Measurement 240 (2025) 115528

Contents lists available at ScienceDirect

Measurement

journal homepage: www.elsevier.com/locate/measurement

Seat to beat: Novel capacitive ECG integration for in-car cardiovascular measurement

Júlia Kafková [a,*], Branko Babušiak [b], Rastislav Pirník [a], Pavol Kuchár [a], Juraj Kekelák [a], Filippo D'Ippolito [c]

[a] Department of Control and Information Systems, University of Zilina, Zilina, Slovakia
[b] Department of Electromagnetic and Biomedical Engineering, University of Zilina, Zilina, Slovakia
[c] Department of Engineering, University of Palermo, Palermo, Italy

ARTICLE INFO

Keywords:
Heart monitoring
ECG
Capacitive electrodes
Long-term monitoring
Passenger car

ABSTRACT

Cardiovascular diseases (CVD) are a leading cause of global mortality, responsible for approximately one in five deaths. In response to these concerning statistics, there is a growing need for innovative solutions enabling cardiac monitoring beyond medical facilities. This paper demonstrates the fabrication and development of a novel system designed to be integrated into passenger car seats, facilitating continuous monitoring of passengers' cardiac activity. The system comprises a capacitive electrode prototype, a custom-designed analog signal processing circuit, and a computing unit. Two ultrathin capacitive electrodes were fabricated with a diameter of 56.42 mm protected by a guard layer to reduce noise. The amplification circuitry was comprised of operational amplifiers tasked with filtering and conditioning the electrocardiogram (ECG) signal. Heart rates calculated from our measurements were comparable with clinical ECG. The experimental setup underwent testing on human subjects while they were in a state of tranquil sitting within passenger cars and also in laboratory conditions. The proposed system offers the flexibility for single and dual power supplies in full capacitive or hybrid mode. Our experimental results confirm the system's feasibility and its capability to record high-quality signals from weak biopotentials, highlighting its potential for real-world applications.

1. Introduction

For many years, doctors have utilized electrocardiography, a vital diagnostic technique, to track and examine the electrical activity of the heart. It helps in the diagnosis and treatment of a variety of cardiovascular problems by offering insightful information about a patient's heart health. To detect the electrical impulses produced by the heart during traditional ECG readings, conductive electrodes are normally applied to the skin.

Alternative ECG acquisition techniques have drawn more attention in recent years, with capacitive ECG measurement being one of the newer approaches. Capacitive ECG measurement uses the concepts of capacitance to detect and record cardiac impulses, in contrast to the traditional conductive method [1]. This non-contact method offers several distinct advantages. One of the primary benefits is enhanced patient comfort. By eliminating the need for direct skin contact, capacitive ECG measurement reduces skin irritation and sensitivity issues often associated with traditional electrodes [2]. Moreover, it enables more discreet and continuous monitoring, as it can be seamlessly integrated into various surfaces, such as car seats, mattresses, or clothing [3,4]. This makes it a promising technology for long-term, unobtrusive cardiac monitoring in both clinical and everyday life settings.

The underlying concept of capacitive ECG measurement involves the detection of changes in the electrical field generated by the heart. As the heart beats, it generates an electric field that extends through various anatomical structures. This signal forms the ECG and it still stands as the predominant biomedical signal in the realm of cardiac diagnostics. It involves the measurement of the electrical potential difference between electrodes placed at standardized locations on the body's surface. This measured signal exhibits various deviations from the baseline isoelectric level, each manifested as distinct partial waves within the ECG trace. Collectively, these waves amalgamate to constitute a singular cardiac cycle signal, which forms the complete ECG record. The temporal intervals between individual waves, their duration, amplitude, polarity, and morphological characteristics harbor critical clinical insights [5].

An electrocardiogram represents the outcome of a multifaceted interplay of physiological and technological processes. At its core,

* Correspondence to: Univerzitna 8215/1, Zilina, 010 26, Slovakia.
E-mail address: julia.kafkova@feit.uniza.sk (J. Kafková).

https://doi.org/10.1016/j.measurement.2024.115528
Received 26 January 2024; Received in revised form 18 July 2024; Accepted 13 August 2024
Available online 22 August 2024

transmembrane ion currents are generated as ions traverse cell membranes and traverse the interconnections between adjacent cells. These currents are intricately synchronized with the heart's activation and recovery sequences, thereby giving rise to the electric field within the cardiac tissue [6].

This electric field is inherently dynamic and it traverses various anatomical structures such as the bloodstream, pulmonary system, and skeletal muscles, all of which introduce perturbations to the heart's electrical field. The electric currents stemming from the heart, which propagate through these structures and ultimately penetrate the skin, are sensed by strategically positioned electrodes situated on the limbs and trunk. Subsequently, the signals detected by these electrodes undergo amplification and filtration using a variety of devices, resulting in the acquisition of an electrocardiographic recording. In modern computerized systems, these analog signals are further subjected to digitization, storage, and sophisticated processing techniques [6].

In response to the challenges associated with traditional ECG acquisition and recognizing that a significant portion of the population spends considerable time inside vehicles [7], we have devised an appropriate non-contact electrocardiogram ECG recording system. This system utilizes custom-designed capacitive electrodes designed to be integrated into the cover of automobile seats or automobile seats themselves. Additionally, we have designed an analog circuit to effectively process the capacitively acquired ECG signals from our proposed electrodes. This analog circuitry includes filtering and amplification stages to ensure that the obtained ECG signals are of high quality and suitable for diagnostic or monitoring purposes.

Capacitive ECG measurement represents a significant advancement in the field of cardiac monitoring, offering a comfortable and non-intrusive alternative to traditional ECG methods. It has the potential to revolutionize how we track and manage cardiovascular health, both within clinical environments and in everyday life. Therefore, the purpose of this paper is to present our novel system for ECG signal acquisition and to explore the differences in the signal when symmetric or asymmetric power supplies in full capacitive or hybrid mode are utilized.

Several methods exist for monitoring heart function, although not all directly measure the electrical activity of the heart like a traditional electrocardiogram (ECG). Impedance cardiography (ICG) is a diagnostic technique that uses the electrical properties of tissues in the chest area to assess heart function [8]. ECG itself reflects the heart's electrical activity generated by ionic currents and influenced by surrounding tissues [6]. Photoplethysmography (PPG) is an optical method that detects changes in blood volume within the microvascular tissue bed, providing an indirect measure of heart activity [9]. Finally, camera-based cardiac monitoring, also known as remote photoplethysmography (rPPG), offers a contactless way to track heart rate by applying the PPG principle remotely to access vital signs [10]. Echocardiography, also known as cardiac ultrasound, is a non-invasive imaging technique that utilizes high-frequency sound waves (ultrasound) to generate detailed images of the heart's structure and function in real-time [11]. Our interest in capacitive ECG stems from its potential for real-world applications. Unlike traditional ECG, capacitive technology offers a comfortable and convenient way to monitor heart activity in various settings — at work, home, even while exercising or sleeping. This is because capacitive electrodes do not require direct skin contact, eliminating the need for conductive gels (which can be irritating) and lengthy preparation routines. Additionally, the contactless nature makes them more hygienic and potentially suitable for integration into clothing for seamless monitoring. While capacitive sensors may not be ideal for detailed clinical diagnosis, they offer a promising avenue for broader health data collection, including respiratory rate, alongside ECG measurements. However, it is important to acknowledge that traditional ECG might still be preferred in situations requiring highly precise diagnostic information.

This paper is organized as follows: A comprehensive review of the existing literature in Section 2 reveals a rich landscape of research and advancements in the field of cardiac monitoring, capacitive monitoring and long-term monitoring as well as the importance of capacitive electrocardiography and theory behind it.

In Section 3, we propose our novel capacitive electrode together with signal recording circuit. Section 4 contains the presentation of the experimental results, and the authors of this paper served as subjects in our experiments. The discussion surrounding capacitive ECG electrodes in Section 5 reveals a range of advantages and disadvantages that are pivotal in evaluating their potential for transforming cardiac monitoring technology. In Section 6 we conclude this paper.

The main contributions of our work can be summarized as follows:

- We provide an in-depth and wide-ranging review of capacitive electrode design principles, specifically for applications in ECG signal measurement. Investigating the core principles of capacitive electrocardiography, this overview explores not only its sensing mechanisms but also offers a critical assessment of its advantages and drawbacks.
- The construction of our novel capacitive electrode is designed to minimize impedance, with the aim to reduce electromagnetic interference (EMI) for cleaner signal acquisition.
- Our design encompasses both a signal processing unit and capacitive electrodes. Both the circuit and the electrodes can be powered by either symmetrical or asymmetrical voltage sources. Notably, the signal processing unit is specifically designed for compatibility with a 12 V car onboard network.
- Compared to indoor measurements, our results demonstrate significant improvement in car environments. This is primarily due to the absence of 50 Hz interference, which is typically not present in a vehicle's electrical network.
- Additionally, to ensure high-quality ECG measurements, we prioritized minimizing impedance throughout the whole system. This was achieved through a combination of design strategies implemented in both the capacitive electrodes and the signal processing unit.

2. Literature overview

Continuous monitoring of cardiac function over extended periods holds significant importance within the realm of preventive medicine focused on cardiovascular diseases. The extensive data generated during prolonged monitoring serves as a valuable resource, greatly enhancing the precision of cardiovascular assessments. The availability of long-term physiological records empowers healthcare professionals to effectively detect and analyze crucial symptoms. The utilization of wearable electronic devices or embedded sensors for the surveillance of human health is experiencing a rapid expansion as they can monitor patients' heart conditions outside the hospital, aiding in both prevention and post-diagnosis care [12,13]. These devices might serve in both clinical and remote monitoring capacities, enabling improved continuous data collection of patient health status [14]. They have the capacity to collect extensive health data, thus facilitating proactive interventions to prevent patient health decline and clinical incidents [15,16].

When it comes to acquiring conventional ECG measurements, traditional wet Ag/AgCl electrodes are the tried and tested tools of choice. These electrodes, made of silver/silver chloride, have been a cornerstone in the field of electrocardiography for decades. They are favored for their reliability and effectiveness in capturing electrical signals emanating from the heart. The Ag/AgCl electrodes are designed to establish a low-impedance connection with the skin, ensuring the efficient transmission of electrical potentials.

Their success in ECG acquisition is owed to their unique properties. Silver/silver chloride is an excellent conductor and exhibits high electrical stability, making it ideal for long-term skin contact with minimum

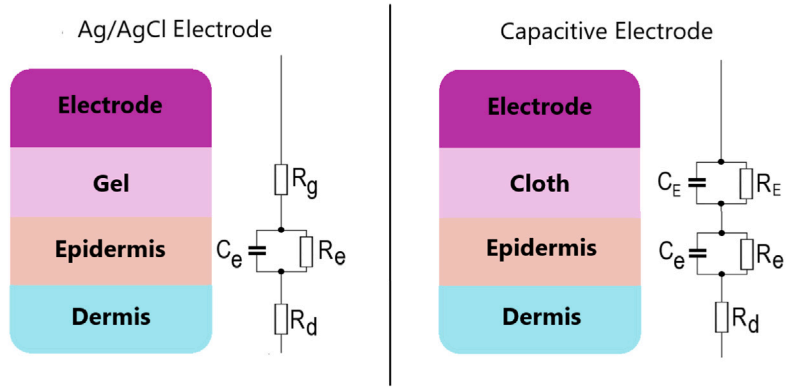

Fig. 1. Equivalent model of: (**left**) The wet contact Ag/AgCl electrode and (**right**) typical capacitive electrode.

cost [17]. The electrodes are typically coated with a conductive gel or paste that improves the conductivity and helps maintain stable electrical contact, minimizing signal noise and interference. This method is favored for extended patient monitoring, although it is not without its limitations. Prolonged contact with conductive gels can lead to skin discomfort or allergic reactions.

Fig. 1 illustrates a comparative examination of the mathematical models for Ag/AgCl electrodes and capacitive electrodes. In these models, the dermis layer is analogously represented as resistance R_d [15]. Within the Ag/AgCl model, the electrodes simulate the epidermis layer through a parallel connection of resistance R_e and capacitance C_e. In contrast, when utilizing a capacitive electrode, the clothing layer is equivalently depicted as a parallel combination of resistor R_E and capacitance C_E, where R_E signifies the resistance of the fabric surface, while C_E characterizes the capacitance of the electrode-skin interface [15].

The skin, fabric, and the sensing component of the electrode are regarded as constituting a capacitor. The capacitance at the interface between the electrode and the skin is determined by factors including the permittivity of the fabric, the thickness of the fabric, and the effective surface area between the electrode and the fabric, as described by the following relationship [15]:

$$C = \mathcal{E}_0 \mathcal{E}_r \cdot \frac{A}{d}, \tag{1}$$

where $\mathcal{E}_0$ stands for the vacuum permittivity, $\mathcal{E}_r$ denotes the relative permittivity of the fabric, A represents the surface area responsible for capacitive coupling between the electrode and the body, and d signifies the thickness of the material.

A wearable capacitive electrode usually has the capability to detect changes in capacitance induced by various types of forces, including pressure, deformation, and torsion. These changes in capacitance are a result of alterations in dimension or permittivity, stemming from physical, chemical, or biological stimuli. An electrode utilizing this sensing approach typically comprises a substrate, an electrode, and an active material situated between two electrodes. Such electrodes offer excellent sensitivity, superior thermal conductivity compared to resistive sensors, and an adequate frequency response [18].

