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Real-Life Skin Impedance Variations with Dry and Wet Electrodes for Bio-Potential Signal Acquisition

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Abstract

Capturing bio-potential signals such as electrocardiograms (ECGs), electromyograms (EMGs), and electroencephalograms (EEGs) is essential for diagnosing and monitoring diverse physiological conditions, including cardiac arrhythmias, neuromuscular disorders, and brain activity. These signals reflect real-time electrophysiological states, enabling timely detection and management of critical medical conditions. Traditionally, Ag/AgCl wet electrodes with conductive gel provide superior signal quality by reducing skin-electrode impedance; however, their limitations such as gel drying, skin irritation, and short usability span hinder long-term and wearable applications. Dry electrodes offer advantages in comfort, biocompatibility, and long-term usability, but face challenges related to high and variable skin impedance, which impacts signal fidelity and analog front-end (AFE) circuit design.

This study presents an empirical characterization of skin-electrode impedance using dry stainless-steel and wet (gel-mimicked) electrodes on the palmar surface of the thumbs in 15 adult subjects (age: 23–60 years). Impedance was measured at six logarithmically spaced frequencies 20 Hz, 100 Hz, 1 kHz, 10 kHz, 100 kHz, and 1 MHz and at two temporal intervals (5 seconds and 1-minute post-application). Results indicate significant differences in impedance magnitude and behavior across electrode types and frequencies. At 20 Hz, the average impedance for dry electrodes was 133.1 kΩ at 5 seconds, increasing marginally to 139.5 kΩ at 1 minute, whereas wet electrodes exhibited 60.2 kΩ and 64.7 kΩ, respectively. Impedance decreased with frequency for both types, with convergence observed at 1 MHz (dry: 1.24 kΩ, wet: 1.13 kΩ), consistent with the behavior of the skin-electrode interface modeled as a dispersive RC network.

Inter-subject variability was also pronounced, with dry electrode impedance at 1 kHz ranging from 9.7 kΩ to 87.5 kΩ (mean: 41.2 kΩ), compared to wet electrodes which ranged from 6.5 kΩ to 31.8 kΩ (mean: 14.3 kΩ). These variations underscore the necessity of designing high input impedance AFEs (>10 MΩ) and adaptive impedance-matching techniques to ensure optimal signal capture and patient safety under diverse physiological conditions.

These findings provide critical input parameters for biomedical circuit designers, particularly in the context of front-end amplifier design, current-limiting protection circuitry, and electrode selection for wearable and implantable bio-potential sensing systems. This work advances the understanding of real-world skin impedance characteristics and supports the development of robust, low-power, non-invasive electrophysiological monitoring technologies.

Keywords: Skin impedance, Bio-potential signals, Dry electrodes, Wet electrodes, LCR Meter, Electrode-skin interface

1. Introduction

The acquisition of bio-potential signals such as electrocardiograms (ECGs), electromyograms (EMGs), and electroencephalograms (EEGs) forms the cornerstone of physiological monitoring, diagnostic evaluation, and closed-loop therapeutic systems. These signals represent electrical activity originating from cardiac, muscular, or neural sources and are captured via electrodes placed on the skin surface or invasively. In non-invasive applications, surface electrodes serve as the interface between the biological medium and electronic acquisition systems, necessitating robust coupling with minimal signal distortion, consistent contact impedance, and high signal-to-noise ratio (SNR).

2. Background on Bio-Potential Signal Acquisition in Medical Systems

Medical-grade instrumentation for capturing bio-potentials demands front-end circuits with high input impedance, wide dynamic range, and strong common-mode rejection to accurately represent microvolt-level signals amid noise, motion artifacts, and electrode-skin interface inconsistencies [7]. These requirements are particularly critical in ambulatory, long-term, or wearable healthcare devices where consistent signal quality must be maintained under dynamic physiological and environmental conditions. Accurate ECG signal acquisition, for instance, is pivotal in early detection of cardiac arrhythmias, ischemic events, and electrophysiological abnormalities. Similarly, EMG and EEG applications depend heavily on stable electrode-skin coupling to characterize muscle activation and cerebral dynamics, respectively.

Traditionally, wet electrodes typically composed of Ag/AgCl sensors embedded in hydrogel have been the gold standard due to their ability to reduce skin–electrode impedance and improve electrical coupling. The hydrophilic gel acts as an electrolyte that bridges the stratum corneum's high impedance barrier, thereby lowering the interface impedance to the range of 5–10 kΩ at 10 Hz and enabling high-fidelity signal acquisition [1]. However, wet electrodes suffer from performance degradation over time due to gel dehydration, leading to rising impedance, increased motion artifacts, and skin irritation during prolonged use. In contrast, dry electrodes eliminate the need for conductive gels, offering advantages in long-term monitoring, user comfort, and mechanical robustness. These electrodes are fabricated from a wide variety of materials, including stainless steel, gold-plated brass, TiN-coated titanium, carbon-based composites, and conductive polymers such as PEDOT:PSS. However, their impedance is typically higher and more variable, often exceeding 100 kΩ at low frequencies, making signal acquisition more susceptible to common-mode interference and thermal noise [1].

