Chapter 1

Wearable sensors for physiological parameters measurement: physics, characteristics, design and applications

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1.1 Introduction

Wearable sensors—which can be worn on the human body, or on livestock and poultry—play an important role in monitoring the physiological parameters that are used in diagnosing health conditions, supporting safe, comfortable and healthy living [1]. These sensors offer enormous potential in detecting the wide range of physiological parameters that need to be monitored regularly to determine a patient’s metabolic state, in particular for those who are hospitalized or need sudden intensive healthcare. Intensive care is one of the most challenging and stressful health services, where a doctor or a team of doctors need to make critical decisions quickly. These decisions often have to be taken when the critical physiological parameters of the patients are unknown. In such situations, wearable sensors that facilitate the measurement of calcium, lithium, lactate, cholesterol, urea, uric acid, oxalate, triglycerides, ascorbic acid, creatinine, oxygen saturation, blood pressure and pulse rate are extremely useful [1–5]. Similarly, it would be extremely helpful to have online monitoring systems for the creatinine, sodium, potassium, chloride and CO₂ levels of kidney failure patients who require frequent dialysis. Today, wearable sensing devices and associated technologies, such as saturated oxygen monitoring devices, heart rate monitoring, smart watches and smart glasses, are experiencing rapid growth [1]. It is possible that in the future these wearable devices may alter present medical healthcare practices and redefine the doctor–patient relationship, also reducing healthcare costs and the difficulties of obtaining such services. The future of wearable technologies is extremely bright as more consumers are opting for this service, and it is expected that healthcare markets will grow at a rapid pace, with a value over US$20 billion by the end of 2017 [1–3]. Wearable sensors with remote parameter monitoring capability can address patient
access issues. At present, nearly 60%–65% of the Indian population lives in rural areas, but only 10%–20% of physicians work in rural areas. Most of the doctors in rural areas are non-specialist physicians. Therefore, a rural patient with a critical condition has to travel long distance to obtain specialized medical services. Often, patients travelling long distances with critical conditions such as serious heart or lung disease die without medical treatment [3]. Remote parameter monitoring through wearable sensors with integrated transmitters can extend the services of specialist doctors in urban areas to rural areas and fill the gap in medical services in those areas. A basic block diagram of the remote vital parameter monitoring of a human or animal through wearable sensors is shown in figure 1.1. There are large numbers of emerging flexible user-friendly wearable sensing technologies that can perform a variety of physical and physiological parameter measurements. The parameters that can be measured and monitored using wearable sensors integrated with/without wireless communication features can be broadly classified into two categories: (i) physical sensors and (ii) chemical sensors (biological parameters). These technologies can be utilized for consumer electronics, medical prosthetics, artificial skin, soft robotics, therapy, drug delivery and health parameter monitoring.

1.1.1 Physical parameters

Physical parameters include motion, stress, vibration, temperature, acceleration, heart rate, neurological or cardiovascular diseases such as seizure or hypertension, and pulmonary diseases such as asthma or chronic obstructive pulmonary disease, etc. Temperature measurements on human skin can provide a lot of useful information on health conditions such as stroke, heart attack, shock lung disease and infections, etc. Human motion is affected by many factors, including physiological, anatomical, psychological, environmental and social effects [1, 3]. Hence, motion provides several health parameters in conditions such as heart attack, osteoarthritis, aging and some autoimmune diseases. For example, for a patient suffering from chronic lung disease, the 6 min walk test is an important experiment to assess the condition of their lungs.
Mobility is also affected by abnormal physical and emotional conditions. Clearly, to achieve the objective of effective monitoring, we must first be able to monitor and quantify movement, identify reduced or impaired movement, and estimate the important parameters affecting the movement. Accelerometers are commonly used to monitor human activity, such as detection of falls, motion and analysis of body movement [1, 3], or postural orientation. Accelerometers can be fabricated using piezoresistive, piezoelectric and capacitive type sensors. An accelerometer employs a mass that responds to the acceleration by causing a spring or some equivalent component to stretch or compress according to the measured parameter [1]. An inertial sensor is another important sensor for physical parameter monitoring. These sensors can be used to detect falls, movement of the body and postural orientations. A low-power, sensitive, compact plaster form of impedance sensor was developed to measure the variation of impedance of the thoracic region and heart for cardiac condition monitoring. A wearable electrocardiograph can be used for initial short-time assessment of heart functionalities [3]. A flexible, thin capacitive sensor fabricated using conductive fabric has been utilized to monitor various human activities such as heart rate, breathing rate, hand gesture recognition, swallowing monitoring and gait analysis [1]. These physical sensors may work on relative variation in their electrical parameters such as capacitance, resistance, magnetic field and piezoelectricity. Depending on the types of active sensing elements, the sensors may be classified as solid state sensors or liquid state sensors. Solid state sensors can be fabricated using bulk or nanomaterials. The nanomaterials may be polymers, carbon, semiconductors, carbon nanotubes (CNTs), metallic nanowires, polymer nanofibers or metallic nanoparticles. In liquid state sensing, the active elements may be ions or liquid metals.

1.1.2 Biochemical parameters

Biochemical parameters that can be measured by wearable sensors include pH, fluoride content, lactate, glucose, different electrolytes, the oxygen saturation of blood, transcutaneous oxygen of the eye, the presence of sodium, ammonium, potassium, keratoconjunctivitis sicca, chloride, uric acid, \( \beta \)-nicotinamide adenine dinucleotide, etc [2, 5, 6]. The body fluids for vital chemical parameter monitoring are liquid samples, which may be in the form of excreted human/animal body fluids such as urine, sweat, saliva or stools. It can also be secreted fluids such as breast milk and bile, or may be obtained by piercing a needle into the body, as in the case of blood or cerebrospinal fluid. Body fluid can also be formed due to pathological process, for example cyst fluid.

1.2 Types of wearable sensors

Two types of sensors can be employed to detect the physiological parameters: invasive (in vivo) and non-invasive (in vitro).

1.2.1. Invasive sensors

Invasive sensors (sometimes called intrusive sensors), require body fluids obtained by penetrating the body through injection or incision. For example, blood is an
important body fluid which can be used to extract most of the vital organ parameters of mammals. Collecting a blood sample to determine the concentration level of different physiological parameters requires penetration of the body by a sharp needle. Similarly, to obtain the status of living organs, some living cells need to be collected by incision, as in the case of bronchoscopy, which requires a lung sample for testing purpose. The invasive nature of the sensors can pose hurdles and cause fear for the patients, and these can be more painful and severe in case of infants, elderly people or haemophobic patients. In case of infants and elderly people, the collection of blood samples is more challenging as finding proper veins can sometimes be difficult. Continuous parameter monitoring using wearable sensors is of great importance in certain situations, such as the glucose monitoring of diabetic patients, fitness level monitoring of athletes, oxygen saturation monitoring for lung patients, cholesterol monitoring of heart patients and monitoring drug effectiveness for many diseases. In such cases, invasive sensors, which require body fluids such as blood, serum, etc, are not suitable for measurement and monitoring purposes. Also, there is a high chance of blood contamination due to improper use of needles.

1.2.2 Non-invasive wearable sensors

Non-invasive sensors do not require body fluids from a human/animal obtained by penetrating the body using injection or incision, therefore they are more attractive and less painful for the user. The body fluids used by non-invasive sensors may be saliva, sweat, tears or skin interstitial fluids [2].

1.2.2.1 Saliva

Saliva is a complex fluid of numerous chemical species permeating from the blood. Such a fluid carries many vital physiological parameters. It is easily available, and does not require pre-treatment steps for testing. Since the samples can be used directly, there is no chance of contamination. The biochemical species present in saliva provide online condition monitoring of emotional, hormonal, nutritional and metabolic condition, pH level, fluoride acidity etc. The installation of a wearable sensor utilizing saliva as a body fluid is easy—it can be fixed on the teeth.