Noninvasive ECG measurement has the potential to revolutionize a wide range of fields across healthcare, sports, technology, and more. This method holds immense potential for enhancing safety and security in critical infrastructure such as tunnels and bridges. By integrating ECG sensors within the vehicle's interior, these systems can continuously monitor the driver's cardiac activity without the need for intrusive or uncomfortable attachments. In the event of a medical emergency, sudden cardiac event, or extreme stress, the noninvasive ECG measurement system can provide real-time data to the vehicle's central monitoring system. This information can be crucial for early detection of driver health issues and can trigger automatic responses,

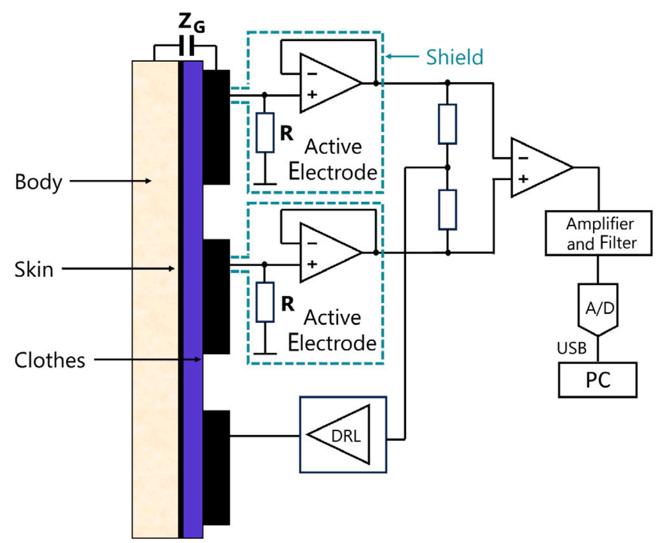

Fig. 2. Schematic representation of a capacitive electrode.

such as slowing down or stopping the vehicle, alerting emergency services, or rerouting traffic.

The approach to capturing biopotentials from the human body without the necessity of direct skin-to-electrode contact is referred to as capacitive sensing. In Fig. 2, we observe a schematic representation of a traditional capacitive electrode setup, complete with a preamplifier and shielding [19]. This setup treats the body's surface and the electrode as opposing sides of a capacitor denoted as Z_G, which facilitates the transmission of bioelectrical signals through this capacitor [20].

Capacitance is established by the proximity of the body's surface to the electrodes, even when a material like clothing is present between these layers. A capacitive electrode is characterized by the inclusion of a substantial insulator, such as clothing. This method, owing to its simplicity has found widespread application in ECG monitoring.

The bottom of the electrode serves as a conductive plate that detects variations in electrical potential on the body's surface through a capacitive link between the skin and the lower sensing surface of the electrode. Reducing the impedance between the skin and this electrode surface enhances signal quality. The preamplifier facilitates the transfer of signals from the high-impedance electrode-skin interface to the subsequent stage, comprising an amplifier and a filter. The preamplifier functions as an impedance transformer, designed with a high input impedance [20].

Given the elevated impedance between the body and the electrode, capacitive electrodes are sensitive to external noise. To mitigate this noise, the entire electrode, excluding the surface in contact with the body, is galvanically shielded. The flexibility of the electrode material allows it to adapt to the body's contours, ensuring the elimination of any air gaps and enhancing the capacitive coupling between the body and the electrode.

It is worth noting that, thus far, there have been no reported adverse effects associated with capacitive measurement methods during non-contact ECG measurements, such as discomfort or skin irritation in patients [15].

The detection of biopotentials stemming from the heart's electrical activity typically requires the use of voltage amplifiers that adhere to two fundamental criteria:

1. High Input Impedance: The amplifier's input impedance should be maximized to match the impedance of the electrode and the current flowing between the body and the amplifiers;
2. Low Current Flow: For safety considerations regarding the subject being measured, it is essential to minimize the current flow [16].

Biopotential sensing is notably susceptible to electromagnetic interference. The most prevalent form of interference arises from the coupling between the subject's body and electrical lines, resulting in the induction of a common-mode voltage (V_{cm}) at the sensing electrodes. This issue can be mitigated through differential biopotential measurement and the enhancement of common-mode rejection ratio (CMRR) [16].

Differential measurement is achieved by applying biopotentials to the input of the instrumentation amplifier. Electromagnetic interference generates a current that traverses the patient's body from a sinusoidal power line source (typically 230 V/50 Hz) via coupling capacitances relative to ground. To facilitate effective ECG measurements, the patient's body must be grounded by the ECG device. This grounding, however, contributes to an increase in the voltage V_{cm} due to its inherent resistance. Consequently, it becomes imperative to either decrease the impedance of the grounding electrode or compensate for the current generated by electromagnetic interference. This compensation is often achieved using a driven right leg (DRL) electrode. Typically placed on the right leg, this electrode senses the standard voltage on the body and provides feedback to the subject's body. Negative feedback serves to decrease V_{cm} values to the microvolt (μV) level [16].

Capacitive recording of ECG signals offers numerous advantages but also presents some drawbacks. Advantages include [18,21–23]:

- Reusable Electrodes: These electrodes can be reused and minimize leakage currents, thereby preventing short circuits between the electrodes and biological tissues;
- Encapsulation Potential: Capacitive connections involve dielectric layers between electrodes and the skin, enabling the encapsulation of electrodes in relation to the surrounding environment;
- Dielectric Insulation: Capacitive electrodes not only serve as non-contact electric field detectors but also feature at least one insulating dielectric layer between the conductive electrode and the skin;
- Signal Customization: These electrodes offer flexibility in choosing dielectric materials to tailor signal properties;
- Wireless Integration: Modern wireless modules, integrated, for instance, in car seats, offer a wireless alternative to conventional cable connections;
- Multi-Functionality: Non-contact methods allow for monitoring several physiological functions simultaneously;
- No Gel Requirement: They eliminate the need for electroconductive gel to improve conductivity at the electrode-skin interface;
- Dry Electrodes: These prevent issues like gel drying and skin irritation during heart function monitoring in the presence of gel.

Disadvantages encompass the following [20–23]:

- Reduced Coupling Capacity: The absence of direct skin contact can lead to an air gap, which significantly reduces coupling capacity, decreasing the sensed signal's amplitude.
- Humidity-Induced Impedance Changes: Variations in clothing humidity can cause unwanted impedance changes.
- Variable Distance: The distance between the body and the electrode surface may fluctuate depending on the pressure applied to the dielectric layer.
- Signal Sensitivity to Clothing and Pressure: Non-contact sensing of biopotentials is susceptible to signal quality degradation due to clothing thickness and pressure, potentially impacting data accuracy.
- Motion Artifacts: Movements can represent artifacts, affecting signal quality, depending on the connection between detectors and the subject.
- Comfort and Fit: Rigid capacitive electrodes may lack proper fit and comfort, especially during subject motion.
- Inadequate Body Contact: Rigid electrode shapes may not effectively conform to the curved surface of the human body.
- Thin and Flexible Electrode Development: The need to develop thin and flexible capacitive electrodes remains a challenge.

The realm of capacitive ECG sensing has been explored by humanity for over half a century [24–26]. Non-invasive bioelectrodes have demonstrated their capacity for prolonged operation without requiring physical contact. Numerous studies by scientists and experts have delved into the realm of long-term ECG monitoring [27–29]. Nevertheless, the persistent challenge remains the interferences arising from muscle movements, skin characteristics, or the inherent design of bioelectrodes [30].

The main component of capacitive ECG electrodes is the dielectric material. However, the existing literature often lacks comprehensive information regarding the suitability of various dielectrics. The current trajectory in this field is directed towards the development of reliable ECG monitoring systems that require minimal or no intervention from medical professionals. Non-invasive electrodes can be categorized based on their degree of skin contact [30]. This categorization is illustrated in Fig. 3.

Gel-based electrodes are the established standard in clinical medicine and ambulatory applications. In contrast, non-gel or dry electrodes are capable of capturing ECG signals without the need for conductive gel or paste. They offer comfort and eliminate the risk of allergies. Capacitive electrodes, due to their varying dielectric properties, represent an intriguing avenue of research [30].

The design of capacitive electrodes typically involves the integration of a flat electrode or conductive fabric with different types of insulators such as polydimethylsiloxane (PDMS), polyimide film, fabric, and silicon dioxide. PDMS, in particular, is a common choice as an insulating layer in electroactive polymer elastic dielectrics [31]. This material offers advantages like high heat resistance, air permeability, elasticity, and extensibility. These electrodes are affixed to the skin using tapes, straps, or can be incorporated into textiles such as caps, t-shirts, or car seats [18].

Flexible capacitive electrodes are engineered for voltage measurement on the skin. For instance, Ullah et al. developed capacitive electrodes adhering to the principles of an epidermal electronic system, considering factors like effective mechanical modulus, thickness, and areal density [18]. Various studies employing capacitive electrodes for long-term monitoring have achieved favorable Signal to Noise Ratio (SNR) values [18].

Capacitive sensing has recently gained prominence and popularity in the field of mechanical stimulus detection, particularly due to its fit, low energy consumption, sensitivity, and adaptive sensing configuration. Capacitive pressure sensors have found extensive applications in both consumer and industrial domains. These applications have been

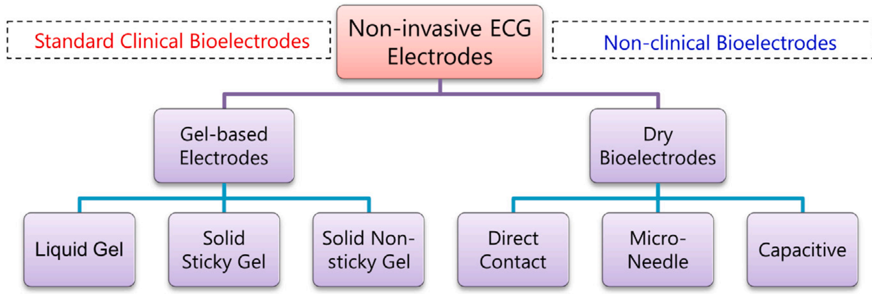

Fig. 3. Categorization of ECG electrodes.

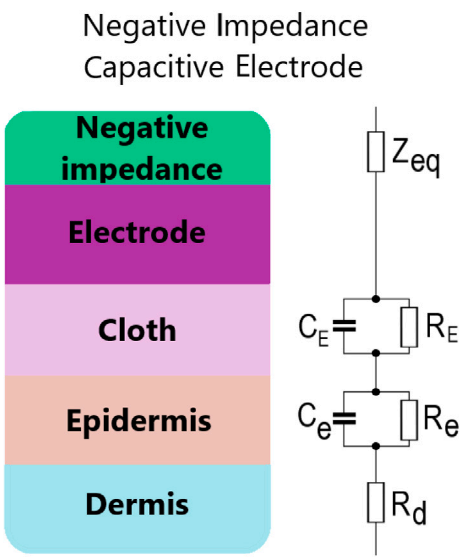

Fig. 4. Equivalent circuit of a capacitive electrode with negative impedance.

extended to diverse human interfaces, encompassing electronic skin (e-skin) replicating human tactile sensation, body pressure mapping, and joint bending detection [18].

Enhancing the flexibility of electrode materials holds paramount importance, as they constitute the core component of wearable electrodes. Typically, conductive nanomaterials and polymers are employed in the fabrication of capacitive electrodes. Bao et al. developed a series of capacitive sensors for electrochemical sensing, demonstrating the relative ease of designing and constructing such models, thus contributing to the widespread adoption of this technique [32,33].

One of the current research directions focuses on the development of capacitive electrodes with negative impedance (depicted in Fig. 4). Wang et al. introduced principle of these capacitive electrodes, aimed at enhancing the quality of non-contact ECG signals. It involves balancing the impedance mismatch at the interface between the skin and the fabric. The study also conducts a comparison of SNR values between the newly designed electrode and conventional capacitive electrodes. Fig. 4 illustrates the schematic connection of a capacitive electrode with negative impedance [15].

The relationship between the original signal from the heart (U_{heart}) and the measured signal (U_{out}) after preamplification can be expressed by Kirchhoff's voltage law (Eq. (2)) [15]:

$$U_{out} = U_{heart} \cdot \frac{1}{\frac{Z_{skin}+Z_{cloth}+Z_{neg}}{Z_{in}} + 1}, \tag{2}$$

where Z_{skin}, Z_{cloth}, Z_{in} are the input impedances of the skin, cloth, preamplifier and Z_{neg} is the negative impedance. Considering relation Eq. (2), the output signal U_{out} can be similar to a cardiac signal with a high impedance Z_{in} and a low impedance Z_{cloth}. The high input impedance of the preamplifier can reduce the attenuation of the ECG signal. The material of the fabric can also affect the coupling capacitance effect during non-contact ECG monitoring, it is the resistance of the fabric R_E and the capacitance C_E at the skin-electrode interface [15].

Lower fabric impedance can improve signal quality in non-contact ECG measurement. Other factors affecting signal quality can be applied pressure and atmospheric humidity. Properly applied pressure can reduce the air gap affected by the curvature of the body to improve the efficiency of capacitive coupling [15].

This non-contact ECG monitoring system was implemented to verify the negative impedance capacitance electrode in order to obtain a better non-contact ECG signal. The system consists of a capacitive electrode with a negative impedance analog signal preprocessing circuit. The hardware is securely applied to the subject's chest using an elastic strap [15].

The capacitive electrode with negative impedance comprises several components: the sensing surface of the electrode, a circuit featuring negative impedance, a preamplifier, and shielding. In this research, a negative impedance, denoted as Z_{neg}, was incorporated behind the sensing electrodes to equalize the impedance between the skin and fabric surfaces during ECG measurements [15].

