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MONITORING TIME-VARYING BIOLOGICAL
IMPEDANCES AT MICROWAVE FREQUENCIES

by

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This is a thesis submitted to the Faculty of Graduate Studies
and Research at Carleton University in partial fulfillment of
the requirements for the degree of Doctor of Philosophy.

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ABSTRACT

Microwave-Doppler-radar and microwave-interferometric methods for monitoring time-varying biological impedances are described in this thesis. Both methods are based on monitoring phase changes of the signal scattered from a biological tissue region. These phase changes are an indication of the net changes within the tissue region due to various physiological processes, for example, pulsations of arteries. A number of specially designed simulators and phantoms were built for testing the developed techniques. It has been found that the detection characteristic of the Doppler radar varies sinusoidally as a function of the antenna-object spacing which is an undesirable occurrence. This occurrence was eliminated with the development of the interferometric system. Experimental results have shown that the interferometer output voltage is proportional to the object displacement in liquid muscle phantom, and to the volume changes of animal artery in the simulated model of a human thigh. The sensitivity of the system depends on the signal attenuation between the antenna and the object. Because of the interference between the signal scattered from the test region of the tissue and the signal reflected from the antenna-tissue interface, the operation of both microwave methods has been found to be limited to within a half-wavelength distance in that medium.

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TABLE OF CONTENTS

	<u>PAGE</u>
ABSTRACT	i
ACKNOWLEDGEMENTS	ii
LIST OF FIGURES	v
LIST OF IMPORTANT SYMBOLS	ix
CHAPTER 1. INTRODUCTION	1
1.1 Aims and Objectives	1
1.2 Medical Aspects	1
1.3 Existing Methods for Interrogating Biological Objects	2
1.4 Electromagnetic Waves in Biological Bodies	9
1.5 Proposed Methods for Monitoring Biological Impedances	13
1.6 Thesis Organization	14
CHAPTER 2. PRINCIPLES OF OPERATION	15
2.1 Basic Principle of Monitoring Biological Impedances	15
2.2 Doppler Radar Method	17
2.3 Microwave Interferometer Principle	19
2.4 Modified Microwave Interferometer System	25
CHAPTER 3. MICROWAVE DOPPLER RADAR FOR MONITORING BIOLOGICAL IMPEDANCES	30
3.1 A Simple Simulation of a Biological Tissue Volume: Instrumentation and Calibration	30
3.2 Results for Human Subjects	41

	<u>PAGE</u>
CHAPTER 4. MICROWAVE INTERFEROMETER SYSTEMS FOR MONITORING BIOLOGICAL IMPEDANCES	45
4.1 Transmission Characteristics of Microwave Interferometers	45
4.2 Implementation of the Microwave Interferometer System	49
4.3 Testing the Interferometer System with an Ideal Simulator	54
4.4 Testing the Interferometer System with a Simple Simulator	59
4.5 Resolution and Stability of Interferometer Systems	67
CHAPTER 5. MICROWAVE INTERFEROMETERS IN MEDICAL APPLICATIONS	74
5.1 Monitoring Changes within a Phantom by the Microwave Interferometer	74
5.2 Results for Human Subjects	85
CHAPTER 6. CONCLUSIONS AND SUGGESTIONS	88
6.1 Summary	88
6.2 Suggestions for Future Research	91
REFERENCES	93

LIST OF FIGURES

<u>FIGURE NO.</u>	<u>CAPTION</u>	<u>PAGE</u>
1.1	Attenuation of the electromagnetic energy in water and selected biological tissues. Data from [40], [54].	11
1.2	Phase constant of the electromagnetic wave in water and selected biological tissues. Data from [40], [54].	12
2.1	Simplified diagram of a monitoring device for measuring time-varying biological impedances. Γ = reflection coefficient, $\theta(t)$ = time-varying phase angle.	16
2.2	Configuration of a CW Doppler Radar as applied to monitoring time-varying biological impedances.	17
2.3	Microwave interferometer for monitoring time-varying biological impedances.	19
2.4	Characteristic curve of the interferometric bridge.	24
2.5	Schematic diagram of a superheterodyne receiver for the interferometer.	27
3.1	Dynamic phantom model of the artery.	31
3.2	Experimental set-up showing a section of cow artery installed in a plastic container.	32
3.3	An engraved 1-mm glass scale, placed close to the artery between a light source and a microscope. The scale is for measuring the displacement of the artery.	32
3.4	Experimental arrangement for calibrating arterial wall movement with an optical system.	34
3.5	Experimental set-up showing a Doppler transceiver with the antenna installed above the artery.	34
3.6	Experimental arrangement for the measurement of the detected signal from the Doppler radar.	35