1.2.2.2 Tears

Tears are another important biological fluid for non-invasive sensors, being composed of a complex extracellular fluid containing proteins/peptides, lipids, electrolytes and metabolites. Metabolites are extracted from the lacrimal glands, ocular surface, epithelial cells, meibomian glands, goblet cells and blood [2]. Most of these species are generated from the blood. Thus, tears can be another important fluid for detecting the presence of many physiological parameters, such as amino acids, antioxidants, metabolites, etc. However, the collection of a tear sample requires extra care, and the sample volume is small and can evaporate during transportation to distant testing laboratories. By placing miniaturized, flexible, thin wearable sensors on the retina, important parameters can be measured and monitored directly without any need for sample transportation.
1.2.2.3 Sweat
A third important biological body fluid is human/animal sweat. This fluid contains many important chemical species for the condition monitoring of mammals. Sweat can be used by non-invasive painless sensors to determine the presence of sodium lactate, ammonium, calcium, cystic fibrosis parameters, physical stress levels, osteoporosis, bone mineral loss, skin fibrosis, alcohol level, signs of drug abuse, etc. Wearable sensors for physiological parameter monitoring using sweat come in two types: flexible plastic-based fabric or epidermal-based fabric. The technique used for collecting sweat samples involves an electric current of carefully determined value, which excites a chemical stimulant into the skin. Some sensing devices utilizing sweat samples now integrate a wireless data communication module for online sharing and monitoring of measured data. This method of parameter measurement is also helpful for non-invasive animal healthcare applications. The presence of metals in the body fluid of an animal can also be measured using a sweat sample.

1.2.2.4 Skin interstitial fluid
This is the fourth important body fluid which can be utilized to extract important physiochemical parameters. It consists of a water solvent containing sugars, salts, fatty acids, amino acids, co-enzymes, hormones, neurotransmitters, white blood cells and waste products from cells. This fluid can be used for the detection of sugar levels, organ failure, drug efficacy, salts, etc [6].

1.3 Wearable sensors for animal health
The traditional methods of livestock monitoring—using a notepad or offline instruments without the capabilities of data storage, analysis and sharing through wireless communications—are time consuming and manual. Modern sensors with wireless feature are now popular for the physiological monitoring of livestock parameters. Radiofrequency identification tags attached or embedded in the animal body have become popular for monitoring health parameters such as fat, weight, milk and location [5]. Precision livestock farming using wireless sensors for health management has affected the livestock industry tremendously in term of the welfare of animals. It also helps to increase the productivity of the farms by efficient utilization of animal fodder and water.

It is important to note that, although humans and cows (for example) are both mammals and have many similarities in terms of physiological parameters, there are many differences in their basic parameters. For example, human blood has four groups, while in cow blood samples a total of 11 blood groups are present [5]. Because of these differences, often the same detection method and instrument are not applicable for different animals, for example the sensor for human β-hydroxy butyrate (β-HBA) detection is not suitable for the detection of ketosis in other animals. Detection of the ketosis parameter is important for the efficient health management of all animals, including humans, but the same sensor cannot always be used for different species. Different techniques can be employed for the detection of β-HBA, but the optical spectroscopic principle of UV absorbance in the range of 445–455 nm can detect the lowest level of 0.05 mM [5]. Electrochemical sensors with catalytic enzyme 3-hydroxybutyrate dehydrogenase can selectively detect β-HBA.
1.3.1 Physiochemical parameters
The essential biochemical parameters to be monitored are the concentrations of sodium, potassium, pH, ammonia, ketosis, progesterone, glucose, ethanol, lactate, cortisol, urea, peptides, calcium, etc.

1.3.2 Physiological/physical parameters
The physiological/physical parameters of animals that can be monitored using wearable non-invasive sensors include weight estimation, animal tracking, behaviour monitoring, respiratory infection using audio/video data and blood saturation level, sound analysis, vocal distress, stress in egg laying, oestrus using audio surveillance, chick thermal comfort using chick noise, pregnancy from body temperature, body temperature, acceleration for fall detection, etc.

1.3.3 Pathogen detection
The influenza virus causes highly contagious disease among birds. It can spread to other birds through saliva, nasal secretion and other excretions from infected birds. Even sharing food and water can spread the influenza virus known as bird flu. This influenza virus has a low pathogen attack, causing small egg production, or a high pathogen attack, causing deadly multi-organ failure and rapid spread through flocks [7]. The effects of the deadly bird flu are high body temperature and abnormal body movement. Both of these physical parameters can be measured using wireless sensor networks using wearable temperature and accelerometer sensors. Other pathogenic viruses which cause reproductive or respiratory disease (such as porcine reproductive and respiratory syndrome virus [5]) can be detected using optical ellipsometry techniques. Bacteria also can cause many diseases. A microfluidic platform that utilizes an electronic tongue consisting of an electrochemical sensor array, and a data analysis scheme employing principal component analysis and the linear discriminant method, can be used to identify and to estimate the concentration of different foodborne pathogenic bacteria. Important bacteria which can be detected by the wearable sensors include Escherichia coli, Listeria monocytogenes, Listeria innocua, Salmonella typhimurium, Salmonella enteritidis, Pseudomonas aeruginosa, methicillin-resistant Staphylococcus aureus (MRSA) 35 and MRSA 86. The sensor array consists of biosensors fabricated using silver/gold nanoparticles with different sensitivity and selectivity. The biosensors with silver or gold nanoparticles embedded into a nanostucture of CNTs can detect anthrax spores or multi-drug-resistant bacteria. Hybrid nanoparticles such as graphene-oxide-based electrochemical sensors can be employed for detecting L. monocytogenes necaterai, a major food pathogen causing listeriosis [8].

1.4 Working principles of wearable sensors
Different techniques can be employed to operate the sensors, such as optical, electrical, piezoelectric and electrochemical. Two important classes of wearable sensors for physiological parameter measurement and monitoring are (i) electrochemical sensors and (ii) impedance sensors. These two classes of sensors are extensively used for
measuring various parameters. Electrochemical sensors can be further classified as (i) conductive, (ii) amperometric and (iii) potentiometric [2]. Impedance sensors, which mostly include resistive and capacitive approaches, are extensively used for the fabrication of different sensors. A capacitive sensor provides high sensitivity, low temperature dependence, small size and small power consumption, with the possibility of sensing large varieties of physical and chemical parameters. Capacitive sensors may be of different types, such as parallel-plate, co-axial cylindrical, cylindrical cross-capacitor and fringing field [9]. The fringing field of the capacitor can also be used for sensing the strength, location and texture of the sample, and many physiochemical parameters. Capacitive sensors are also suitable for non-invasive parameter measurement [10]. The sample for the measurement of parameters may be in gaseous form or in liquid form. Electrochemical sensors play a dominant role in measuring physiochemical parameters because of their high sensitivity, portability, simplicity of construction and low cost. Many commercial hand-held analysers, such as ACCU-CHEK (Roche Diagnostic Ins), Lactate Scout (Sport Resource Group) and iSTAT (Abbot, Inc), measure metabolites and electrolytes using electrochemical sensors. Most of these sensors use blood samples and are thus invasive in nature. Many wearable physical sensors have been developed for monitoring vital physical fitness levels, but few non-invasive electrochemical sensors using body fluids have been developed, with very few available for commercial applications [2].

1.4.1 Characteristics of wearable sensors

When a sensor works in some environments to detect certain parameters, it may encounter three different inputs: the (i) target input, (ii) interfering input and (iii) modifying input. The target input is the parameter which is intended to be measured by the sensor. Interfering inputs refer to those inputs to which the sensors are unintentionally sensitive. Modifying inputs are those that cause a change in the input–output relation of the sensor to the target and the interfering inputs. For example, in a thermometer which is used to measure body temperature, the change in temperature causes a change in resistance of the sensor, which is finally converted into an electrical signal. Body temperature is the desired input. One interfering input that often causes error in the measurement is the 50 Hz electromagnetic field produced by electrical apparatus in the surrounding environment causing an unwanted induced voltage in the circuit. The self-heating effect of the thermistor and fluctuation of the excitation voltage may act as modifying inputs. Both of them change the actual input–output relation of the sensor. The characteristics of a wearable sensor in general are divided into two categories: static characteristics and dynamic characteristics. There are several physiological parameters which are almost constant or vary very slowly. The body temperature of a human remains almost constant except for an increase in the case of fever. The performance characteristics of the sensor for such inputs are the static characteristics. The important static characteristics are sensitivity, span, accuracy, resolution, threshold, tolerance, linearity, hysteresis, short-term and long-term drift, response time, recovery time, cross-sensitivity, yield ratio and interchangeability [10, 11].
However, there are many parameters that vary rapidly. Dynamic characteristics study the performance characteristics of the sensor for such inputs. The dynamic characteristics of an instrument are studied with some standard input signals such as step, ramp, parabolic and sinusoidal inputs. The output response to a step input of a first-order sensor is the transient response reaching a steady state value and then returning to the initial base value during recovery. For a ramp input, the input signal changes linearly and the output response is also linear. However, for a second-order sensor, the output transient response to the step input is either under damped, over damped or a critically damped response. For a sine input, the signal is harmonic, and the response is the frequency response. From the dynamic step response, the rise time, recovery time, reproducibility, settling time and peak overshoot can be determined [10]. The electrical output of a sensor can be related to the input parameter by an $n$th order polynomial mathematical equation. The order of the equation may change according to the complexity of the sensor. The electrical output may be voltage, current, frequency (time period) or phase angle.