To address the issue of deteriorated ECG signal quality stemming from impedance mismatches between the skin and electrode interfaces, a negative impedance circuit was introduced [15]. Its purpose is to mitigate the attenuation of the signal captured by the electrode's sensing surface. Fig. 5 illustrates the negative impedance circuit theorem, which employs an OP07 operational amplifier and three resistors. Notably, resistors R_1 and R_2 are connected to the inverting input of the operational amplifier, non-inverting input is directed towards resistor R_3.

The analog signal processing circuit plays a crucial role in signal amplification and filtration. This setup comprises an AC coupling circuit, an instrumentation amplifier, an active bandpass filter, and a DRL circuit. In a study conducted by Ng et al. [34], the focus was on mitigating mains interference with textile-insulated electrodes using five digital filters. They explored six types of textile materials: cotton, linen, rayon, nylon, polyester, and PVC. The authors developed a system that included TEX-C (Textile-Insulated Capacitive) biomedical sensors, front-end amplifiers, a band-pass filter, a data acquisition unit, a computer, and incorporated negative feedback.

In a related study from 2017, the authors investigated the optimal textile insulator [35]. The findings indicated that natural fabrics like cotton or linen exhibited superior properties for integration with a capacitive biosensor, delivering a minimal noise level of 2 mV and demonstrating signal measurement repeatability and consistency throughout the testing period.

Ueno et al. explored the capacitive method of ECG sensing from the dorsal surface of the body [36]. They insulated the electrode with varying thicknesses of cotton and employed two 50 Hz analog filters to suppress Power Line Interference (PLI). The experimental results were influenced by different cotton thicknesses and pressure.

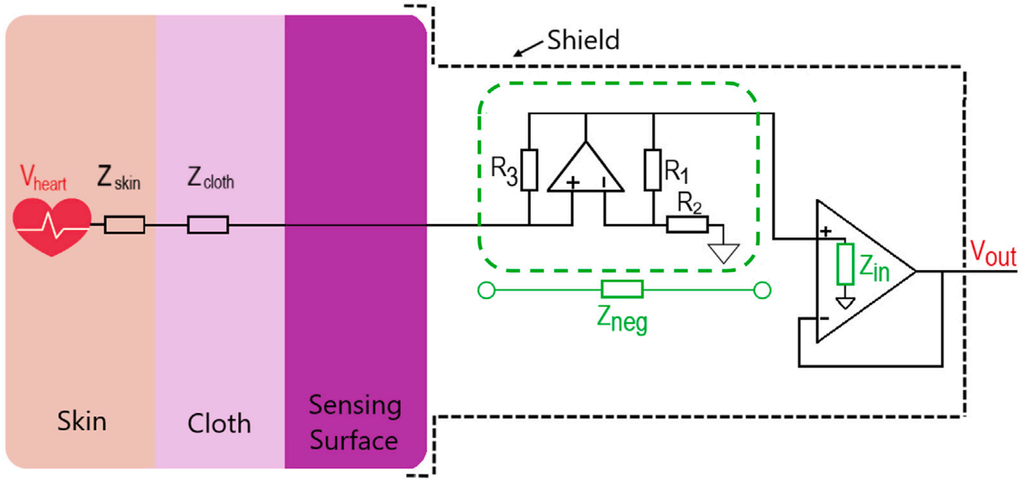

Fig. 5. Diagram illustrating a capacitive electrode with negative impedance.

Eight years later, Li et al. conducted experiments using cotton T-shirts of different thicknesses, including 1.25 mm, 0.51 mm, and 0.78 mm [37]. Their findings revealed that thicker T-shirts led to a decrease in the quality of the recorded ECG signal, primarily due to increased noise interference.

In 2019, Pfeiffer et al. published a study in which they investigated various types of capacitive electrodes, including copper plate electrodes, conductive foam, and electroconductive substances [38]. They demonstrated that each electrode exhibited different levels of impedance imbalance during muscle movement, and the level of PLI also varied accordingly.

Takano et al. conducted measurements of ECG and EMG signals on the neck region [39]. They opted for 100% polyester with a thickness of 0.45 μm as the insulating material. Their findings revealed that EMG signals could be detected during coughing by the test subjects. However, the interference was substantial, resulting in a low SNR. Notably, most research groups have predominantly utilized cotton as an insulating material for capacitive biomedical electrodes, leaving a wide array of other textile materials relatively unexplored or underexamined.

Prior studies have primarily focused on assessing the effectiveness of ECG devices based on the capacitance principle. Still, the comprehensive diagnostic utility of these novel devices, particularly their potential to support preliminary arrhythmia diagnosis, has not been thoroughly investigated.

Wu et al. designed an ECG cushion and assessed its functionality on four healthy individuals and seven patients with myocardial arrhythmia [40]. The electrode unit comprised three square CC (Capacitively Coupled) electrodes labeled as E_1 to E_3, along with one square reference electrode. The CC electrode consisted of three layers: the sensing layer, the deformation layer, and the insulating layer. The sensing layer monitored potential changes on the clothing's surface, while the foam-based deformation layer conformed to the body's contours, ensuring reliable contact and high signal quality. A larger reference electrode was positioned beneath the three electrodes, separated by an insulating layer. A reference electrode connected to the DRL circuit served to enhance Common Mode Rejection (CMR) and reduce electromagnetic interference. Both the reference and sensing layers were constructed using conductive silver fiber fabric. The schematic depicting the entire electrode system's connection design can be seen in Fig. 6.

The presented cushion has the capability to provide information on par with a conventional medical holter monitor. This cushion excels at distinguishing between normal, ventricular, and premature supraventricular contractions during prolonged monitoring. The authors demonstrated the system's robustness in capturing the intricate details of Q, R, and S waves. Its elastic and soft nature makes it well-suited for extended ECG monitoring, without causing allergic reactions [40].

In the application of cardiac sleep monitoring using capacitive ECG, obtaining high-quality raw signals can be challenging due to the low relative dielectric constant of bed sheets and pajamas. Feng et al. tackled this issue by designing a system incorporating a hydrogel-textile electrode to monitor cardiac activity during sleep [31]. Additionally, the system integrates eight membrane pressure sensors to detect body positions during sleep and capture ECG signals from various sleeping positions. The sensing system, shown in Fig. 7, consists of three flexible electrodes, which are composed of a multilayer composite electrode (MLCE). Similarly, noncontact measurements of electrocardiogram utilizing a capacitively coupled electrode array based on flexible circuit board has been recently investigated by Wang et al. [41].

This MLCE primarily comprises a hydrogel layer, a sensitive and insulating layer, a reference electrode, and eight position-detection array pressure sensors labeled A_0-A_7 [31]. Twelve hydrogels are distributed within the sensitive textile layer, organized into two groups. It is assumed that the presence of cotton in pajamas and bedding will enhance water absorption by the hydrogel, thereby increasing its binding capacity and significantly improving signal quality. The sensitive layer is crafted from flexible conductive fabric to enhance conductive contact and comfort during sleep. The insulating layer is constructed using a PDMS membrane, while the reference electrode, made from the same material, is connected to the DRL circuit. Given the assumption that the SNR of the measured heart signal can be improved due to the hydrogel layer, the typical DRL electrode is dispensable and is replaced by the aforementioned reference electrode, positioned directly beneath the insulating layer [31].

Pani et al. developed new electrodes for bio-potential measurement using textile fabric with a highly conductive solution [42]. They utilized one of the most commonly used conducting polymers named poly-3,4-ethylenedioxythiophene doped with poly(styrene sulfonate) (PEDOT:PSS). This polymer stands out for its biocompatibility, low band gap, and superior electrochemical and thermal stability. Electrodes created from this fabrication show a decreased electrochemical impedance mismatch between the tissue and electrode, without the need for using any conductive gel.

They primarily focused on developing an innovative textile electrode rather than its implementation in smart cloth. Dried fabrics were cut into pieces measuring 35 mm × 65 mm. The conductive fabric was then sewn onto a nonconductive layered structure of foam (4 mm) and polyester, sized at 35 mm × 35 mm, using simple cotton yarn. The resulting active area of the electrodes, after sewing, measures 30 mm × 30 mm — an appropriate dimension chosen for ECG monitoring. The authors selected sorbitol, ethylene glycol, and glycerol for testing as they have the highest reported conductivities in preparing conductive

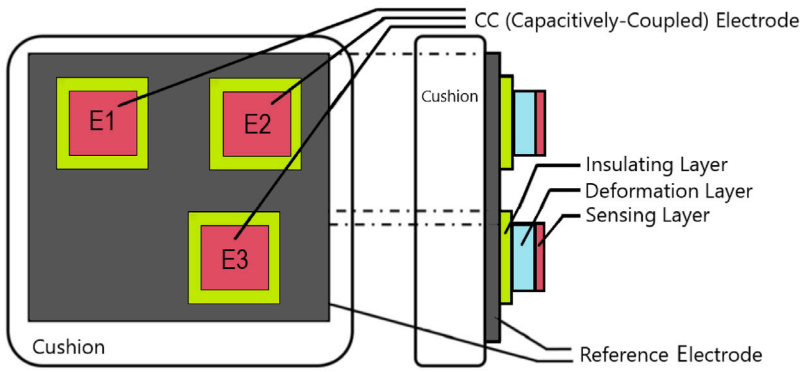

Fig. 6. Configuration of the CC electrode unit.

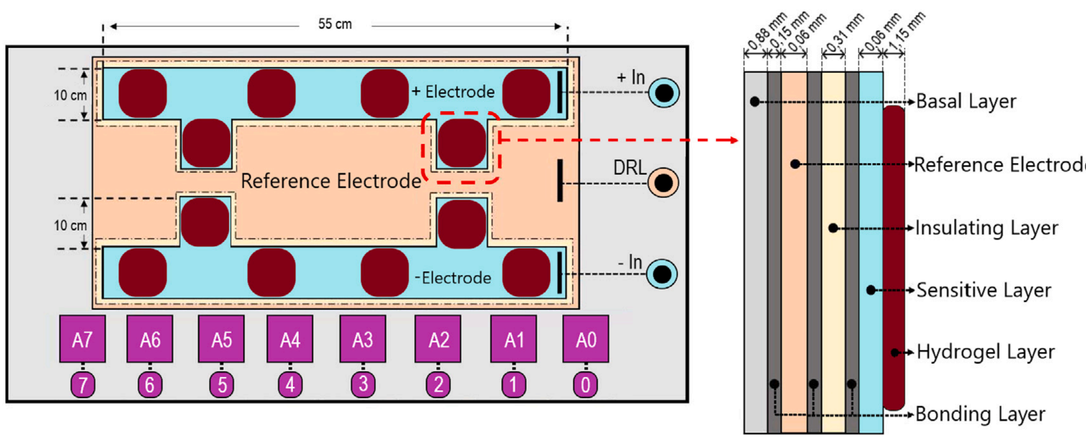

Fig. 7. Diagram of the multilayer composite (MLC) electrode.

fabrics. A snap fastener was positioned on the excess portion of the conductive fabric, fixed in place with silver conductive paste, and further secured by sewing it with a silver-coated yarn. The textile electrodes created exhibit performances that are nearly comparable to commercial electrodes. They underwent testing in two conditions. In the first condition, involving measurements at rest, the results were comparable to the usage of commercial electrodes. However, in the second condition, which involved motion during measurements, the results were compared to the usage of commercial electrodes.

Xu et al. focused on creating a dry electrode made of printed silver nanowires (AgNWs) for the acquisition of ECG signals [43]. This flexible electrode, with excellent electrical conductivity, is suitable for healthcare monitoring. The authors demonstrated a wearable, non-invasive, and transparent electrode using AgNW and graphene oxide (GO) as components. This hybrid electrode was created in individual layers on polyethylene terephthalate (PET) substrates through a screen printing process. Oxidation reduction occurred, thanks to the GO layer, which is considered a protective layer. This layer also prevents contact at the interface of the skin and the metal electrode. The acquired ECG signals by the AgNWs/GO hybrid electrode show comparable performance. Also, the key characteristics of transparent electrodes are presented, such as stability and mechanical flexibility. They are able to acquire a suitable ECG signal using AgNWs/GO electrodes with obvious features of ECG. In comparison with AgNWs and Ag/AgCl electrodes, the AgNWs/GO electrodes are able to acquire high-quality ECG signals. These electrodes exhibit very good resistivity against human motions, which is crucial for long-term real-time monitoring during daily life. The authors also demonstrate that the method for full-screen printing electrodes is not difficult. It is effective, low-cost, and enables the creation of various patterns. Thanks to the combination of AgNWs and GO sheets, they developed flexible AgNWs/GO electrodes. These electrodes stand out for their high conductivity as well as transparency. Additionally, another area is flexible, printed and skin-mounted sensors, which are becoming more popular [44].

A recent paper by Zhang et al. proposed a front-end amplifier structure with a fast reset scheme [45]. It facilitates the recovery of motion artifacts before reaching saturation, thereby preventing an extended recovery process that might last for several seconds and result in signal loss in conventional capacitive sensing systems. A reconstruction algorithm is utilized to restore the waveform, consequently expanding the dynamic range beyond the initial saturation level. They utilized a 2 cm^2 PCB with a thin oxide layer for insulation.

Despite many advantages, capacitive ECG systems are susceptible to motion artifacts [46], a challenge arising from the relative movement of electrodes in relation to the skin. A state-of-the-art overview has been recently published, encompassing diverse approaches based on materials and construction, analog circuits, and digital signal processing [47]. Ding et al. investigated motion artifacts in noncontact ECG for wearable monitoring. Their approach uses a novel ECG electrode structure for simultaneous ECG and reference data collection. This design, coupled with an adaptive filter, efficiently suppresses motion artifacts. Experiments under normal walking conditions show a significant improvement in the ECG signal quality [48].

