ORGANIC BIOCOMPATIBLE NANOLAYERED POLYMERIZATION OF SOLID-STATE DEVICES

by

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ABSTRACT

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Nanotechnology plays a major role in today's society due to its convergence of nanoscale which is the level of atoms and molecules as a part of miniaturization trend. The interference between biomedical and nanotechnology are of intense research. It concerns about the utilization of miniature biological systems such as nucleic acids, proteins, cells and cellular components to fabricate functional organic and inorganic nanostructures.

This research work focuses on the process of developing a simple method to obtain fluorinated organic nanocoating which can be used to coat 3D structures and also details the extended applications of the coating in the fields of biotechnology and medicine. These nanocoatings are made using two simple non-toxic chemistry and materials having different chemical and physical properties; 3-Aminopropyltrimethoxysilane (APTMS) and 1H, 1H, 2H, 2H-Perfluorooctyl-trichlorosilane (PFTS). The phenomenon is demonstrated using a simple vapor-phase approach which allows the monomers react in gaseous state and directly form the nanofilm. The main objective behind this work is to develop a simple time efficient and cost efficient method which helps in the deposition of nanolayers, over the pre-existing techniques which involve harsh chemical and plasma treatment such as Plasma polymerization, Chemical

Vapor Deposition (CVD), biomineralization and Self Assembled Monolayers (SAM). Further, detailed research demonstrates about the two important goals of such coatings; biostability and biocompatibility; especially for the surfaces of medical implants. These properties are attained by modifying the surface characteristics of the substrates. With the help of chemical characterization and spectroscopic analysis, these nanocoatings were proved to be biostable and bio-compatible organic porous nanofilm making it applicable in biochemical/medical areas.

Further research involves the formation of medicated nanocoatings which could be used to coat bio-implants which needs to have medicated surface.

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CHAPTER 1

INTRODUCTION

Nanobiotechnology refers to a field which uses nanoscale principles and techniques to understand and create new devices using biological principles. It is an interface between the biotechnology and nanoscience; two interdisciplinary areas which combine advances of science and engineering [1, 2]. New approaches on nanodevices provide high sensitivity and specificity at the very basic level which involves in the modification of their chemical and physical properties [3, 4]. The fabrication of the primary materials for any biomedical system is done with the help of biological principles and the nanoscale principles helps in the modification of the physical and chemical properties which further show desired outputs at macro scale. Integration of nanotechnology with biomedical micro/nanoelectromechanical systems (Bio-MEMS/NEMS) offers tremendous potential to tackle medical problems in the areas of diagnostics, therapy, surgical implants and drug delivery [5, 6].

Surface modification of MEMS/NEMS has become one of the most desirable aspects in medical related devices. With the help of current advancements, to develop a micro/nano scale device is of ease but the challenging aspect is to modify its surface and chemical characteristics according to the needs in the fields of medicine. Fluorinated coatings have been useful in many applications in the fields of biochemistry and tissue engineering. These coatings help to attain low surface energy and corrosion resistance properties in nano/micro structured devices. These fluorinated organic coatings can be attained using many pre-determined techniques which are discussed in later chapters but the most important criteria is to obtain a biologically stable and compatible film which can be attained by modifying the surface characteristics [7-9].

1.1 Objective

The major objective of this work is to design and fabricate an organic porous biocompatible nanolayered film without harsh chemical or plasma treatments with better bio-sensitivity and low cost approach than current available nanolayer deposition methods. Along with the deposition of the nanolayer, it is important to have low surface energy coatings to make them applicable in medicine. Therefore, fluorinated organic porous nanofilms are to be obtained with most efficient factors that would promote biocompatibility, biostability/corrosion resistance and also support cell/protein adhesion.

1.2 Synopsis of Research Work

Nanotextured fluorinated polymer surfaces play major role in barrier properties of medical implants. The vapor-phase reactions of hydrophilic and hydrophobic layers bead up on exposure to obtain low surface energy. The common methods involved in the deposition of fluorinated polymers on silicon substrates include chemical vapor deposition, plasma polymerization spin coating and self assembled monolayer formation. These methods provide good surface deposition but the control on surface properties of the films is critical and these methods require special equipment and are cost prohibitive in large scales.

To overcome these issues, a novel technique was introduced which uses vacuum reaction chamber to deposit the organic nanocoatings [10]. Biological behavior of the cells and proteins were studied which showed better protein adsorption on lower surface energized films. The chemical/physical properties analyzing the nanocoating application in the fields of medical and engineering are shown in S D Vidyala et al. [11]. This research work details about the nanocoating depositions using hydrophilic-hydrophobic polymers which are allowed to react in vapor-phase to deposit fluorinated organic nanolayers. The process takes place in a simple vacuum reaction chamber. Chemical characterization is one of the most important factors to analyze the chemical properties of the layer which determines the deciding factors and the applications of the obtained layer in the field of large scale medical devices. The chemical

characterization of the deposited film was done using Fourier Transform Infrared Spectroscopy (FTIR) and X-ray Photoelectron Spectroscopy (XPS). Films were made with different concentrations of hydrophilic and hydrophobic polymers and the chemical composition of each layer was analyzed.

The chemical stability of the films was also tested using different pH solutions. The silicon substrates coated with these thin films were immersed in the various pH solution baths and were left for 15 hours. Later the coated silicon substrates were taken out and the surface morphology was analyzed using Scanning Electron Microscope (SEM). It was seen that the film was capable of surviving various chemical surroundings. These layers were also used to coat a 3D MEMS structure to study the coverage performance in coating such devices.

1.3 Structure of Thesis

The content of the thesis spans around the design and fabrication of organic porous nanofilm which shows biologically stable and bio-compatible behavior. The changes in the depositiong process change the chemical and physical behavior of the primary materials making the resulting film to be applicable in vast areas of biochemistry and medicine. The breakdown of the chapters is given below:

Chapter 1: This chapter is meant to introduce the reader to the drive and objective behind the entire research work. It also explains the benefits of using nanotechnology in medicine.

Chapter 2: This chapter provides background and literature review covering current nanolayer deposition methods; Chemical Vapor Deposition (CVD), Plasma polymerization, Self Assembled Monolayers (SAM) and Biomineralization. This chapter also explains their advantages and limitations.

Chapter 3: This chapter expands on the approach adopted to fabricate the nanofilm in vapor phase using two monomers; *3*-Aminopropyltrimethoxysilane (APTMS) and *1H, 1H, 2H, 2H*- Perfluorooctyl-trichlorosilane (PFTS) which have different chemical and

physical properties. The outline of the experiments spans characterizing main parameters involved in the process, namely:

i. Ratio of Concentration of APTMS and PFTS

ii. Deposition Time

iii. Vacuum Pressure

Chapter 4: The chapter 4 describes chemical and physical characterization of the nanolayer specially: Fourier Transform Infrared Spectroscopy (FTIR) and X-ray Photon Spectroscopy (XPS) analysis are described in this chapter. It also details the stability of the coating in various chemical surroundings. Surface morphology, thickness in correspondence to the deposition time and concentration of the monomers and surface energy properties involving mechanical strength and stability are also explored in detail. It concludes the thesis and details the applications of the nanolayers in the fields of biotechnology and medicine and describes the coating of 3D Structures/BioMEMS.

Chapter 5: This chapter describes some future directions that can be spanned out of this work. It includes the scope of more work that could supplement/complement this work.

CHAPTER 2

BACKGROUND AND LITERATURE REVIEW

2.1 Medical Device Coatings

Modern technologies have always been a major driving force in medical device research. Due to their high economical values, biomaterials are incorporated into new products at an early stage of development/invention. Considering these factors, nanotechnology has iconic influence on medical device technology. Integration of nanotechnology with medical device technology can result in device size reduction and produce new properties at the very basic levels of atoms and molecules. This enabling technology involves chemists, physicists, biologists, engineers and physicians and many more areas of specialists [12, 13]. New problems and innovative solutions have been brought to attention with the confluence of these technologies, one of which is the coating of implants. The coating materials which were originally developed to improve outer surface resistance are now being used in biomedical body implants for example in stents [1, 13-15]. In past few decades, various medical conditions are being treated with medical device implants into the human body. When such a device is introduced, the patient is placed at high risk of a variety of complications including heart stroke. These problems were solved with technical advancements in implantable medical devices that provide controlled release of an agent, drug or bioactive material into the body where a stent of other medical device is positioned. This agent/bioactive material can be degraded during the application. Researchers have discovered that this loss of degradation of an agent/bioactive material can be avoided by coating them with porous layers of biocompatible polymers that are applied without the use of solvents, heat or other chemicals that would degrade or damage the

agent/bioactive device. These biocompatible polymers can be applied by vapor deposition/plasma deposition [14, 16].

2.1.1 Bio-Coatings

Biomaterials and tissue engineering technologies are becoming highly important in biomedical practice with high pace of advancements in medical appliances. Cellular responses depend on topographical properties of the biomaterial at the nanometer scale. Structures on biomaterial surfaces are used as powerful tools to influence/control interactions among implants and biological systems. The influence of the surface structure of a biomaterial on the biological system includes from the initial steps of protein interactions on the biomaterial surfaces to the final steps like cell orientation which involves cell type specific reactions to nanostructured surfaces [17, 18].

Implants are basically of two types; Temporary implants; needles, catheters, etc and Permanent implants; stents, dental implants, joint implants, etc. Every medical implants needs to be coated majorly to modify their surface characteristics in order to attain biocompatibility and biostability. These are further coated with required growth enhancing materials such as hydroxylapatite which helps in bone in-growth in prosthetic implants [19, 20]. Incorporation of medical implants into the body depends on the occurrence of the disease/emergence for the need of implanting a foreign material due to injuries [17, 21]. Considering these factors, it is important to develop new improved techniques to coat medical implant devices with required materials which could last longer reducing the number of incision times.

2.2 Micro/Nanoelectromechanical Systems in Medical Implants

MEMS and polymer based 3D devices have always been in major applications for local drug delivery. In order to use them as body implants, they need to fulfill certain criteria, the foremost of which is the biocompatibility [1, 22].

A Microelectromechanical system (MEMS) is a technology which involves very small mechanical devices with sizes ranging in micrometers. Further smaller dimensions of

objects/devices and modifications at nanoscale form Nanoelectromechanical systems (NEMS) and nanotechnology. These are basically a subset of MEMS devices for application in biomedical research and medical micro devices [23]. Polymer based medical implants for example; stents, dental implants, etc are also used with their primary polymeric 3D structure with modified surface parameters to attain biocompatibility and stability in chemical surroundings. Medical implants developed are usually not biocompatible until their surface properties are modified to low surface energy, selectivity and corrosion resistance [16, 22]. Therefore, in order to attain such properties, the micro/nano structures are often coated with various molecules which help in forming organic composite films. Highly hydrophobic surfaces can have very low surface energy and such low surface energy biological interfaces can be obtained using fluorinated coatings on surfaces. Deposition of such fluorinated organic films on solid-state devices can be achieved using many currently available techniques such as chemical vapor deposition [24], plasma polymerization [25], biomineralization [26], pulsed laser deposition [27], self assembled monolayers [28], etc. Some of the widely used important deposition techniques are explained below. The films are used in biosensing, photoluminescence, and also to import biocompatibility to the surfaces [29-31].

2.3 Current Organic Film Deposition Techniques

Following is the detailed explanation of the available deposition techniques, their advantages and limitations.

2.3.1 Chemical Vapor Deposition (CVD)

Chemical vapor deposition refers to the processes of deposition of thin-films with the help of molecular gases and vapors rather than solid evaporants or sputter targets as their source materials [32]. CVD involves passage of precursor gas/gases into a chamber which holds objects that are to be coated. The temperature inside the chamber is maintained high to certain degree depending on the type of object present inside the chamber. Chemical reactions occur near the hot surfaces of the chamber resulting in the deposition of the thin film on the

surface of the object. During this process, chemical by-products are formed that are later allowed to exhaust out of the chamber along with unreacted precursor gases (figure 2.1). This process is categorized depending on the various factors; Type of Application, Process/Reactor needed and Precursor and chemical reactions used [33, 34].

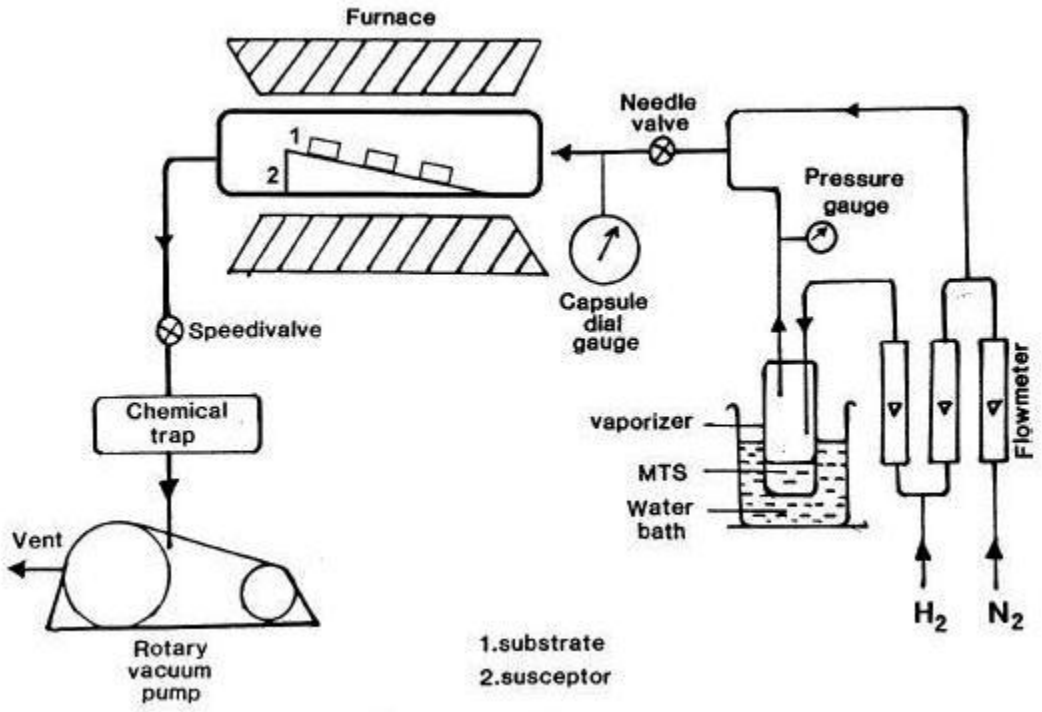

Figure 2.1 Schematic diagram of typical laboratory CVD [35]

The fundamental processes of CVD involve in the mass transport of electrons, thermodynamics and chemical kinetics. Depending on the type of application and materials to be deposited, there are many variants of CVD. Some help in the deposition using hot wall chambers, few involve the cold wall chambers and pressure controlled chambers. The choice of reactor is determined by the application based on the requirements of deposition substrate, material used for coating, resulting surface morphology, desired thickness of the film and attaining uniformity and the estimated cost of the whole process [36]. Most of the CVD

processes use hot wall chambers for the best results. In this, the chamber holding the object on which the thin-film needs to be deposited is surrounded by the hot furnace. The chamber is heated to the desired temperature and then the precursor gases are introduced into the chamber. This type of process helps in depositing the desired film on large batches of substrates and having uniform coating thickness [37].

Cold wall reactors are the next major options of CVD. Here, the substrates used for deposition are heated with constant cooling of the walls. These often run at high pressures and the reactive precursors are usually diluted in carrier gas. Cold wall reactors are mostly used in compound semiconductor CVD processes [36].

CVD has a number of advantages and also dis-advantages as a method for depositing thin films. Table 2.1 explores the advantages and disadvantages of this technique.

Table 2.1 Advantages and Disadvantages of Chemical Vapor Deposition

Chemical Vapor Deposition		
#	Advantages	Disadvantages
1	CVD films are deposited on objects having any shape. They coat the entire surface area including sidewalls and any curved surfaces. Quite conformal.	The precursor gases involved in the CVD process like Nickel Tetracarbonyl ($Ni(CO)_4$), Diborane (B_2H_6) and Silicon Tetrachloride ($SiCl_4$) are highly toxic, explosive and corrosive.
2	Wide variety of materials can be deposited	Films are usually deposited at high temperatures resulting in coating of limited materials which are resistant to higher temperatures than they are usually stored.
3	Purity in the deposited thin-film; as all the impurities are removed along with the unreacted gases being exhausted.	The byproducts of CVD reactions; Carbon Monoxide (CO), Hydrogen (H_2) or Hydrogen Fluoride (HF) are hazardous.
4	High deposition rates.	Precursors used for the deposition are quite costly.
5	Doesn't require high vacuum pressures.	Causes mechanical instabilities in the deposited films due to stress occurred through different thermal expansion coefficients.
6	Large batch of substrates can be coated at once using hot-wall process of CVD.	In bulk coating process, hot walls are used which get heavily coated resulting in frequent cleaning which involves higher energy usage.
7	Usage of cold walls results in less cleaning and less thermal load.	Cold wall process fails to deposit uniform thickness of the film all over the substrate.