<u>FIGURE NO.</u>	<u>CAPTION</u>	<u>PAGE</u>
3.7	Detected signal (RMS value) as a function of the antenna-artery distance.	37
3.8	Output signals from the Doppler radar for different antenna-artery distances (z). Chart speed = 10 mm/s.	38
3.9	Mapping arterial pulsations for different operating points on the characteristic curve of the Doppler radar.	39
3.10	Characteristic curve of the Doppler radar which operates as a converter. λ_w is the wavelength in water.	40
3.11	Recorded signals from various parts of a normal male, each of which contains a major superficial artery (10.5-GHz Doppler radar).	43
3.12	Signals from the abdominal aorta of a normal human subject (3-GHz Doppler radar). Chart speed = 25 mm/s.	44
4.1	Transmission coefficient of the 9.3-GHz magic tee vs phase difference of reflected signals at collinear arms. L_R is the return loss in the collinear arms.	47
4.2	Transmission coefficient of the 915-MHz hybrid tee vs phase difference of reflected signals at collinear arms. L_R is the return loss in the collinear arms.	48
4.3	Phase detection unit (second detector) in the microwave interferometer system.	51
4.4	Phase-locked circuit for the local oscillator.	52
4.5	Schematic diagram of the interferometer system for monitoring biological impedances.	53
4.6	A dial gage indicator for static calibration, and a LVDT for dynamic calibration in the displacement measurements of the moveable short circuit.	56

<u>FIGURE NO.</u>	<u>CAPTION</u>	<u>PAGE</u>
4.7	Amplitude spectra of the LVDT output signal (above) and the interferometer output signal (below). Object displacement = 1 mm. Attenuation setting on arm 1 = 20 dB.	57
4.8	Detection characteristic of the experimental 9.3 GHz interferometer system. The noise level is 3.6 μ V.	58
4.9	Input reflection coefficient versus the antenna-reflector spacing for the 9.3-GHz antenna (dielectric loaded cylindrical waveguide) facing the muscle phantom.	60
4.10	Input reflection coefficient versus the antenna-reflector spacing for the antenna (flanged rectangular waveguide mounted with a capacitive screw and an inductive diaphragm) facing the muscle phantom.	62
4.11	Input reflection coefficient versus the antenna-reflector spacing for the 915-MHz antenna (microstrip slot radiator) facing the muscle phantom.	64
4.12	Interferometer output voltage versus the antenna-reflector spacing. Reflector displacement = 0.5 mm. Amplifier gain = 20.	66
4.13	Simplified diagram of the receiver.	68
4.14	Relative phase resolution versus the return loss in the collinear arms for different values of input power at the H-port. The receiver bandwidth = 10 Hz.	71
4.15	Relative phase resolution versus the receiver bandwidth for different values of the return loss in the collinear arms. Input power at the H-arm = 10 mW.	72
5.1	Cross-sectional view of the middle part of human thigh (right side) [74]. Scale 3:4.	75

<u>FIGURE NO.</u>	<u>CAPTION</u>	<u>PAGE</u>
5.2	Input reflection coefficient versus the antenna-artery spacing for the cylindrical waveguide antenna (9.3 GHz) facing the muscle phantom.	77
5.3	Input reflection coefficient versus the antenna-artery spacing for the antenna (915 MHz microstrip slot radiator) facing the muscle phantom.	78
5.4	Tissue phantom around the femoral artery at the middle part of human thigh.	80
5.5	Experimental arrangement for the measurement of changes within the simulated tissue model resulting from varying the volume of the cylinder.	82
5.6	Output voltage from the 915-MHz interferometer system versus the volume increase of the artery in the simulated biological tissue model for different values of the fat-layer thickness.	84
5.7	Interferometer output signal from a radial artery (9.3-GHz system).	85
5.8	Interferometer output signal from the abdominal aorta of a normal human subject (915-MHz system). Chart speed = 10 mm/s.	86

LIST OF IMPORTANT SYMBOLS

Γ	Input reflection coefficient of the antenna
θ_o	Constant phase angle of the input reflection coefficient
$\theta(t)$	Time-varying phase angle of the input reflection coefficient
$v_T(t)$	Transmitter signal
V_T	Amplitude of the transmitter signal
ω_o	Frequency of the transmitter signal
θ_T	Phase of the transmitter signal
$v_R(t)$	Return signal
V_R	Amplitude of the return signal
$\theta_R(t)$	Time-varying phase angle of the return signal
Z_{IN}^r	Reference impedance at arm 2 of the magic tee
Z_{IN}^m	Measured impedance at arm 1 of the magic tee
[a]	Vector representing normalized incident waves at the hybrid junction ports
[b]	Vector representing normalized reflected waves at the hybrid junction ports
[S]	Scattering matrix of the hybrid junction
T	Transmission coefficient defined as the ratio of the output signal from the E-arm to the input signal at the H-arm of the hybrid junction, also temperature in Kelvin
Γ_1	Reflection coefficient at port 1
θ_1	Phase of Γ_1
Γ_2	Reflection coefficient at port 2
θ_2	Phase of Γ_2
Γ_4	Input reflection coefficient of the receiver
v_D	Detected output voltage