The Table 1 presents an overview of capacitive electrode studies found in the existing literature. Each entry in the table encapsulates the essence of the respective study, offering a glimpse into the exciting and innovative contributions made in this ever-evolving field. More can be found for example in [20,49].

3. Materials and methods

Upon reviewing the currently available literature and examining the various methods and solutions applied, it becomes apparent that

Table 1
This table presents capacitive electrode studies found in the literature.

Group	Size and shape	Mitigation of EMI	Face of the electrode
Gao et al. [21]	Square 20 × 20 mm	Guard ring and conductive foil tape	Nanolaminate dielectrics
Lee et al. [50]	Flexible electrode 30 mm × 150 mm	Shielding plate with guarding feedback	Electrically insulating paint 0.01 mm
Harland et al. [51,52]	Disc of diameter 1 to 20 cm	Using a differential signal from two probes	N/A
Lim et al. [53]	Square plate 40 × 40 mm	Using aluminum shield	PCB clad with copper and plated with gold
Arcelus et al. [54]	PCB 70 × 40 mm	Soldering PCB directly to sensing surface	Copper tape (3M) rectangles of 85 × 55 mm
Maruyama et al. [55]	Copper plate 50 × 20 mm	Circuits driven by batteries	Paper
Pani et al. [42]	Square 30 × 30 mm	Data acquisition in controlled environments	PEDOT:PSS
Xu et al. [43]	Rectangle 20 × 8 mm	N/A	AgNWs and GO are components of each layer
Zhang et al. [45]	Rectangle 2 cm^2	Fast-reset function	Thin oxide layer
Fujita et al. [56]	Copper 70 mm × 120 mm	A differential amplifier circuit and a filter	Rubber sheet placed between the electrode and the body
Xiao et al. [57]	Rectangle 45 cm × 45 mm	Bandpass filtering	11%–14% silver
Our proposed electrodes[a]	Disc of diameter of d = 56.42 mm	Guard layer and additional shielding case	Silver fiber with yarn

[a] Detailed description of our proposed capacitive electrode design is in Section 3.1.

a reliable and consistent solution unaffected by external factors such as electromagnetic interference (EMI), humidity, electrode pressure, insulating material quality and thickness, among others, is currently unavailable. In order to improve the SNR, we have introduced several measures:

- As mentioned in Section 3.2, the power supply of signal processing unit includes filtering components like inductors and capacitors to enhance its resilience against fluctuations and interference on the onboard network.
- In the signal processing unit, capacitive electrodes were connected through an AC coupling circuit (as described in [58]). This circuit removes DC voltage, essential for biopotential measurements, by acting as a high-pass filter. It enhances a common-mode rejection ratio (CMRR) by eliminating DC components without a grounding resistor.
- We minimized impedance between GND-marked areas by using multiple vias connections between the top and bottom layers. These transitions are parallel coupled impedances with a value close to zero.
- The shortcomings of the state of the art in designing a capacitive electrode with just enhanced resistance to external factors are eliminated by the capacitive electrode with impedance minimization presented herein. Its greatest contribution is primarily its ability to mitigate the influence of the external factor, which is electromagnetic interference.

All the measures described above should contribute to the improvement of signal processing in terms of shielding and immunity to interfering signals.

Given the considerations of technology accessibility and complexity required for electrode development and its impact on the ultimate solution, we have chosen to outline a system in the subsequent sections.

3.1. Capacitive electrode design

A capacitive electrode for measuring bioelectric potential consists of a conductive material that forms a capacitor together with the body surface. The electrical charges on the body surface due to the body's electrical potentials interact with the charges of the conductive electrode, which is separated from the body surface by an insulating material (dielectric) [59]. The design of capacitive electrode is shown in Figs. 8, 9.

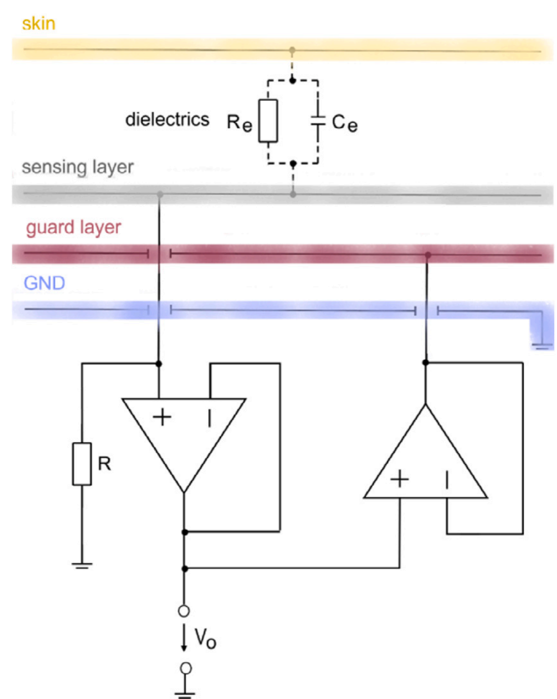

Fig. 8. Design of the proposed capacitive electrode.

An AD8642ARZ operational amplifier was used to process the bioelectric signals appropriately. The integrated circuit contains two embedded operation amplifiers [60]. The AD8642ARZ is a precision and low-power JFET input amplifier with extremely low input bias current and rail-to-rail output. It is suitable for medical applications that require low power due to easier heat dissipation. R_e represents resistance of the fabric, C_e represents bonding capacitance between the skin and the surface of the active electrode, R represents bias resistance and V_o represents output voltage.

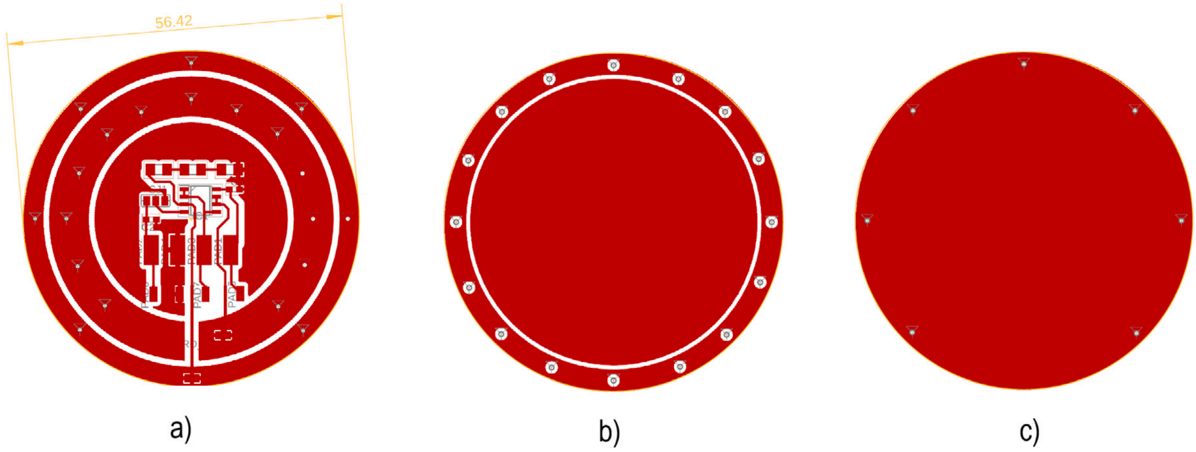

Fig. 9. PCB design of the capacitive electrode. **(a)** First layer with electronic components, **(b)** second – guard layer, **(c)** sensing layer.

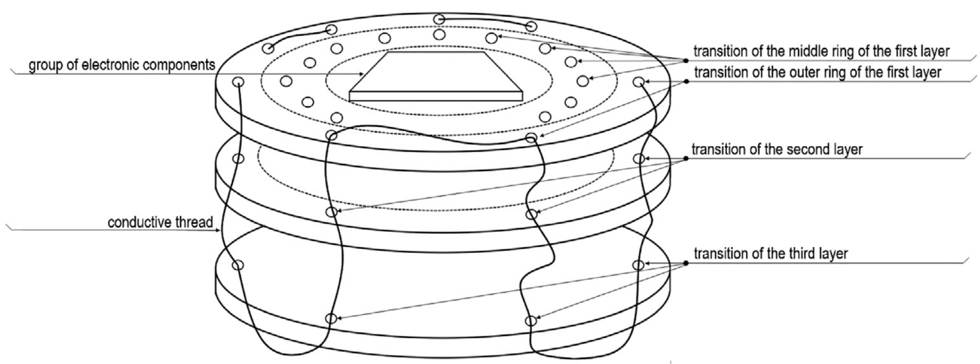

Fig. 10. Different layers of a capacitive electrode.

A critical challenge in our capacitive electrode construction is maintaining a guard layer while minimizing its impact on signal transmission. Our novel electrode is designed to overcome this by reducing the impedance between the guar layer and the circuitry. The capacitive electrode is also designed to minimize the impedance coupling of the guard layer to the electrode circuitry. The impedance reduction is achieved by using multiple transitions between the inner surface of the second layer and the middle ring of the first layer, and by using transitions between the surface of the third layer and the outer ring of the first layer. The reduction of impedance between the layers is achieved by multiple interconnections of the respective surfaces of each layer of the PCB through transitions.

The first two layers of the printed circuit board are interconnected by soldering the transitions of the middle ring of the first layer to the inner surface of the second layer, which serves as a guard layer. Transitions are provided in the outer ring of the first layer for multiple interconnections of the first layer with the surface of the third layer, reducing the impedance of the interconnection to a value close to zero. The first layer and the third layer are galvanically coupled not by soldering, but by an electrically conductive thread through the transitions of the outer ring of the first layer and the transitions of the third layer. The electrically conductive thread also passes through the outer ring of the second layer, but a galvanic connection of the electrically conductive thread to this ring is not necessary. A shielding cap is placed over the first layer containing the electrical components, including resistors, capacitors, and operational amplifiers, to reduce the influence of electromagnetic interference on the operational amplifier circuitry. Fig. 10 schematically shows the different layers of a capacitive electrode.

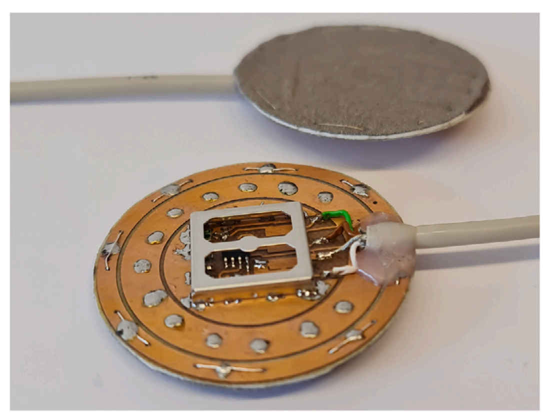

Fig. 11. Actual configuration of capacitive electrodes.

The technical design of the capacitive electrode meets the requirements for functionality, practicality, and user comfort, making it suitable for medical applications. The benefits of the technical design of the electrode include increased patient comfort due to:

- The size of the electrode, which does not bother the user.
- The flexibility of the electrode, allowing it to adapt to the curvature of the body and eliminating the air gap.

Advantages of the technical solution:

- The technical design of the capacitive electrode ensures improved contact with the skin and guarantees reliable signal acquisition.

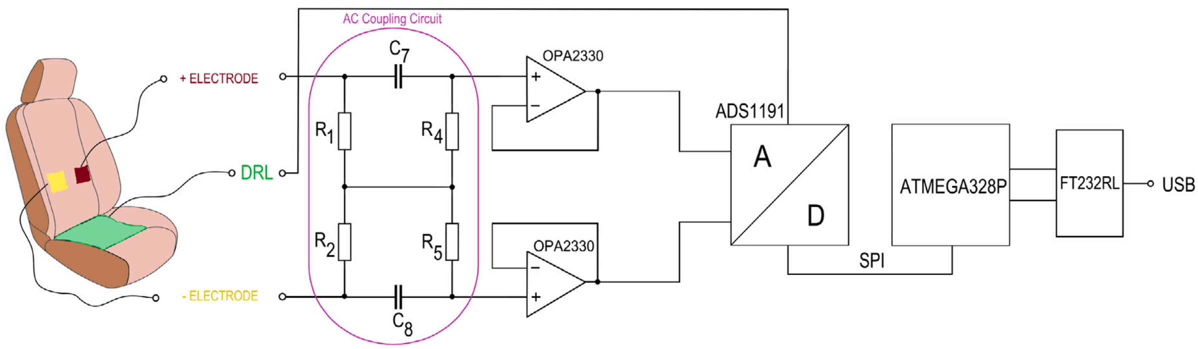

Fig. 12. Capacitive ECG acquisition system.

- The technical design makes it easy to integrate into a variety of monitoring platforms, including wearables, smart clothing, and even car seats.
- In its final operation, the capacitive electrode is suitable for continuous and non-invasive monitoring of the heart, which has the potential to improve early detection of cardiovascular disease and enhance the overall quality of patient care.

The image of the proposed capacitive electrode design is depicted in Fig. 11.

3.2. ECG signal processing unit

The ECG Signal processing unit is a vital component of our system. Our proposed electrodes are responsible for capturing the ECG signal generated by the heart and conducting them to the circuit for further analysis and processing. Simplified diagram showing our whole proposed system is shown in Fig. 12.

The outputs from the capacitive electrodes are linked to a coupling circuit featuring a high-pass filter function [58].