This process uses relatively low temperature resulting in reduced deposition with less cleaning and lower usage of thermal loads. Here, the film deposited is not uniform and during some cases, it also fails to coat the deeply curved surfaces on the depositing substrate.

Most of the MEMS devices are fabricated from polysilicon films which are deposited on silicon substrates with intermediate SiO_2 layers with the help of plasma enhanced chemical vapor deposition (PECVD) polymerization method. In this technique, electrical power is used at persistent critical pressure in order to perform the chemical activation. As a result, plasma is discharged having equal concentrations of ions and electrons [34]. During this process, one can observe glow in plasma due to the pressures applied and this visible glow is caused due to charge in the excited atoms and molecules. With the help of these plasma electrons, the electrical power is coupled into gas which further ionizes very small amount of gas. The remaining gas is chemically activated by the electrons which results in dissociation of the molecules into smaller particles called radicals. Radicals are chemically unsaturated and therefore lead to chemical reactions at surface level and cause film formation [38]. This method involves three different processes; Chemical activation of the gas molecules, Transport of radicals to the substrates and Chemical reaction at the surface of the film.

Lighter electrons present in the plasma are easily accelerated by the electric field compared to the slow acceleration of heavier ions. Due to the variation in weights, electrons do not lose much of their energy unlike the ions and therefore these plasma electrons help in the accumulation of the energy in the electric field. Further, inelastic collisions occur making the electrons loose energy and at the same time low energized gas molecules gain energy resulting in the increase of chemical activity of the gas molecules. At typical pressures, a radical is created in the plasma which undergoes many collisions before it reaches to the surface of the film with the help of source gas. After a few collisions, the chemical reaction comes to an end between the radical and the source gas. There is a possibility of the molecule to grow continuously and become a macromolecule later to a particle. In this PECVD process, only the

degree of gas-phase reactions can be controlled but, film formation at the substrate competes with the undesirable tendency of the neutral radicals to react with each other in the gas phase. These gas-phase reactions are inherent to PECVD process and determine the properties of the film [34].

2.3.2 Plasma Polymerization

Plasma polymerization refers to the deposition of thin films showing a broad range of properties using different monomer gases under varying plasma conditions such as pressure, discharge power and temperature without any fabrication. It takes place in a low pressure and low temperature plasma that is produced by a glow discharge through organic gas [39, 40]. The elemental reactions occurring are due to the fragmentation of monomer molecules which occurs in plasma phase resulting in the formation of radicals which are also known as active sites. Cross linking during film growth determines the properties of the film ranging from soft highly functional coatings to hard crosslinked films [34, 38]. The growth of monomers having low molecular weight into polymers with high molecular weight molecules occurs with the assistance of the plasma energy which involves activated electrons, ions and radicals. However, the starting gaseous molecules that are fragmented into activated small fragments depends on the level of plasma and the nature of the starting molecules [41].

Plasma polymerized coatings are applied inside the vacuum chamber. A monomer gas is pumped into the vacuum chamber where it is polymerized in plasma phase to form a thin, clear coating. This deposition occurs when a glow discharge of plasma initiates the polymerization. The electrons which get excited during glow discharge help in ionizing monomer molecules. The electrons which get excited during glow discharge help in ionizing monomer molecules. These molecules break down into radicals [42]. These radicals further absorb, condense and polymerize on the substrate (figure 2.2). Crosslinking of these radical and electrons and ions create a chemical bond with the surface of the substrate to form thin coating (figure 2.3).

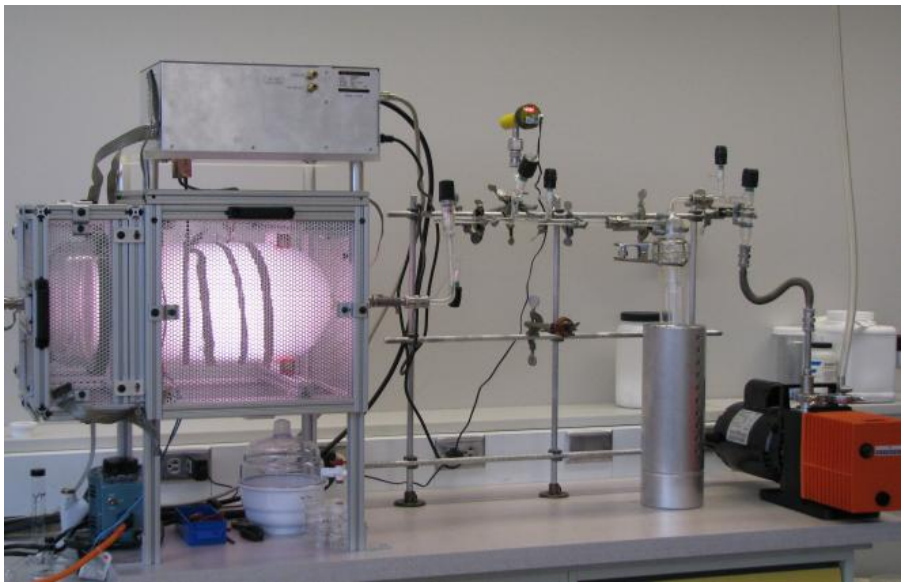

Figure 2.2 High Vacuum Pulsed Plasma Polymerization Chamber [Dhinojwala et al., 2010]

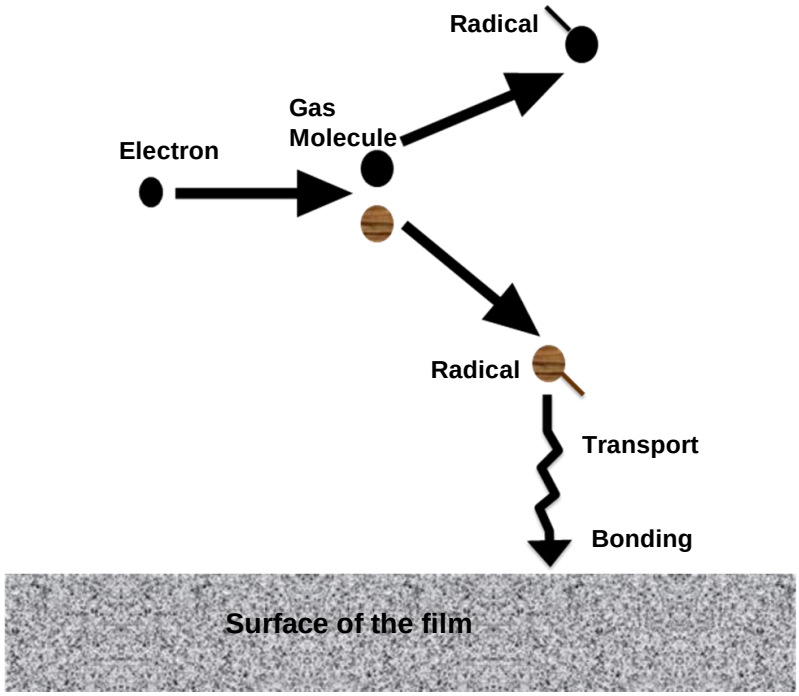

Figure 2.3 Plasma Polymerization process (Inspired from [34])

Table 2.2 details the advantages and disadvantages of using plasma polymerization technique for the deposition of organic films [42-45].

Table 2.2 Advantages and Disadvantages of Plasma Polymerization

Plasma Polymerization		
#	Advantages	Disadvantages
1	Polymeric films are made of organic compounds that do not polymerize under normal chemical polymerization conditions.	This process is very expensive.
2	Almost all types of monomers; organic compounds and saturated hydrocarbons can be polymerized using this technique.	Plasma polymerization process involves harsh chemical mechanisms and is very complex.
3	It coats the substrates potentially with less time compared to any other conventional coating process.	Surface morphology of the resulting polymer surface cannot be predicted due to its complexity.
4	Requires no solvent during the polymer preparation and no cleaning of the resultant polymer.	Good control over the chemical composition of the deposited surface is challenging.
5	The film deposited with this technique is smooth, non-porous and very dense.	Polymerized coatings have low abrasion resistance.
6	Thickness can be controlled depending on the requirement.	This process is limited to deposition of layer on single substrate unlike other techniques which can perform batch processing

2.3.3 Self Assembled Monolayers (SAMs)

Self assembled monolayer is an organized layer of chemical compounds on top of a substrate. These chemical compounds usually have a head group and a tail group (figure 2.4). The two-dimensional molecular organization is a key ingredient for SAM stability and function. Their primary function is to reduce surface tension which is attained by the adsorption which is driven by a chemical reaction of hydrophilic head groups on the substrate from vapor/liquid phase. The tail groups are organized by hydrophobic properties. These two groups are selected depending on the type of application of SAM [28, 46]. SAMs are formed when surfactant molecules spontaneously adsorb in a monomolecular layer on surface. Briefly, their formation involves two steps; adsorption and monolayer formation. The initial step of adsorption of the molecules and materials is a quick process and hence it is also termed as initial fast adsorption

and the monolayer formation; referred as second slower step of monolayer organization. Adsorption occurs due to interface between solid, liquid and vapor forms of materials while, the transport of molecules to the surface occur due to diffusion. These offer a unique combination of physical properties that allow fundamental studies of interfacial chemistry, interactions between molecules and solvents and self-organization. The final film provides a closely packed adsorbate molecules with relatively uniform molecular orientation and conformation. This simple function of formation of well organized arrays makes them ideal model systems in many fields. They provide the design flexibility both at the molecular and material levels and also provide unique opportunities to increase fundamental understanding of self organization, structure-property relationships and interfacial phenomena. The ability to tailor both head and tail groups makes SAMs excellent systems for applications including wetting, chemical resistant, solvent interactions like ordering and growth, adhesion, lubrication, biocompatibility, sensitization, corrosion and molecular recognition for sensors and nanofabrication [47].

2.3.3.1 SAMs and Organic Surfaces

SAMs are organic assemblies formed by the adsorption of molecular constituents from the solution/gas phase onto the surface of solids or in regular arrays on the surface of liquids which organize into crystals with high affinity. They provide key to modify the chemical properties of solid substrates that can be used for biosensing, friction, wetting purposes and also provide deformability of materials required in implantable biomaterials that require certain degree of mechanical flexibility. The structure of SAMs also depends on the curvature of the substrate.

SAMs themselves are nanostructures with a number of useful properties; the thickness of a SAM is 1-3 nm (organic thin film material). The molecular component of SAM determines the atomic composition of the SAM, which directs the organic synthesis to tailor organic and organometallic structures at the surface. SAMs can also be fabricated as patterns having 10-

100 nm dimensions. Therefore, application of SAMs in micro medical devices is basically of two categories; Thin Film SAMs and Patterned SAMs.

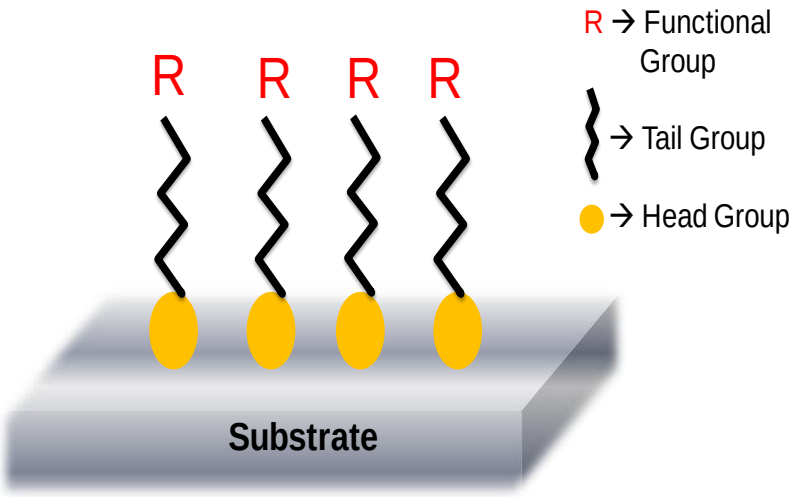

Figure 2.4 SAM structure (Inspired from [48])

Thin-film SAMs are of major use in the areas of biology, bio/electro chemistry, MEMS/NEMS. Membrane properties of cells and cell organelles can be studied with the help of thin-film SAMs. The surface modification of biosensors and also bio-implantable MEMS/NEMS devices can be achieved with the help of thin-film SAM coatings [31]. They help in the functionalization of the nanostructures which help in adhesion properties for cell/protein attachment. For example, magnetic nanoparticles are used to remove fungus from a blood stream with the help of SAMs coating. The contaminated blood is filtered through the coated MEMS device and the fungus in the contaminated blood is driven out with the help of the magnetic nanoparticles [47, 49]. The formation of thin-film SAM surfaces can be done in several ways out of which the most commonly used methods involve vapor-phase deposited SAMs and Silane SAMs. Vapor-phase depositions of SAM layers involve ultra high vacuum substrate cleanliness and in-situ surface characterization. The growth of Silane SAMs involves an irreversible covalent cross-linking step. The kinetics of this step causes implications on the

growth mechanisms and final film structure. Because of the molecular packaging determined by covalent siloxane network, formation of SAM is limited to certain range. These layers are highly sensitive to temperature, pH and abrasion. Table 2.3 details the advantages and disadvantages of using SAMs.

Table 2.3 Advantages and Disadvantages of Self Assembled Monolayers

SAM		
#	Advantages	Dis-advantages
1	It is relatively simple method which uses a beaker, solvent and molecules	Choice of substrate to deposit SAM is limited.
2	SAM layers have high density.	SAMs have low abrasion resistance.
3	Large variability of molecules can be used.	Defects in thermodynamics of SAM formation.
4	Design Flexibility.	Temperature sensitive.
5	Ability of surface modification of biosensors.	Polymerized coatings have low abrasion resistance.

To compete with the advantages and overcome the disadvantages of these above techniques, a new approach has been presented in this thesis work which uses a simple reaction chamber and hydrophilic/hydrophobic polymers to deposit nanotextured film. The deposition, chemical/physical characterization and applications of the nanolayer film are detailed in the next chapters.

CHAPTER 3

ORGANIC POROUS NANOFILM DEPOSITION

This chapter details the approach adopted to deposit a nanofilm having potential features of biocompatibility and biostability. The goal of the research work is to develop an inexpensive and time-efficient deposition method, with control on surface modification of the coatings useful for the applications in micro medical devices.

3.1 Introduction

From earlier chapters, it has been explained that the biocompatibility and biostability of a micro medical device such as biosensors or BioMEMS/NEMS can be achieved by modifying the surface characteristics of the substrates. Self-organized silane films having fluorinated surfaces have great structural stabilization which can be achieved from multiple covalent and hydrogen bonds [50]. These surfaces have been studied to form complete and uniform deposition and also modify the surface properties such as cell adhesion, protein adhesion, and growth of organic-inorganic hybrid alloys [51]. Biomaterials such as micro-fabricated immunoisolation bio-capsules and retinal implants require flat substrates [52].

The method involves the fabrication of the nanolayered in vapor-phase using two monomers; *3*-Aminopropyltrimethoxysilane (APTMS) and *1H,1H,2H,2H*- Perfluorooctyl-trichlorosilane (PFTS). APTMS and PFTS are two vastly used monomers for the deposition of organic films on silicon substrates. Silicon wafers have become, over years, basic substrates used before experimenting on any biomedical devices. The X-ray photoelectron spectroscopy (XPS) spectra results of APTMS on silica substrates prove the formation of acid-base bonding mechanisms due to the interactions between silanols and amino groups of APTMS [53, 54]. Covalent bonds between organic/inorganic components lead to the formation of a cross-linked

structure which is phase separated on micro/nano scale resulting in a macroscopically uniform material. These structures when modified with PFTS possess extremely super-amphiphobic properties. These fluorinated surfaces were characterized for their lower surface energies [55], reduced cell adhesion [56], storage of hybrid alloys and protein aggregation [57]. These can be used as functional coatings by simple coating process. This thesis work involves in the characterization of the polymer nanocoatings and coating was 3D microstructures using gas-phase reactions which succeed in coating of inner structures of a nano-textured surface which is idealistic compared to many other currently available thin-film deposition methods used in biomedical applications.