1.5 Issues in the fabrication of wearable sensors

1.5.1 Toxic nature of fluids

The biofluids which are used to measure important physiological parameters are often chemically corrosive or toxic. There is always the possibility of a chemical reaction that changes the sensor irreversibly through acid attacks. Sensors employing electrolytes (fluids) for conduction of current lose a small amount of analyte, requiring that the electrolyte be replenished.

The sensors are often in contact with large numbers of analytes in the medium, which introduces interferences in the sensor output. For a porous catalytic electrode, the adsorption of species that cannot be removed causes modification of pore morphology. The change in effective surface area due to the modification of morphology changes the calibration of the sensor.

There may be some species that reduce the reaction rate thus reducing the sensitivity. A filter with suitable material and pores helps to eliminate those analytes. The analytic selective film that is used to improve the selectivity and the sensitivity of a sensor can be poisoned by some non-removable species causing drift in the output. For example, plaques develop within minutes over teeth and this biofilm hinders the performance of wearable salivary sensors. A protective coating with antimicrobial material can minimize this problem [2].

1.5.2 Electrode material

Recent publications show that materials such as copper, silver, nichrome, chromium, aluminium, stainless steel, gold and platinum are being used in various electrochemical sensors for biological conductivity and impedance measurement [2, 6]. The contact electrode of the sensor sometimes reacts with the analyte resulting in a change in the resistance of the wire. The resistance of the wire also changes due to oxidation of the electrode material. The electrolyte samples which are used to determine body parameters sometimes promote oxidation of the electrode.
Compared to other electrode materials, platinum electrodes are the most frequently used in electrochemical cells because this metal has a much lower impedance and does not oxidize easily [6].

1.5.3 Electrode polarization

Another important error for this type of sensor is the polarization of electrodes. A Faradic or charge transfer process occurs at the electrode surface. The contact area of the electrodes with the electrolyte plays a role on the electrode’s polarization impedance, as the electrode polarization impedance is inversely proportional to the electrode surface area. If the sensor is excited by direct current (DC) while the electrode is immersed in a conducting liquid, ions are neutralized at the electrode surface area and depositions are formed. The depositions of electrolyte can be seen from bubble formation and are caused by the presence of organic salts or the ionic nature of the electrolyte. Such behaviour reduces the contact area of the electrode. Therefore, it is important for the sensor to give due consideration to keeping the electrode area constant during measurement. By applying a small alternating current (AC), the electrode polarization can be minimized because the alternating electric field at the interface keeps on changing. Another important consideration is the formation of a double layer adjacent to the electrode when the potential is applied. The effect of the Faradic process can be minimized by selecting a high cell constant \((L/A)\). This implies selection of a small electrode surface area \((A)\) and a large separation distance between the electrodes \((L)\). This, however, reduces the sensitivity of the Wheatstone bridge circuit that is often used to interface the sensors. A better solution is to use a multi-electrode, as discussed in section 1.5.4. Both Faradic and double-layer capacitance can be minimized using a high-frequency, low-amplitude AC signal. Another good technique is to balance the capacitance and resistance of the cell by connecting a variable capacitor in parallel to the resistance of the bridge area adjacent to the cell [9].

1.5.4 Electrode geometrical arrangement

The electrode configuration and techniques implemented significantly contribute to the accuracy of conductance and impedance measurements. In practice, conductance measurements using two, three, four or more electrodes have been reported in the literature [12]. The four-electrode measurement, sometimes called tetrapolar, is a good choice in the case of impedance measurements. Figure 1.2 shows a schematic diagram of different electrode configurations for conductivity measurement of a blood sample.

In a four-electrode configuration, the excitation current passes through the outer two electrodes and the voltage drop is measured between the inner two electrodes. In the two-electrode configuration, current passes through and the voltage drop is measured across the same two electrodes. The two-electrode conductivity measurement method is not accurate due to the frequency and current-density dependence of the impedance–electrolyte interface. Only the AC method with suitable signal frequency is applicable, not DC, because of electrode polarization. The use of four electrodes can eliminate this issue through its separate exciting electrodes and drop
voltage measuring electrodes. However, by choosing AC excitation with a higher signal frequency, the two-electrode configuration is also suitable for conductivity measurements. A multi-electrode arrangement with more than four electrodes can also be used for accurate measurement of conductivity. Other factors to be considered are the shape of the electrode (rectangular, circular, etc), the placement of electrodes (vertical, horizontal, etc) and the distance between the electrodes.

1.6 Fabrication of wearable sensors using electrical properties

1.6.1 Impedance sensors

Impedance spectroscopy involves measurement of the impedance of a sensor in the presence of biological species by applying a small, time-varying AC signal of different frequencies. It is simply the ratio of the root mean square (rms) value of the output voltage to the rms value of the input excitation current. This method allows the measurement of the physiological parameters of living cells. The change in impedance caused by some pathogens can be associated with variations of essential biochemical species such as physiological structure or ionic compositions and their concentration. This method of measurement is an effective technique for indirect non-invasive determination of vital physiological parameters in a variety of wearable sensing applications. The important passive electrical components of the impedance are the resistance, capacitance and inductance. According to the
compositions of physiological parameters, the passive components may have resistance and capacitance connected either in series or in parallel. Impedance sensors can be used to study the dielectric and bulk conductive properties of body fluids. Planar interdigitated microelectrodes (IDE), consisting of alternate arrangements of sensing and working electrodes, are the most commonly used electrode configuration for impedance sensors. Important features of this configuration are miniaturization of the electrode, simple fabrication, mass fabrication at low cost, the ability to use them over a wide range of applications without significant change in their construction, and the ability to integrate the electrode with interface electronics to develop autonomous wearable chips for wireless measurement purposes. When the microelectrodes of the sensors are in contact with a liquid sample such as blood, a much higher impedance value compared to a macroelectrode is normally observed. This high value of impedance is due to the formation of interfacial capacitance or double-layer capacitance ($C_{dl}$) at the interface between the electrode surface and the liquid sample [13]. Thus the interactions of ions and molecules present in the liquid sample cause the formation of $C_{dl}$. This $C_{dl}$ is directly proportional to the contact area of the electrodes with the liquid sample. In addition, $C_{dl}$ is frequency-dependent and increases the measurement error at low frequencies (below 1 kHz) [14]. Therefore, the configuration of the IDE and the excitation frequency needs to be optimized to reduce the interfacial impedance, to improve the performance of the sensor and to increase the frequency towards a higher value. Figure 1.3(a) shows a schematic of a conventional planar IDE electrode impedance sensor for measuring the conductance and capacitance of a fluid. The configuration consists of $N$, the number of fingers; $G$, the gap between the fingers; $W$, the width of the finger; $L$, the length of the finger; and $A$, the area of the electrode surface. The excitation AC source is connected between the working and the sensing electrodes. The response characteristics of the IDE sensors are influenced by the number of electrodes, the wavelength, the distance between two adjacent working (positive) electrodes, the area of the IDE pattern, and the parameters to be measured. The IDE electrodes can be deposited on the top or bottom layer of the sensing film. The simplified equivalent circuit of the sensor is shown in figure 1.3(b). This type of equivalent circuit is reported in the literature by several authors [13, 15]. The circuit has a series combination of