K	Sensitivity of the square-law detector
L_R	Return loss in the collinear arms of the hybrid junction (magic tee)
v_o	Interferometer output voltage
A	Amplitude of the interferometer output voltage
G	Amplifier gain
A_s	Amplitude of the RF source signal
A_L	Amplitude of the local oscillator signal
ω_L	Frequency of the local oscillator signal
θ_L	Phase of the local oscillator signal
ω	Intermediate frequency (IF) and the center frequency of the IF amplifier
$v_{IF}(t)$	Output signal of the IF amplifier
A_1	Amplitude of $v_{IF}(t)$
K_1	Constant depending on the mixer characteristic
ϕ_0, ψ_0	Constant phase
z	Antenna-artery distance
λ_w	Wavelength in water
λ_m	Wavelength in the medium
$v_{REF}(t)$	Reference signal
A_2	Amplitude of $v_{REF}(t)$
F	Noise figure of the receiver
B	Receiver bandwidth
S_1	Input signal power
N_1	Input noise power
S_o	Output signal power
N_o	Output noise power
k	Boltzmann's constant = 1.38×10^{-23} J/K

E_o	Time-average signal amplitude
E_n	Time-average noise amplitude
K_D	Sensitivity of the phase detector
ϵ_r^*	Relative complex permittivity
ϵ_r'	Relative dielectric constant
ϵ_r''	Relative loss factor

CHAPTER 1

INTRODUCTION

1.1 Aims and Objectives

In comparison with other techniques used in medical diagnostics [1] - [3], electromagnetic methods have only recently been reported and most of them are still in research stages. Lack of understanding of the nature of the interaction between the electromagnetic waves and the human body seems to be a major factor that has affected the development of this area. More experimental and theoretical work is therefore required in order to bring the electromagnetic methods to a level suitable for clinical applications. New areas of applications of electromagnetic radiation in medicine should also be explored so that its potentially beneficial use can be evaluated.

The objectives of this research program are to develop techniques to monitor time-varying biological impedances at microwave frequencies and to ascertain their potential use as diagnostic tools in medicine.

The emphasis of this research work is placed on experimentation in order to achieve these objectives.

1.2 Medical Aspects

One of the major causes of morbidity and mortality in man is an arterial systemic disease called atherosclerosis [4], [5]. The primary lesion of this widespread disease is the plaque which deposits on arterial walls. The plaque originates from the infiltration of cholesterol into the arterial walls [6]. As the plaques increase, the lumen of blood vessels becomes narrower and the arterial walls become increasingly more

rigid. Thick plaques can occlude vessels and cause impaired circulation of blood. Atherosclerosis is severe in certain segments of the arterial system. This includes the abdominal aorta, the thoracic aorta and common iliac arteries (right more than left). Raised plaques in carotid arteries and cerebral arteries result in neurologic dysfunction and stroke. Those in coronary arteries result in disabling cardiac disease and infarction, and those in vessels of lower limbs may result in pain, disability and possibly amputation.

Atherosclerosis is a very slowly progressing disease requiring several years to reach a clinical problem. Modern surgical techniques permit the improvement of the arterial system once the illnesses are characterized. A diagnostic instrument or method for localizing the vascular disease and quantifying its progress would facilitate the surgeons to select appropriate procedures and patients. The search for such objective methods remains an area of active research.

1.3 Existing Methods for Interrogating Biological Objects

A wide range of techniques have been developed and used for studies of disease states in the living body. It is becoming clear that many disease states result from disorders of internal organ motility. Of greatest clinical interest are the disorders of the cardiovascular system for which many new diagnostic techniques are being sought. Measurement parameters of the internal organ motility include peripheral pulse propagation speed, local temperature changes, arterial bruit¹⁾, limb volume changes, segmental blood flow and local blood velocity. In the following discussion, the main existing methods for monitoring internal motion are reviewed.

1) A sound heard when the stethoscope is placed over an artery.

Information about the vascular system may be obtained by using electromagnetic (EM) flowmeters, intravascular methods, and angiography.