3.2 Materials Used

Deposition of the nanocoating on the solid-state substrates is done by using two monomers; APTMS/PFTS which are different in their chemical properties. APTMS is hydrophilic in nature and PFTS is a hydrophobic monomer. Hydrophilic polymers contain polar/charged functional groups which render them to solve in water. Most hydrophilic polymers are grouped by the chemistry of their structure. Amine functional polymers include allylamine, ethyleneimine, oxazoline and other polymers in their main/side chains [58]. A hydrophilic polymer molecule contains a large number of covalently bonded units whose chemical composition may be different but they would contain proteins, polyurethanes, acrylate copolymers, etc. In an aquesous solution, weakly cross-linked chains and isolated macromolecules are characterized by controlled polymer-polymer interactions and polymer-solvent interactions [59, 60]. Hydrophobic molecules tend to be non-polar and therefore prefer neutral and non-polar solvents. They exhibit high contact angles with lower surface energies. These substances are usually lipophilic but silicones and fluorocarbons are also considered to be hydrophobic [61].

3.2.1 3-Aminopropyltrimethoxysilane (APTMS)

APTMS is a hydrophilic polymer which is used as a silylation reagent for coating glass and silica surfaces to add primary amines. It is first coupled to glass or silica with the help of

silane and then the compounds of interest are coupled to the newly added amino groups either directly or by using additional chemistries. This polymer is hydrophilic in nature with a molecular weight of 179.29 g mol^{-1}. The molecular formula of APTMS is $C_6H_{17}NO_3Si$. Fig 3.1 shows the structure of the polymer. This polymer plays a major role in today's medical advancements and is used in many in- vitro and in- vivo diagnostics approaches. It helps in the formation of uniform films, antibody attachment, cell adhesion, DNA/protein attachment, etc [62]. With the help of this polymer, different types of structure can be obtained with slight modification of the reaction conditions such that lateral/vertical polymerization or the combination of both can accompany in the cell attachment. Parameters such as concentration of APTMS, temperature and reaction time are important for the final deposition. The film morphology obtained from this depends on the method of deposition of the layer. APTMS is non-toxic to cells and hence is used for cell growth and carries highly reactional functional groups [63].

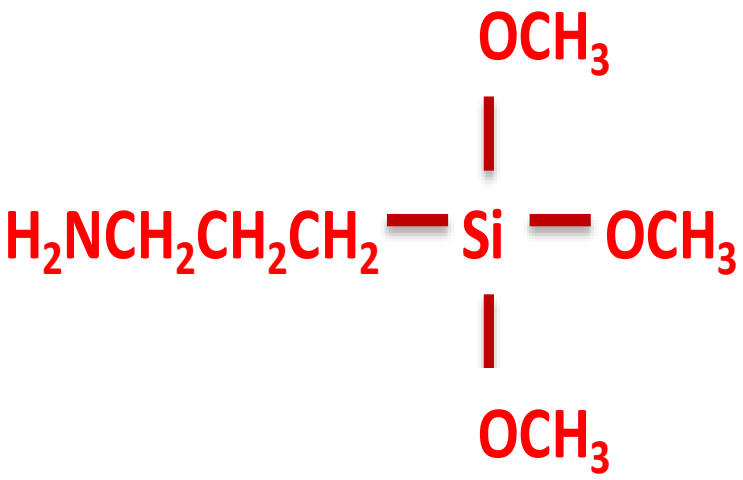

Figure 3.1 Structural formula of APTMS

The chemisorptions of molecules on silicon surfaces is one of the most important processes in current research which involves interaction between finite units and periodic substrates. Growth of homogeneous organic films onto microelectronic substrates such as silica

surfaces is one of the major ongoing research [51]. Amino-terminated surfaces form hydrophilic surface that can strongly bind to other materials [64, 65]. These promote mineral growth that can be useful for attaching and spreading of neurons, for antigen-antibody attachments, to improve cell adhesion, to enhance biocompatibility ad also to maintain bioactivity of enzymes on alloy surfaces. The methoxy groups of APTMS makes it one of the most useful monomers for the preparation of amino-terminated layers on silicon surfaces. It exhibits high coagulation activity which helps as a linker for various nanoparticles such as those made of gold [66, 67].

3.2.2. 1H, 1H, 2H, 2H- Perfluorooctyl-trichlorosilane (PFTS)

PFTS is a hydrophobic polymer which is used as a releasing agent for polydimethylsiloxane (PDMS). The molecular weight of PFTS is 481.54 g mol^{-1} and the molecular formula is $C_8H_4Cl_3F_{13}Si$. Fig 3.2 shows the structure of PFTS. It is non-toxic and is used mostly to generate fluorine rich coatings which show specific properties such as low surface energy, chemical inertness and thermal resistance [63].

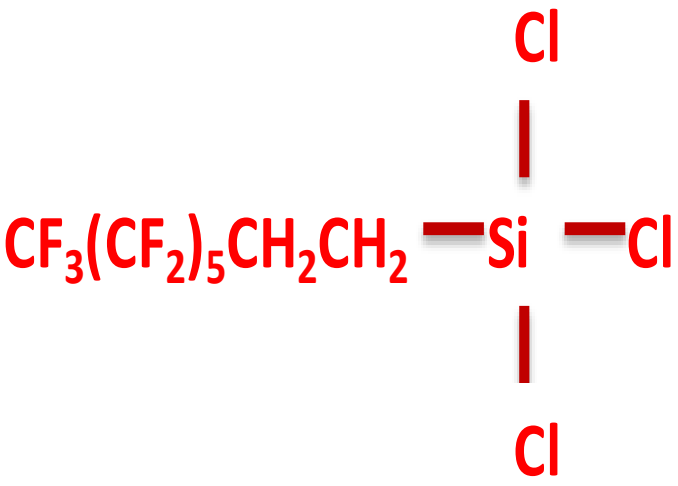

Figure 3.2 Structural formula of PFTS

Due to these properties, the polymer shows relatively low level of coagulation activity.

Surface hydrophobicity is defined as the interplay between the surface roughness and its chemical composition. There are several methods which are used to fabricate artificial hydrophobic surfaces such as lithographic patterning [68, 69] and electrodeposition [70]. With the use of PFTS, the surfaces formed show contact angle higher than 130^0 and often a low hysteresis.

3.3 Experimental Procedures

In this work, silicon wafers were used as solid substrates to deposit the nanolayered coatings. APTMS and PFTS were used as received from Sigma Aldrich.

3.3.1 RCA Cleaning Process

The RCA cleaning process was performed to clean silicon wafers prior to their oxidation, diffusion or chemical vapor deposition. This process involved three procedures RCA -1, RCA -2 and RCA -3. The RCA -1 (organic clean) involves in the removal of organic contaminants, RCA -2 involves in the removal of oxides (oxide strip) and RCA -3 removes the metallic wastes i.e., ionic contamination present on the surface of the silicon wafers.

3.3.1.1 Removal of Organic Contaminants

The organic contaminants (RCA -1) from the surface of the silicon wafers were removed by immersing them in chemical solution bath containing 1:1:5 ratio of NH_4OH (Ammonium Hydroxide), H_2O_2 (Hydrogen Monoxide) and DI water at 75^0C for 5 minutes. This treatment resulted in the formation of a thin silicon dioxide layer on the surface of silicon wafer. After 10 minutes of chemical bath, the wafers were washed using DI water. During this process, ionic contamination occured was removed by further ionic clean (RCA -3) process.

3.3.1.2 Removal of Oxides

Oxides deposited on the surface of the wafers during removal of organic wastes were removed by immersing the wafers in chemical solution having 1:50 concentration of HF (Hydrogen Fluoride) and DI water at a temperature of 25^0C for 30 seconds. Wafers were then

washed with DI water. During this process, small fractions of ionic contaminants were washed away.

3.3.1.3 Removal of Ionic Contaminants

The third step called Ionic clean was performed where all the metallic deposits were removed by immersing the wafers in a solution having 1:1:6 concentrations of HCl (Hydrochloric Acid), H_2O_2 (Hydrogen Peroxide) and DI water for 10 minutes at a temperature of 75^0C.

In the above cleaning process, always reactive compounds (acid/base) were added to water. These solutions were prepared in polypropylene beakers and teflon rod was used to stir them. These beakers were placed into the temperature controlled water baths and the power was maintained to required temperatures (75^0C for RCA -1, 25^0C for RCA -2 and 75^0C for RCA -3) [71].

3.3.2 Oxidation of Silicon Wafers

Silicon wafers were <100> orientation p-type doped, oxidized in a thermal oxidation furnace. The thermal oxidation is a way to produce a thin layer of oxide on the surface of the wafer. Oxidized silicon wafers i.e., silicon dioxide served as a mask against an implant or diffusion of dopant into silicon. In this work, thermal wet oxidation was.

Silicon's surface has a high affinity for oxygen and thus an oxide layer rapidly forms upon exposure to the atmosphere. Equation 3.1 details the chemical reactions involved in this wet oxidation of silicon.

$$Si_{(Solid)} + 2H_2O_{(Gas)} \Rightarrow SiO_{2(Solid)} + 2H_{2(Gas)} \quad 3.1$$

Wet Thermal oxidation is performed in furnace at a temperature of 1200^0C. This involves a tedious process since the oxide layer growth must be uniform and pure and takes long time to acquire enough thickness. Wafers were placed in a horizontal boat (rack/tube) made of quartz material and are loaded vertically into the furnace where the steam source flows over the wafers. Figure 3.3 shows schematic diagram of the quartz tube with wafers for oxidation.

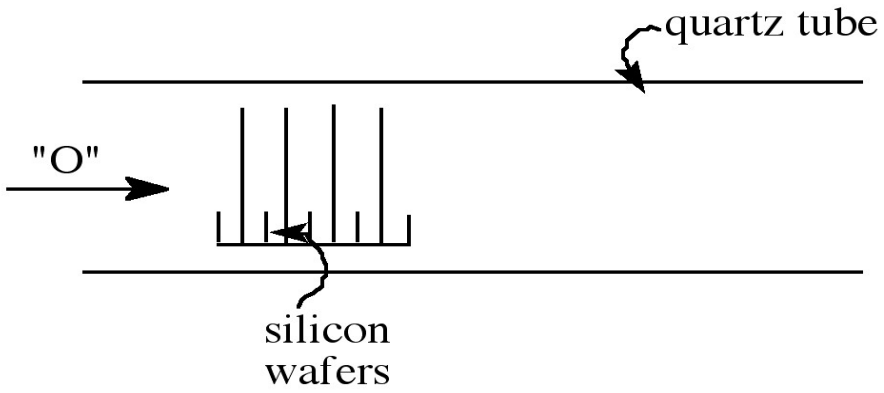

Figure 3.3 Horizontal diffusion tube showing the oxidation of wafers

These oxidized wafers were covered using photo resist in order to avoid contamination during the storage of wafers. In order to coat the substrates, the stored wafers were taken and cleaned using piranha solution cleaning process which involves the removal of photo resist and organic wastes deposited during storage. A glass beaker which fit the desired number of wafers to be cleaned was taken and filled with DI water. Another beaker of similar size was taken and silicon wafers with photo resist were placed in it. Piranha solution is composed of 1:1 ratio of Sulfuric acid (H_2SO_4) and Hydrogen Peroxide (H_2O_2). First H_2O_2 was poured into the beaker. Then H_2SO_4 was poured into the basic solution. The wafers were left in the piranha solution bath for 20 minutes. These wafers were then immersed in three DI water baths each for 5 minutes and dried using nitrogen gas flow. The piranha solution was allowed to cool down before it was transferred to the respective waste container. The cleaned wafer(s) were then diced into small dyes (0.5x0.5 cm dimensions) and were used as solid substrates to deposit the nanolayered coatings.

3.3.3 Nanotextured Polymer Film Deposition

The formation of a thin nanolayer-film was attained when the two chemicals; APTMS and PFTS were allowed to react in vapor-phase. The diced silicon wafer chip which was used as substrate was placed in a vacuum reaction chamber and the two monomers were allowed to react at controlled vacuum and reaction time allowing the consecutive nanolayer deposition.

Schematic diagram of this set up is shown in figure 3.4 and the image of the vacuum chamber loaded with samples is shown in figure 3.5.

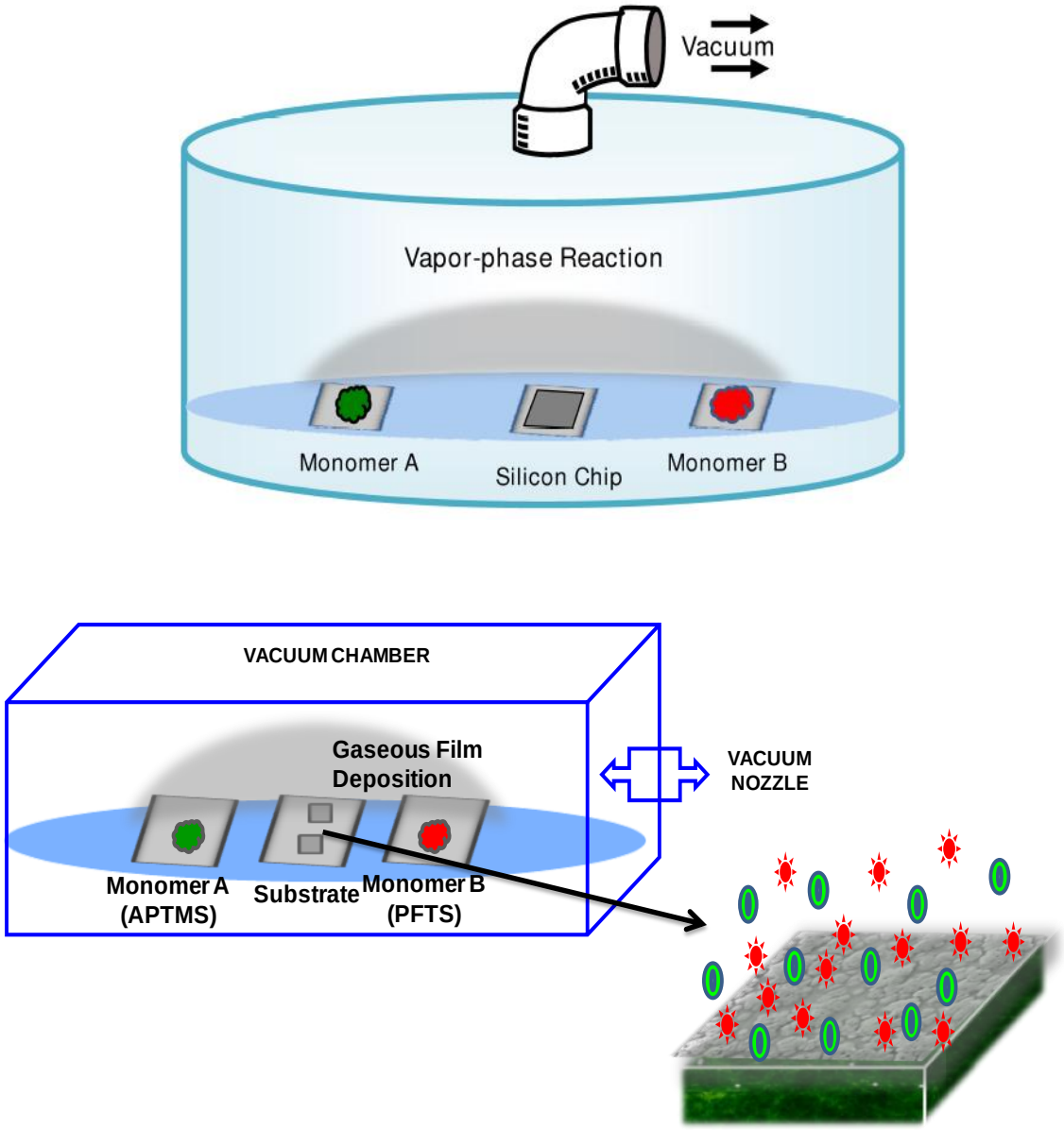

Figure 3.4 Schematic illustration of the vacuum chamber set and vapor-phase nanocoating

Prior to the nanolayer deposition process, the vacuum chamber must be cleaned with acetone and vacuum must be turned on for several hours. Once the reaction chamber was cleaned, glass slides were arranged as the chemical/substrate support bases. The two

monomers were placed on each glass slides on two sides and the glass slide with silicon wafer chip which was used as substrate to deposit the nanolayer was placed in between. The lid of the reaction chamber was closed and vacuum was turned on by turning the knob and was maintained at 22 – 25 mmHg inside the chamber throughout the deposition time.