Recent efforts in dry electrode development include the use of micro-structured surfaces, spring-loaded pin arrays, and flexible conductive substrates (e.g., graphene-infused elastomers and e-textiles) to improve mechanical contact and reduce interface impedance [2]. Still, the performance of dry electrodes remains significantly influenced by the physical and electrical properties of human skin, particularly the outermost stratum corneum layer, which exhibits capacitive and resistive behavior varying with hydration, pressure, age, and anatomical location

One of the foremost challenges in designing front-end acquisition circuits for dry electrode-based systems is the high inter-subject and intra-subject variability in skin impedance. Human skin impedance is not only nonlinear but also frequency-dependent, often modeled as a parallel RC network whose parameters change with sweat gland activity, motion, and electrode pressure. Impedance magnitudes can vary by orders of magnitude between individuals, and impedance phase angle introduces additional complexity in signal phase accuracy. Such variability directly impacts the analog front-end (AFE) design, particularly the input impedance requirements of instrumentation amplifiers, bias resistor sizing, and protection circuitry. Mismatched impedance may lead to signal attenuation, phase distortion, and potential safety concerns due to inadvertent current paths. Moreover, high skin–electrode impedance increases the risk of saturation in low-noise amplifiers and reduces common-mode rejection ratio (CMRR), degrading overall signal quality.

3. Objective and Contributions of This Study

This study aims to systematically investigate the **real-life skin impedance characteristics** of human subjects across a diverse population sample, comparing **dry stainless steel electrodes** with **wet electrodes simulated using electrode gel** under controlled conditions. By capturing impedance spectra across a relevant frequency band (e.g., 0.1 Hz–10 kHz), the study quantifies inter-subject impedance variability and its influence on signal acquisition.

Key contributions of this research include:

- **Empirical quantification** of skin impedance differences between dry and wet electrodes in a clinical-like, real-world setting across 15 subjects aged 23–65.
- **Identification of trends** in impedance variability with respect to electrode type, age group, and gender, providing statistical insight into design parameters.
- **Guidelines for analog front-end circuit optimization**, including input impedance thresholds, biasing strategies, and protection design to accommodate impedance fluctuations.
- **Implications for patient safety**, especially in wearable and implantable diagnostic systems where electrode–skin interaction governs both performance and safety margins.

By elucidating these impedance variations and their practical consequences, this work seeks to aid biomedical engineers and device designers in creating more robust, user-friendly, and high-performance bio- potential acquisition systems suitable for next-generation healthcare technologies.

4. Materials and Methods

4.1 Measurement System for Skin Impedance Recording

To accurately measure skin impedance across selected anatomical sites, the Keysight E4980A Precision LCR Meter was employed. This instrument offers a frequency range from 20 Hz to 2 MHz with a basic impedance accuracy of 0.05%, ensuring precise and reliable measurements across the impedance spectrum relevant to bio-potential signal acquisition [3]. Some of the main key specifications of the E4980A LCR meter are:

- **Test Frequency Range:** 20 Hz to 2 MHz with 4-digit resolution
- **Measurement Speed:**
    - SHORT mode: 5.6 ms at 1 MHz

    - MED mode: 88 ms at 1 MHz
    - LONG mode: 220 ms at 1 MHz
- **Test Signal Levels:** Adjustable from 100 μV to 2 Vrms, accommodating various electrode-skin interface conditions
- **DC Bias Capability:** Standard 1.5 V/2 V; with Option 001, extends to ±40 V DC bias with 0.3 mV resolution
- **Cable Extension Support:** 1/2/4-meter extensions with Open/Short/Load compensation, facilitating flexible test setup.

The LCR meter was configured to measure impedance at discrete frequencies ranging from DC (0 Hz) and spanning 20 Hz to 10 kHz. This frequency range encompasses the spectral components of typical bio-potential signals such as ECG, EMG, and EEG, thereby providing insights into the frequency-dependent behavior of skin impedance.

The measurement procedure involved:

1. Electrode Placement: Electrodes were positioned at the palmar surface of both the thumbs
2. Connection to LCR Meter: The electrodes were connected to the LCR meter's test ports using shielded cables to minimize electromagnetic interference and ensure signal integrity.
3. Impedance Measurement: For each electrode configuration (dry and wet), impedance measurements were recorded at the specified frequencies. The instrument's open/short/load compensation was utilized to correct for any fixture-related errors, ensuring measurement accuracy.

4.2 Specifications of Dry and Wet Electrodes Used

Dry Electrodes: The material used for dry electrode is Stainless Steel 316L and the dimensions of the electrode were 1 inch × 1 inch (25.4 mm × 25.4 mm). Stainless Steel 316L was chosen for the dry electrodes due to its favorable electrochemical properties and biocompatibility [4]:

- Corrosion Resistance: 316L stainless steel exhibits excellent resistance to corrosion, particularly against chlorides and other industrial solvents, ensuring durability in prolonged skin contact scenarios.
- Biocompatibility: Widely used in medical implants, 316L stainless steel is known for its compatibility with human tissue, minimizing the risk of adverse reactions.
- Electrical Conductivity: While not as conductive as noble metals like gold, stainless steel provides sufficient conductivity for bio-potential signal acquisition when designed appropriately.

Wet Electrodes: Application of Spectra 360 Electrode Gel was used to the stainless steel electrodes to emulate wet electrode conditions. Spectra 360 is a chloride-free, salt-free electrically conductive gel, distinct from traditional electrode pastes:

- Mechanism of Action: Operates by wetting the skin to reduce impedance, rather than relying on ionic (salt-based) conduction [5].
- Prolonged Use: The non-gritty STAY-WET formula allows for extended application without the need for re-application, making it suitable for long-term monitoring [5].
- Biocompatibility: Non-irritating and bacteriostatic properties enhance patient comfort and safety.

- Application Suitability: Recommended for ECG, TENS, monitoring, and pediatric applications; however, it is not advised for defibrillation purposes [5].