In the EM flowmeter, a magnetic field is applied at right angles to the flow of blood in a vessel and an electric potential generated at right angles to the vessel and to the field is measured [1], [7]. The magnitude of the electric potential is proportional to the velocity of blood flow. To apply the instrument, the vessel has to be surgically exposed. The intravascular methods for studies of the pulse pressure wave in a blood vessel require pressure transducers to be inserted into the vessel lumen [8], [9]. This causes some discomfort and danger to the patient. Their applications are therefore limited to specific clinical situations. The angiographic technique involves injecting a radio-opaque substance into a blood vessel and producing X-ray images to observe the functional changes in the vascular system [10]. The technique gives high resolution [11], but carries some risks to the patient and does not always give a good correlation between the angiographic appearance and the functional changes [12]. The invasive nature of the technique makes it inappropriate to be used as a routine diagnostic method.

Of the diagnostic techniques, the non-invasive ones seem to be interesting because they can monitor and evaluate the internal changes or movements without invading the living subjects. There are two groups of non-invasive techniques, namely, active and passive. The active means are concerned with the use of an interrogating source which is transmitted to interact with internal structures and is subsequently detected outside the body. Examples of currently used energy sources for diagnoses are ultrasound and electromagnetic radiation which are coherent sources, and X-rays and gamma rays emitting tracers which are non-coherent. The passive

techniques involve the measurement of changes brought about in some external parameters due to the internal changes. Palpation and auscultatory methods for measurements of arterial blood pressure, and plethysmographic method for measurements of blood flow in human extremities [6] are examples of the passive means.

The familiar X-ray system has been used extensively in medical diagnoses and its principle of operation is well known [1]. Using computer reconstruction techniques, anatomical information can be formed from X-ray transmission data obtained by scanning an area from many directions [1], [13], [14]. These techniques have currently been used in many hospitals for diagnoses of human brain, chest and abdomen as well as the cardiovascular system. Even though the X-rays can provide very good spatial resolution in medical imaging, they are quite hazardous because they can cause ionization of biological media. The X-ray source also has an additional limitation since most of the energy is absorbed in some tissues such as bone as demonstrated by many radiographs. Consequently, it is difficult to differentiate other tissues which look more or less similar. Due to the lack of contrast, the X-ray techniques are not widely used in the detection of pulmonary edema¹⁾ [15] - [17].

Scintigraphy has been used in nuclear medicine for the diagnosis of disease and for assessment of the internal changes of the patient [1], [18]. In this techniques, a scintillation camera is used to detect the gamma rays emitted from a radioactive source which is initially introduced into the body. The blue light, scintillated from crystals in the camera due to striking of gamma rays upon them, is photo-multiplied and displayed as an image. This technique has recently been used to study heart motion [19] and lung ventilation [20].

1) Excessive accumulation of fluid in the lung tissues, thus causing swelling.

The computer reconstruction techniques can also be applied in nuclear medicine for tomographic reconstruction of images [1], [21]. In this case, the images are formed using the data obtained from the detection of radioactive distributions inside the body. The techniques have recently been used to study topographical changes of regional cerebral blood flow and metabolism [22], [23]. The nuclear-medicine imaging techniques provide information on organ functions rather than simple organ morphology¹⁾ which is obtained from the X-rays. Therefore, they are more suitable for the dynamic studies of internal changes. However, there is still some hazard due to the ionizing radiation.

Nuclear magnetic resonance (NMR) techniques have been used to measure blood flow in the extremities²⁾, the brain and certain regions of the heart [24], [25]. The technique is based on the fact that the hydrogen nuclei in the water fraction of blood act as small magnets which are usually oriented in a random fashion. In the presence of an external magnetic field, they become aligned. Blood entering the NMR flow meter is first fully magnetized by a steady field. The magnetization of the blood which is proportional to the blood flow is then detected in the distal artery by a receiver coil. A NMR imaging method for the display of cross sectional anatomy has been developed recently [26]. Even though the NMR is non-invasive and has no known hazard, its performance and applicability for clinical use remain to be improved.

Optical methods can also be used in medical diagnoses. This includes the techniques based on Doppler shift principle for measuring cutaneous blood flow [27] and photoplethysmography for detecting occlusions in carotid arteries [1], [28]. Due to the high attenuation of the skin at optical frequencies [29], these methods are limited to superficial

1) The science of anatomy.

2) Arms or legs.

arteries.