![Figure 1.3](image_url)

Figure 1.3. (a) Conventional interdigitated electrode configuration. (b) Equivalent circuit of the impedance sensor.
double-layer capacitance $C_{dl}$ and medium resistance $R_x$, which is in parallel with the dielectric capacitance $C_x$. Each working and sensing electrode forms a $C_{dl}$ capacitance at the electrode–electrolyte interface to facilitate physical modelling of the sensor. The $C_{dl}$ depends on the material of the electrode and the composition of the liquid medium. For coplanar IDE, the $C_{dl}$ can be approximately given by \[ C_{dl} = 0.5 AC_{dl\_surface}, \]
where $A = WLN$ and $C_{dl\_surface}$ is equal to the characteristic capacitance of the Stern layer for a medium with high ionic concentration. The Stern layer capacitance is approximately equal to $C_{Stern\_surface} = 10^{-2} \mu F/cm^2$. However, $C_{dl}$ can be replaced by constant phase impedance (CPI), which is a measure of non-Faradaic impedance. The impedance of a CPI in the Laplace domain can be represented by \[ Z(s) = QS^{-\alpha} = Q(j\omega)^{-\alpha} < (-\frac{\pi}{2} \alpha). \] The phase angle of a CPI does not change with signal frequency but depends on the fractional exponent value $\alpha$. $R_c$ is the series contact resistance of the lead wire and its value is negligible. $R_x$ and $C_x$ are the main components of the equivalent circuit representing the physiological parameters.

The resistance value of the sensors depends on the conductivity of the liquid and the sensor constant $K_Z$, which are related by \[ R_x = K_Z/\sigma_x, \] where $K_Z$ can be given by \[ K_Z = \frac{2}{(N-1)L} \times \frac{P(k)}{P_l(1-k^2)}, \] where $P(k) = \int_0^1 \frac{1}{\sqrt{(1-t^2)(1-k^2 t^2)}} dt$, $k = \cos(\frac{\pi}{2} \cdot \frac{W}{G+W})$ and $t$ is the thickness of the electrode.

The function $P(k)$ is an incomplete elliptical integral. The sensor constant $K_Z$ entirely depends on the geometrical parameters of the sensor ($N, L, G, W$). If $K_Z$ is known, then by measuring the resistance of the sensor, the conductivity of the fluid can be determined using equation (1.1). The capacitance value of the sensor with permittivity medium $\varepsilon_x$ can be determined by \[ C_x = \frac{\varepsilon_0 \varepsilon_x}{K_Z}. \] (1.2)

### 1.6.2 Design of an optimum IDE electrode configuration
Neglecting the contact resistance of the sensor, the impedance of the sensor can be written as \[ Z = \frac{Z_R}{1 + j\omega C_x Z_R}, \] (1.3)
where $Z_R = R_x + \frac{2}{j\omega C_{dl}}$. Replacing $Z_R$ in equation (1.3), the total impedance can be written as \[ Z = \frac{R_x + \frac{2}{j\omega C_{dl}}}{1 + \frac{C_x}{C_{dl}} + j\omega C_x R_x}. \] (1.4)
The frequency response of the impedance is shown in figure 1.4. The response has three regions. In region 3, when the signal frequency is more than \( f_h \), the dielectric capacitance \( C_x \) is dominating and the total impedance is close to the impedance of the dielectric capacitance \( C_x \). In region 1, when the frequency is less than \( f_l \), the double layer capacitance mainly contributes to the total impedance. In region 2, when the frequency is more than \( f_l \) but less than \( f_h \), the resistance of the solution mainly contributes to the total impedance. Total impedance at a frequency below \( f_h \) is equal to \( Z \approx Z_R \) and the lower frequency is given by

\[
f_l \approx \frac{1}{\pi R_x C_{dl}} = \frac{\sigma_x}{0.5\pi WLN C_{dl, surface} K_Z} = \frac{\sigma_x}{0.5\pi C_{dl, surface} K_G K_Z},
\]

where, \( K_G = WLN \).

In region 2, the conductive sensitivity is higher and the conductivity is independent of the signal frequency.

In region 3, at a frequency above \( f_h \), the double-layer capacitance is negligible and the impedance

\[
Z = \frac{R_x}{1 + j\omega R_x C_x}
\]

and the frequency at the boundary

\[
f_h = \frac{1}{2\pi R_x C_x} = \frac{\sigma_x}{2\pi \epsilon_0 \epsilon_x}.
\]

The higher frequency is independent of the geometrical parameters but the lower frequency depends on the geometrical parameters. Therefore, by optimizing the geometrical configuration, the lower frequency can be reduced and the frequency range can be extended.

To maximize the conductive sensitivity in equation (1.5), \((K_G \times K_Z)\) should be maximized. By increasing \( W \), \( K_G \) can be increased but the sensor constant \( K_Z \) is also reduced. If a square IDE structure is selected, then the length of the IDE finger can be written as

\[
L = N \times (W + G) - G.
\]

Since \( L \approx \mu m \gg G \approx \mu m \), \( L \approx N \times (W + G) \approx NW(1 + r) \), where \( r = G/W \).

![Figure 1.4. Frequency response curve of the impedance sensor in an ionic medium.](image-url)
Then

\[ K_G K_Z = WLNK_Z = \frac{2L}{(N - 1) (r + 1)} \frac{1}{P\left(\sqrt{1 - k^2}\right)} = D(N, L) \times Y(r), \] (1.6)

where \( D(N, L) = \frac{2L}{(N - 1)} \) optimizes the size of the sensor, \( Y(r) = \frac{1}{(r + 1)} \frac{p(k)}{p(\sqrt{1 - k^2})} \) is related to the sensor constant.

For maximization of \( K_G \), \( D(N, L) \) should be increased, which indicates increasing \( L \) and decreasing the number of fingers. But the sensitivity of the sensor also depends on the number of fingers and the sensitivity increases by increasing the area of the contact surface of the IDE with the medium under study. Therefore, a higher number of fingers leads to high sensitivity. The second factor \( Y(r) \) is related to the ratio of finger gap to finger width. This factor can be optimized by calculating \( Y(r) \) with different values of \( r \). It is shown that by calculating \( Y(r) \) for different values of \( r \), the optimum value of \( r \) is found to be 0.66. Thus optimum square IDE structure for maximum frequency range is \( W = 1.5G \) and \( L = (2.5N - 1)G \).

In order to increase the sensitivity of the wearable sensor to physiological parameter measurement, it is preferable to use the electrode structure with the lowest impedance in the frequency range where the total impedance of the sensor is close to the resistance of the medium \( R_v \). The IDE which has a minimum \( K_Z \) is more suitable because the impedance variation due to medium is significant and can be easily detected. Taking an IDE configuration with \( W = 1.5G \), the sensor constant \( K_Z \) can be optimized for different numbers of fingers \( N \), and it is reported that the \( K_Z \) value decreases with increasing \( N \) up to 20. Thus, the optimization rule to be followed in designing a square co-planar IDE sensor with a fixed surface area is \( W = 1.5G \) and \( N \leq 20 \) [13].

For example, for a square IDE of area 1 mm\(^2\) and \( N = 4 \), the length of an IDE finger is \( L = N(W + G) \), \( W + G = L/N = 1000 \mu m/4 = 250 \mu m \), then taking \( W = 1.5G \), \( G = 250/2.5 = 100 \mu m \) and \( W = 150 \mu m \).

### 1.6.3 Perfect capacitive IDE sensors

If the IDE sensor is intended to test a sample with almost perfect relative permittivity (conductance is negligible), then a capacitive IDE sensor will be required. There are several applications where such types of sensor can be used to test a dielectric sample. For example, testing the quality of transformer oil or edible oil can be treated with such a sensor. Consider a planar half-wavelength cross-section of an IDE capacitive sensor for testing a pure dielectric sample, as shown in figure 1.5.