The main role of AC coupling circuits is to ensure the elimination of DC voltage when measuring biopotentials. The circuit operates as a high-pass filter without a grounding resistor to eliminate DC components and obtain a higher CMRR. This circuit has been designed with a cutoff frequency of 0.3386 Hz. The cutoff frequency of the coupling circuit:

$$f_m = \frac{1}{2\pi R_4 C_7} = \frac{1}{2\pi R_5 C_8}, \tag{3}$$

is determined by the resistors R_4 and R_5, each with a value of 4.7 MΩ, and the capacitors C_7 and C_8, each with a value of 100 nF (Fig. 12). R_1 and R_2 also have a value of 4.7 MΩ and provide a continuous polarization path (I_{bias}) to the subsequent stages.

Subsequently, the outputs from the voltage followers are directed to the inputs of the differential amplifier found in the front-end circuit, ADS1191. ADS191, in addition to the features described later in this paragraph, effectively reduces the common-mode voltage with a CMRR of −95 dB. The DRL circuit also increases the CMRR. As discussed in [61], the DRL circuit can reduce the common-mode voltage by 10 to 50 dB. The ADS1191 is a two-channel, 16-bit, low-power analog front-end circuit designed for biopotential measurements [62]. The ADS1191 accomplishes the conversion of an analog signal to a digital one through a 16-bit A/D converter. The ATmega328P reads the data processed by the ADS1191 circuit and transmits it to the PC through a serial channel via a USB converter. To enable communication between the microcontroller ATMega328P's UART interface and the superior system's UART/USB interface, the FT2322RL converter is essential. The connection involves linking the FT2322RL to the ATMega328P through its UART interface, and simultaneously connecting it to the superior system via the USB interface using a mini B type connector, which also serves as its power source.

The analog part of the system can be powered by a single supply or a dual supply. To achieve the conversion from a positive to a negative supply voltage, a CMOS voltage converter is employed.

In designing this printed circuit board, we adhered to commonly accepted PCB creation principles. The PCB is composed of two layers, with all surface-mount (SMD) and through-hole (THT) components situated on the top layer. This includes jumpers, screw terminals, a mini B USB connector, and an external power supply connector responsible for providing the 12 V supply voltage to the circuit.

We strategically arranged the components to meet our specific requirements. Placing the screw terminals on the left side of the board ensured easy access during assembly. These terminals are used for connecting the output from individual electrodes and the DRL connection, serving to shield the supply cables and the supply voltage, which can be a single supply (asymmetrical voltage power source) of +5.04 V or a dual supply (symmetrical voltage power source) of ±2.66 V based on the chosen jumper configuration. This design also means that the ground of the supply voltage is available at the terminals, ensuring appropriate connections can be made according to the selected supply voltage type.

The processing unit is supplied with a voltage of 12 V from the vehicle's onboard electrical network. Considering that the voltage in the onboard network is less than ideal, and it may contain interference and voltage fluctuations, the protective components of the power supply have been adapted to accommodate these real-world conditions. These protective components include rectifier diodes, voltage stabilizers, capacitors, and LEDs.

Fig. 13 displays the layout of the printed circuit board (PCB) for analog signal processing, showing both the top (on the left) and bottom (on the right) layers.

In response to the device's miniaturization, we selected the smallest available cases for active components and utilized 0603 cases for passive components. In some instances, capacitors with a CG0 designation were chosen due to their superior dielectric material, known for its reduced microphonic interference compared to other materials. The default thickness of power and ground paths is 1.524 mm, although it can be adjusted as needed. Data signals are routed through 0.3048 mm thick paths. To maintain a clear separation between analog and digital components, including GND and AGND potentials, we ensured they remained topologically distinct. However, these potentials are joined at a single point in close proximity to the external power supply voltage's negative input, to which they are connected.

To minimize impedance in the GND and AGND potentials, we incorporated two copper polygons on both layers, facilitating a copper pour to enhance grounding. Each layer features distinct, non-overlapping polygons representing the AGND and GND potentials. Surfaces with matching potentials on both layers are interconnected via multiple over-plated holes, commonly referred to as vias. Additionally, three strategically positioned holes along the board's edges serve the essential function of securing the PCB within the enclosure.

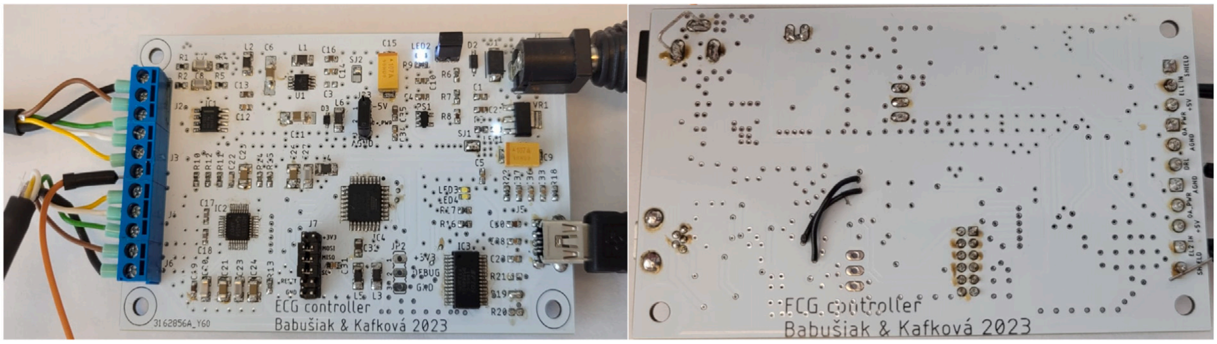

Fig. 13. Actual PCB configuration: (**left**) top and (**right**) bottom layer.

While these design choices effectively enhance grounding and minimize impedance, we have not studied coupling interference in detail. The coupling interference has not been studied in detail in our work because the capacitive system is dedicated to measurements in the car, where the coupling interference from the electricity is minimal. This fact was confirmed by demonstrative measurements in the vehicle. This problem arises only in laboratory (indoors) measurements. There may be some other interferences from the car surroundings that are effectively eliminated by our system or are out of the frequency range of the measured signal. We recognize the importance of the interference, and we plan to address the problem in future research to provide a more comprehensive analysis.

4. Experimental results

The experimental results for our capacitive ECG monitoring system are highly promising and demonstrate its effectiveness in capturing accurate and reliable cardiac data during ideal conditions. In a series of rigorous tests and trials, we assessed the system's performance in various scenarios. The results revealed that the capacitive electrodes, when integrated onto different surfaces such as car seats or clothing, provided great signal quality. The system successfully recorded ECG signals with low noise and artifacts, showcasing its potential for real-world applications.

In the first round, we compared the capacitive ECG output signal under both single and dual power supply conditions. We visualized the recorded ECG signal using the UniversalGraph by BB application throughout the measurement process [63]. This application offers flexibility in adjusting the sampling rate and signal gain. According to the Nyquist theorem, the sampling frequency should be at least double the maximum signal frequency, which, in our case, is 120 Hz [64]. In the application, we could choose between two sampling frequencies: 125 Hz or 500 Hz. We selected a 125 Hz sampling frequency in the UniversalGraph application [63] to align with the Nyquist theorem for our expected signal spectrum. Additionally, we configured the gain to a value of 6 within the application. The application stores the measured ECG signal in a binary format suitable for subsequent processing, such as in the MATLAB environment (data visualization, axis configuration, frequency spectrum calculation etc.).

Through experiments, we have demonstrated that employing a dual supply significantly enhances signal quality, rendering it clearer and more interpretable. As a result, we have made the decision to exclusively employ a dual supply for all subsequent measurements. We conducted measurements by monitoring the ECG of authors while they were seated in the driver's seat of the car. Instead of connecting the DRL to the Ag/AgCl electrode on the foot (hybrid mode), we employed an electrically conductive fabric (Fig. 14) on which the subjects sat (fully capacitive mode). In total, we conducted over 100 measurements across various conditions, including different months or car types. The measurements were conducted during system assembly (over 70 measurements) and testing with human subjects (over 30 measurements). Notably, we observed no significant differences in the results despite variations in participants' clothing types, thicknesses, and the number of layers worn. We also conducted measurements on the certain subjects over a period of 3 months under various conditions (Fig. 23).

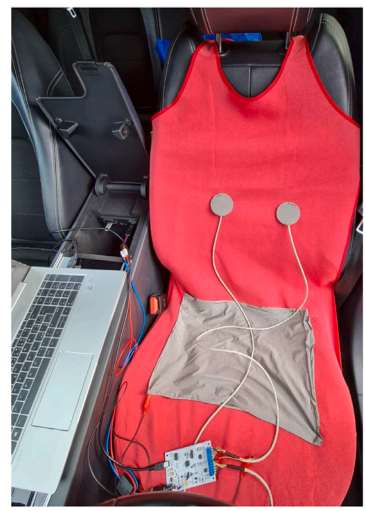

Fig. 14. Capacitive electrodes integrated into the car seat cover.

The car seat cover featured in Fig. 14 is non-electric and possesses hydroscopic properties. It incorporates PUR foam for a seamless fit to both the seat and the subject's body, facilitating accurate measurements. This innovative car shirt has the potential to serve as a prototype for future commercial applications, such as a car cover designed for the long-term monitoring of ECG signals for both drivers and passengers, utilizing the capacitance method. Moreover, electrodes can be seamlessly integrated within the car seats, forming an imperceptible measuring system that remains hidden from the user's view. Regrettably, owing to resource constraints, we chose to utilize a seat cover rather than directly integrating electrodes into the seat. However, in our upcoming experiments, we plan to incorporate electrodes into the car seat and report our findings in our future paper.

The experiments were conducted under two different settings: room conditions and inside a vehicle. Figs. 15 and 18 show the recorded waveforms of ECG signals, allowing for a comparison of how the power source of our capacitive ECG system impacts the resulting ECG signal. We processed the useful ECG signals using the MATLAB environment, without employing any additional software filtering.

In order to compare the different power supply methods, we chose to examine the signal in the frequency domain. Consequently, we have generated frequency spectra for both signals, as illustrated in Figs. 16 and 17.

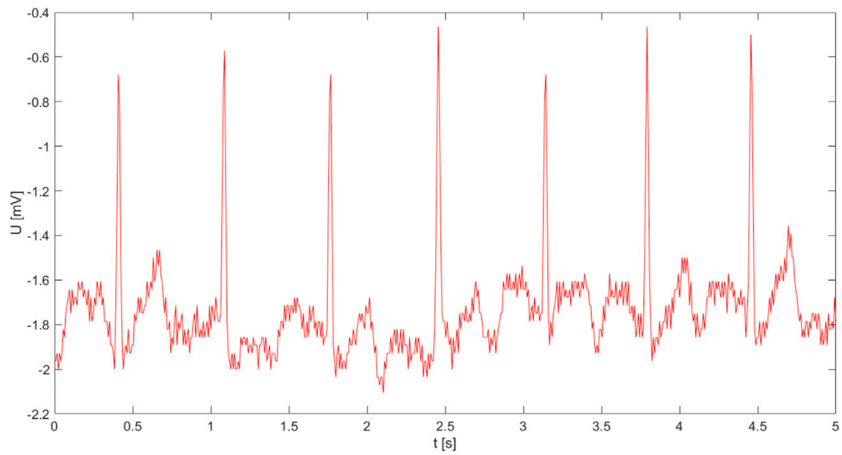

Fig. 15. ECG signal obtained from capacitive electrodes in room environment.

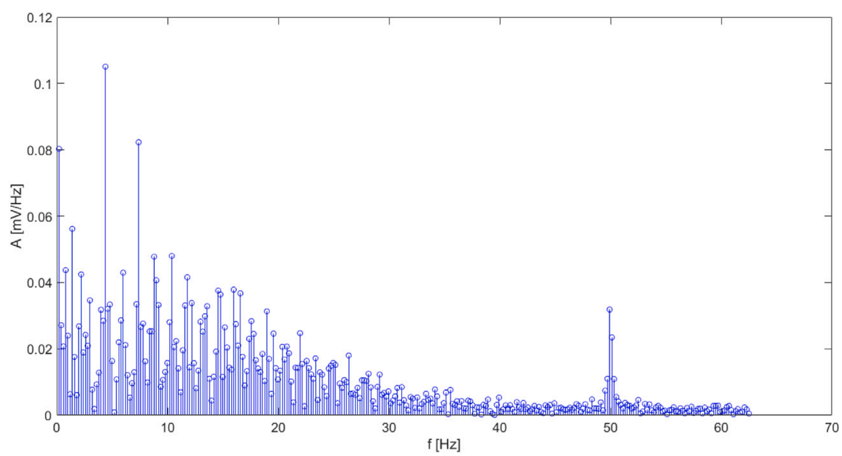

Fig. 16. Frequency spectrum of the ECG signal obtained under room conditions.

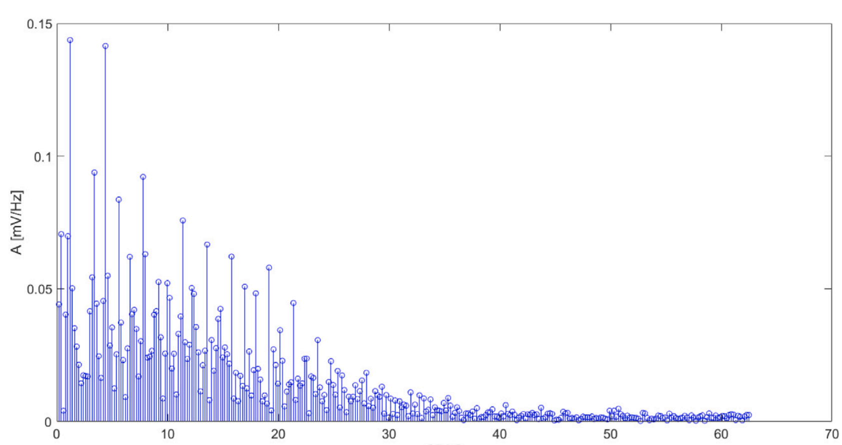

Fig. 17. Frequency spectrum of the ECG signal obtained within a motor vehicle.