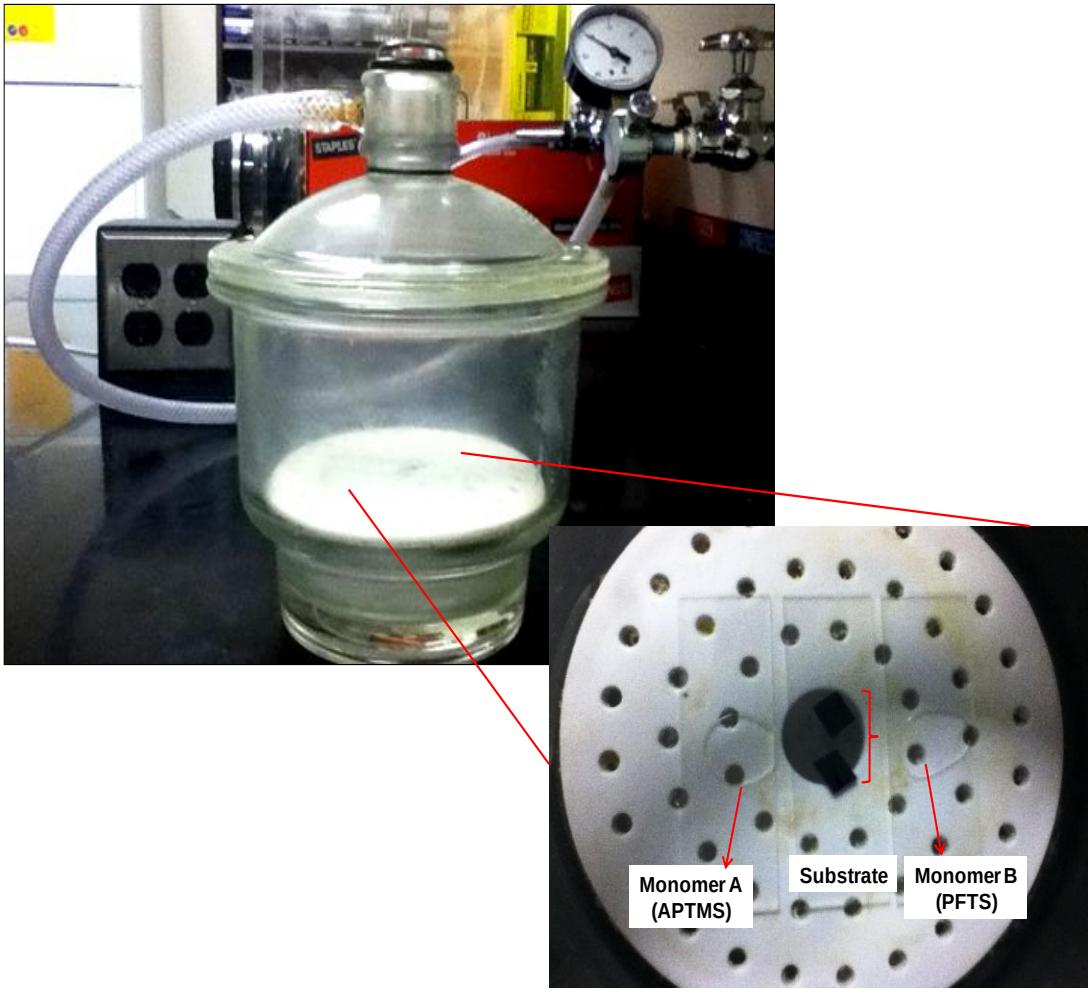

Figure 3.5 Image showing the design and setup of the process

The chemical reactions among the monomers in vacuum led to monomer volatility resulting in consecutive deposition/reaction on the substrate. The reacting monomers were volumetrically used in different ratios resulting in varied surface properties of the film.

The contact angle measurements and surface energy calculations are discussed in next chapter under physical characterization. APTMS being air-sensitive, PFTS was first introduced into the reaction chamber and soon after APTMS was placed on the glass slide. The lid of the chamber was then closed. Various samples were made with different volumetric ratios of APTMS and PFTS to analyze the pattern of nanocoating. For each ratio combination the film porosity also changed as the film grew thicker. The process was done at room temperature which one of the advantage of this process. Temperature sensitive substrates can be successfully coated using this method in contrast to typical polymerization processes which involve high temperatures. Once the deposition time is elapsed, the vacuum was turned off by closing the knob and lid was kept close until the pressure indicator showed 0 mmHg.

3.4 Nanolayered Film Properties

The surface morphology and the smoothness of the film varied with respect to the changes in ratios of APTMS and PFTS monomers in the vacuum reaction chamber. Samples were made with different ratios of APTMS and PFTS and also with varied deposition times. Biocompatibility and biostability of the film can be determined by the critical factors involving surface chemistry. The vapor-phase deposition resulted in a smooth continuous film. The thickness of the layer formed was measured with respect to time.

For thickness study, samples were made from 2.5:1 ratio of APTMS:PFTS at deposition times of 20, 30, 40, 50 and 60 minutes. The samples were used to analyze the chemical and physical properties of the film. Figure 3.6 shows the schematic illustration of thickness variation with time. As the deposition time increased, the grown layer was thicker. The gaseous phase interactions of the two monomers resulted in the deposition of the vapor-phase coating on the substrates.

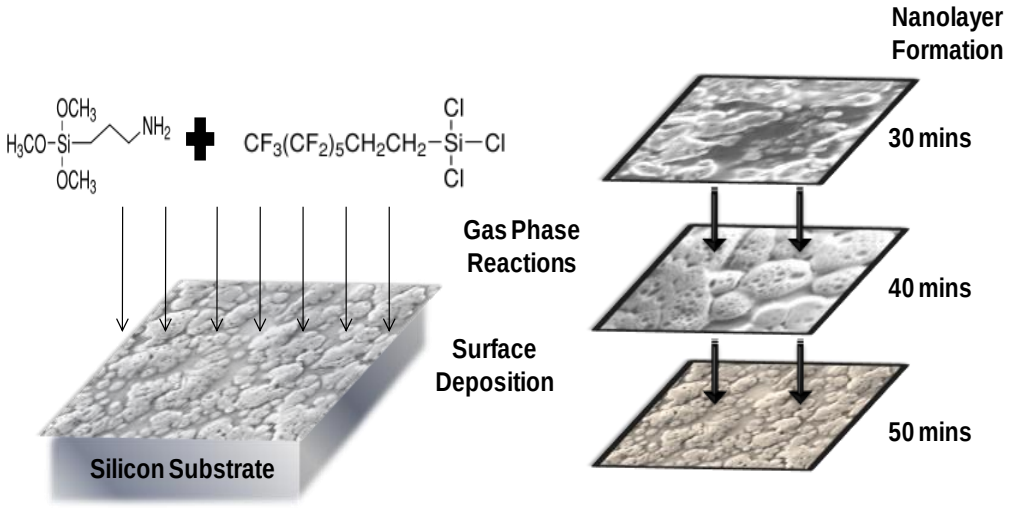

Figure 3.6 Schematic illustration of nanolayered film deposition with time

The morphology of the films were studied using ZEISS supra 55 VP Scanning Electron Microscope (SEM). Figure 3.7 and 3.8 show the SEM micrographs of the deposited.

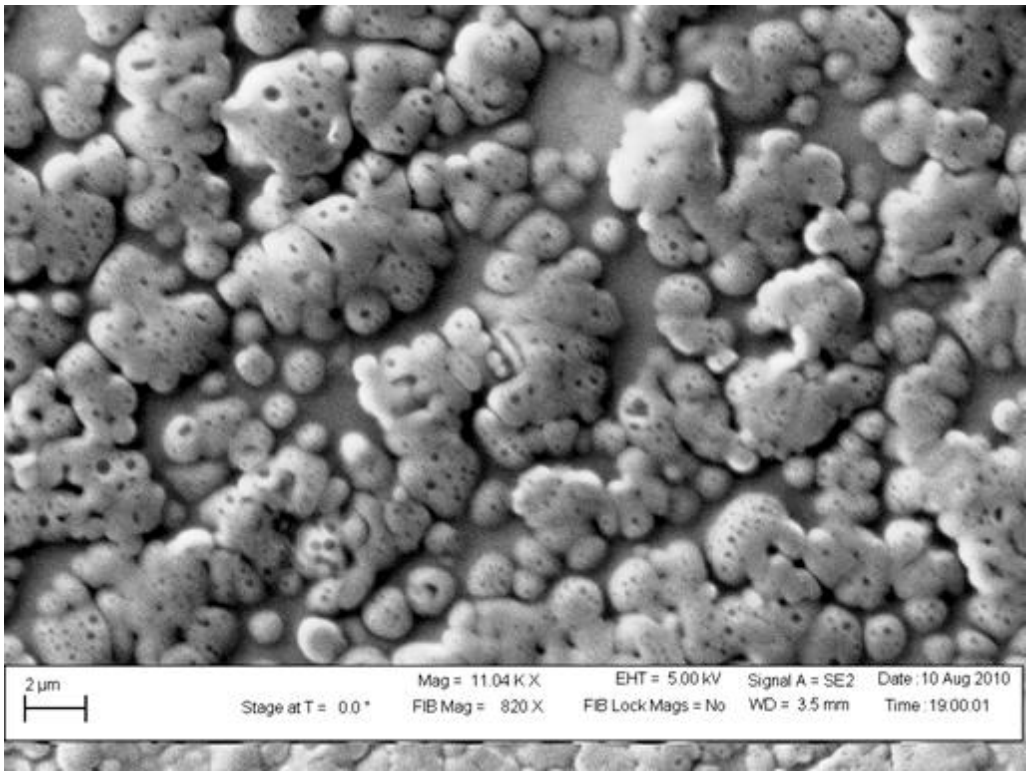

Figure 3.7 SEM micrographs of the porous nanocoating with 2.5:1 volumetric ratio of APTMS:PFTS for 40 mins deposition time at 11.04 KX magnification

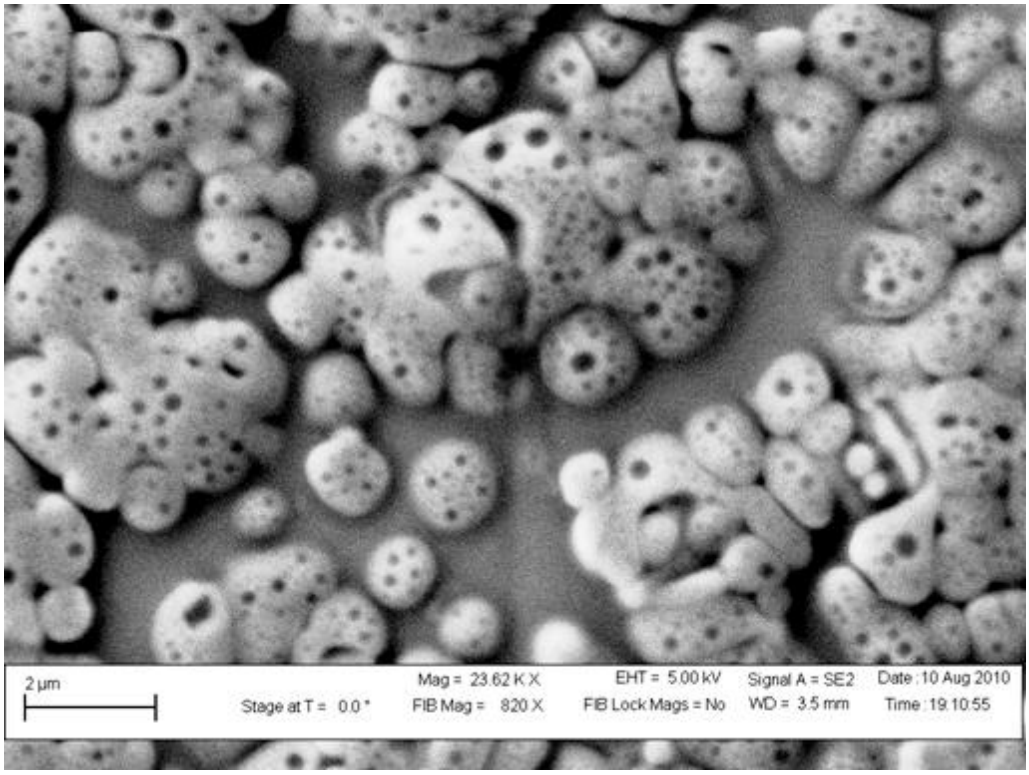

Figure 3.8 SEM micrographs of the porous nanocoating with 2.5:1 volumetric ratio of APTMS:PFTS for 40 mins deposition time at 23.62 KX magnification

CHAPTER 4

NANOLAYERED FILM TEXTURE CHARACTERIZATION

Biocompatible nanolayered coatings are the most desirable properties for a number of device applications in the fields of medical and engineering. Characterization of the coatings is an important factor to ascertain the chemical and physical properties which define biocompatibility and biostability. Spectroscopic analysis was performed to characterize the films using Fourier transform infrared spectroscopy (FTIR), X-ray photoelectron spectroscopy (XPS) and scanning electron microscopy. Films were made with different volumetric ratios and for various deposition times. With respect to these factors, the film thickness, porosity, morphology and chemical behavior was analyzed.

4.1 Physical Characterization of the Nanolayered Films

The surface properties of the films were analyzed to determine the applicable parameters in medical device implants. Four different films were coated each with specific volumetric ratio of monomers for a deposition time of 40 minutes at a temperature of 40^0C. Vacuum was maintained at 22 mmHg. The smoothness of the films showed a trend with increase in relative volume of APTMS in reaction mixture. From table 4.1, film A with 1:1 ratio of APTMS and PFTS showed interconnected layer which was slightly porous. Film B with 1:2 ratio of APTMS and PFTS showed highly hydrophobic layer due to higher concentration of PFTS. Film C made at 2:1 ratio of APTMS and PFTS showed porous and continuous layer with pores in the film that ranged in size between 100–200 nm, while film D showed similar morphology with the pore size in the range of 100–500 nm. From these results, the film morphology was analyzed to be dependent on the relative monomer volume. Fig 4.1 shows the SEM

micrographs of film C with the film grown for 30 minutes. Fig 4.2 shows the SEM micrographs of film D with 40 minutes deposition time.