Ultrasonic methods have found widespread applications in medicine and the techniques are well known [2], [30]. Ultrasound is a mechanical wave which can be produced coherently by exciting a piezoelectric transducer with a signal of frequency higher than that which is normally audible. For motion detection, it can be used non-invasively in either transmission, reflection or Doppler reflection mode. These all involve transmitting a beam of ultrasonic energy (usually in the range 5 - 20 MHz) and measuring the properties of the transmitted or reflected waves. Examples of ultrasound diagnostic methods include the study of the motion of mitral valves [31], the measurement of instantaneous blood flow [1], the measurement of the speed of contraction of the heart wall [32], the monitoring of arterial wall movements [33], the measurement of urine velocity and the detection of obstructions in the urethra [34]. Methods of ultrasonic scanning for studies of the internal structure of the human body are very well described in the literature [1], [2], [30]. Newly developed techniques of ultrasonic computerized tomography for detection and diagnosis of myocardial infarction¹⁾ and cancer in the breast have been reported recently [35], [36].

The health hazards involved in ultrasonic methods are minimal as very low power levels are used in diagnostic applications. Ultrasound is highly attenuated in the soft lung tissue, air and bone [2], [30] and thus it is incapable of monitoring some internal disorders, such as lung disease [3]. However, it does have shorter wavelengths than microwaves in biological media and hence can provide better spatial resolution.

Diagnostic ultrasound can be used to obtain images of most internal objects except at locations where there are some bone or gas-contained tissues.

- 1) Damage of heart muscle secondary to the loss of its blood supply, as in coronary thrombosis.

Volumetric changes inside the body can be measured by using low-frequency electromagnetic methods (20 - 100 kHz) known as electrical-impedance plethysmographic (EIP) techniques [1], [3]. The methods are based on attaching electrodes to the segment of the test tissue and measuring its impedance. The changes in impedance provide a rough estimate of internal changes. Applications include monitoring the fluid accumulation in the lungs, measuring blood flow in the limbs and detecting changes in the cardiac output. The EIP techniques are non-invasive and relatively simple to use, but their sensitivity and accuracy leave much to be improved in regard to their applications. The hazard involved in the use of these techniques is minimal since very low currents (1 - 4 mA) are employed.

Radio frequency (RF) and microwave methods for diagnostic applications in medicine have emerged in recent years. The methods make use of the semi-transparent property of biological tissues at certain frequencies. Knowledge about electromagnetic properties of biological materials [29], [37] - [40] has been utilized extensively for these purposes. The RF method for diagnostic applications was first suggested by Moskalenko [41]. In his work, the changes of RF absorption within the human body were measured and correlated to significant parameters such as blood or respiratory volume changes.

Lung diseases can be monitored by using microwave methods [15] - [17]. Low intensity microwave energy is transmitted into the body chest via an applicator and the diseases which modify the permittivity and conductivity of the lung tissues are diagnosed from the amount of microwave energy reflected from or transmitted through the lung.

Microwave methods based on the Doppler principle have been used

successfully to detect apnea spells in sleeping infants [42] and also to record apexcardiogram [43].

Microwave thermographic techniques for the detection of breast cancer and tumors in the neck and brain area have been reported recently [44] - [47]. The methods are based on the detection of thermal radiation at microwave frequencies. Several medical conditions such as cancer and vascular occlusions which can result in local changes in temperature of the order of 1°C above the surrounding healthy tissue are characterized from the body's emitted radiation.

Several attempts have been made to view a body section using microwave imaging techniques [3], [48], [49]. However, these are still in the development stage and only algorithms and numerical results have been presented so far.

All RF and microwave methods for diagnostic applications require matched antennas for transmitting and/or receiving purposes. Failure to achieve this would result in a radiation leakage and a multipath propagation as demonstrated by Yamaura [50]. A number of antennas for medical diagnostic applications have been reported recently [51].

The limited number of diagnostic applications of electromagnetic energy at microwave frequencies for biological interrogation may be somewhat surprising in comparison to its various applications in other areas. This may be due to the fact that at short wavelengths in the microwave region, where spatial resolution of an imaging system would be best, the attenuation of energy in the high water-content tissues is so great that the detection of transmitted or returned signal becomes impractical. However, if the frequencies of operation are lowered or if

the internal objects are moving, the detection of a transmitted or returned signal seems to be possible. The second problem which creates difficulties for microwave interrogating systems is the complexity and lack of understanding of the nature of the interaction between electromagnetic waves and the human body. This problem has slowed the development in this area. Other problems include lack of a reasonably adequate human model for theoretical and experimentation studies.

However, the RF and microwave techniques possess some advantages in diagnostic applications over the X-ray and ultrasound methods; for example, monitoring changes in lung water content [15] - [17]. The superior sensitivity of the microwave techniques in this area has encouraged scientists to develop the microwave imaging [3], [48], [49]. It is now possible to continuously monitor internal changes without any hazard. It should be underlined that risks involved in using microwaves for medical diagnosis are comparable to those associated with ultrasound, and are substantially smaller than those resulting from the X-ray and nuclear methods.