A uniform layer of Spectra 360 gel was applied to the contact surface of the stainless steel electrodes before placement on the skin. This ensured consistent electrode-skin interface characteristics, facilitating accurate simulation of wet electrode conditions.

5. Subject Recruitment and Data Collection Protocol

5.1 Subject Recruitment

Fifteen healthy adult volunteers (12 males, 3 females), aged between 23 and 65 years, were recruited to evaluate skin impedance characteristics across a representative population. Subjects exhibited diverse anthropometric profiles, with heights ranging from 140 cm to 190 cm and body weights from 54 kg to 91 kg, encompassing a wide spectrum of body types. The cohort was selected to ensure statistically relevant impedance data across sex, age, and somatotype distributions. All procedures complied with ethical guidelines, and informed consent was obtained from each participant prior to measurement. Table 1 shows the distribution of their gender, age and body physical sizes.

Gender	Age	Weight	Height
Male	35	159	178
Male	39	175	167
Male	25	164	176
Male	51	171	171
Male	63	155	165
Female	24	108	155
Male	34	181	176
Male	22	169	183
Female	41	113	154
Male	29	168	177
Male	54	165	161
Male	26	185	191
Male	28	155	167
Male	65	171	180
Male	48	112	155

Table 1: Subjects Physiological Trait Distribution

5.2 Data Collection Protocol

Impedance measurements were acquired under controlled laboratory conditions (ambient temperature: 25 °C, relative humidity: 30%) using a Keysight E4980A precision LCR meter interfaced via 18 AWG shielded copper leads to dual 1-inch × 1-inch 316L stainless steel plate electrodes. Subjects were instructed to press the palmar surfaces of the thumb and index finger from each hand against the electrodes, forming a consistent contact area for all trials. No pre-cleaning or skin abrasion was performed to emulate real-world usage conditions. For each subject, five impedance measurements were recorded at two temporal intervals: 5 seconds, and 1 minute post initial electrode contact, reflecting the dynamic evolution of the electrode-skin interface impedance over time. After these dry-electrode readings, a rest period of 3 minutes was enforced to allow electrodermal recovery. Subsequently, a thin film of Spectra 360 conductive gel was applied to the same contact surfaces to simulate wet electrode conditions. Identical impedance measurement intervals were repeated (5 s and1 min). All impedance readings were conducted at frequency points of 20Hz, 100Hz, 1KHz, 10KHz, 100KHz, and 1MHz using a constant 1 V RMS test signal in series mode configuration (Cs-Rs). This protocol ensures data fidelity across time, frequency, and electrode modality, enabling high-resolution characterization of interfacial impedance dynamics.

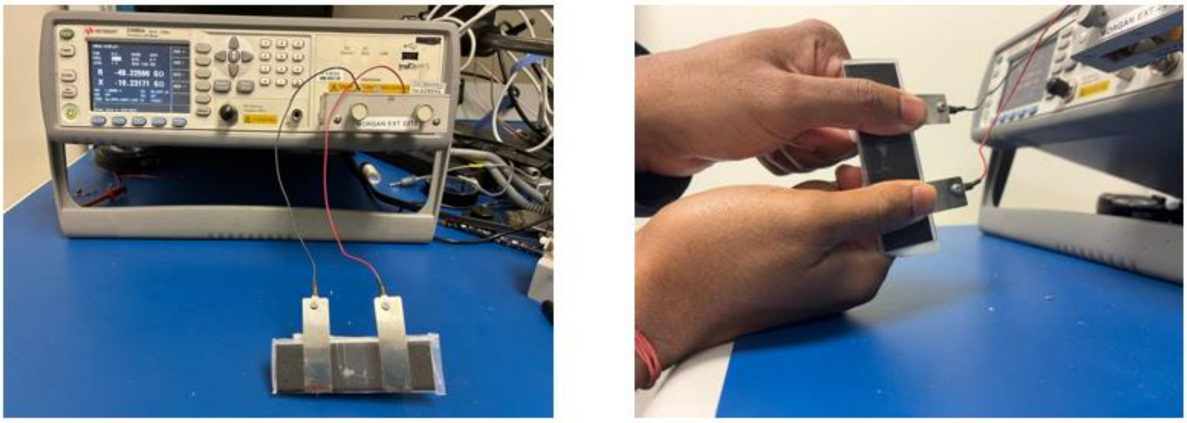

Figure 1: Test Setup for Impedance Measurements

5.3 Data Processing and Analysis

Raw impedance data captured from the Keysight E4980A LCR meter across the frequency domain (20, 100, 10KHz, 100KHz, and 1MHz) and time series (5 s, 1 min) were initially denoised and smoothed using a zero-phase digital Butterworth filter to remove high-frequency fluctuations and environmental artifacts. For each measurement interval and electrode modality (dry and wet), five consecutive impedance readings were acquired per subject to characterize intra-subject run-to-run variability and transient skin-electrode interface dynamics. The arithmetic mean of these five values was computed and used as the representative impedance for subsequent analysis at each time point.

To assess central tendency and dispersion of skin impedance across the population, bar plots were generated for all 15 subjects at 5 s, and 1 min using paired data for both dry and wet electrode conditions. These visualizations enabled direct patient-to-patient impedance comparisons and highlighted the stabilizing effect of conductive gel on interfacial resistance.