The structure consists of two working and one sensing electrodes on an insulating substrate of dielectric constant \( \varepsilon_s \) (conductivity is negligible) separated by \( G \). The thickness of the planar electrode is \( t \). The space between the electrodes is filled with dielectric sensing film of dielectric constant \( \varepsilon_f \). On the top boundary of the electrode, there is another dielectric of permittivity \( \varepsilon_v \). In this structure, three capacitances \( C_1 \), \( C_2 \) and \( C_3 \) with relative permittivities \( \varepsilon_{y1}, \varepsilon_s \) and \( \varepsilon_f \), respectively, will be formed. Using a complete elliptic integral, of the first kind, \( K[x] \), the sum of the capacitances \((C_1 + C_2)\) per unit length can be given by [16]
The capacitance $C_3$ formed with relative permittivity $\varepsilon_f$ is approximately given by

$$C_3 = \varepsilon_0 \varepsilon_f \frac{t_e}{G}.$$  \hspace{1cm} (1.8)

Total capacitance for this half-wavelength configuration between the working and the sensing electrodes can be given by

$$C_T = C_1 + C_2 + C_3.$$  \hspace{1cm} (1.9)

When the IDE sensor is excited by an AC signal, the penetration depth of the fringing electric field depends on the spatial wavelength, i.e. the gap between the working and sensing electrodes, and is independent of frequency [16]. However, the dielectric property of the biochemical species may change due to signal frequency. In wearable sensing applications, where only a few µL of the sample body fluid is available and is required to be exposed on the sensing area, conventional IDE sensors are a good choice. Another important feature of the IDE sensor is the ability to test the sample from one side, i.e. the test sample need not be placed between two sensing and working electrodes. The electric field passes through the sample under test (SUT). Hence the impedance, comprising the capacitance and the resistance, depends on the SUT properties and the geometrical configuration of the sensor. These IDE sensors have most conveniently been used for the detection of food borne pathogens, humidity, volatile organic vapours, inspection of sea food, meat, and biological species, etc [10]. In the conventional structure, the number of working and sensing fingers is the same. Sometimes, a guarding plate is provided on the back side of the substrate (polyimide) as shown in figure 1.6. Such an arrangement can improve the performance of the IDE.
sensor by nullifying the unwanted parasitic lead and earth capacitance. The guard electrode is maintained at ground potential or at the potential of the sensing electrode. Neglecting the contact resistance, the equivalent circuit of the configuration is also shown in figure 1.6. The SUT is represented by \((R_x, C_x)\), the impedances at the working and sensing electrodes are represented by \((R_{wg}, C_{wg})\) and \((R_{sg}, C_{sg})\), respectively. \(V_s\) is the AC voltage source applied to the working electrode, and the output is taken from the inverting opamp circuit; then the output voltage \(V_0 \) almost depends on the unknown resistance and the capacitance of the SUT. Because of the virtual ground of the opamp, \(R_{sg}\) and \(C_{sg}\) are minimized. Similarly, since the \(V_s\) with low source impedance appears across the working impedance, the output will be least affected, provided the signal frequency is not very high.

Some modified forms of IDE sensors with a greater number of sensing electrodes than working electrodes have been proposed in the literature [10]. Having more sensing electrodes between two working electrodes enhances the magnitude of the field penetration depth parameter in the sensor. This enhanced field penetration depth of modified IDE sensors over conventional sensors results in better impedance profiling of the test sample. The modified design of the sensor thus shows better performance and sensitivity. Figure 1.7(a) shows a schematic diagram of the modified configuration of the IDE sensor. It consists of four sensing electrodes between two working electrodes. Figure 1.7(b) shows the field distribution of an IDE sensor with fingers with different wavelengths. The field distribution of the small wavelength covers a small sample area while the large wavelength covers a larger area. The design and simulation results of such an IDE structure are shown in figure 1.7(a), with different sensing electrodes reported in the literature [17]. The optimum configuration was designed by keeping the effective area and the distance between two adjacent electrodes unchanged. The only parameter varied was the number of sensing electrodes between the two working electrodes. It is shown that optimum numbers of negative electrodes between positive electrodes will give a better distribution of the electric field, but results in a comparatively weak field strength.
The electric field strength is sufficient to interact with the SUT. The highest capacitance with a uniform electric field distribution results in better sensitivity of the configuration for the sample under test. A study was also carried out to decide the optimum sample thickness for better sensitivity. Figure 1.8 (a) shows the spiral electrode structure and (b) shows the top electrode with mesh.

Figure 1.7. (a) Schematic of modified design of the IDE configuration. (b) Field penetration depth of the structure with different wavelength fingers [10].

Figure 1.8. (a) Spiral electrode IDE configuration and (b) grid electrode configuration.
1.7 Electrochemical wearable sensors

The sensor converts the physiochemical parameter into a measurable electrical signal proportional to the analyte concentration. The electrical signal may be change in current or voltage or impedance. Electrochemistry implies the transfer of charge or ions from one electrode to another electrode through solid or liquid electrolyte [2]. The electrode reactions or charge transport can be modulated by the biochemical fluids and hence form the basis of sensing. The electrochemical sensor is always required to form a close circuit to flow electrical current. A typical electrochemical sensor consists of a sensing electrode (working electrode), a counter electrode (auxiliary electrode), a reference electrode and the electrolyte, the analyte test sample for detecting the parameters. The sensing electrode of this sensor is often made of catalytic metals such as platinum, or palladium or carbon coated metal. The sensing electrode is the one where the actual reaction takes place, and is specially designed to increase the surface area or catalytic to enhance the reaction rate with analyte and hence the sensitivity; above all it needs to be chemically inert. The sensing electrode is catalysed and many times nanoporous to increase the effective surface area. Depending on the design of the sensor, all three electrodes may be of different materials. The electrical signal is measured between the sensing and the working electrodes. It is essential to maintain a stable and fixed potential at the sensing electrode. In practice, the sensing electrode potential with respect to the counter electrodes does not remain constant due to the continuous reaction at its surface. A fixed potential is applied to the sensing electrode. The reference electrode is used to maintain the potential of the sensing electrode at the fixed value. It is placed towards the sensing electrode between the counter and the sensing electrode. No current flows to or from the reference electrode. In some electrochemical sensors, no external voltage is applied to the sensing electrode, hence there is no reference electrode. Figure 1.9 shows the schematic of a typical electrochemical sensor. A semipermeable membrane is sometimes used to cover the catalytic sensing electrode or, in some cases, to control the amount of analyte reaching the electrode surface. Such sensors are called membrane clad sensors. The membrane is made of thin, low-porosity Teflon. The membrane provides mechanical protection and

![Diagram](image_url)

**Figure 1.9.** (a) Basic electrochemical sensor and (b) typical set-up of electrochemical sensor.
additionally filters out unwanted particulates. Some manufacturers provide screen printed electrodes with contact pads using different metals such as copper, silver, carbon, gold, platinum and nickel. Screen printed electrodes on a single substrate for multiple electrochemical sensors are also available on the market [18]. These electrodes are cheap, disposable, require micro-volumes of the liquid sample and can be useful for environmental, healthcare and agro/food processing applications. The electrolyte of the sensor helps the cell reaction and transports the ionic charge across the electrode. It forms a stable potential at the reference electrode. The composition of electrolyte should be compatible with the electrode materials as well as the analytes. It should not evaporate, otherwise the performance of the sensor will deteriorate. It forms the basis of selectivity towards a target analyte to be detected by causing a specific chemical reaction, thus generating ions in the cell. The electrochemical sensor is affected minimally by changes in the pressure of the ambient. The sensor is affected by ambient temperature variation; hence, temperature error needs to be compensated. It is better to maintain the sensor working temperature as stable as possible.