Table 4.1 Film Morphology with Different Monomer Concentrations

Films	APTMS/PFTS Ratio	Morphology
A	1:1	Porous interconnected layer
B	1:2	Interconnected non-porous highly hydrophobic layer
C	2:1	Porous continuous film
D	2.5:1	Uniform and smooth continuous porous film

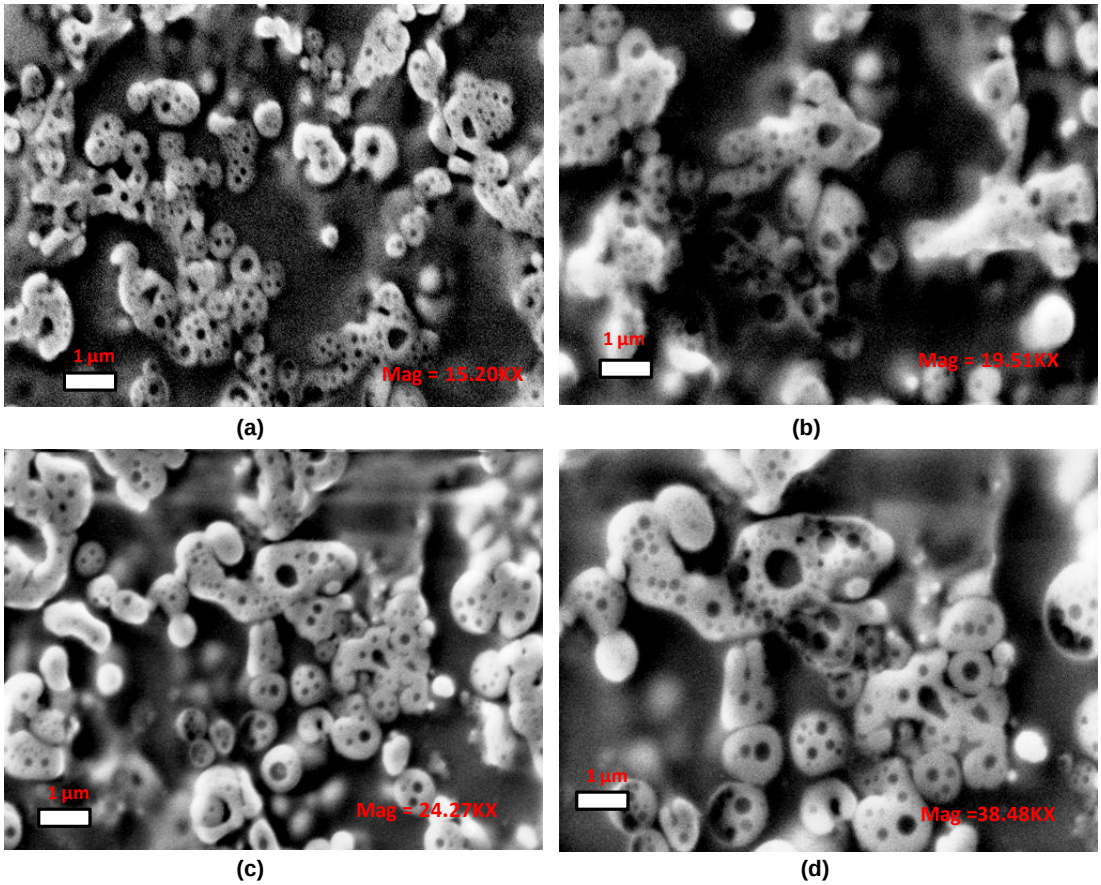

Figure 4.1 SEM micrographs of nanolayered coating having 2:1 APTMS and PFTS (a)-(d) show the surface of the coating at different magnifications as highlighted in the figure

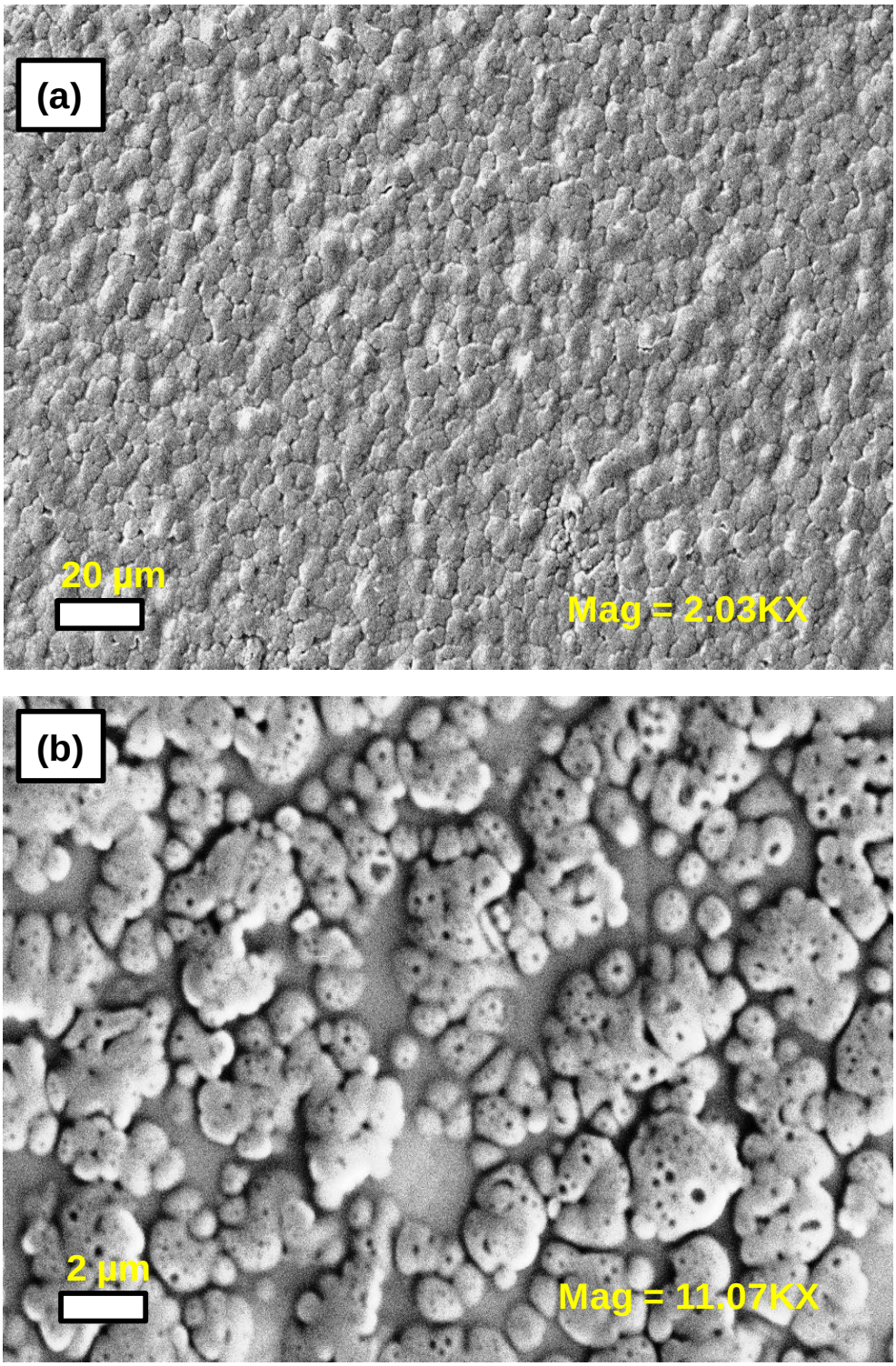

Figure 4.2 SEM micrographs of 2.5:1 APTMS and PFTS (a) and (b) are at different magnifications [11]

4.1.1 Thickness Measurements

The thicknesses of the deposited coatings were measured using KLA-Tencor Alpha-Step IQ Profilometer. Several samples were coated at volume ratio of 2.5:1 between APTMS and PFTS for 20 minutes to 1 hour to analyze the thickness pattern. Table 3.1 shows the thickness measurements of the nanolayers. After 20 minutes of deposition, the film showed very thin surface which could not be analyzed accurately using the profilometer. As the deposition time increased, the films grew thicker and more uniform and continuous. The films formed after 60 minute deposition showed thick, continuous and porous structure at nanoscale. Fig 4.3 shows the graphical representation of the thickness curve.

These measurements were taken using 4 samples of each type and the thickness was averaged (Number of samples n=4).

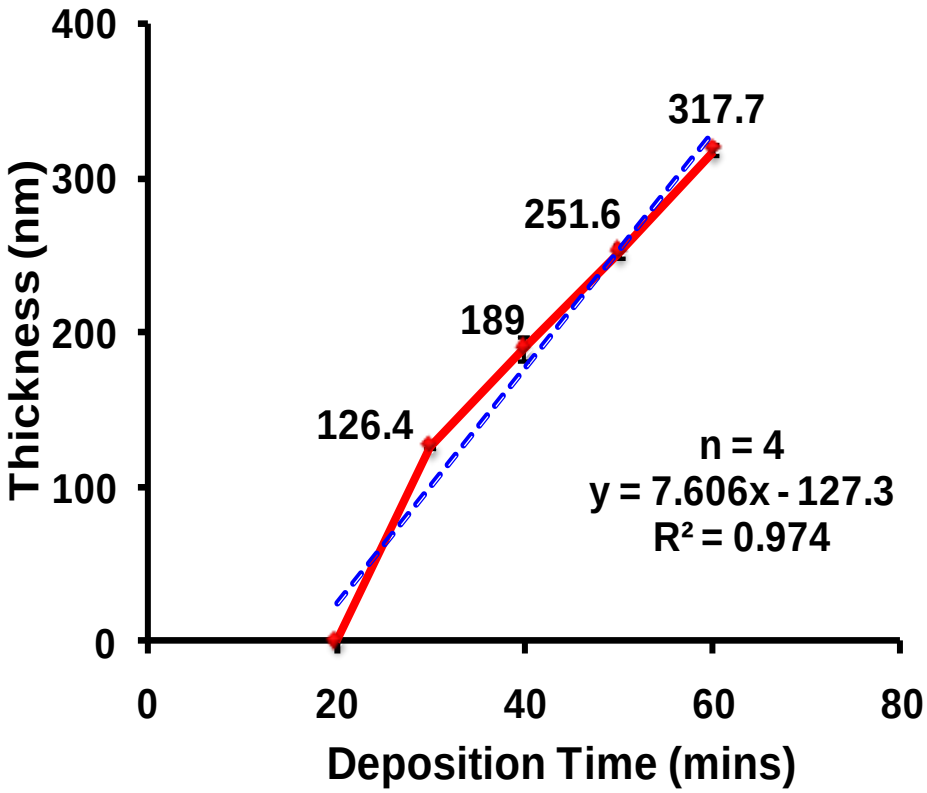

Figure 4.3 Graphical representation of the thickness curve

Table 4.2 Thickness of the Nanolayer with respect to Deposition

Sample	APTMS to PFTS Ratio	Deposition Time (minutes)	Average Thickness of the film (nm)
1	2.5:1	20	-
2	2.5:1	30	126.4
3	2.5:1	40	189
4	2.5:1	50	251.6
5	2.5:1	60	317.7

4.1.2 Surface Energy Measurements

Surface energy was measured with respect to the variations in the relative volume of the monomers. Fifteen silicon chips were taken and each of these were coated with at different ratios; 2:1, 1:2 and 1:1. This means that for 2:1 case, 100 µl of PFTS was placed on one glass slide while 200 µl of APTMS was on the other. Uncoated silicon chips were used as controls. The contact angles of the surfaces of the films were measured using Contact Angle Goniometer. Neumann's approach was adopted to calculate the surface energy values based on the obtained contact angle data [72]. Water was used as the probe liquid. The contact angle (θ) values were used to calculate the surface energy ($S.E$) using standard relation for Neumann's approach (Equ 4.1). The calculated data is shown in (Table 4.3).

$$S.E = 2.9 * 10^{-5}(\theta)^3 - 0.00652(\theta)^2 - 0.1326(\theta) + 72.8 \quad 4.1$$

Uncoated silicon wafer substrate was used as control. The average contact angle of the nanolayered coating deposited with 1:2 ratio of APTMS/PFTS showed a relatively very high contact angle of 133^0 indicating low surface energy of the film. The coatings from 2:1 and 1:1 ratios gave almost similar contact angle readings resulting in surface energy values in the range of 10-12 mJ/m^2. These results indicate that the surface energy values are dependent on the

fluorinated properties of PFTS in the coating. Fig 4.4 shows the bar graph representation showing the low surface energy property of the deposited nanocoatings.

Table 4.3 Average Surface Energy Data for the Nanolayered Coatings

Sample	APTMS to PFTS Ratio	Contact Angle (degrees)	Average Surface Energy (mJ/m^2)
1	No Coating (Control)	35	62.005
2	1:2	133	5.664
3	2:1	120	11.585
4	1:1	122	10.566

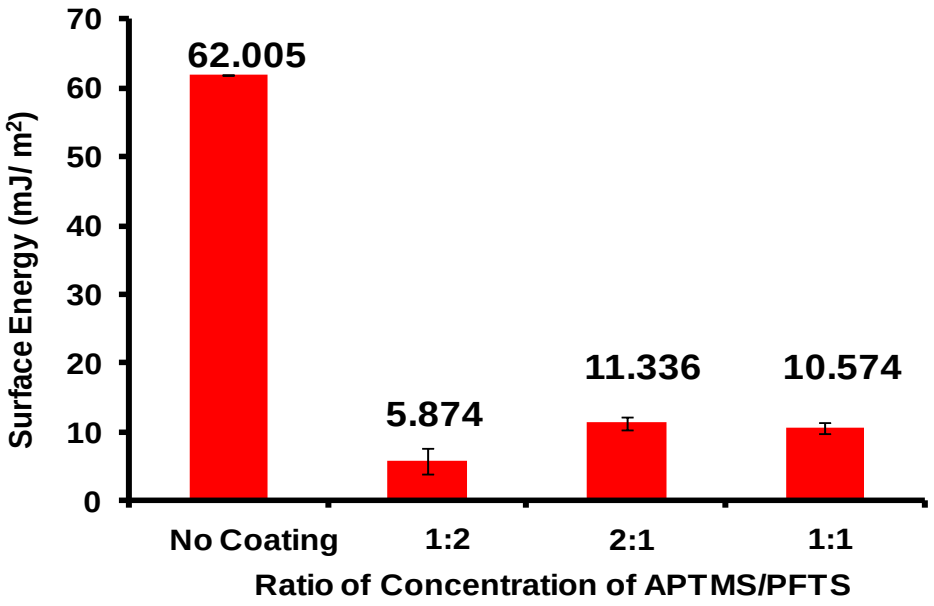

Figure 4.4 Bar graph showing the surface energy values at different monomer ratios

4.2 Chemical Characterization of the Nanofilm

FTIR and XPS were used for chemical characterization. Similar samples were made and the process is repeated at least 5 times to analyze the results to determine the properties of the film.

4.2.1 Fourier Transform Infrared Spectroscopic Analysis

Nanolayer films were made with 2:1, 1:1 and 1:2 ratios of APTMS and PFTS and the chemical composition of each was characterized using FTIR spectroscopy. In this analysis, Nicolet 6700 FTIR spectrophotometer was used. These results showed remarkable differences with multiple absorption bands between the three types of samples. The spectrum was recorded in transmission mode on kBr crystals at a resolution of 4 cm^{-1}. Fig 4.5 shows the FTIR spectra of the nanolayered coatings deposited with 2:1 ratio of APTMS/PFTS in40 minutes at a vacuum of 22 mmHg and at 40 ^{0}C temperature. Table 4.4 lists the numerical FTIR peak data.

Table 4.4 FTIR Peak Data of the Nanolayer with 2:1 Ratio of Concentration Of APTMS:PFTS

Sample	**O-H and C-H bonds (cm^{-1})**	**-C=O (cm^{-1})**	**Si-O-Si (cm^{-1})**	**Halogen Si-C/C-Cl (cm^{-1})**
1	2630-3630	1635	1139-1010	749, 688
2	2645-3211	1701	1135-1006	670, 641
3	2649-3335	1652	1135-1005	782, 650
4	2573-3202	1506	1137-1013	755, 698

These results explain the chemical properties of the film and their organic behavior showing the presence of various chemical bonds. In figure 4.5, which shows the FTIR spectra of the nanocoating deposited at 2:1 APTMS/PFTS with 40 minutes deposition time, various peaks were recorded at different specific regions which explain their chemical properties. A broad stretching of the peaks was observed in the range of 2500-3200 cm^{-1}. This stretching occurs when O-H and C-H bonds are present on the surface. Therefore, this reading indicated that there were strong oxygen/hydrogen and carbon/hydrogen bonds on the surface of the deposited nanocoating. At 1700 cm^{-1}, a small peak was recorded which explained the presence of –C=O bonds. The spectra below the wave number 1400 cm^{-1} is known as finger print region. Any

peaks recorded under this region indicate the presence of siloxane bonds (Si-O-Si). High absorption peaks were observed in the range of region at 1000 - 1200 cm^{-1}. Therefore, the presence of siloxane bonds was recorded. FTIR spectra for halogens showed peaks in the range of 800-600 cm^{-1}. Sharp peaks were observed at 650-740 cm^{-1} which indicated the halogen peaks due to fluorine contributed by PFTS in the coating. These groups can have high protein/cell adsorption due to their inertness and thermal resistance.

Fig 4.6 shows the FTIR spectra of the film made at 1:2 ratio of APTMS:PFTS and fig 4.7 shows the FTIR spectra of the film from 1:1 ratio of APTMS/PFTS. When there was more concentration of PFTS polymer in the nanocoating mixture, the FTIR spectra (figure 4.6) showed different results compared to the one from fig 4.5. Sharp peaks were observed in the halogen region near 780 cm^{-1} due to higher concentration of fluorine in PFTS. No specific peaks were seen under the finger print region when compared to 2:1 APTMS/PFTS polymer ratio. C-H bonds were indicated with the broad stretching near 3500-2500 cm^{-1}. In this spectrum, most of the bonds formed due to PFTS halogen bonds dominance over APTMS.