1.4 Electromagnetic Waves in Biological Bodies

Biological bodies are known as complex electromagnetic media whose characteristics are randomly varying in time, frequency and space [52]. The propagation of electromagnetic waves in these media involves reflection, transmission, diffraction and scattering, depending on the dielectric properties of various organs and tissues, as well as their sizes and geometries [29], [53]. Also, the wave propagated in a given tissue medium is both attenuated in amplitude and changed in phase; that is the wave energy is absorbed within the medium and the wave

velocity is altered. The absorption and velocity of wave in the medium are commonly described in terms of the attenuation and phase constants.

Using the available dielectric data [40], [54], the values of the attenuation and phase constants for different tissues as well as for water are plotted as a function of frequency in Figures 1.1 and 1.2.

The attenuation is high in tissues with high water content such as skin, muscle and blood. Fat and bone have lower water content and have lower attenuation.

The studies on the diagnostic applications of electromagnetic waves require good understanding of the interaction between these waves and biological tissues. Such interaction has been studied extensively for different tissue models exposed to plane waves [29], [53]. However, the situation is different and much more complex in the diagnostic applications in which the antenna is placed in contact with the skin to transmit the electromagnetic energy into the tissue and receive signals reflected from the underlying tissue discontinuities. In this case, the radiation from the antenna is predominantly in the near-field and as a result the waves are highly attenuated.

The interaction between electromagnetic fields and human tissues in the vicinity of the antenna (near-field) is very complicated and its nature is not clearly understood. The complication of the interaction arises from the complexity of the body structure which varies from person to person. Also, the near field irradiation in the tissues depends on the type of radiating antenna. In an attempt to solve this complicated problem one may start by studying the scattering of waves at tissue interfaces such as biological objects imbedded in fat or muscle.

Near-field scattering from simple geometrical objects in free

Figure 1.1. Attenuation of the electromagnetic energy in water and selected biological tissues. Data from [40],[54].

Figure 1.2 Phase constant of the electromagnetic wave in water and selected biological tissues. Data from [40],[54].

space has been studied theoretically and experimentally by a number of investigators [55], [56]. Analysis of the scattered fields from these objects is quite complicated and involves a lot of computation. The situation becomes even worse if scatterers are biological objects imbedded in a lossy medium.

Since the study of the near-field interaction between electromagnetic waves and human tissues is not an objective of this research, the investigation will not be pursued here.

1.5 Proposed Methods for Monitoring Biological Impedances

Two techniques have been developed to monitor time-varying biological impedances. The first one is based on application of a microwave Doppler radar to measure the phase change of the signal backscattered by the biological tissue region adjacent to the antenna. In the second method, a microwave interferometer system is used to monitor the phase change of the reflection coefficient at the input of the antenna placed in contact with the test region. The backscattered signal and the reflection coefficient are related and their phase changes vary proportionally to the net changes within the tissue region.

Microwave Doppler radar has been used for many years in navigation, traffic control and various military applications [57]. The potential applications of microwave Doppler radars in medicine have been reported recently [58], [59].

Microwave interferometers based on hybrid junctions have also been used in many applications, among which are the detection of surface irregularities of the commutators and slip rings during the operation of the associated machines [60], the monitoring of the thickness of metal

tape [61], and the measurement of mechanical displacements of stationary or moving objects [62], [63]. The applications of the microwave interferometer in measuring the movements associated with human physiological functions such as respiration, muscle action, etc., have also been reported [64] [65]. However, its applications for monitoring changes within the human body which could be useful for medical diagnosis have not been explored.

1.6 Thesis Organization

This thesis is divided into two parts. The first part, described in Chapter 2, is a discussion of the basic concept of monitoring biological impedances and the basic principles and mathematical analysis of the proposed methods. The second part is concerned with the descriptions of the experimental work and includes the experimental arrangements with specially designed simulators, phantoms, and on normal human subjects as well as the experimental results. The purpose of this work was to determine the capabilities and limitations of both methods as applied to monitoring biological impedances and to ascertain their potential applications in medicine. This part of the thesis is presented in the subsequent Chapters. Chapter 3 is devoted to the descriptions of the work done with microwave Doppler radar, whereas Chapter 4 discusses the details of the implementation and experimentation with microwave interferometer systems. A detailed description of testing the interferometer system using a phantom is given in Chapter 5 together with experimental results for human subjects. The capabilities and limitations of both techniques for monitoring biological impedances and, therefore, changes within a tissue region are summarized in Chapter 6, together with final conclusions and suggestions for future research.