To statistically interrogate inter-subject impedance variation and to evaluate skin impedance distribution patterns, box-and-whisker plots were constructed for each time interval (5 s and 1 min) for both electrode types. These plots captured key distributional metrics including median, interquartile range (IQR), and potential outliers, offering insight into variability in cutaneous electrical properties. Signal quality metrics were derived by calculating the signal-to-noise ratio (SNR) at each measurement point, defined as:

$$\mathrm{SNR(dB)} = 20log10\frac{\mu z}{\sigma z}$$

where μZ is the mean impedance and σZ is the standard deviation across the five runs per timestamp. Stability of contact impedance was assessed using the coefficient of variation (CV):

$$\mathrm{CV} = \frac{\mu z}{\sigma z} * 100\%$$

These metrics quantify the temporal stability and reproducibility of the skin-electrode interface.

This robust analysis pipeline ensures a comprehensive characterization of skin impedance dynamics under realistic clinical conditions and informs front-end circuit designers on interface variability that may affect gain stability, impedance matching, and noise suppression strategies.

6. Results

This study investigated skin–electrode impedance characteristics across different frequencies (20 Hz to 1 MHz), two electrode types (wet and dry), and two measurement timestamps (5 seconds and 1 minute) across 15 human subjects. The data reveal several consistent trends in impedance variation as a function of frequency, time, and electrode type.

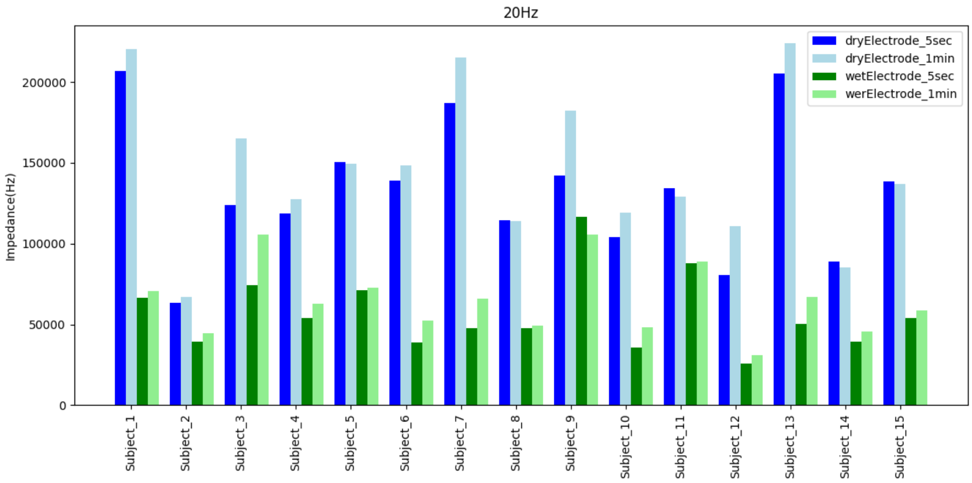

Figure 2: Impedance at 20Hz across 15 patients with wet electrode and dry electrode configuration

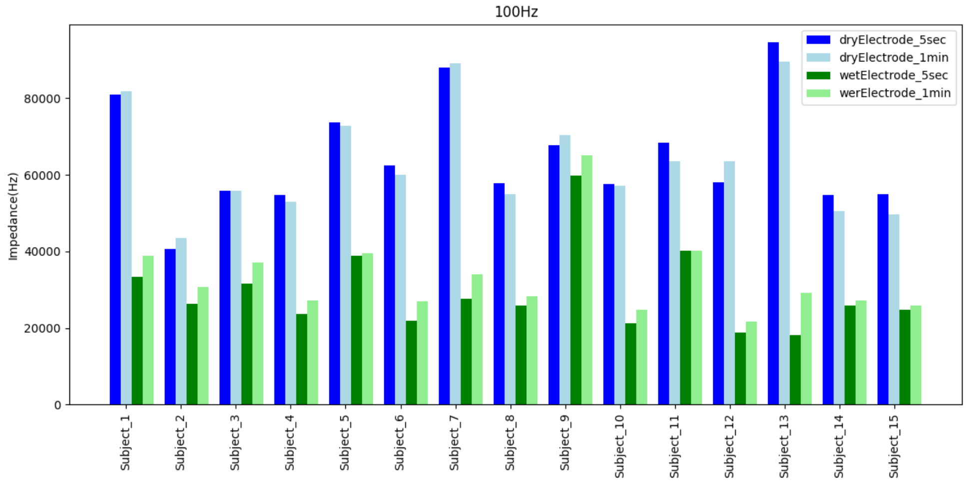

Figure 3: Impedance at 100Hz across 15 patients with wet electrode and dry electrode configuration

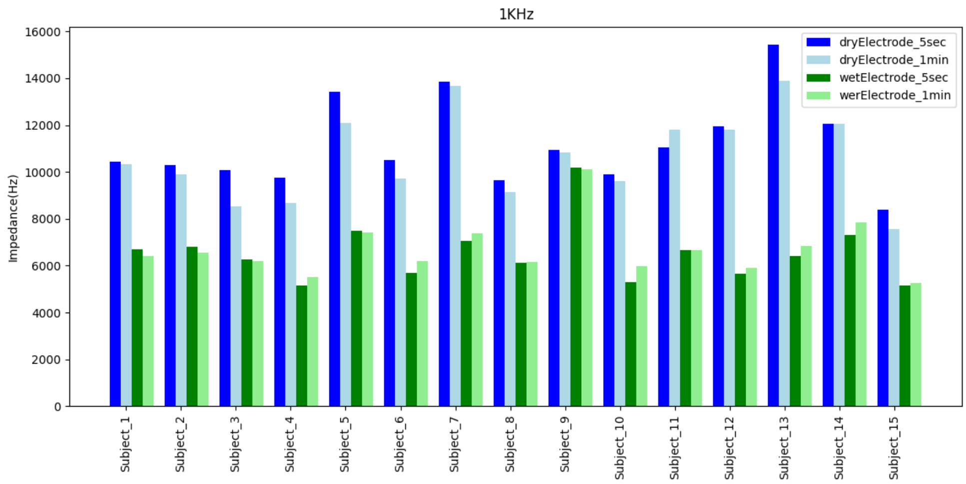

Figure 4: Impedance at 1KHz across 15 patients with wet electrode and dry electrode configuration