As evident from the timeline presented in Fig. 15, recorded under room conditions, the signal demonstrates adequate quality, with well-visible QRS complexes and recognizable ECG waveform waves. The horizontal axis represents the time in seconds, while the vertical axis represents the signal's amplitude (voltage).

When monitoring the ECG signal within a vehicle environment, notably better results were achieved. A consistent and suitable signal could be obtained even with slight variations in electrode positioning concerning the subjects being measured (as illustrated in Fig. 18).

The improved measurements are a result of the absence of 50 Hz interference, which is typically absent in the vehicle's electrical network. This absence allows for the continuous and consistent acquisition of high-quality ECG waveforms, as illustrated in Fig. 17.

In Fig. 19 we are comparing our capacitive system with a professional clinical electrocardiograph, the SEIVA EKG practitioner. Clinical

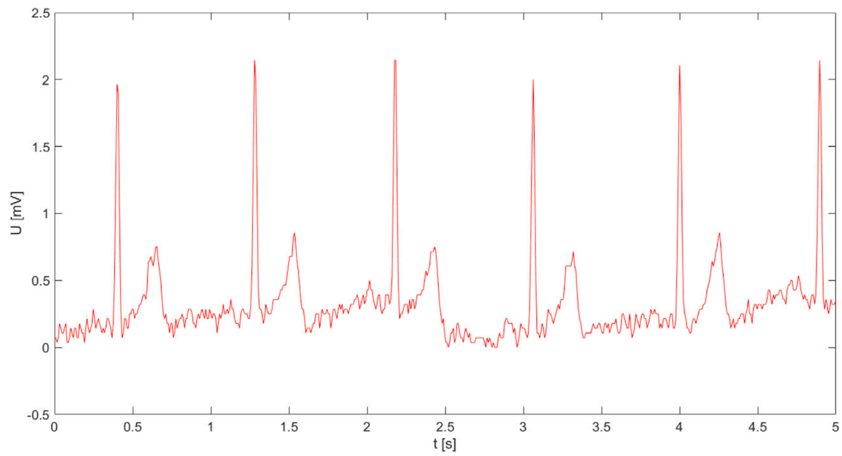

Fig. 18. ECG signal obtained from capacitive electrodes within a motor vehicle.

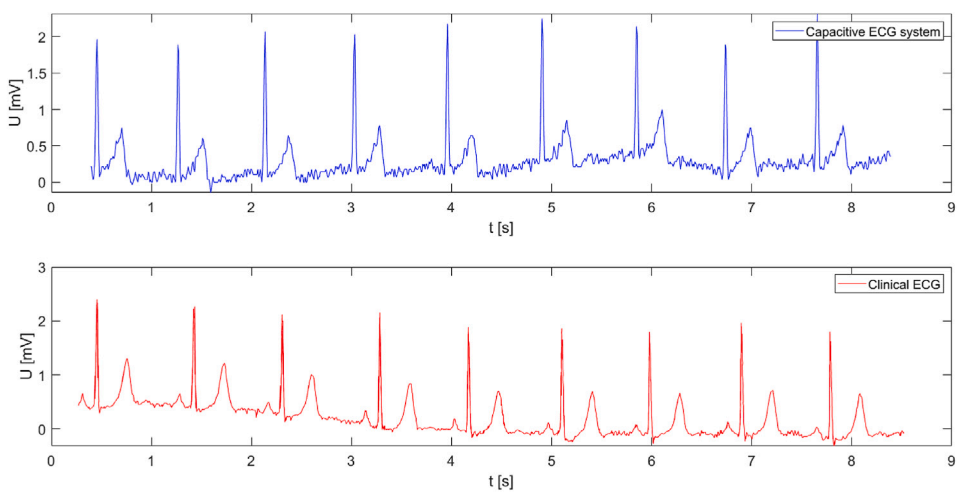

Fig. 19. The ECG signal obtained through the capacitive system (blue) and the SEIVA ECG practitioner (red).

ECG measurements were conducted with the subject in a supine position, while capacitive ECG measurements were performed with the subject seated. Since the two devices could not monitor heart function simultaneously, primarily due to the absence of a T-shirt during clinical ECG monitoring, our aim was to assess the ECG waveform quality and compare heart rates between the two methods.

To facilitate this comparison, we extracted an approximately 8-second segment from the ECG signal obtained by our system. In this period, we recorded 9 QRS complexes, resulting in 8 RR intervals for both measurement types. The average length of the RR interval was 950 ms during capacitive ECG recording and 906 ms when using clinical equipment. Using this data, we calculated the BPM value from our system as 63 beats per minute, while the BPM calculated from the clinical ECG was 66 beats per minute.

The higher heart rate recorded by the SEIVA EKG practitioner in comparison to the capacitive system could be attributed to various factors. These may include the stress associated with the examination, the act of undressing in the presence of medical personnel, and the focus on maintaining a state of calm. It is worth noting that emotional or physical stress is known to cause an increase in heart rate [65]. Additionally, it is important to consider that the ECG signal obtained from our capacitive system has not undergone digital filtering, whereas the ECG waveform from the clinical device has been processed with three filters (35 Hz, AC 50/60 Hz, 0.3 Hz).

One of the inherent challenges in ECG monitoring is the presence of breathing artifacts. These artifacts can obscure the underlying electrical activity of the heart, potentially hindering accurate interpretation. As Fig. 19 demonstrates, even professionally acquired clinical ECGs measured with SEIVA ECG equipment can exhibit this phenomenon, where a breathing pattern is superimposed on the ECG signal.

Our proposed ECG system prioritizes minimizing the impact of breathing artifacts through various design and manufacturing techniques. A crucial element in achieving this is the high-pass filter incorporated into the signal processing unit. This filter specifically targets and attenuates low-frequency components, which include those associated with respiration. Since the typical human respiratory rate falls within the range of 10–20 breaths per minute (0.16–0.33 Hz) [66], we strategically set the filter's cut-off frequency at 0.3386 Hz. This effectively removes the breathing artifact, resulting in a cleaner and more accurate representation of the cardiac signal.

4.1. Signal quality analysis

In this study, the signal quality of the capacitive ECG was examined by evaluating two morphological parameters, including duration of the RR interval and amplitude of the R wave [67]. The primary objective of this analysis is to assess the fidelity of the recorded signals by examining their clarity, consistency, and absence of artifacts. Sample image showing detection of R wave and S wave is shown in Fig. 20 and calculated duration of the RR interval and amplitude of the R wave is shown in Table 2.

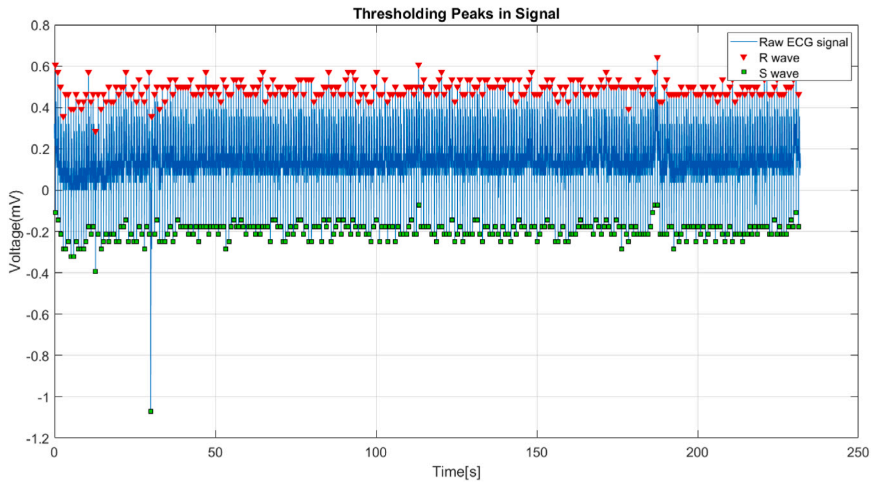

Fig. 20. Analysis of signal no. 1 in MATLAB environment.

Table 2
Analysis of duration of RR interval and amplitude in subjects seated inside the vehicle.

Signal	Duration (s)	N (–)	Mean RR (s)	BPM	Avg. amplitude (mV)
1	232	296	0.783	76	0.694
2	192	209	0.920	65	0.963
3	260	314	0.829	72	0.458
4	256	329	0.778	77	0.562
5	352	456	0.772	77	0.610

5. Discussion

Capacitance electrocardiography continues to evolve, proving its utility across diverse industries. Ongoing research in this field encompasses several key areas. These areas involve efforts to mitigate the impact of electromagnetic interference and variations in clothing thickness, the selection of suitable dielectric materials, the development of effective topological and structural designs for high-quality and versatile capacitive electrodes, as well as advancements in signal processing units [68].

We successfully demonstrated the feasibility of using a capacitive system for ECG recording. However, it is crucial to acknowledge that this system is not devoid of fundamental limitations, which we will explore in more depth in the subsequent discussion. Through our measurements, we observed that the recorded output signal is notably affected by the diversity of clothing, the type of fabric used, and its thickness. Furthermore, the quality of the measurement is influenced by the chosen power supply method. Finally, the behavior of the subjects during the measurement process is a critical consideration, as inadvertent contact with others or conductive objects can significantly alter the conditions during the acquisition of a capacitive ECG.

5.1. Effect of electromagnetic interference (EMI)

The proposed capacitive system offers numerous advantages, as outlined in previous chapters. However, these advantages, while significant, may not alone suffice to widely promote the capacitive ECG measurement in medical facilities. The primary limitation of capacitive ECG recording lies in its susceptibility to electromagnetic interference (EMI). As depicted in Fig. 21, the proximity of a mobile device to the SPU has a noticeable impact on the output ECG signal.

While we can employ various methods to mitigate this interference, complete elimination often remains unattainable. Interference from sources like mobile devices, computers, wearable electronics, and other electrical equipment in the vicinity of the patient tends to be unpredictable. Consequently, it becomes imperative to place special emphasis on the effective suppression of EMI to ensure the reliability of capacitive ECG recordings.

5.2. Power supply

Modern vehicles are equipped with an increasing array of electronics and devices that consume significant power from the vehicle's battery. In the course of our measurements, we examined the impact of activating and deactivating seat heating on the recorded signal. The undesired artifacts are depicted in Fig. 22.

It is evident that the operation of various onboard systems within the vehicle has a detrimental effect on the entire ECG measurement system. When the seat heating is operated at maximum power, the signal exhibits higher levels of noise compared to when the heating is completely turned off. Similar artifacts are present in the recorded signal even when powered from a 50 Hz electrical network. Thus, the significance of the power supply's quality in capacitive ECG signal recording becomes evident. It is advisable to delve deeper into these influences and endeavor to eliminate them entirely in our future research endeavors.

5.3. Type of clothing

The quality of the detected signal is contingent on the textile's adherence. In cases where the clothing (typically a T-shirt in our context) is loose-fitting, it results in surface irregularities that adversely affect signal quality. Material quality worn by the subject is another determining factor. Our observations reveal that recording the ECG signal through cotton fabric yields a more consistent signal compared to fabrics crafted from polyester or those with polyester blends. Additionally, the thickness of the clothing material plays a pivotal role. Notably, it becomes challenging to capture a discernible ECG signal from subjects wearing thicker T-shirts or multiple layers of clothing. Consequently, the applicability of this system in vehicles, especially during colder months when motorists commonly wear thick winter jackets and coats, appears less feasible. Nevertheless, a capacitive ECG sensing system may offer significant advantages in warmer climates, where consistent warmth and the use of lightweight, moisture-absorbing clothing by vehicle occupants contribute to more favorable outcomes for the developed system.

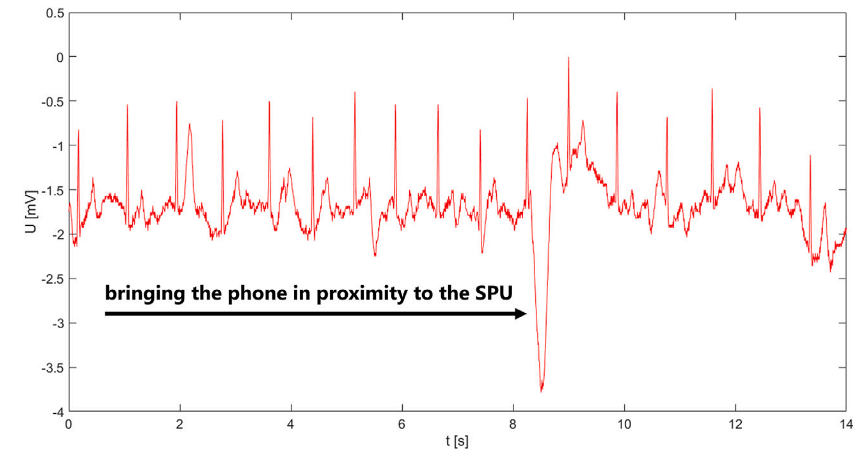

Fig. 21. Interference resulting from the close proximity of a mobile device to the SPU.