CHAPTER 2

PRINCIPLES OF OPERATION

2.1 Basic Principle of Monitoring Biological Impedances

The term *biological impedance* is introduced here in order to describe the net change of properties within a region of biological tissue. When an electromagnetic wave is transmitted through a matched antenna into a tissue region adjacent to the antenna, backscattered waves are generated and are returned to the antenna. These waves originate at discontinuities in the tissue region. Such discontinuities exist at interfaces between tissues of different electromagnetic properties. The returned waves may be described in terms of impedance or reflection coefficient seen at the antenna input plane. The impedance at this plane is defined as the biological impedance. The complex nature of this impedance or reflection coefficient is derived from the complex electromagnetic properties of biological tissues in the region in which electromagnetic waves propagate.

Biological tissue regions contain arteries, the walls of which are in motion due to the pulse pressure wave transmitted from the heart. The arterial wall movement results in changes in the amplitude and phase of each return wave and hence, of the biological impedance or the reflection coefficient. The change in the magnitude of the return wave is attenuated in the tissue region while the change in phase remains almost unaltered.

The method of monitoring time-varying biological impedances described in this thesis is based on the measurement of phase changes of the return waves or the reflection coefficient defined at the input of a matched antenna placed in contact with the given tissue region. The basic configuration of a monitoring device as applied to this measurement is

shown in Figure 2.1. The monitoring device is basically a converter which provides an output signal proportional to the phase change. The antenna must be matched to the surface of the test tissue region so that electromagnetic energy can be coupled without reflections. The antenna length should be as short as possible to avoid multiple reflections.

Figure 2.1 Simplified diagram of a monitoring device for measuring time-varying biological impedances.

Γ = reflection coefficient.

$\theta(t)$ = time-varying phase angle.

2.2 Doppler Radar Method

The basic principle of a continuous wave (CW) Doppler radar as applied to monitoring time-varying biological impedances is shown in Figure 2.2. In this configuration a transmitter sends out a continuous

Figure 2.2 Configuration of a CW Doppler radar as applied to monitoring time-varying biological impedances.

microwave signal through a matched antenna and a receiver continuously detects the return signals which carry information on the phase change due to variations within a test medium [57]. The receiver consists of a circulator

which isolates the outgoing and incoming signals, and a mixer which is normally a Schottky barrier diode. The transmitter signal may be expressed as

$$v_T(t) = V_T \exp[j(\omega_o t + \theta_T)] \quad (2.1)$$

where V_T , ω_o and θ_T are amplitude, frequency and phase of the transmitter signal, respectively.

At the antenna input, all return signals may be represented by a single signal which can be written as

$$V_R(t) = V_R \exp[j(\omega_o t + \theta_R(t))] \quad (2.2)$$

where V_R and $\theta_R(t)$ are its amplitude and phase, respectively. The amplitude V_R depends upon V_T , the structure of biological tissues adjacent to the antenna and tissue attenuation characteristics at ω_o .

The signal $v_R(t)$ arrives at the mixer input via the circulator and is heterodyned in the mixer with a portion of the transmitter signal [57]. After filtering unwanted components out, the resulting signal is [66]

$$\begin{aligned} v(t) &= V \cos(\pi + \theta_T - \theta_R(t)) \\ &= V \cos(\pi + \theta_o + \theta(t)) \end{aligned} \quad (2.3)$$

where V and θ_o are constants, and $\theta(t)$ is a time-varying phase change.

The constant V depends upon the sensitivity of the receiver, V_T and V_R . The constant θ_o depends upon frequency ω_o and the structure of the test biological tissue.

For a small phase change, $\sin\theta(t) \approx \theta(t)$, and (2.3) may be rewritten as

$$v(t) \approx -V \cos\theta_o + (V \sin\theta_o) \theta(t) \quad (2.4)$$

It is evident from (2.4) that the output signal of a Doppler

radar is directly proportional to the phase change $\theta(t)$ when $\theta_0 = (2n+1)\frac{\pi}{2}$, where $n = 0, 1, 2, \dots$. However, when θ_0 is in the vicinity to $n\pi$, a large D.C. component and the distortion of the A.C. component of $v(t)$ will occur. A similar result was found by [67] for the case of a Gunn diode oscillator acting as a load variation detector. In the latter case, the D.C. (bias) current was analysed and measured as a function of the phase of the load impedance and was found to be sinusoidal.