1.8 Piezoelectric wearable sensors

The piezoelectric effect is a property of a particular class of materials. When a mechanical force is applied, an electric charge which is proportional to the force is generated. The effect is reversible, meaning that when an electric force is applied a mechanical vibration is induced. This property is due to the asymmetrical nature of the structure of the material. When force is applied, the centres of the positive and negative charges move with respect to each other, causing the formation of an electric dipole. A large number of naturally available or synthetically fabricated materials, such as quartz, zinc oxide (ZnO), lithium niobate, lead zirconate titanate, etc, show such effects. This property can be used for various transduction applications. The surface acoustic wave (SAW) microgravimetric sensor is based on the piezoelectric effect. SAWs are the phenomenon of propagating mechanical waves produced by electric force along a solid piezoelectric surface that is in contact with a medium of lower density such as air. The SAW sensor fabricated using a piezoelectric substrate consists of three parts: (i) the piezoelectric transmitter IDE, (ii) the transmission line with chemically selective sensing film, and (iii) the piezoelectric receiver IDE [18]. An electric oscillator circuit connected to the transmitter produces a mechanical wave of particular wavelength, which propagates along the transmission surface towards the receiver. During transmission, it interacts with the toxic species SUT deposited at the sensing film, modulates the mechanical wave and the modified wave is reconverted into electrical form. Since the receiving IDE is deposited on the same piezoelectric substrate, the mechanical vibration is converted into an electric signal. The electrical signal at the receiver is different from the electrical signal applied to the transmitter IDE. The schematic of the SAW sensor is shown in figure 1.10. It is desired that the mechanical wave at the transmitter moves towards the receiver, but it moves towards the left as well. The absorber is used to arrest the mechanical wave moving towards the left. A similar absorber is placed towards the edge of the receiver. Often, there is
another reference SAW sensor without sensing film, whose signal is subtracted from the actual sensor to minimize the ambient temperature error as well as drift. At initial condition without SUT, both the reference and the actual SAW sensors operate at the same frequency, so the output of a mixer signal conditioning circuit is zero. The configuration of the IDE of both transmitter and receiver decides the phase, frequency, delay and the amplitude of the output signal with respect to the input signal. The piezoelectric substrate decides the transverse or longitudinal wave, the wave velocity and the temperature dependence of the output signal. The change in frequency ($\Delta f$) of the SAW device due to change in chemical species mass ($\Delta m$) of the SUT is given by the well-known Sauerbrey equation [19]:

$$\Delta f = -2\Delta mf_0^2 / A(\mu\rho)^{1/2},$$  \hspace{1cm} (1.10)

where $f_0$ is the fundamental resonant frequency, $A$ is the area of the electrode, and $\mu$ and $\rho$ are the shear modulus and the density of quartz, respectively.

However, the capacitance and the conductance of the SUT also change the frequency of the SAW sensor, but the frequency change due to mass is more significant. The sensing film in the mass loading area depends on the target chemical species to be detected. To improve the sensitivity as well as the selectivity, the mass loading area is coated with target analyte selective polymer film. A nanostructure of the metal oxide sensing film with suitable pore morphology is desirable in certain applications to improve the performance of the sensor. SUT has a large number of chemical species; it is almost impossible to fabricate a single SAW device to detect different parameters present in the sample. The SAW-based electronic nose (e-nose) with different SAW sensors, in the form of an array with different sensitivities and selectivities, is suitable to detect multiple chemical species in SUT. An e-nose is a system that uses the pattern of responses using a pattern recognition engine from an array of SAW sensors to identify the desired chemical sample and then quantify its concentration. The SAW e-nose is used to detect a wide range of chemical parameters, including toxic chemical warfare agents such as dimethyl methyl phosphonate (DMMP), mammal health parameters, environmental parameters,
beverages and humidity. The important features of SAW sensors are their low-noise operation, low detection limit, IC planar fabrication technology, mass producibility, low cost, small size, robusticity, reproducibility, reliability, fast response time and the fact that they can be placed in confined and inaccessible places. They operate in harsh environments, such as explosive, corrosive and radiated environments, and are hermetically sealed, so there is no effect from environmental conditions.

1.9 Fabrication of wearable sensors

1.9.1 Substrate selection

Wearable sensors can be fabricated using thin film or thick film technology. Important parts required for the fabrication of the sensors are: (i) the substrate, (ii) fabrication of electrode patterns on the substrate and (iii) deposition of sensing film. Sensing film can be deposited either on the electrode pattern or below the pattern. These micro-sized sensors can reduce the size of conventional analytical clinical laboratory and can fulfil the patient’s needs of convenience, comfort, small size, simplicity of operation, flexibility and, most importantly, timely availability of physiological parameters. Instead of depending fully on doctors to know their health status, patients know their physical conditions. Substrates play an important role in terms of wearer comfort and convenience. A flexible, thin, chemically inert, biocompatible, mechanically strong (to avoid wear-and-tear) and low-cost substrate is always desirable for the fabrication of the sensors. Ambient temperature, humidity and toxic gases present in the environment can affect the performance of the sensors and these parameters affect to varying degrees the performance of the substrate. Therefore every possible care should be taken to select materials to minimize errors due to environmental factors. A hydrophobic substrate such as Teflon is a suitable choice to avoid humidity errors.

Polyimide substrate. Polyimide is a favourable candidate for the fabrication of flexible microelectronic devices, including sensors, tactile force sensors for artificial skin and printable circuit boards [20]. This substance is mechanically strong, chemically inert and can withstand large working temperatures up to 350 °C. However, it is hydrophilic and absorbs moisture up to 3% relative humidity of its dry weight. This property gives the sensors cross-sensitivity to humidity in the intended applications. There are several manufacturers of polyimide substrate, popular ones include Dupont (USA), Upilex (Japan) and Polyflon (USA) [16]. Another important flexible substrate is Teflon. When the sensor needs a minimum cross-sensitivity for humidity and is required to operate at relatively high working temperatures, then Teflon is an excellent substrate material. A hybrid polyimide and Teflon substrate exploiting the advantages of both is also available on the market (Dupont). A flexible substrate can also be made using textiles. These fabric substrates, with different physical and chemical properties, are manufactured from natural wool and cotton, and synthetic nylon and polyester. These fabrics can be used to fabricate printable electrode wearable sensors, which provide stable operation over an extended period under normal and heavy wear-and-tear conditions. Many of these
substrates are available with a metal coating on a single side or on both. Thus there is no need for metallization of the electrode for the fabrication of electrode patterns. Different substrates used for the fabrication of wearable sensors include polyethylene terephthalate (PET), polyethylene naphthalate (NET), polyimide polypropylene, poly urethane (PU), cotton yarn, polyester, parylene skin patches, mouth guards, dentures and paper [20, 21]. Another important class of substrates which has drawn the attention of engineers is soft silicon elastomers such as polydimethylsiloxane (PDMS), silicon rubber. These materials have a high degree of flexibility, compatibility to different surfaces, and are chemically inert and biocompatible. However, most of these substrates are not CMOS compatible. Therefore, an integrated sensor-on-chip for smart sensing cannot be fabricated using these substrates. Other non-flexible substrates popular in sensor fabrication are alumina, fluorine-doped tin oxide (FTO), indium tin oxide (ITO), piezoelectric materials and silicon.

1.9.2 Substrate pre-processing

Pre-processing is an important requirement for the success of sensor fabrication. To laminate the electrode and the sensing film, the substrate must be properly cleaned. Unclean particles may reduce the surface energy of the substrate for proper adherence of the film. A polymer substrate has low surface energy; the same methods and chemicals which are used for cleaning inorganic substrates such as alumina, glass and silicon may not be suitable for polyimide or other substrate materials. Organic and inorganic contaminants may be present in the form of particles, thin films and island peak. Normally, wet and dry cleaning methods can be used to clean the substrate. Wet cleaning methods using chemicals remove the metallic and organic impurities. Solutions for cleaning include dilute hydrochloric acid, sulphuric acid, acetone, alcohol (ethanol, methanol, propanol, etc) and deionized water of superior quality. For inorganic substrates, the acid-based piranha solution with a 3:1 ratio of H$_2$SO$_4$:H$_2$O$_2$ is a popular choice. The reader should see [22] for different methods for cleaning substrates. A UV cleaner which has a UV light fixed in a stainless steel chamber with a closing lid is most often used to remove organic impurities. For some metal-coated polyimide substrates, metal particles such as nickel are strongly adhered to the polyimide surface. Plasma dry etching can be employed to remove such particles. A radio-frequency plasma at a pressure below 0.01 Pa is generated by ionizing inert gases such as argon gas, trifluoromethane, etc. This ionized gas will strike the surface of the substrate with very high kinetic energy, removing any particles on the surface. Immediately after plasma cleaning, the sensing film is to be deposited. A commercial low-pressure plasma cleaner (model PDC-32G-2) manufactured by Harrock Plasma Ins., USA, can be employed for plasma cleaning.