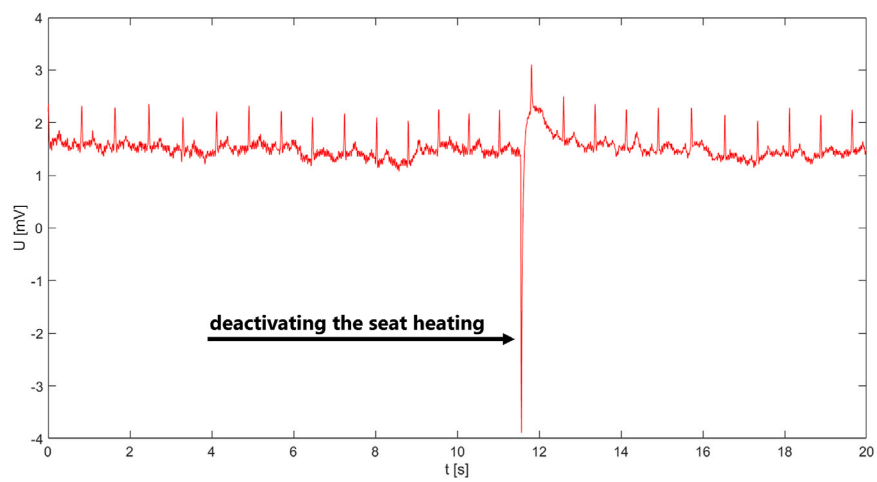

Fig. 22. ECG signal subsequent to deactivating the seat heating.

5.4. The advantages of long-term monitoring in vehicles

Given the high mortality rates associated with heart diseases, early detection of symptoms pointing to these conditions is crucial. Monitoring cardiac function in vehicles remains a relatively underexplored field. Thus, any research dedicated to monitoring vital functions can play a pivotal role in enhancing road safety [22].

A capacitive ECG sensing system within a motor vehicle can offer significant benefits. It can function as a valuable identifier, capable of detecting changes in the electrocardiogram or an increase in heart rate, which may indicate sudden cardiac disorders such as an impending heart attack or atrial fibrillation. Moreover, the system's capacity for remote health monitoring is a notable advantage. In today's busy world, regular medical check-ups are often neglected, making early symptom detection a valuable prompt for individuals to seek medical attention.

Long-term monitoring within a motor vehicle also addresses the issues of a shortage of medical professionals and prolonged waiting times for examinations. By analyzing various additional parameters, doctors can gain valuable insights into a wide range of heart conditions. Here are examples of diagnoses based on specific ECG features [69,70]:

- P wave, QRS complex, and T wave — diagnostics of sinus rhythm,
- QRS complex presence — analysis of the QRS complex presence or absence serves as a fundamental indicator for identifying potential cardiac rhythm disturbances,
- heart rates — possible diagnostics of the rhythm abnormalities (very slow or fast heart rates, extreme irregularity, or presence of frequent extrasystoles might be identified),
- P wave — diagnostics of the absence of distinct P waves can suggest atrial fibrillation,
- QRS complex — wider or narrower QRS complexes might suggest ventricular or supraventricular extrasystoles. Additionally, the coupling interval, which is the time between the preceding normal QRS and the premature QRS, can provide some information about the origin of the extrasystole.

These features allow early identification of heart diseases by the user themselves. From the perspective of continuous monitoring and ongoing care for patients diagnosed with heart disease, implementing a capacitive system in vehicles offers a promising approach to enhancing patients' quality of life.

Additionally, HRV analysis (time-domain, frequency-domain, and non-linear) might offer insides into sympathetic and parasympathetic nervous system activity. Time-domain analysis quantifies variation magnitude (e.g., SDNN) for cardiovascular risk assessment. Frequency-domain analysis (PSD) partitions ECG rhythms into frequency bands, revealing cardiac control via autonomic modulation. This might offer insights also into sleep autonomic regulation and potential sleep apnea indicators [71].

6. Conclusions

The growing interest in non-clinical heart monitoring is rapidly expanding. This article explores the development of the novel capacitive electrode prototype and custom-designed analog signal processing circuit. The capacitive electrodes were produced with a guard

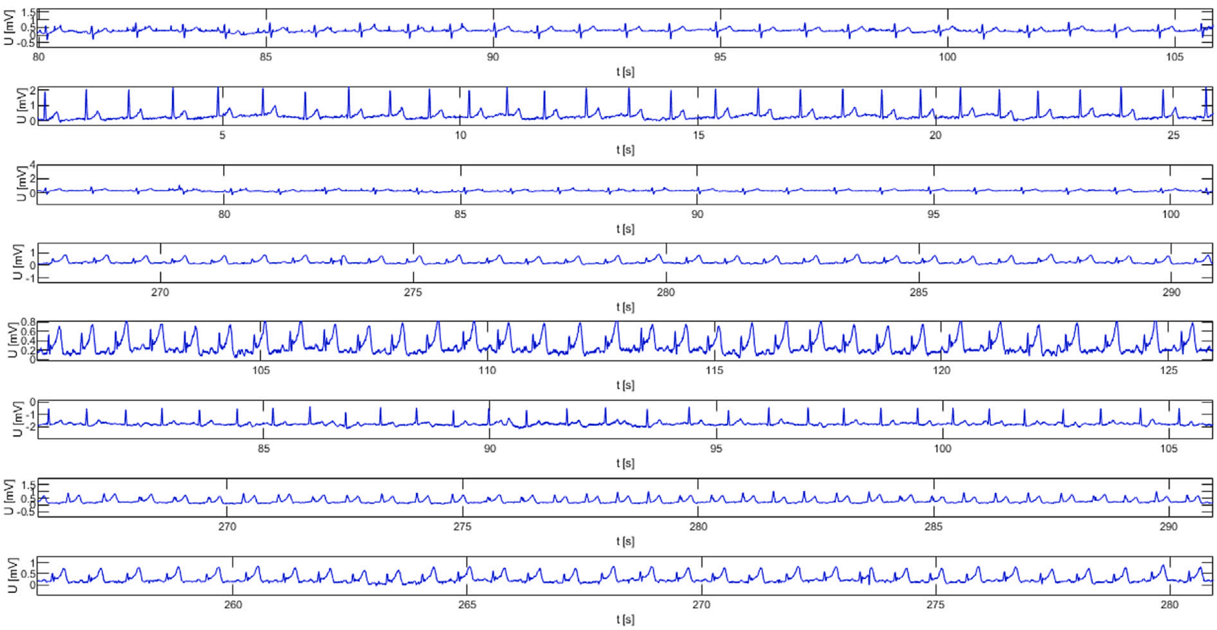

Fig. 23. Eight distinct ECG measurements showcasing varying heart rhythms and patterns indicative of diverse cardiac activity of different subjects.

layer to minimize input parasitic capacitance. We employed an optimized electrode geometry designed to minimize the size of the electrodes. Rounded edges minimize the risk of skin irritation or pressure points, making them well-suited for long-term monitoring applications. Through our experimentation, we have demonstrated that this innovative circular electrode design offers several key advantages, including great signal quality and possible adaptability for various monitoring platforms. The elimination of the need for skin contact and the reduction of potential skin irritation make this design particularly promising for long-term monitoring applications.

Experimental testing was conducted involving human subjects to verify the system's performance within motor vehicle. All subjects provided written informed consent for the publication of the results in this study. Our findings indicate that the system performs effectively when measuring subjects under ideal conditions. However, in scenarios where conditions deteriorated, such as due to improper electrode placement, electromagnetic interference, or exaggerated subject movements, the measured signals exhibited deviations from ideal results.

Overall, the experimental results indicate that our capacitive ECG monitoring system is a viable and promising technology for continuous and non-invasive cardiac monitoring. Its potential makes it a compelling option for enhancing cardiac healthcare and improving the quality of patient monitoring in clinical and everyday life settings.

CRediT authorship contribution statement

Júlia Kafková: Writing – original draft, Visualization, Validation, Resources, Investigation, Formal analysis, Data curation, Conceptualization. **Branko Babušiak:** Validation, Resources, Project administration, Investigation, Conceptualization. **Rastislav Pirník:** Supervision, Resources, Project administration, Funding acquisition. **Pavol Kuchár:** Writing – review & editing, Writing – original draft, Validation. **Juraj Kekelák:** Writing – review & editing, Formal analysis. **Filippo D'Ippolito:** Writing – review & editing, Writing – original draft.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Data availability

No data was used for the research described in the article.

Acknowledgment and funding sources

This research was supported by VEGA, Slovakia 1/0241/22 Mobile robotic systems for support during crisis situations. This research was also supported by the Operational Program Integrated Infrastructure, Slovakia for the project: Independent research and development of technological assemblies based on wearable electronics products as tools for increasing hygiene standards in society exposed to the virus causing the disease COVID-19, code ITMS2014+ 313011ASK8.

All authors have read and agreed to the published version of the manuscript.

Appendix. My appendix

See Fig. 23

References

[1] A. Arcelus, M. Sardar, A. Mihailidis, Design of a capacitive ECG sensor for unobtrusive heart rate measurements, in: 2013 IEEE International Instrumentation and Measurement Technology Conference, I2MTC, 2013, pp. 407–410, http://dx.doi.org/10.1109/I2MTC.2013.6555449.

[2] D. Doci, M. Ademi, K. Tuvshinbayar, N. Richter, G. Ehrmann, T. Spahiu, A. Ehrmann, Washing and abrasion resistance of textile electrodes for ECG measurements, Coatings 13 (9) (2023) http://dx.doi.org/10.3390/coatings13091624, URL: https://www.mdpi.com/2079-6412/13/9/1624.

[3] C.M. Vidhya, Y. Maithani, J.P. Singh, Recent advances and challenges in textile electrodes for wearable biopotential signal monitoring: A comprehensive review, Biosensors 13 (7) (2023) http://dx.doi.org/10.3390/bios13070679, URL: https://www.mdpi.com/2079-6374/13/7/679.

[4] O. Linschmann, D.U. Uguz, B. Romanski, I. Baarlink, P. Gunaratne, S. Leonhardt, M. Walter, M. Lueken, A portable multi-modal cushion for continuous monitoring of a drivers vital signs, Sensors 23 (8) (2023) http://dx.doi.org/10.3390/s23084002, URL: https://www.mdpi.com/1424-8220/23/8/4002.

[5] C. Böck, P. Kovács, P. Laguna, J. Meier, M. Huemer, ECG beat representation and delineation by means of variable projection, IEEE Trans. Biomed. Eng. 68 (10) (2021) 2997–3008, http://dx.doi.org/10.1109/TBME.2021.3058781.

[6] D.M. Mirvis, A.L. Goldberger, Electrocardiography, Heart Dis. 1 (2001) 82–128.

[7] S. Leonhardt, A. Aleksandrowicz, Non-contact ECG monitoring for automotive application, in: 2008 5th International Summer School and Symposium on Medical Devices and Biosensors, IEEE, 2008, pp. 183–185.

[8] G. Cybulski, A. Strasz, W. Niewiadomski, A. Gąsiorowska, Impedance cardiography: recent advancements, Cardiol. J. 19 (5) (2012) 550–556.

[9] J. Allen, Photoplethysmography and its application in clinical physiological measurement, Physiol. Meas. 28 (3) (2007) R1.

[10] E. Allado, M. Poussel, J. Renno, A. Moussu, O. Hily, M. Temperelli, E. Albuisson, B. Chenuel, Remote photoplethysmography is an accurate method to remotely measure respiratory rate: A hospital-based trial, J. Clin. Med. 11 (13) (2022) 3647.

[11] H. Geyer, G. Caracciolo, H. Abe, S. Wilansky, S. Carerj, F. Gentile, H.-J. Nesser, B. Khandheria, J. Narula, P.P. Sengupta, Assessment of myocardial mechanics using speckle tracking echocardiography: fundamentals and clinical applications, J. Am. Soc. Echocardiogr. 23 (4) (2010) 351–369.

[12] E. Nemati, M.J. Deen, T. Mondal, A wireless wearable ECG sensor for long-term applications, IEEE Commun. Mag. 50 (1) (2012) 36–43.

[13] V. Randazzo, J. Ferretti, E. Pasero, ECG WATCH: a real time wireless wearable ECG, in: 2019 IEEE International Symposium on Medical Measurements and Applications, MeMeA, IEEE, 2019, pp. 1–6.

[14] W. Zhang, Z. Li, J. Gryak, P. Gunaratne, E. Wittrup, K. Najarian, Evaluation of capacitive ECG for unobtrusive atrial fibrillation monitoring, IEEE Sens. Lett. 7 (10) (2023) 1–4, http://dx.doi.org/10.1109/LSENS.2023.3315223.

[15] T.-W. Wang, S.-F. Lin, Negative impedance capacitive electrode for ecg sensing through fabric layer, IEEE Trans. Instrum. Meas. 70 (2020) 1–8.

[16] A. Zompanti, A. Sabatini, S. Grasso, G. Pennazza, G. Ferri, G. Barile, M. Chello, M. Lusini, M. Santonico, Development and test of a portable ECG device with dry capacitive electrodes and driven right leg circuit, Sensors 21 (8) (2021) http://dx.doi.org/10.3390/s21082777, URL: https://www.mdpi.com/1424-8220/21/8/2777.

[17] M.S. Lee, A. Paul, Y. Xu, W.D. Hairston, G. Cauwenberghs, Characterization of Ag/AgCl dry electrodes for wearable electrophysiological sensing, Front. Electron. 2 (2022) 700363.

[18] H. Ullah, M.A. Wahab, G. Will, M.R. Karim, T. Pan, M. Gao, D. Lai, Y. Lin, M.H. Miraz, Recent advances in stretchable and wearable capacitive electrophysiological sensors for long-term health monitoring, Biosensors 12 (8) (2022) 630.

[19] B. Babusiak, S. Borik, L. Balogova, Textile electrodes in capacitive signal sensing applications, Measurement 114 (2018) 69–77.