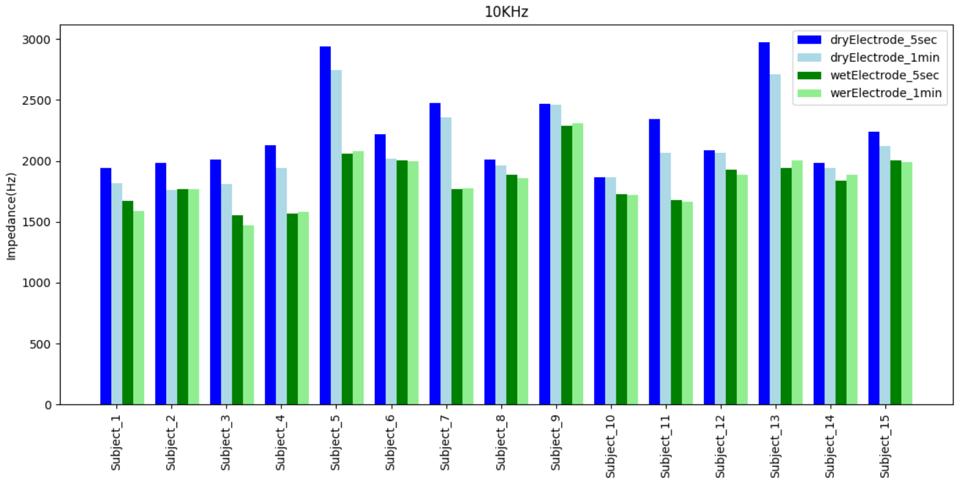

Figure 5: Impedance at 10KHz across 15 patients with wet electrode and dry electrode configuration

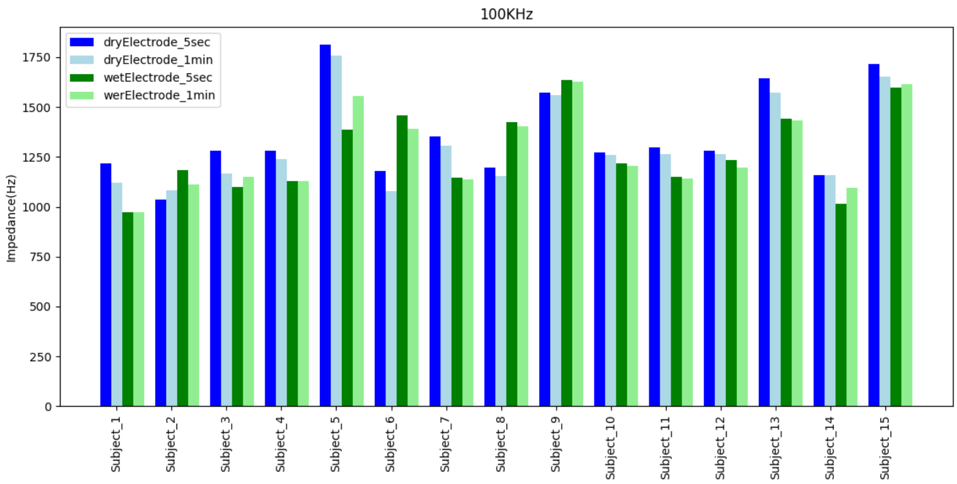

Figure 6: Impedance at 100KHz across 15 patients with wet electrode and dry electrode configuration

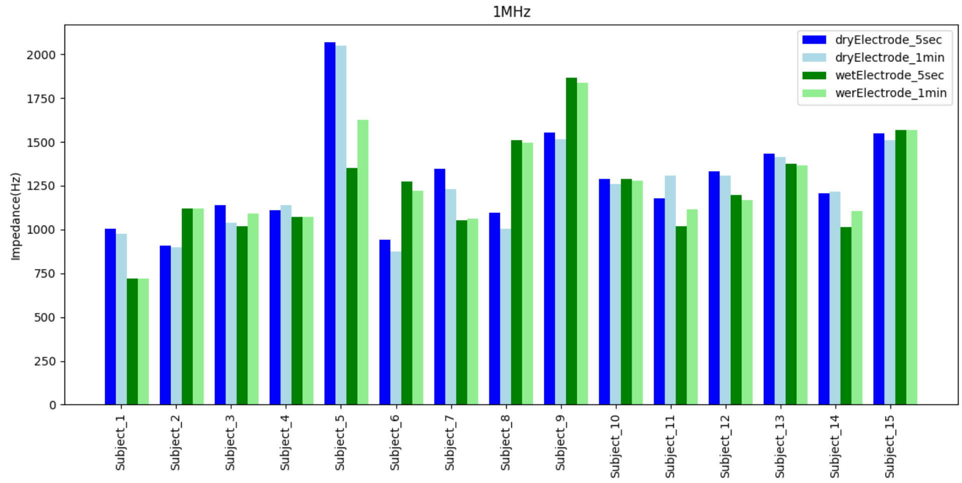

Figure 7: Impedance at 1MHz across 15 patients with wet electrode and dry electrode configuration