Fig 4.7 shows the FTIR spectra with equal volume of APTMS/PFTS. The data obtained from the spectra show similar peaks recorded as of fig 4.6. Peaks under halogen region and fingerprint region formed due to PFTS polymer dominance over the carbon/oxygen bonds formed from APTMS polymer. This again showed that the PFTS was dominant over APTMS.

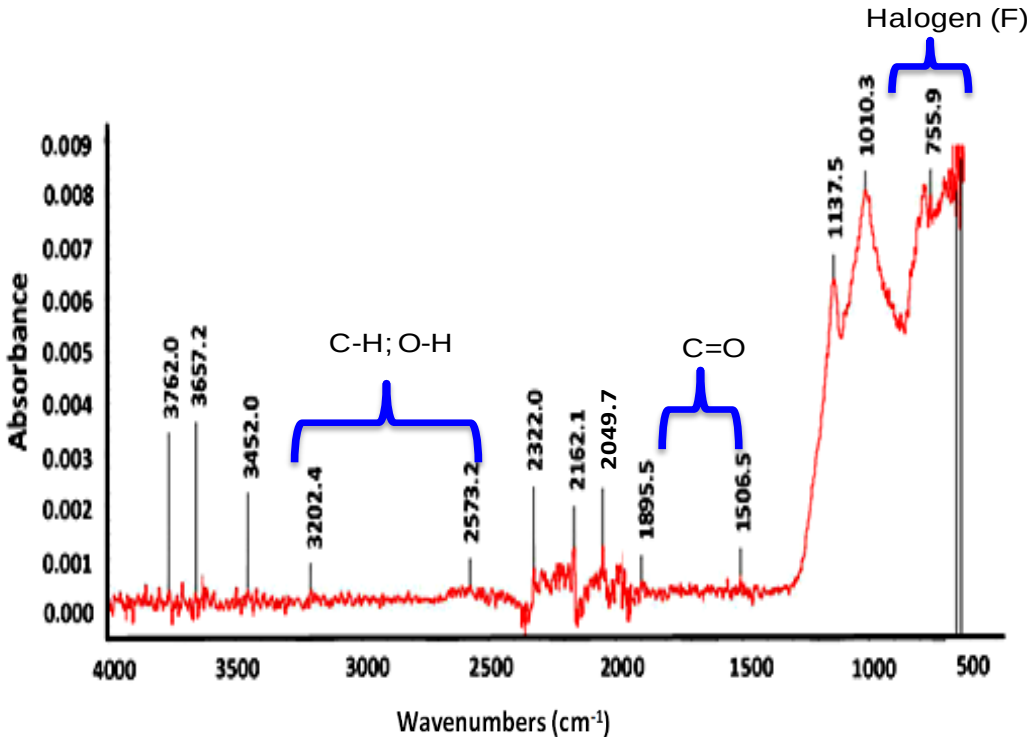

Figure 4.5 FTIR spectra of the organic film coated with 2:1 ratio of APTMS/PFTS

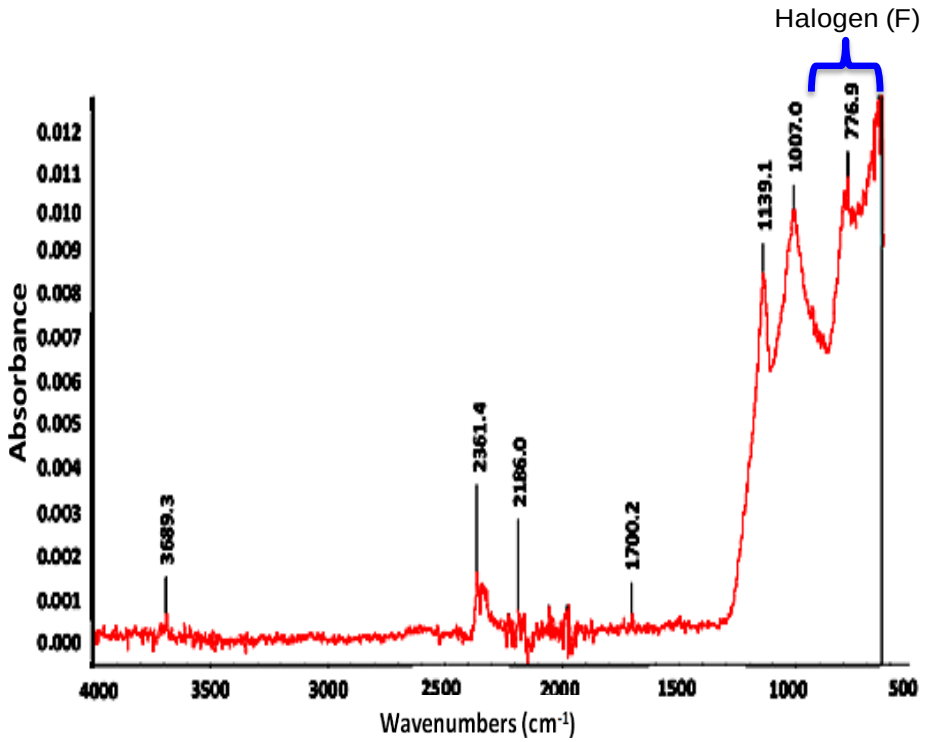

Figure 4.6 FTIR spectra of the nanolayered coating made by 1:2 ratio of APTMS/PFTS

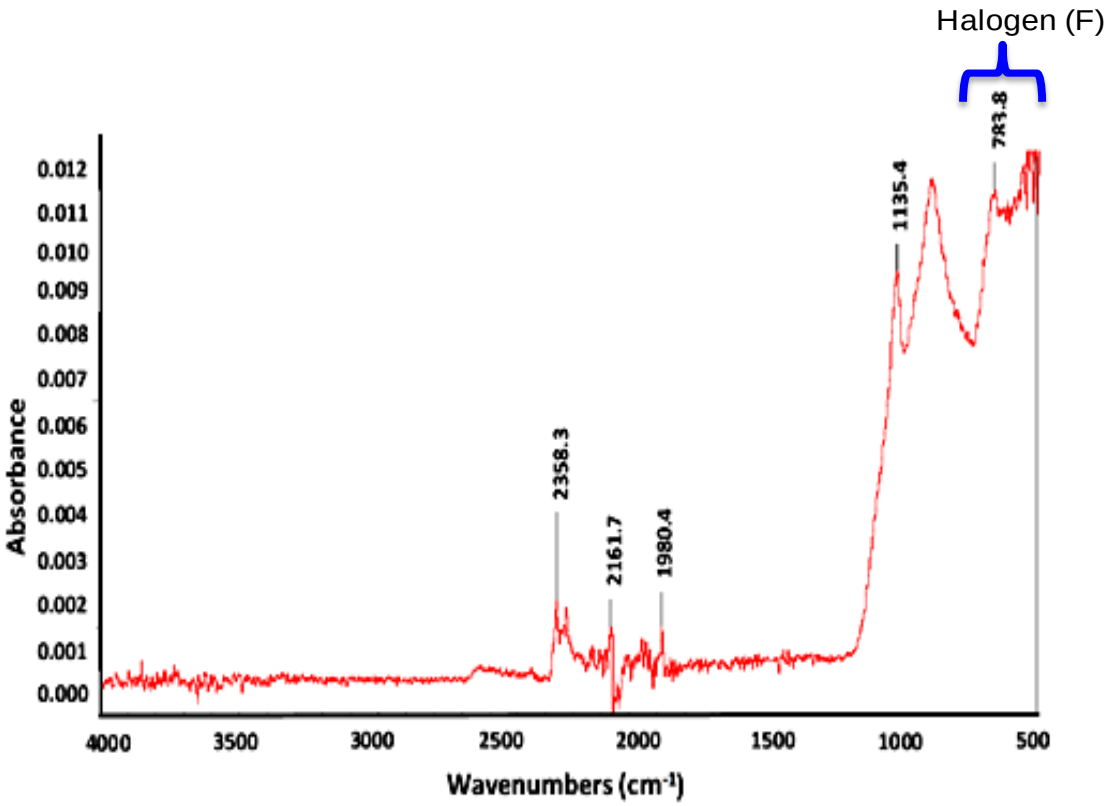

Figure 4.7 FTIR spectra of the organic film coated with 1:1 ratio of APTMS/PFTS

These results showed the differences in the chemical compositions of the organic nanolayered films when opposite volumetric ratios of the polymers were used.

4.2.2 X-ray Photoelectron Spectroscopic Analysis

The organic property of the film was determined using XPS analysis. It is a quantitative spectroscopic technique that measures the elemental composition of the elements that exist within the top 1 to 10 nm of the surface of material. The goal of this analysis also includes the uniformity of the elements on the entire top surface. Any contamination of the biomaterial can be analyzed using this approach. The biocompatibility of the nanolayer coating depends on the specific binding energy of carbon and oxygen. These studies were done using Al Kα radiation at 1489.9 eV. A nanolayer formed at 2:1 ratio of APTMS:PFTS for 40 minutes was used for XPS analysis. The elemental composition obtained by high resolution XPS peaks of carbon (C), oxygen (O), Fluorine (F), Chlorine (Cl) and Silicon (Si) are shown in table 4.3.

Table 4.5 Elemental Composition of the Nanolayer Derived from XPS Analysis

Elemental Compound	**Atomic Weight %**
F	44.56
O	8.76
C	36.15
Si	6.51
Cl	1.89

From these results, fluorine had the highest elemental atomic weight percentage compared to the other elements identified on the polymeric coating. This again showed the domination of PFTS over APTMS. Chlorine showed the least atomic weight % of 1.89. Fig 4.8 shows the survey spectrum of the XPS analysis showing the presence of elemental compounds. The components of carbon with 36.15 % atomic weight was observed in the range of 280-298 eV and oxygen (8.76% atomic weight) was observed in the range of 530-534 eV confirmed the film to be organic. Fig 4.9 shows the high resolution peaks of Si, F, O and C elements.

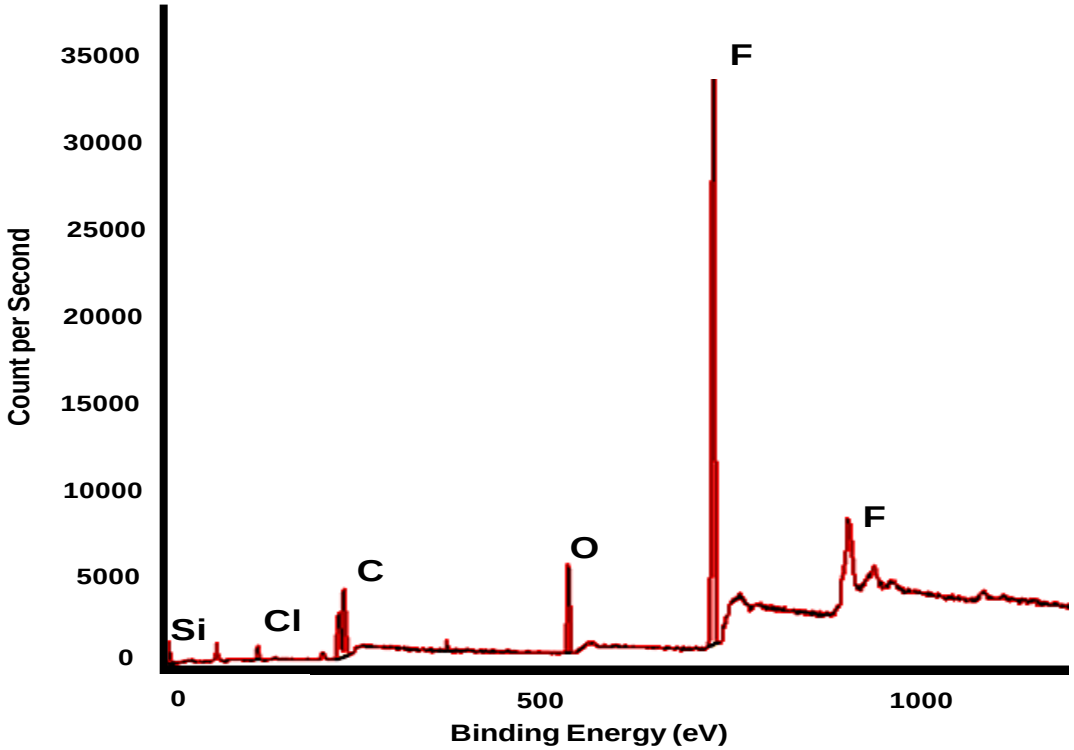

Figure 4.8 Survey Spectrum of XPS analysis with elemental compound peaks

The data of fig 4.9 shows that silicon is deconvoluted into two distinct peaks at 103.56 eV and 105.4 eV (Si-O-Si) [73]. Siloxane bonds due to silane molecules were indicated with stretching of spectra without peaks at 101.6 eV which is characteristic of Si-O-C. The C peak is deconvoluted into five components that are located at 284.6 eV (CH_2-CH_2), 286.6 eV (-C-O), 288.75 eV (-C=O), 292.25 eV (CF_2-CF_2) and 294.58 eV (CF_3-CF_3) [10, 74]. The data showed that 36% of C (288.74 eV) were carboxylic component, which also accounts for high percentage of oxygen in the film, indicating oxidation of the molecules. Sharp peak at 689.17 eV (C-F) determined fluorine where C-F corresponds to 88% of total fluorine present [74, 75]. Oxygen peaks having high resolution were observed at 533.45 eV. A small Cl peak was observed at 200 eV. A weak nitrogen peak was also recorded, indicating that either the PFTS was predominantly present in the top layer or the amine group of APTMS reacted with Cl of PFTS leaving trace amount of nitrogen in the top layer [10].

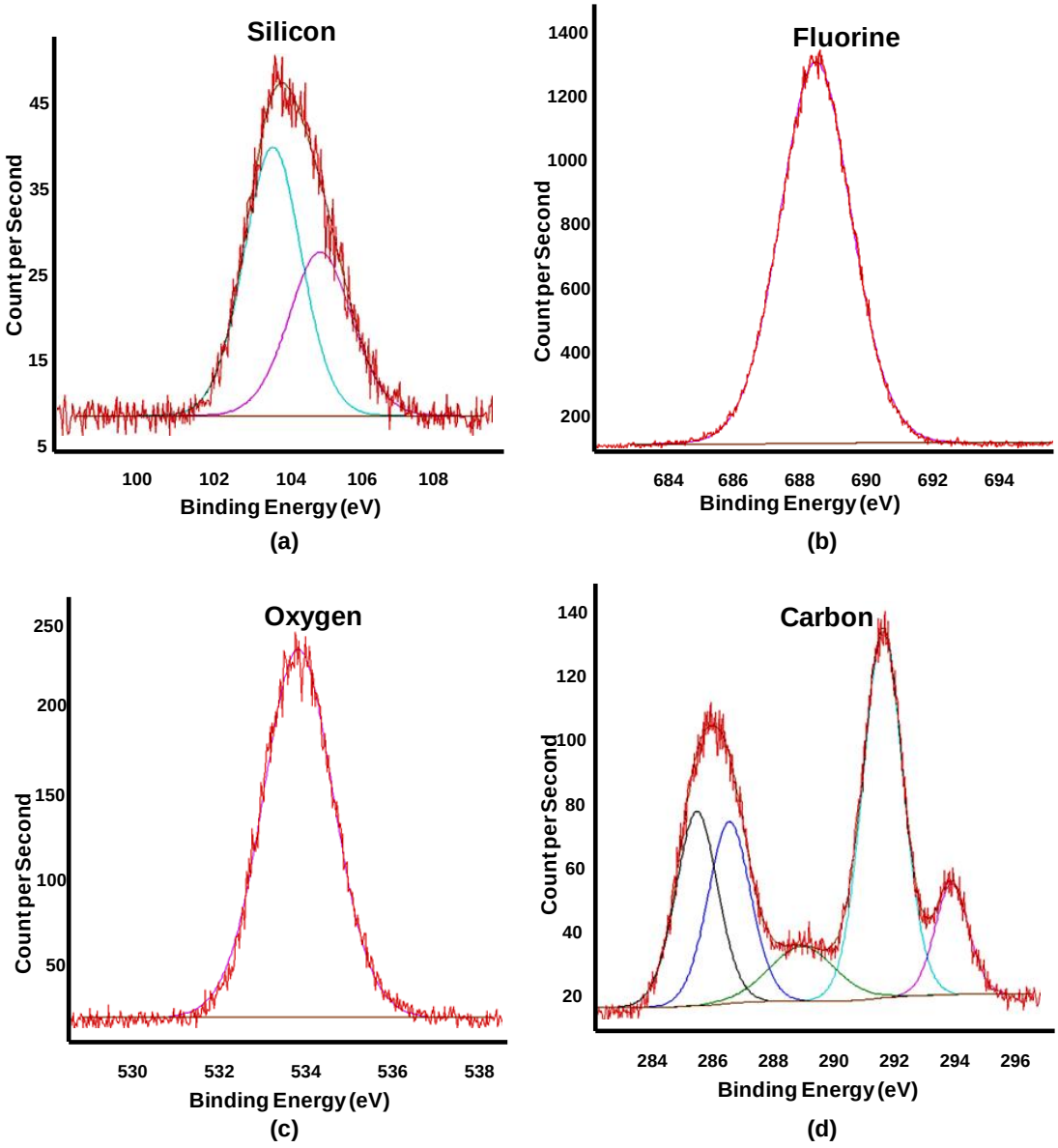

Figure 4.9 Zoomed in view of the XPS spectra of elements in nanolayered film made from 2:1 ratio of APTMS:PFTS (a)-(d) Spectra with characteristic peaks of Si, F, O and C

4.3 Organic Film Biocompatibility and Stability

Biomedical implants are coated with thin organic films to modify the surface characteristics and to make the surface of the implant biocompatible. The degradation of the

coating is an important factor for medical device coating applications. Whenever an organic biomaterial is coated or Bio-MEMS/NEMS devices are used in- vivo/in- vitro, the coating needs to be stable to withstand the varying chemical surroundings [76-78]. To qualitatively verify the chemical stability of nanolayered organic polymer coatings, the silicon substrates were coated and exposed to various solutions having variety of chemical compositions. The samples were first washed with DI water, Acetone and Ethanol and then were immersed in DI water for 24 hours. The surface energy of the film was measured which showed no difference from the un-washed coated substrate. Different pH solutions; pH 2, pH 4, pH 7 and pH 10 were taken in separate dishes and the substrates coated with 2:1 ratio of APTMS and PFTS for 40 minutes at 22 mmHg were immersed for 15 hours. Buffer solutions having pH 4, pH 7 and pH 10 were taken directly. pH 2 solution was prepared using HCl and NaOH [79]. Fig 4.10 shows four glass petri dishes filled with different pH solutions and immersed coated silicon substrates.