1.9.3 Fabrication of electrodes

1.9.3.1 Screen printing method

Various fabrication techniques can be adopted to form various patterns of electrodes on the substrate for the interdigitated impedance, electrochemical sensors and other...
physical sensors. These methods are the screen printed electrode, conductive inkjet printed electrode and photolithographic electrode. These printable sensors can be fabricated at a mass scale, with robust and attractive responses at low cost. The electrode can be made either manually or using an automatic screen printer with a conductive metal paste. The paste/ink that is employed for printing an electrode may be modified and tailored to provide selective detection or to improve the catalytic behaviour. Dispersion of enzymes into the ink can enhance the selectivity for biological species. For electrode fabrication, first the electrode configuration should be designed and optimized by choosing suitable dimensions of the electrodes according to the application needs with the help of dedicated software, such as Ledit, Autocad, ANSYS or PCB making, etc. A screen stencil of the design will be prepared on laser cut stainless steel or a chemically etched polymers mesh sheet. With the help of the screen frame, the pattern on the electrode will be transferred to the substrate (flexible/rigid) using a thick film screen printer. The flexible or textile substrate is fixed on the substrate. Metal paste (silver, silver chloride, silver palladium paste, gold, platinum or aluminium) can be employed for the fabrication. The operator adjusts the height and stroke of the embedded squeeze. However, the screen printing method of electrode fabrication is only suitable for a flat and uniform surface.

1.9.3.2 Electrode fabrication by elastomeric stamp
Fabrication of the sensor for a non-planar substrate requires an alternative approach that enables the fabrication of the electrode according to the surface contours and non-uniformity. There are various applications of wearable sensors that need electrode formation on a non-uniform surface. For example, most places on the human body are non-uniform. In such situations, elastomeric stamps (like rubber stamps) with the desired pattern of electrode and insulating layer are very useful [6]. These stamps can be used to form an electrode pattern on an irregular surface such as mammal skin. Such stamps can be used to fabricate electrochemical/impedance-type sensors using textile fabric as a substrate. Some research groups have developed elastomeric stamps which can directly print the electrode pattern on the human epidermis. In this method, the stamp uses conductive ink with suitable viscosity to print the electrode of a wearable sensor on the skin [6]. A new design of epidermis sensor is the fabrication of tattoo sensors, which involves printing the electrode/insulating layer on commercial tattoo paper with selective enzyme sensing film. Such tattoo-based electrodes can work as a biofuel cell to non-invasively harness electrical energy from the lactate present in human sweat. This energy can be used to provide electrical energy to the wearable sensors.

1.9.3.3 Ink-jet printing technique
Printed electronics, where the sensing devices are integrated with electronic circuits for wearable applications, can be fabricated using the ink-jet printing method in a printed form on PCB or paper [21]. These printed electronic devices are easy to handle and portable, and have found significance in recent years in different applications, including toxic vapour detection, food and beverage quality, and physiological parameter monitoring, etc. This technology involves the deposition of
electrode/sensing film ink by printing equipment in the form of jets of fluids. A typical application was to develop an enzyme-functionalized nanoparticle sensor on a paper substrate to detect the presence of bacteria. In the fabrication, gold nanoparticles were functionalized with a particular enzyme, and then ink-jet printing was used to print the material on the paper substrate. For example, β-galactosidase and gold nanoparticles formed a sensing material and the chlorophenol red β-galactopyranoside was used as a substrate. In the presence of bacteria, the colour of the sensor changes from pale yellow to purple.

Metal nanoparticles such as gold, silver, copper, etc, currently play an important role in the fabrication of functional electrochemical/impedance sensors. Different methods can be used for their synthesis. In one of these methods, silver nanoparticles were synthesized by stirring a mixture of silver acetate and oleic acid at 80 °C. Then tin acetate was added to the mixture and was heated at 120 °C. The resultant solution was then mixed with acetone/methanol and particles were precipitated [21].

1.9.3.4 Electrode formation by vapour deposition technique
In this case, the microelectrode of proper geometry can be formed by thermal evaporation or sputtering of different metals such as Al, Au, Pt, Ag and NiCr. For example, to fabricate IDE sensors on a silicon wafer, initially the wafer is properly cleaned to remove impurities and contaminants using the standard methods as discussed above. Metal film is deposited on the substrate by any one of the metal vapour deposition techniques. Then a thin film of polymer photoresist is deposited on the electrode, and the design of the IDE is transferred onto the wafer with the help of a photolithography mask under UV exposure. The unmasked area is removed by dipping the patterned substrate in etching solution (wet or dry). The wafer is then cleaned in ethanol and dried in dry nitrogen gas.

1.10 Deposition of sensing film on the electrode
A bare electrode pattern without sensing film on the substrate for the IDE impedance or electrochemical sensors may work to detect some body parameters. However, sensing film is always desirable to detect specific biological species in order to improve the sensitivity and the selectivity of the sensor. The catalytic electrode enhances the sensitivity as well as selectivity. One can use sensing film of polymer, enzyme or metal oxide. The selection of the sensing material to fulfil the application needs, such as selective target physiological parameters and stability to the working environment, is important. Since most wearable sensors and SUTs are affected by the variation of ambient temperature and humidity, precautions are necessary to select the materials which are thermally and chemically stable and hydrophobic. The effects of these parameters are to be studied carefully and the errors due to them compensated by analog/digital/software-based techniques. To improve the sensitivity and the selectivity of the sensor, a sample selective nanostructured film can be deposited. The nanostructure film in the form of nanopores, nanowires, nanorods and nanofibers provides a very high effective surface area and helps produce better functioning than in the bulk material. The nanostructure material has a rough
surface, which helps proper adhesion of the SUT. However, the morphology of such a surface area, the thickness of the film, average pore size and pore distribution must be optimized to enhance sensor performance. Different methods can be adopted to fabricate nanostructure film, such as electrochemical anodization, sol–gel, sputtering, etc. Electrochemical anodization and sol–gel are the low-cost simple chemical routes. These two methods are extensively used to prepare nanostructures of the inorganic materials, such as zirconium titanate, magnesium chromate titanate, alumina, zinc-oxide, titanium oxide, graphene oxide (GO) and many others [20]. The fabrication of nanowires of alumina involves (i) sonication of the cleaned anodized aluminium oxide template (AAO), (ii) electropolishing in the presence of electrochemical solution, (iii) anodization (potentiostatic/galvanic), (iv) etching with a suitable chemical solution, (v) nanopore opening, (vi) sputtering, (vii) electrode deposition and (viii) release of nanowires. All the steps are to be performed carefully in a clean room. The important parameters are current density, concentration of electrolyte, anodization time and the resistivity of the base material. Anodization is normally performed in a single- or double-pond Teflon cell with electrolyte using an inert cathode electrode such as platinum or gold. The electrode is in the form of a mesh or solid. The anode electrode material may be of copper, brass or graphite. These parameters may vary from material to material. For example, for the fabrication of porous alumina from pure aluminium tape, the electrolyte is dilute sulphuric acid or oxalic acid, but for the fabrication of porous silicon, the electrolyte solution is a mixture of hydrogen fluoride acid (HF) and ethanol/methanol solution. The sol–gel method of alumina nanoporous structure fabrication involves (i) the hydrolysis of the mixture solution of water (excess) and aluminium alkoxide, (ii) peptization in the presence of a minute amount of hydrochloric acid (HCl) or nitric acid (HNO₃), (iii) refluxing the solution to remove the volatile organic impurities, (iv) the addition of a binder such as polyvinylchloride and (v) deposition of the film on the glass or alumina substrate (alpha). Different phases of alumina such as gamma (γ), beta (β) and very stable alpha (α) can be prepared by sintering alumina film at different temperatures. The pore morphology of the materials can be controlled by the sol formation parameters as well as the sintering conditions. The important features of the alumina ceramic sensor are high sensitivity, low response time, low hysteresis, high reproducibility, high stability due to temperature, the possibility of batch fabrication at low cost, and the possibility of an integrated sensor with an integrated signal conditioning unit [14].