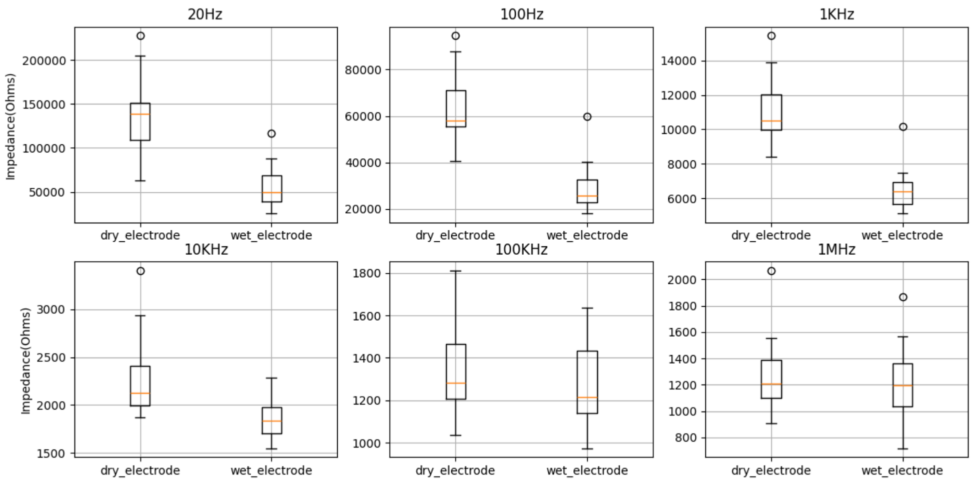

Figure 8: Impedance spread across 15 patients with wet electrode and dry electrode configuration

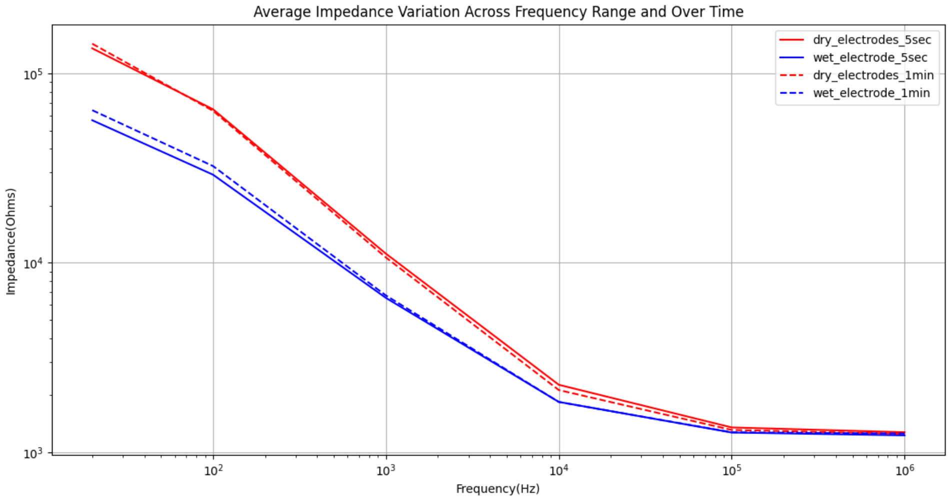

Figure 9: Variation of average impedance (across 15 subjects) with respect to frequency with wet electrode and dry electrode configuration

6.1 Impedance Variation Across Frequencies

Impedance measurements as plotted in Figure 9 show a strong inverse relationship with frequency. Across both dry and wet electrodes, the impedance significantly decreased as frequency increased from 20 Hz to 1 MHz. For instance, the average impedance using dry electrodes at a 5-second timestamp dropped from 135,487 Ω at 20 Hz to 1,248 Ω at 1 MHz. Similarly, the wet electrodes dropped from 58,382 Ω at 20 Hz to 1,222 Ω at 1 MHz. Specifically, the dry electrode average impedance values (at 5 seconds) were:

- 20 Hz: 135,487 Ω
- 100 Hz: 65,037 Ω
- 1 kHz: 11,181 Ω
- 10 kHz: 2,263 Ω
- 100 kHz: 1,332 Ω
- 1 MHz: 1,248 Ω

Wet electrodes at the same timestamps had lower values:

- 20 Hz: 58,382 Ω
- 100 Hz: 29,339 Ω
- 1 kHz: 6,625 Ω
- 10 kHz: 1,850 Ω
- 100 kHz: 1,243 Ω
- 1 MHz: 1,222 Ω

This trend is expected due to the capacitive nature of the skin–electrode interface, where impedance decreases with higher frequencies due to capacitive reactance reduction. The data illustrate this principle clearly and consistently across subjects.

6.2 Impedance Variation Over Time (5 seconds vs 1 minute)

Comparing the impedance values at 5 seconds and 1 minute from Figure 2 to Figure 7 reveals temporal stability patterns. In general, dry electrodes showed a tendency for impedance to increase or remain similar after 1 minute, while wet electrodes often showed slight increases or stabilization depending on the frequency.

At 20 Hz, dry electrode impedance increased from an average of 135,487 Ω (5 s) to 145,670 Ω (1 min)—a 7.5% increase. In contrast, wet electrodes at the same frequency showed a smaller change from 58,382 Ω to 62,246 Ω, a 6.6% increase. This may reflect drying effects or improved skin contact in dry electrodes over time.

At 100 Hz, dry electrode impedance rose slightly from 65,037 Ω to 65,429 Ω, while wet electrodes increased from 29,339 Ω to 32,447 Ω—a slightly more pronounced 10.6% increase, possibly due to gradual fluid redistribution or evaporation.