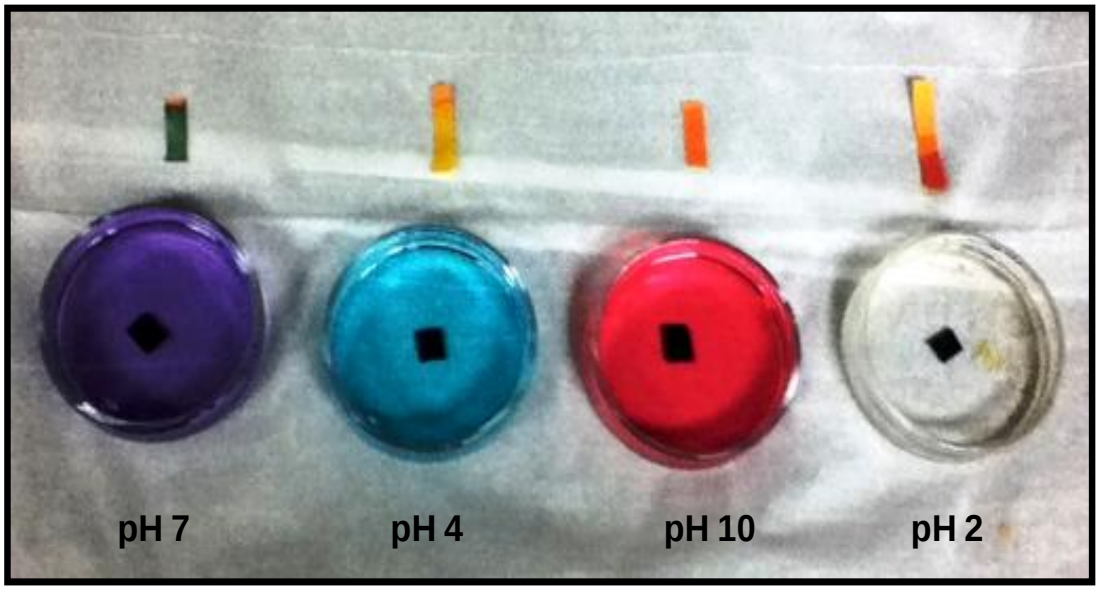

Figure 4.10 Coated silicon substrates immersed in different pH solutions

4.3.1 DI Water effect on Nanolayer Coating

The surface of the coated silicon wafer chip showed no change when it was washed with DI water, ethanol and acetone. Chips were then dried using nitrogen gas flow for further

measurements. There was no difference in the contact angle and surface energy before and after DI water/ethanol/acetone wash of the layer. Slight increase in the surface energy was noted when the coated substrate was left immersed in DI water for 24 hours. This could be due to hydrolysis of the film in the DI water. Fig 4.11 shows the SEM micrographs of the coated silicon substrate used as control and fig 4.12 shows the SEM micrographs of the coated silicon substrate chips immersed in pH solutions.

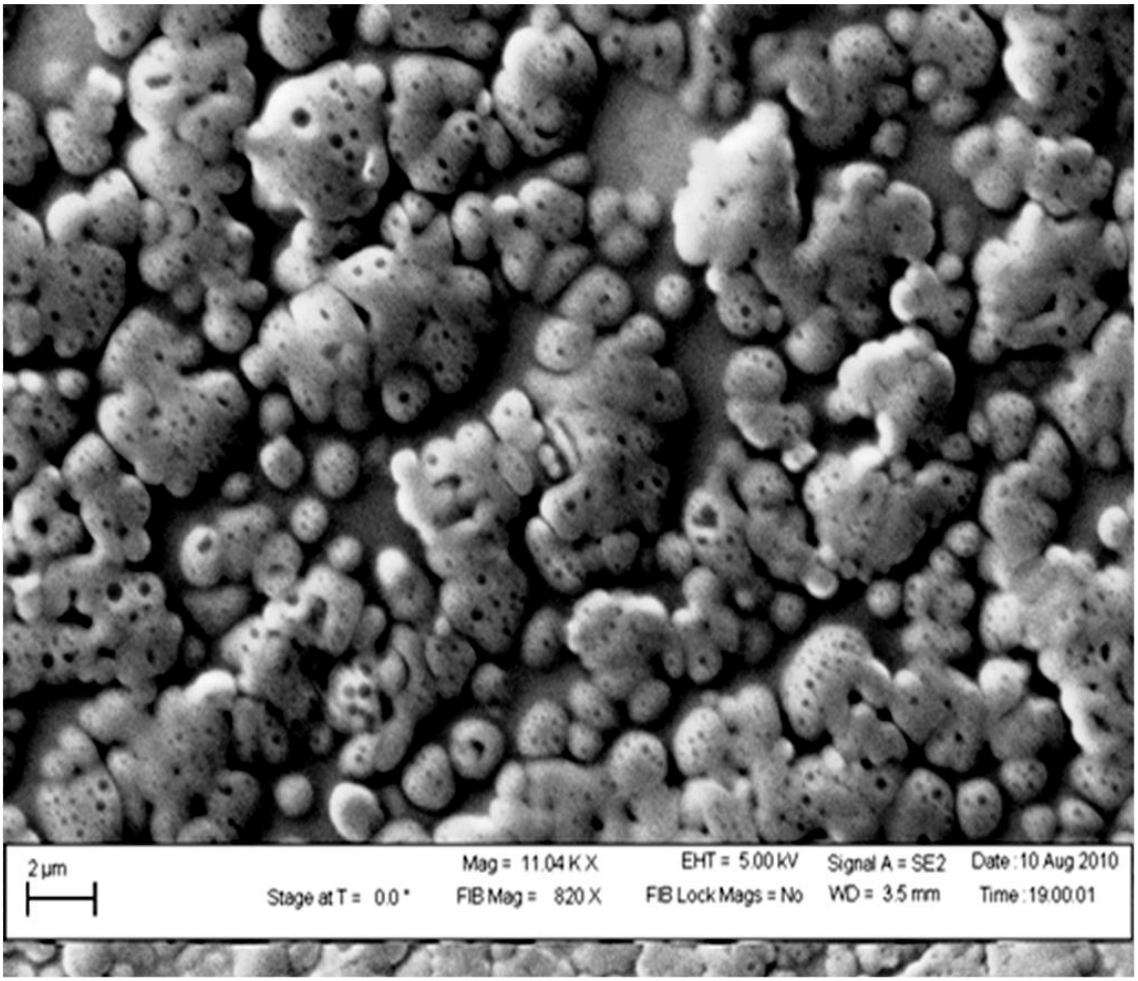

Figure 4.11 SEM micrograph of the nano-layered coating used as control for stability test

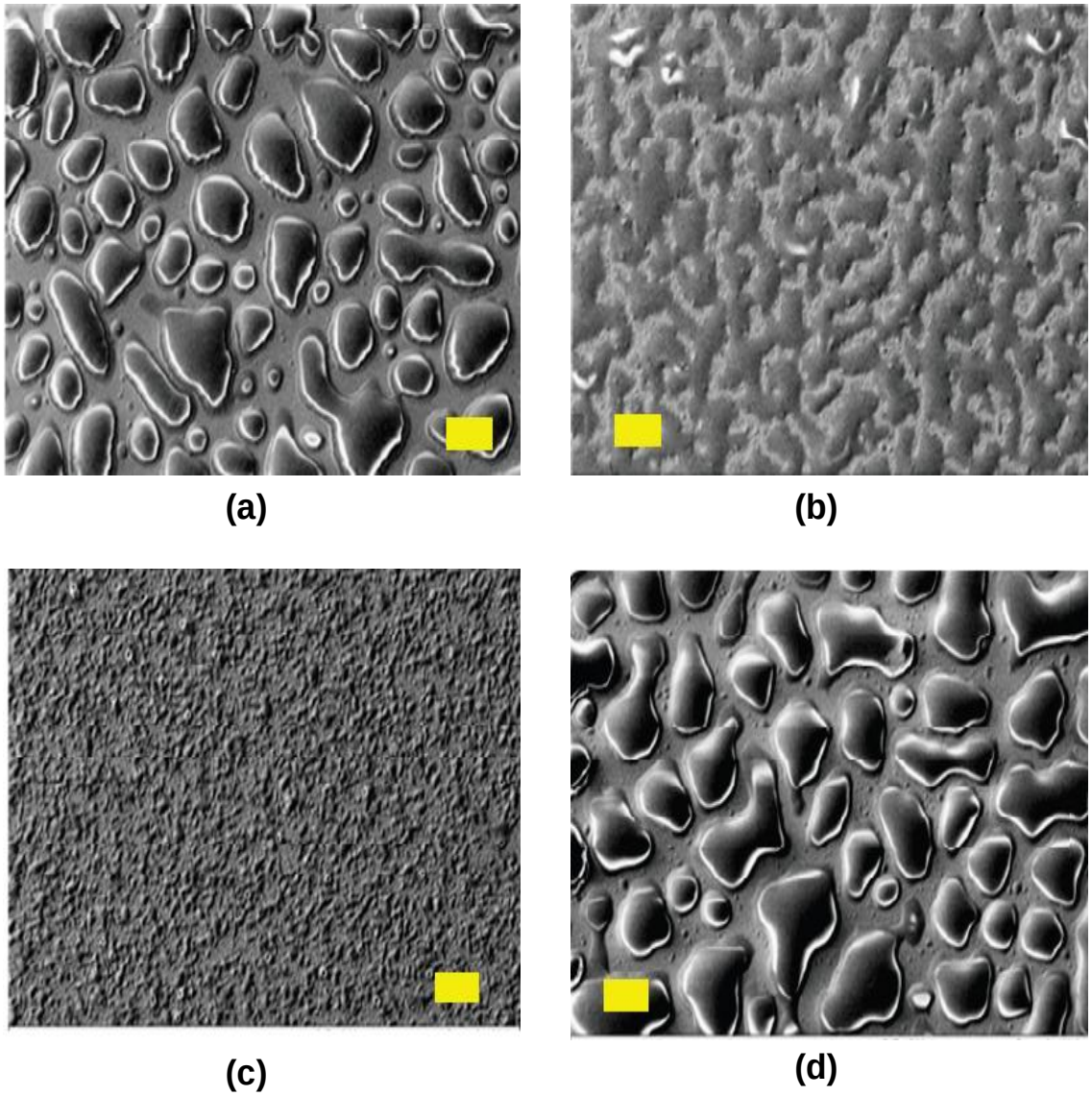

Figure 4.12 SEM micrographs of the coated silicon chips immersed in pH solutions. All scale bars are 20 µm (a) nanocoating surface immersed in pH 2 solution (b) pH 4 (c) pH 7 solution and (d) pH 10 solution

4.3.2 Effect of Different pH Solutions on Nanolayer Coating

Coated substrates immersed in different pH solutions for 15 hours were taken out and dried in nitrogen gas flow. These were analyzed from SEM micrographs. It was observed that the nanocoatings were still intact on the chips but the stability varied with the pH solutions used. Fig 4.12 (a) shows the chip dipped in pH 2 solution. It showed low contact angle measurement

resulting in increase in surface energy. Substrate immersed in pH 10 solution also showed similar results (fig 4.12(b)). Substrates immersed in pH 4 and pH 7 (≈ body pH 7.45) (fig 4.12 (c) and (d)) solutions did not show much variations in their surface properties [80]. This indicated that the nanolayer coatings were stable at various chemical surrounding showing high resistivity further proving to be biostable and biocompatible.

4.4 Conclusion of the Research Work

Nanolayers of biocompatible coatings are the most desired properties for a number of device applications in medicine and engineering. Depositions of nanotextured thin films are reported using a simple method of vapor-phase vacuum chamber reaction and biochemical and physiological parameters have been analyzed. The data show that the ratio of APTMS:PFTS, deposition time and the total reactor pressure are the important parameters to achieve desired morphology of the films. The chemical and physical characterization of the coating showed biocompatible, low surface energy fluorinated layers which are ideal for many biomedical applications. However, only a range of molecules are analyzed that could promote low surface energy on the modified substrate. The presented method requires simple equipment and is performed at room temperature which extends the range of temperature-sensitive substrates. As the surface coating is done in vapor-phase, the coating can also be used for applications where inner surfaces of the implants need to be coated with desired polymeric films.

4.5 Coating of Biomaterials

Medical device surface properties are modified to inhibit body compatibility and corrosion resistance along with cell/protein adhesion. Alterations in surface hydrophobicity/hydrophilicity, antimicrobial impregnation, coating of biomaterials and host protein, and cell coating of biomaterials have been used with limited to variable success in both the short and the long term. Determining efficacy of these methods has always been tedious and in- vitro efficacy often does not correlate well with in- vivo effectiveness in vivo [81, 82]. There are many types of medical coatings which can be divided into categories based on the

type of applications. For example, biomedical implants are a viable option for patients in need of joint replacements. Having permanently fixed cement less implant is a challenging aspect. Therefore, porous and bio-active coatings are implemented on orthopedic implants. Pure titanium and hydroxyapatite coatings are used for such applications as the materials are wear-resistant [83, 84]. Based on the kinds of implants needed such as cardio stents, dental implants, the use of stainless steel, cobalt, chromium, titanium and its alloys, bioceramics and polymers which get in constant contact with the aggressive body fluid, they often fail and fracture due to corrosion inside the body. To overcome this, it is important to have a corrosion resistance material. Therefore, surface modification of implants with the best solution to combat corrosion and to enhance the life span of the implants and longevity of the human beings is important [85, 86]. In the case of biological interfaces, the surface properties place stringent requirements for the selective detection of analytes [87]. Based on the study of polymeric nanolayered coatings in this thesis, medical implants that require body fluid compatibility and high corrosion resistance can be coated with this approach. For any medical implant, it is important to have uniform coverage of surface modification. To analyze the uniformity of the coating in inner surfaces of the biomaterials, a silicon 3D structure was coated and the uniformity of the coating was analyzed using SEM [88].

4.6 Coating of 3D structures

A solid-state micropore was coated with APTMS/PFTS nanofilm to analyze the uniformity of the coating in the inner structures of the 3D micropore. A silicon chip with micropore of 11.7 µm diameter was coated using vapor-phase reaction with 2.5:1 ratio of APTMS:PFTS for 40 minutes deposition at a controlled vacuum of 22 mmHg.