[20] Y.G. Lim, J.S. Lee, S.M. Lee, H.J. Lee, K.S. Park, Capacitive measurement of ECG for ubiquitous healthcare, Ann. Biomed. Eng. 42 (2014) 2218–2227.

[21] Y. Gao, V.V. Soman, J.P. Lombardi, P.P. Rajbhandari, T.P. Dhakal, D.G. Wilson, M.D. Poliks, K. Ghose, J.N. Turner, Z. Jin, Heart monitor using flexible capacitive ECG electrodes, IEEE Trans. Instrum. Meas. 69 (7) (2019) 4314–4323.

[22] M. Sidikova, R. Martinek, A. Kawala-Sterniuk, M. Ladrova, R. Jaros, L. Danys, P. Simonik, Vital sign monitoring in car seats based on electrocardiography, ballistocardiography and seismocardiography: A review, Sensors 20 (19) (2020) 5699.

[23] J.H. Kim, S.M. Lee, S.-H. Lee, Capacitive monitoring of bio and neuro signals, Biomed. Eng. Lett. 4 (2014) 142–148.

[24] T. Matsuo, M. Esashi, K. Iinuma, Capacitive electrode for biomedical use. The use of barium titanate ceramics for biomedical sensing electrode (Japanese), Japan. J. Med. Electron. Biol. Eng. 11 (3) (1973) 10–16.

[25] T. Matsuo, K. Iinuma, M. Esashi, A barium-titanate-ceramics capacitive-type EEG electrode, IEEE Trans. Biomed. Eng. BME-20 (4) (1973) 299–300, http://dx.doi.org/10.1109/TBME.1973.324197.

[26] H.G. Rylander, Capacitive Electrocardiograph Electrodes, University of Texas, Electronics Research Center, 1970.

[27] Y.M. Chi, S.R. Deiss, G. Cauwenberghs, Non-contact low power EEG/ECG electrode for high density wearable biopotential sensor networks, in: 2009 Sixth International Workshop on Wearable and Implantable Body Sensor Networks, IEEE, 2009, pp. 246–250.

[28] A.B. Nigusse, B. Malengier, D.A. Mengistie, G.B. Tseghai, L. Van Langenhove, Development of washable silver printed textile electrodes for long-term ECG monitoring, Sensors 20 (21) (2020) 6233.

[29] A.B. Nigusse, D.A. Mengistie, B. Malengier, G.B. Tseghai, L.V. Langenhove, Wearable smart textiles for long-term electrocardiography monitoring—A review, Sensors 21 (12) (2021) 4174.

[30] A.H. Umar, M.A. Othman, F.K.C. Harun, Y. Yusof, Dielectrics for non-contact ECG bioelectrodes: A review, IEEE Sens. J. 21 (17) (2021) 18353–18367.

[31] B. Feng, H. Wei, B. Shi, D. Zhao, S. Ye, G. Wu, R. Wang, G. Zuo, Z. Wu, J. Chen, et al., Sleeping heart monitoring using hydrogel-textile capacitive ECG electrodes, IEEE Sens. J. 22 (10) (2022) 9255–9267.

[32] G. Schwartz, B.C.-K. Tee, J. Mei, A.L. Appleton, D.H. Kim, H. Wang, Z. Bao, Flexible polymer transistors with high pressure sensitivity for application in electronic skin and health monitoring, Nat. Commun. 4 (1) (2013) 1859.

[33] S. Park, H. Kim, M. Vosgueritchian, S. Cheon, H. Kim, J.H. Koo, T.R. Kim, S. Lee, G. Schwartz, H. Chang, et al., Stretchable energy-harvesting tactile electronic skin capable of differentiating multiple mechanical stimuli modes, Adv. Mater. 26 (43) (2014) 7324–7332.

[34] C.L. Ng, M. Bin Ibne Reaz, S.H. Bin Md Ali, M.L. Crespo, A. Cicuttin, M.E. Hoque Chowdhury, S. Kiranyaz, N.B. Kamal, Powerline interference suppression of a textile-insulated capacitive biomedical sensor using digital filters, Measurement 207 (2023) 112425, http://dx.doi.org/10.1016/j.measurement.2022.112425, URL: https://www.sciencedirect.com/science/article/pii/S0263224122016220.

[35] C.L. Ng, M.B.I. Reaz, Characterization of textile-insulated capacitive biosensors, Sensors 17 (3) (2017) http://dx.doi.org/10.3390/s17030574, URL: https://www.mdpi.com/1424-8220/17/3/574.

[36] A. Ueno, Y. Akabane, T. Kato, H. Hoshino, S. Kataoka, Y. Ishiyama, Capacitive sensing of electrocardiographic potential through cloth from the dorsal surface of the body in a supine position: A preliminary study, IEEE Trans. Biomed. Eng. 54 (4) (2007) 759–766, http://dx.doi.org/10.1109/TBME.2006.889201.

[37] H. Li, X. Chen, L. Cao, C. Zhang, C. Tang, E. Li, X. Feng, H. Liang, Textile-based ECG acquisition system with capacitively coupled electrodes, Trans. Inst. Meas. Control 39 (2) (2017) 141–148.

[38] S. Pfeiffer, S. Meyer, O. Amft, D. Anzai, J. Wang, G. Fischer, J. Kirchner, Motion-induced imbalance of contact impedance in ECG capture: Comparison of electrode materials in capacitive coupling, in: 2019 IEEE SENSORS, 2019, pp. 1–4, http://dx.doi.org/10.1109/SENSORS43011.2019.8956609.

[39] A. Takano, H. Ishigami, A. Ueno, Non-contact measurements of electrocardiogram and cough-associated electromyogram from the neck using in-pillow common cloth electrodes: A proof-of-concept study, Sensors 21 (3) (2021) http://dx.doi.org/10.3390/s21030812, URL: https://www.mdpi.com/1424-8220/21/3/812.

[40] Y. Wu, K. Xu, S. Peng, Y. Cao, Y. Huang, R. Cui, H. Zhu, Y. Guo, Z. Wang, C. Dai, et al., Clinical validation of a capacitive electrocardiogram cushion utilized for arrhythmias monitoring, IEEE Trans. Instrum. Meas. (2023).

[41] K. Wang, B. Feng, D. Zhao, K. Xiao, G. Wu, B. Shi, R. Wang, J. Chen, K. Li, L. Li, et al., Non-contact in bed measurements of electrocardiogram using a capacitively coupled electrode array based on flexible circuit board, IEEE Trans. Instrum. Meas. (2023).

[42] D. Pani, A. Dessì, J.F. Saenz-Cogollo, G. Barabino, B. Fraboni, A. Bonfiglio, Fully textile, PEDOT: PSS based electrodes for wearable ECG monitoring systems, IEEE Trans. Biomed. Eng. 63 (3) (2015) 540–549.

[43] X. Xu, Z. Liu, P. He, J. Yang, Screen printed silver nanowire and graphene oxide hybrid transparent electrodes for long-term electrocardiography monitoring, J. Phys. D: Appl. Phys. 52 (45) (2019) 455401.

[44] L.S. Johnson, A.J. Casson, Using a paper carrier to screen print on polyurethane film and fabricate ECG probes, in: 2023 IEEE International Conference on Flexible and Printable Sensors and Systems, FLEPS, IEEE, 2023, pp. 1–4.

[45] Y. Zhang, L. Xu, M. Van der Ven, S. Ouzounov, M. Meftah, M. Mischi, E. Cantatore, P. Harpe, A front-end amplifier with fast reset scheme for capacitive heart rate monitoring, IEEE Sens. J. (2023).

[46] T. Škorić, Stress level detection based on the capacitive electrocardiogram signals of driving subjects, Sensors 23 (22) (2023) http://dx.doi.org/10.3390/s23229158, URL: https://www.mdpi.com/1424-8220/23/22/9158.

[47] V.G. Sirtoli, M. Liamini, L.T. Lins, M. Lessard-Tremblay, G.E.R. Cowan, R.J. Zednik, G. Gagnon, Removal of motion artifacts in capacitive electrocardiogram acquisition: A review, IEEE Trans. Biomed. Circuits Syst. 17 (3) (2023) 394–412, http://dx.doi.org/10.1109/TBCAS.2023.3270661.

[48] J. Ding, Y. Tang, R. Chang, Y. Li, L. Zhang, F. Yan, Reduction in the motion artifacts in noncontact ECG measurements using a novel designed electrode structure, Sensors 23 (2) (2023) http://dx.doi.org/10.3390/s23020956, URL: https://www.mdpi.com/1424-8220/23/2/956.

[49] J. Sakuma, D. Anzai, J. Wang, Performance of human body communication-based wearable ECG with capacitive coupling electrodes, Healthc. Technol. Lett. 3 (3) (2016) 222–225.

[50] S.M. Lee, K.S. Sim, K.K. Kim, Y.G. Lim, K.S. Park, Thin and flexible active electrodes with shield for capacitive electrocardiogram measurement, Med. Biol. Eng. Comput. 48 (2010) 447–457.

[51] C. Harland, T. Clark, R. Prance, Electric potential probes-new directions in the remote sensing of the human body, Meas. Sci. Technol. 13 (2) (2001) 163.

[52] C. Harland, T. Clark, R. Prance, Remote detection of human electroencephalograms using ultrahigh input impedance electric potential sensors, Appl. Phys. Lett. 81 (17) (2002) 3284–3286.

[53] Y.G. Lim, K.K. Kim, S. Park, ECG measurement on a chair without conductive contact, IEEE Trans. Biomed. Eng. 53 (5) (2006) 956–959.

[54] A. Arcelus, M. Sardar, A. Mihailidis, Design of a capacitive ECG sensor for unobtrusive heart rate measurements, in: 2013 IEEE International Instrumentation and Measurement Technology Conference, I2MTC, IEEE, 2013, pp. 407–410.

[55] T. Maruyama, M. Makikawa, N. Shiozawa, Y. Fujiwara, ECG measurement using capacitive coupling electrodes for man-machine emotional communication, in: 2007 IEEE/ICME International Conference on Complex Medical Engineering, IEEE, 2007, pp. 378–383.

[56] R. Fujita, T. Ohhira, H. Hashimoto, Stable electrocardiogram measurement using capacitive-coupled electrodes, in: 2023 IEEE International Conference on Mechatronics, ICM, IEEE, 2023, pp. 1–6.

[57] Z. Xiao, C. Ma, Y. Xing, C. Yang, M. Hao, J. Li, C. Liu, Atrial fibrillation monitoring based on noncontact capacitive ECG using an integrated micro-humidity fabric electrode-sheet sensing scheme, IEEE Trans. Instrum. Meas. (2023).

[58] T.-W. Wang, H. Zhang, S.-F. Lin, Influence of capacitive coupling on high-fidelity non-contact ECG measurement, IEEE Sens. J. 20 (16) (2020) 9265–9273.
[59] S.N. Asl, F. Ludwig, M. Schilling, Noise properties of textile, capacitive EEG electrodes, Curr. Dir. Biomed. Eng. 1 (1) (2015) 34–37.
[60] T. Bednar, B. Babusiak, M. Smondrk, I. Cap, S. Borik, The impact of active electrode guard layer in capacitive measurements of biosignals, Measurement 171 (2021) 108740.
[61] A. Metting van Rijn, A. Peper, C. Grimbergen, High-quality recording of bioelectric events: part 1 interference reduction, theory and practice, Med. Biol. Eng. Comput. 28 (1990) 389–397.
[62] T. INSTRUMENTS, ADS1191, ADS1191 data sheet, product information and support | TI.com, 2012, URL: https://www.ti.com/product/ADS1191.
[63] B. Babušiak, UniversalGraph by BB. Žilinská univerzita v Žiline : s.n, 2016.
[64] E. Ajdaraga, M. Gusev, Analysis of sampling frequency and resolution in ECG signals, in: 2017 25th Telecommunication Forum, TELFOR, IEEE, 2017, pp. 1–4.
[65] J.M. Torpy, A.E. Burke, R.M. Glass, Acute emotional stress and the heart, JAMA 298 (3) (2007) 360, http://dx.doi.org/10.1001/jama.286.3.374, arXiv:https://jamanetwork.com/journals/jama/articlepdf/208031/jpg0718_360_360.pdf.
[66] M.A. Russo, D.M. Santarelli, D. O'Rourke, The physiological effects of slow breathing in the healthy human, Breathe 13 (4) (2017) 298–309.
[67] J.-J. Wang, S.-H. Liu, C.-H. Tsai, I. Manousakas, X. Zhu, T.-L. Lee, Signal quality analysis of single-arm electrocardiography, Sensors 23 (13) (2023) 5818.
[68] A. Bujnowski, K. Osiński, P. Przystup, J. Wtorek, Non-contact monitoring of ECG in the home environment;selecting optimal electrode configuration, Sensors 22 (23) (2022) http://dx.doi.org/10.3390/s22239475, URL: https://www.mdpi.com/1424-8220/22/23/9475.
[69] M.M. Butt, M.U. Akram, S.A. Khan, Classifying normal sinus rhythm and cardiac arrhythmias in ECG signals using statistical features in temporal domain, in: 2015 9th Asia Modelling Symposium, AMS, IEEE, 2015, pp. 28–31.
[70] H. Sohal, S. Jain, et al., Interpretation of cardio vascular diseases using electrocardiogram: A study, in: 2018 Fifth International Conference on Parallel, Distributed and Grid Computing, PDGC, IEEE, 2018, pp. 159–164.
[71] H. Qin, N. Steenbergen, M. Glos, N. Wessel, J.F. Kraemer, F. Vaquerizo-Villar, T. Penzel, The different facets of heart rate variability in obstructive sleep apnea, Front. Psychiatry 12 (2021) 642333.