However, at higher frequencies such as 1 kHz and above, changes were minimal. For example, at 1 kHz, dry electrode impedance changed from 11,181 Ω to 10,661 Ω, and wet electrodes changed from 6,625 Ω to 6,678 Ω, both less than a 5% variation. This stability indicates frequency dominance over time effects at higher frequencies.

6.3 Impedance Differences Between Dry and Wet Electrodes

Figure 8 shows the impedance spread across 15 patients with wet electrode and dry electrode configuration. Wet electrodes consistently demonstrated significantly lower impedance across all frequencies and timestamps. At 20 Hz (5 s), dry electrodes had an average impedance of 135,487 Ω, more than twice the wet electrode impedance of 58,382 Ω. This difference persisted at 100 Hz, with dry electrodes averaging 65,037 Ω versus 29,339 Ω for wet electrodes. At 1 kHz, dry electrodes registered 11,181 Ω, while wet electrodes showed 6,625 Ω, a 40.7% reduction. At 10 kHz, dry impedance averaged 2,263 Ω vs. 1,850 Ω for wet—a smaller but consistent margin. Interestingly, at 1 MHz, the dry and wet impedance values converged, with dry electrodes averaging 1,248 Ω and wet electrodes 1,222 Ω. This convergence suggests that at high frequencies, the skin–electrode interface impedance becomes less influenced by the electrode's hydration or material properties and more dominated by intrinsic tissue properties.

6.4 Subject Variability

Standard deviation and range analysis as shown in Figure 2 to Figure 8 show significant subject-to-subject variability, especially at low frequencies. For example, dry electrode impedance at 20 Hz (5 s) ranged from 63,158 Ω to 206,784 Ω, showing over 140,000 Ω difference. Wet electrodes at the same point varied from 25,709 Ω to 116,619 Ω, a 4.5× difference between minimum and maximum.

At 1 MHz, variation narrowed: dry electrodes ranged from 906 Ω to 2,068 Ω, while wet electrodes ranged from 719 Ω to 1,867 Ω. This reduction in variability at higher frequencies again underscores the dominant capacitive coupling effect over resistive skin–electrode contact.

The impedance analysis across 15 subjects demonstrates three key findings:

1. **Strong frequency dependence**, with impedance decreasing from hundreds of kilo-ohms at 20 Hz to ~1.2 kΩ at 1 MHz.
2. **Minor time dependence**, particularly in dry electrodes at low frequencies, where impedance can increase slightly over one minute.
3. **Consistently lower impedance in wet electrodes**, especially pronounced at low frequencies, with differences diminishing as frequency increases.

These results confirm the electrochemical expectations of electrode-skin interfaces and support the use of wet electrodes for lower impedance requirements, while also validating the predictable performance of dry electrodes, particularly at higher frequencies.

7. Discussion

7.1 **Interpretation of Results and Implications for Bio-Potential Signal Acquisition**

The impedance measurements across various frequencies and time intervals reveal critical insights into the performance of dry and wet electrodes in bio-potential signal acquisition, specifically for the palmar side of the thumbs. At low frequencies (e.g., 20 Hz), dry electrodes exhibit significantly higher impedance values compared to wet electrodes. For instance, at the 5-second timestamp, the average impedance for dry electrodes is approximately 133,000 Ω, whereas wet electrodes average around 60,000 Ω. This disparity is primarily due to the absence of conductive gel in dry electrodes, leading to higher skin-electrode interface impedance [8]

Over time, the impedance values for both electrode types tend to stabilize. At the 1-minute mark, dry electrodes show a slight increase in average impedance to about 140,000 Ω, while wet electrodes increase to approximately 65,000 Ω. This stabilization is crucial for long-term monitoring applications, as consistent impedance ensures reliable signal acquisition [8].

As the frequency increases, the impedance values for both electrode types decrease, which is consistent with the capacitive nature of the skin-electrode interface. At 1 MHz, dry electrodes exhibit an average impedance of approximately 1,200 Ω, while wet electrodes average around 1,100 Ω. This convergence at higher frequencies suggests that dry electrodes can perform comparably to wet electrodes in applications involving higher frequency signals.

7.2 Comparison with Existing Literature

The observed impedance values align with findings from previous studies. For example, Kusche et al. reported that dry electrodes could achieve impedance levels comparable to wet electrodes under optimized conditions, such as increased contact pressure and surface area . Furthermore, studies have shown that dry electrodes, when designed with materials like conductive elastomers or carbon-based composites, can offer impedance values suitable for reliable signal acquisition [9].

However, it's important to note that while dry electrodes can achieve comparable impedance values, they are more susceptible to motion artifacts and may require additional design considerations to ensure stable contact with the skin [10].

7.3 Advantages and Limitations of Dry and Wet Electrodes

Advantages of Dry Electrodes:

- **Ease of Use:** Dry electrodes eliminate the need for skin preparation and conductive gels, simplifying the setup process.
- **Long-Term Monitoring:** They are more suitable for long-term monitoring as they do not dry out over time.
- **Reduced Skin Irritation:** The absence of gels reduces the risk of skin irritation, making them more comfortable for prolonged use [11].

Limitations of Dry Electrodes:

- **Higher Initial Impedance:** As observed, dry electrodes exhibit higher impedance at lower frequencies, which can affect signal quality.
- **Susceptibility to Motion Artifacts:** Without adhesive gels, maintaining consistent contact can be challenging, leading to motion artifacts.
- **Material Degradation:** Some dry electrode materials may degrade over time, especially with exposure to sweat and repeated use [12].