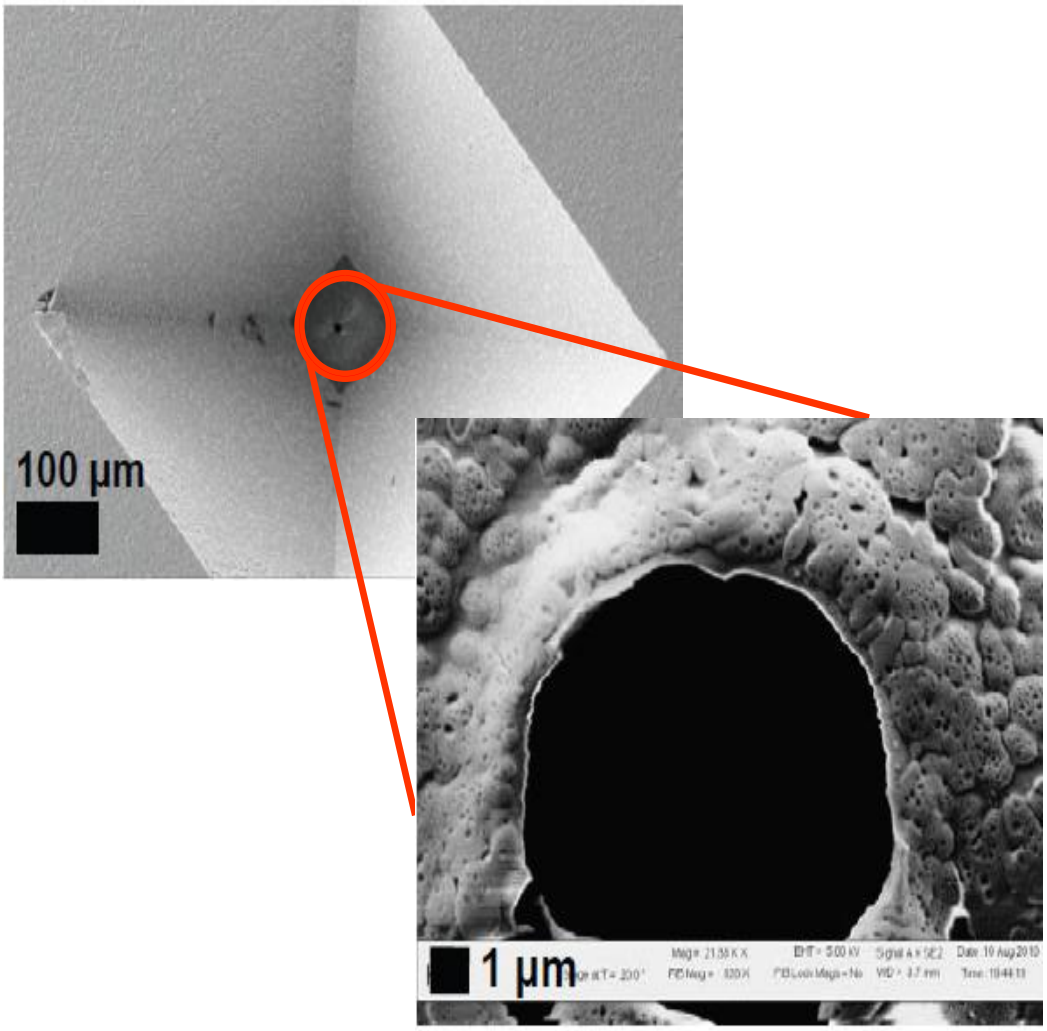

Figure 4.13 SEM micrographs of the coated micropore detailing the uniformity of inner coating of the structure

The coating formed a nanolayer on the pore covering all the sides. Fig 4.13 shows the SEM micrograph of the coated 3D surface of the micropore. The data showed that the micro/nano sized structures can be coated evenly on all sides using this simple approach. Depending on the coating needed, the concentrations of the monomers can be varied. Fig 4.14 shows the SEM micrographs of the inner structures of the coated micropore.

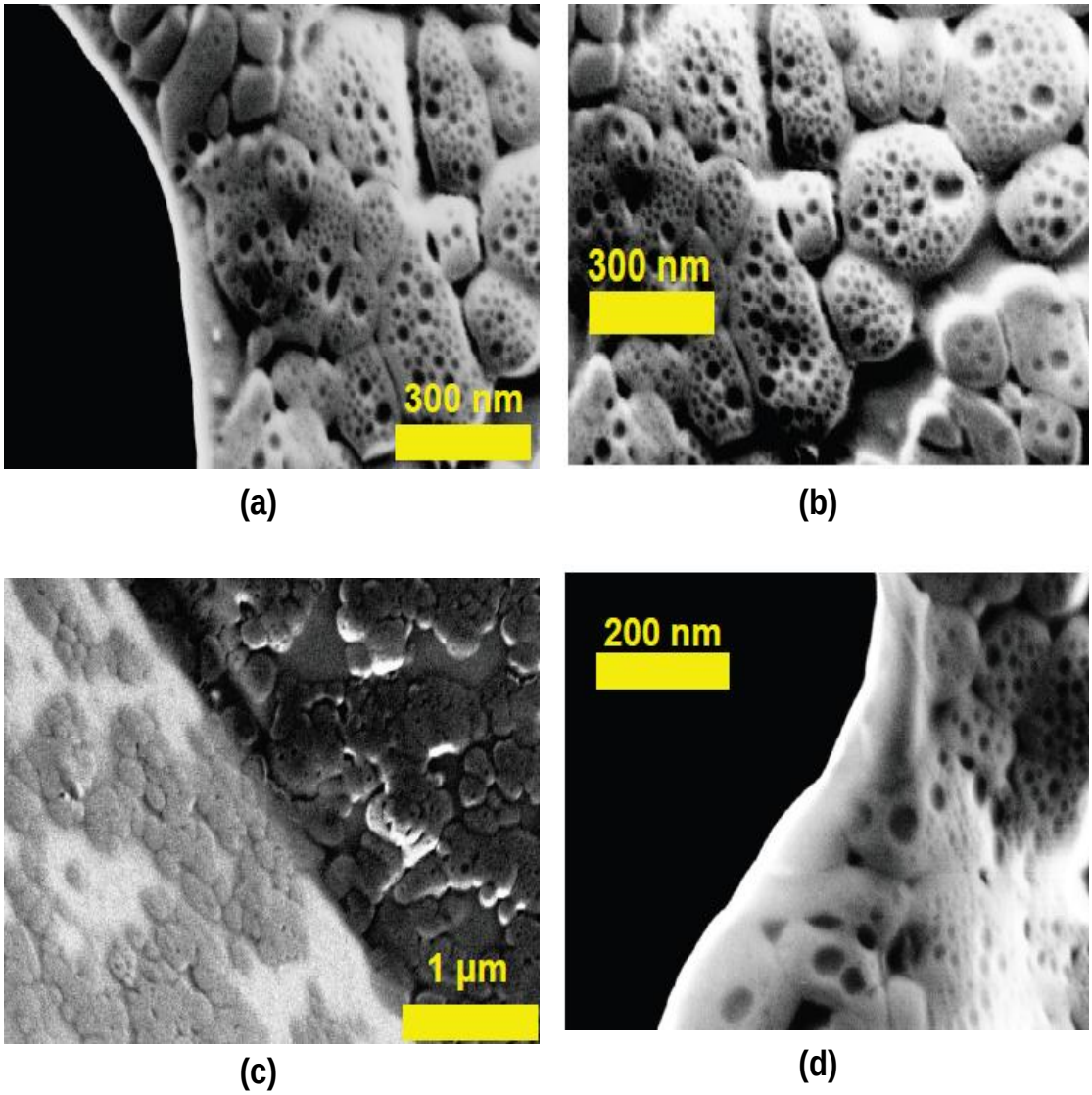

Figure 4.14 SEM micrographs detailing inner structures of the coated micropore. (a) and (d) shows the periphery of the pore. (b) shows the magnified view of the interconnected coating of the surface and (d) shows the coated inclined wall of silicon substrate (bright part) and the coated membrane of the pore (dark part)

No significant change was observed in the micropore size as the coating thickness was ≈ 196 nm. From fig 4.13 it is evident that the vapor-phase approach coats micro structured devices with complete uniformity. Fig 4.14 (a) and (d) show the coated periphery of the micropore. The thickness of the coating was even all over the micropore. Fig 4.14 (c) shows the coated surface of the silicon substrate along with the membrane of the pore. From these results,

it can be seen that micro-structured biomaterials can be coated to modify surface characteristics to attain biocompatibility. Stability test detailed in previous chapter proved the corrosion-resistance of the coating in various chemical surroundings. The fluorinated low surface energy property of the coatings also helped promote the cell growth [10]. Therefore, the coating is applicable in many biomedical appliances such as medical implants which require biocompatibility, corrosion resistance, lower surface energy, uniformity and also support cell/protein adhesion.

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CHAPTER 5

FUTURE WORK

Over the past few decades of developing artificial implants for humans, most synthetic prostheses have consisted of material particles having grain sizes with conventional dimensions. For example, for orthopedic applications, the majority of these early implants were made of stainless steel, cobalt alloys, and titanium. For neural implants, silicon is most widely used material in electrodes for diagnosis and treatment. But the lack of sufficient bonding of these synthetic implants to desirable surrounding tissue is a major challenge. Mechanical characteristics that mimic surrounding bone are a necessity for implant success. Likewise, electrical properties to stimulate neuron cell axonal outgrowth are required to heal damaged areas. The common priorities for these types of implants are the surface properties since proteins and cells will interact immediately with the biomaterial surfaces after implantation. The long-term functionality of the device is dependent on the healing response of the biomaterial.

5.1 Drug Enhanced Medicated Nanocoating for Biomedical Implants

For any medical implant, it is important to modify the surface characteristics of the biomaterial to attain the critical properties involving biocompatibility and biostability. Coating of the medical device is based on the type of organ treatment. For some implant applications, it is important to have a surface which supports protein/cell adhesion for the treatment purposes. When such a device is coated with medicated coating, it might not only provide compatibility and corrosion resistance but may also deliver drug to the surrounding area which treats the infected cells/tissues and help in tissue regeneration.

To improve the device design, the coating needs to be tested if it can also support the drug release parameters. The first step would be to examine if there is any effect of the nanolayered coating of the standard drug release curve.

5.2 Polymeric Biosensor Devices

Biomedical sensors are one of the most important parts of biomedical engineering that enables the detection of biologic events and their conversion to signals. It serves as an interface between biological and electronic systems. They take signals representing biomedical variable and usually convert them into an electrical/optical signals. Conducting and semiconducting polymers can be synthesized and deposited onto a conductive surface. Use of this method might have great prospects in biosensors where enzymes may be entrapped into conducting polymer films. In order to implement this idea using the present thesis work, the primary step would be to find the conductivity of the nanolayer. This could be known by using current-voltage and capacitance-voltage measurements. Fig 5.1 shows the proposed method for current-voltage (*I-V*) measurements to determine the conductivity of the film and its applications in biosensors.

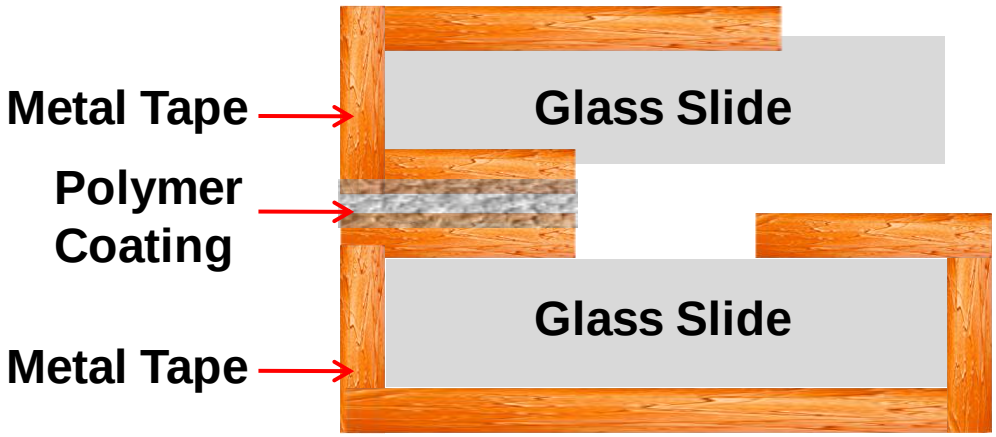

Figure 5.1 Schematic illustration of the device for capacitance measurements

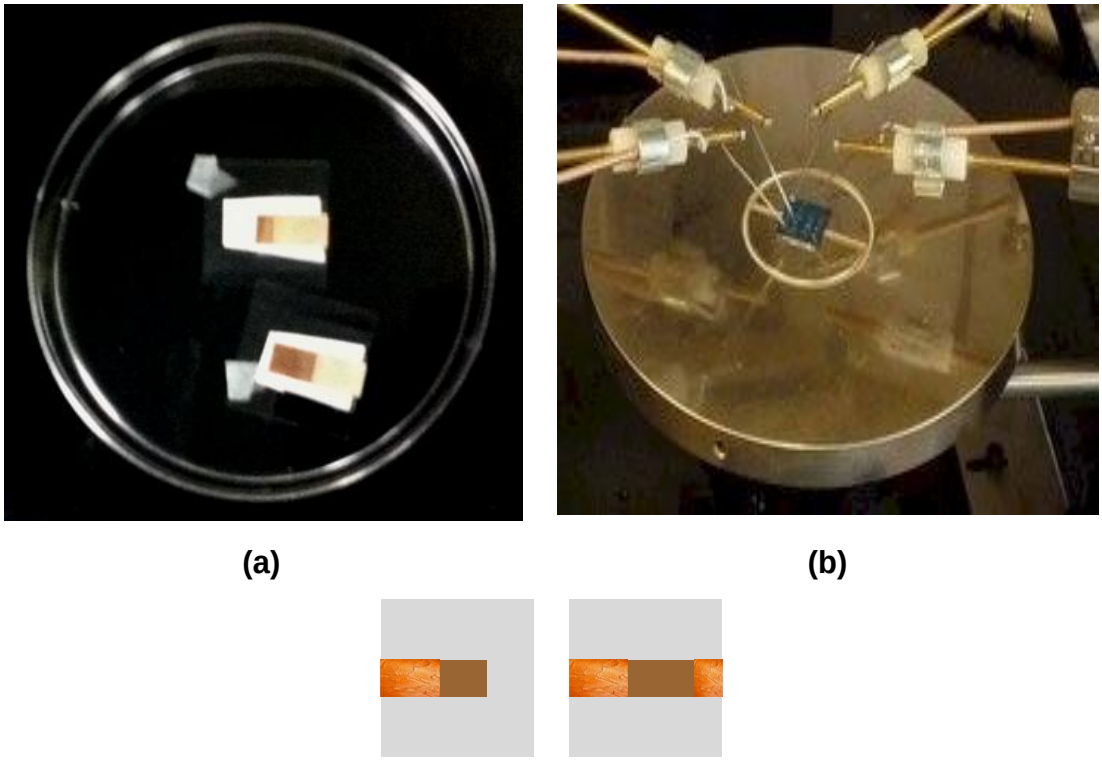

Figure 5.2 (a) shows the coated nano layered film on metal substrate for *I-V* measurement (b) shows the *I-V* and *C-V* measurement probes and (c) zoomed image of (a)

Fig 5.2 (a) shows the image of the microscopic glass slide which is taped with metal foil. APTMS/PFTS nanolayered coating is deposited on the outer side of the foil of both the slides and the device is arranged such that its makes a capacitor design (fig 5.1). May be this kind of device is useful for biosensor study which uses biochemical reactions mediated by isolated enzymes, immunosystems, tissues, cells, etc to detect chemical compounds.

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BIOGRAPHICAL INFORMATION

Sri Divya Vidyala was born in March 1988 in Kakinada and brought up in Hyderabad, India. She completed her high school in Hyderabad. With her keen interest towards education, she has excelled in her undergraduate studies ranking in top five from the Gokaraju and Rangaraju Institute of Engineering and Technology, Hyderabad, India, 2007. She was interested in pursuing a career in biomedical engineering research that could lead to biomaterial implants. Towards this goal she joined for Master's Degree in Bioengineering at University of Texas at Arlington, 2009. Her interests towards nano-biotechnology led her join Nano-Bio Lab in August 2009. Currently she is working on a novel organic nanofilm deposition method and characterized a device for coating medical implants.