saavedra, "Microstrip sensor and methodology for the determination of complex anisotropic permittivity using perturbation techniques," Scientific Reports, vol. 12, p. 2205, 2022. 10.1038/s41598-022-06259-8.

21. P. Chokenukul and D. Worasawate, “Simulation of a permittivity measurement system with microstrip structure at ISM frequency incorporate with lookup table,” Engineering and Technology Horizons, vol. 40, no. 4, p. 400403, 2023. 10.55003/ETH.400403.
22. A. Nella, R. W. Aldhaheri, J. B. Kamili, and A. A. Jiman, “A novel microstrip band-stop filter at 5.5 GHz for non-invasive blood glucose monitoring,” Applied Sciences, vol. 16, no. 7, p. 3197, 2026. 10.3390/app16073197.