Advantages of Wet Electrodes:

- **Lower Impedance:** The conductive gel facilitates lower skin-electrode impedance, enhancing signal quality.
- **Stable Contact:** The adhesive nature of the gel ensures stable contact, reducing motion artifacts.

Limitations of Wet Electrodes:

- **Short-Term Use:** The gel can dry out over time, limiting their suitability for long-term monitoring.
- **Skin Irritation:** Prolonged use can lead to skin irritation or allergic reactions.
- **Complex Setup:** Requires skin preparation and careful application of the gel [11].

7.4 Potential Improvements and Alternative Electrode Materials

To address the limitations of both electrode types, research has focused on developing hybrid and novel materials:

- **Conductive Elastomers:** These materials combine the flexibility of elastomers with the conductivity of metals or carbon-based fillers, offering comfortable and stable contact with the skin [13].
- **Carbon Nanotube Arrays:** Incorporating carbon nanotubes can enhance conductivity and reduce impedance, while maintaining flexibility [13].
- **Metal-Coated Fabrics:** Textile-based electrodes coated with metals like silver can offer a balance between comfort and conductivity, making them suitable for wearable applications [12]
- **3D-Printed Electrodes:** Advancements in 3D printing allow for the customization of electrode shapes and sizes, improving fit and contact stability.

Implementing these materials and technologies can lead to electrodes that combine the advantages of both dry and wet types, offering low impedance, comfort, and stability for long-term bio-potential signal acquisition.

The study's findings underscore the trade-offs between dry and wet electrodes in bio-potential signal acquisition. While wet electrodes offer lower impedance and stable contact, their limitations in long-term use and potential for skin irritation are significant. Dry electrodes, though initially exhibiting higher impedance, show promise for long-term monitoring applications, especially with advancements in materials and design. Future research should focus on optimizing dry electrode materials and configurations to achieve impedance levels comparable to wet electrodes while maintaining user comfort and signal stability.

8. Conclusion and Future Work

This study systematically investigated the electrical impedance characteristics of wet and dry electrodes when applied to the palmar side of the thumbs across 15 human subjects. Impedance measurements were acquired at two time intervals (5 seconds and 1 minute post-application) and six logarithmically spaced frequencies (20 Hz to 1 MHz), allowing comprehensive analysis of both temporal and spectral behaviors.

The results highlight substantial impedance differences between wet and dry electrodes, particularly at lower frequencies. At 20 Hz, the average impedance for dry electrodes was approximately 133 kΩ at 5 seconds, increasing slightly to 140 kΩ at 1 minute, reflecting stabilization over time. In contrast, wet electrodes demonstrated significantly lower impedance ~ 60 kΩ at 5 seconds and ~65 kΩ at 1 minute due to the presence of conductive gel facilitating better skin-electrode coupling.

With increasing frequency, both electrode types exhibited decreased impedance, consistent with the skin-electrode interface acting as a parallel RC network. At 1 MHz, dry electrodes showed impedance values averaging ~1.2kΩ, while wet electrodes reached ~1.1kΩ, effectively converging, which suggests improved signal coupling at higher frequencies.

These findings emphasize the trade-off between ease of use and electrical performance. While dry electrodes offer convenience and better long-term viability due to the absence of gel, their elevated low-frequency impedance could compromise the fidelity of low-frequency bio-signals such as ECG or EEG. Wet electrodes continue to be superior for applications requiring low impedance at frequencies under 1 kHz.

Given the observed impedance ranges, wet electrodes are better suited for clinical settings where short-term signal fidelity is critical**,** whereas dry electrodes are promising for wearable and long-term monitoring solutions**,** provided additional compensation mechanisms (e.g., impedance adaptation or high input impedance amplifiers) are used.

Although this study provides important foundational data, it is constrained by several limitations that can be addressed in future investigations. The sample size of 15 subjects, while sufficient for initial trend identification, limits statistical generalizability. Future studies should include larger and more diverse cohorts, ideally 50–100 subjects, to account for inter-subject variability arising from factors such as skin hydration, age, sex, and baseline skin conductance. Larger samples will also enable robust statistical modeling of impedance characteristics and their variance over time and frequency.

Moreover, the palmar side of the thumbs while hairless and relatively homogeneous does not reflect the complexity of other body regions often used in clinical or wearable applications. For instance, areas such as the forearm, upper arm, chest, thigh, or scalp present more challenges due to hair density, skin oil content, and varying stratum corneum thickness, all of which can significantly alter the skin-electrode interface impedance. Hair on skin has shown to increase electrode-skin contact impedance due to reduced effective contact area and air gaps. Future work should involve systematic impedance characterization across such regions, with careful electrode placement and cleaning protocols to isolate anatomical effects.

Another dimension for future exploration is electrode material innovation. Our study used standard silver/silver chloride (Ag/AgCl) wet electrodes and commercial dry electrodes, but emerging materials like graphene, carbon nanotubes (CNTs), and metal-coated textiles have shown potential in reducing impedance and improving comfort in dry electrodes.

Additionally, the time-based impedance evolution observed in our study—where impedance increases slightly from 5 seconds to 1 minute warrants further temporal analysis. Extended recordings over 30 minutes to several hours could provide insight into electrode settling dynamics, long-term comfort, and adhesion performance. This would be particularly relevant for ambulatory ECG monitoring or sleep-stage analysis, where electrode detachment and impedance drift can severely degrade signal quality.

Finally, combining impedance measurements with machine learning-based classifiers may enable real-time prediction of electrode failure or need for reapplication, especially useful in remote or unsupervised monitoring environments.

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