1.10.1 Methods of thin sensing film deposition

The sensing film, which should be physically or chemically bound to the sensor surface, may be a solid adsorbent, a chemical reagent, or a sorptive liquid or polymer [19]. The film acts as a biofluid sensitive and selective element that immobilizes a finite mass of some fluids from the environment. Resultant changes in physical and/or chemical properties of the coating film in turn produce electrical signal. Stimulation of electrical signal from interactions of the sensor with one or more of the physiological parameters constitutes the basis for detection and
quantification of the desired species. The thickness of the film should be uniform without cracks and it should be properly adhered to the surface. Three important coating methods are (i) the solution technique, (ii) vacuum deposition technique and (iii) vapour phase deposition technique. The solution method is perhaps the simplest and lowest cost coating method. It requires that the coating material be soluble in a solvent that does not chemically attack the device surface. Once the coating material is dissolved, the solution is deposited over the surface and the solvent is evaporated leaving the desired coating material. Popular solution methods are syringe deposition, painting with small brushes or Q-tips, dip coating, spraying coating, spin coating, electrospinning and ink-jet printing [6, 18–21]. Viscosity is an important parameter of the solution for film deposition. Spray coating is performed by spraying a dilute coating through an atomizing nozzle using a compressed-gas propellant (an inexpensive tool available at art supply stores, the air brush, is often utilized for this process) to the sensor surface. The fine, atomized mist of solution droplets stick and the solvent evaporates, thereby leaving a non-volatile coating. Like the syringe- and paint-brush-deposited films, the coatings formed by this procedure often have somewhat irregular texture and coverage, but uniform thickness is possible. Spin coating generally offers the highest degree of film uniformity and the greatest degree of film thickness reproducibility. A commercial spin coater holds the substrate on a motor-driven vacuum chuck. The vacuum chuck can spin at hundreds to several thousands of rpm, evaporating solvent and depositing the film. The uniformity and thickness reproducibility of the resulting film are often excellent. Film thickness is controlled by varying the spinning speed and the solution viscosity. Dip coating can be used to deposit thin films on large surface areas, but there is a possibility of non-uniformity which should be addressed. Another technique is drop coating, where the thin film is deposited by a few drops of solution.

1.11 Applications of wearable sensors

The applications of wearable physical and chemical sensors are numerous. There are a large number of research articles that report the fabrication of wearable sensors and their characterization of body parameter measurements [23]. A smart wearable garment integrating multiple sensors has been developed for monitoring different physical parameters, including electrocardiograph, electromyograph and movement, by printing sensing electrodes using an elastomeric stamp on a garment. Additional sensors embedded in the fabric help to record and monitor thoracic and abdominal signals related to respiration and movement [3]. Figure 1.11 shows the fabrication of different physical sensors on flexible and biocompatible substrates. Figure 1.11(a) shows a flexible and stretchable pressure sensor with high sensitivity fabricated on a PDMS substrate. The sensing layer of single-walled CNTs (SWCNTs) was sandwiched between two conductive composite elastomers comprising poly(3,4-ethylenedioxythiophene-poly(styrenesulfonate)) (PEDOT:PSS) PU-PEDOT layers. The sensor can easily be attached to any body part for the detection and monitoring of skin strains and muscle movement. The multiparametric all-carbon skin sensor shown in figure 1.11(b) was reported to detect multiple
physical parameters such as tactility, humidity, temperature and some physiological parameters [4]. Panel (b) shows a schematic diagram and photograph of the sensor. The working of the sensor was based on piezocapacitance, the structure of which consisted of CNT microyarn circuitry and a stretchable elastomer dielectric on PDMS substrate. Figure 1.11(c) shows a pressure sensor fabricated by sandwiching a gold nanowire impregnated tissue paper between a blank PDMS substrate and interdigitated electrode pattern substrate. Figure 1.11(d) shows a soft, flexible, breathable skin sensor for temperature measurements on a PU substrate using gold sensing film. The sensor was hydrophobic with excellent air and water vapour permeability, but was impermeable to water droplets and bacteria. Microfluidics-based wearable sensing technology has progressed rapidly in recent times due to their high sensitivity, adaptability, minute sample volume, low-power requirement and low fabrication cost. Compared to solid state sensors, liquid state sensors are
more attractive for wearable applications because of the flexibility of the liquid sample to take any shape. Solid state sensors suffer from plastic deformation, delamination of sensing film and formation of cracks. One of the most commonly used liquid metals is the eutectic gallium indium (eGaIn), the conductivity of which is similar to copper. It is also non-toxic. Figure 1.12 shows some different sensors fabricated using liquid active elements. Figure 1.12(a) shows a liquid-state pressure sensor with electrofluidic circuits fabricated using ionic liquid-filled microfluidic channels. The sensor consisted of a thin PDMS membrane sandwiched between a bottom microfluidic channel layer and a top electrofluidic circuit layer. Figure 1.12(b) shows a soft strain sensor fabricated using two liquid samples of different resistivity for measuring the strain of prosthetic hand movement. The high-resistivity liquid solution of NaCl and glycerol works as an active element and very low resistivity eGaIn works as a soft conductive wire. Figure 1.12(c) shows a liquid-ion-based microdroplet array for flexible tactile sensing. The sensor is based on a droplet enabled interfacial capacitive sensing mechanism. Each sensing element consists of a nanolitre droplet resting between two layers of flexible polymer membranes with patterned transparent electrodes [4].

Figure 1.13 shows an artificial electronic skin sensor array capable of detecting three-axis tactile and slip/friction forces as well as temperature variation. Due to changes in force and temperature, the resistance of the sensor arrays changes. A device with $3 \times 3$ arrays of fingerprint-like structures sandwiched between the strain and temperature sensors on polyester film substrate was fabricated using the screen printing method. Each unit of arrays consisted of four strain sensors and one
temperature sensor. A miniaturized multiparametric capacitive sensing platform on a flexible PDMS substrate was developed using functionalized multi-walled CNTs. The sensor was employed for breadth analysis along with limb movement by placing the sensor on the skin [24]. For chemical parameter sensing, a non-invasive mouth guard amperometric sensor has been fabricated with wireless data transmission capability to measure and monitor the lactate content of human saliva [2]. The sensor has been fabricated on a PET flexible substrate with polymer sensing film deposited on carbon metal electrodes. A resistive tattoo sensor fabricated on a graphene modified silk tattoo substrate has been developed by bio-functionalizing antibacterial peptides for wireless monitoring of bacteria. The sensor has excellent sensitivity and selectivity and a fast response time.

A resistive dental wireless tattoo sensor utilizing saliva fluids for the detection of *S. aureus* bacteria has been developed in [2]. This sensor was fabricated on an IDE structure by bio-functionalizing antibacterial peptides on a graphene modified silk substrate and showed excellent sensitivity and selectivity. The recent development of

![Image](image_url)

**Figure 1.13.** Artificial electronic skin with four strain sensors and one temperature sensor: (a) schematic view of the sensor, (b) photograph of the fabricated sensor with an enlarged view, (c) photograph and schematic showing the two-dimensional force and temperature mapping capability of the sensor in response to external stimuli, such as finger touch [4].
a synthetic soft tissue artificial retina that closely matches the natural retina in its functionality can replace the mostly rigid hard-material-based retina. Since the human eye is sensitive, hard-material-based artificial retinas can damage the eye, leading to inflammation and/or scarring. The newly designed soft synthetic-material-based retina has hydrogels and biological cell proteins. These cell proteins work as a tiny pixel of a camera detecting and reacting to light to create a greyscale image of an object. Over the years, several excellent research articles have been published in various journals on wearable physiological parameter measurements. Some of the papers are cited in this chapter. The reader should see these articles for further reading.

1.12 Conclusions

This chapter reviewed the literature on wearable sensing technologies for online health monitoring. Key parameters of healthcare are identified. Today, this is an important area of research and a lot of commercial products are already available for monitoring certain body parameters and many are expected to arrive on the market in the near future. The importance of non-invasive chemical parameter measurement using wearable impedance and electrochemical sensors is discussed. Issues related to fabrication on the basis of working principle, electrode material, electrode structure, substrate material, flexibility and deposition of sensing film for safe, durable, sensitive and selective detection of chemical species are also discussed. New wearable sensors are desired to be developed for measuring some of the body parameters which are yet to be investigated. Furthermore, wireless-device-integrated wearable sensors are needed to enable users to wirelessly transmit data to their smartphone in a more user-friendly manner.

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