2.3 Microwave Interferometer Principle

An effective method of measuring time-varying biological impedances is to use a microwave interferometer as a monitoring device. A block diagram of this method is shown in Figure 2.3. The microwave inter-

Figure 2.3 Microwave interferometer for monitoring time-varying biological impedances.

ferometer is an impedance bridge consisting of an RF source, a test element, a reference element and a detector-amplifier (receiver), all connected to a 4-port hybrid junction which may be a magic tee. The test element connected to arm 1 of the hybrid junction is a matched antenna which is in contact with a biological tissue region. The reference element of variable input impedance, consisting of a variable attenuator and a moveable short circuit, is connected to arm 2 of the hybrid junction. A signal from the R-F source is transmitted into arm 3 and is divided equally in amplitude and phase into two waves which enter the collinear arms (arms 1 and 2). The hybrid junction is made such that there is no direct transmission of the wave between arm 3 and arm 4. Therefore, only reflected waves from the collinear arms enter arm 4 with opposite phase. When the impedances seen looking into arms 1 and 2 are equal, no power reaches the detector and the bridge is balanced. The reference impedance Z_{IN}^r seen looking into arm 2 can be varied by changing the value of attenuation and positioning the moveable short circuit. This arrangement allows for operation on a linear portion of the characteristic of the bridge so that the output voltage of the receiver is linearly proportional to the phase change of the reflected wave in arm 1 - a controllable performance which can not be obtained in the Doppler radar.

The interferometric bridge described in this thesis operates in an off-balance mode at which the difference between Z_{IN}^m and Z_{IN}^r is measured, where Z_{IN}^m is the impedance seen looking into arm 1.

The reflected waves and incident waves at the hybrid junction ports are related by [68]

$$[b] = [S] [a] \quad (2.5)$$

where

$$[b] = \begin{bmatrix} b_1 \\ b_2 \\ b_3 \\ b_4 \end{bmatrix} \quad (2.6)$$

is the vector representing normalized reflected waves b_1 , b_2 , b_3 and b_4 at ports 1, 2, 3, and 4, respectively.

$$[a] = \begin{bmatrix} a_1 \\ a_2 \\ a_3 \\ a_4 \end{bmatrix} \quad (2.7)$$

is the vector representing corresponding normalized incident waves, and

$$[S] = \begin{bmatrix} S_{11} & S_{12} & S_{13} & S_{14} \\ S_{21} & S_{22} & S_{23} & S_{24} \\ S_{31} & S_{32} & S_{33} & S_{34} \\ S_{41} & S_{42} & S_{43} & S_{44} \end{bmatrix} \quad (2.8)$$

is the general-form scattering matrix of the hybrid junction. The transmission of power from the source via arms 1 and 2 to the detector (receiver) may be described in terms of the transmission coefficient which is defined as

$$T = \frac{b_4}{a_3} \quad (2.9)$$

If the collinear arms have reflection coefficients Γ_1 and Γ_2 , seen looking into ports 1 and 2, respectively, then

$$a_1 = \Gamma_1 b_1, \quad (2.10)$$

and
$$a_2 = \Gamma_2 b_2. \quad (2.11)$$

Using equations (2.5) - (2.11), the transmission coefficient for

the hybrid junction is

$$T = \frac{S_{41}\Gamma_1[S_{12}S_{23}\Gamma_2 + S_{13}(1-S_{22}\Gamma_2)] + S_{42}\Gamma_2[S_{13}S_{21}\Gamma_1 + S_{23}(1-S_{11}\Gamma_1)] + S_{43}M}{(1-S_{44}\Gamma_4)M - S_{41}\Gamma_1[S_{12}S_{24}\Gamma_2\Gamma_4 + S_{14}\Gamma_4(1-S_{22}\Gamma_2)] - S_{42}\Gamma_2[S_{14}S_{21}\Gamma_1\Gamma_4 + S_{24}\Gamma_4(1-S_{11}\Gamma_1)]} \quad (2.12)$$

where $M = (1-S_{11}\Gamma_1)(1-S_{22}\Gamma_2) - S_{12}S_{21}\Gamma_1\Gamma_2$, and $\Gamma_4 = \frac{a_4}{b_4}$ is the input re-

flection coefficient of the receiver.

If the hybrid junction is a magic tee, the scattering matrix is given by [68]

$$[S] = \frac{1}{\sqrt{2}} \begin{bmatrix} 0 & 0 & 1 & 1 \\ 0 & 0 & 1 & -1 \\ 1 & 1 & 0 & 0 \\ 1 & -1 & 0 & 0 \end{bmatrix} \quad (2.13)$$

Substituting scattering parameters from (2.13) into (2.12) yields

$$T = \frac{\frac{1}{2}(\Gamma_1 - \Gamma_2)}{1 - \frac{\Gamma_4}{2}(\Gamma_1 + \Gamma_2)}$$

or

$$b_4 = \frac{\frac{a_3}{2}(\Gamma_1 - \Gamma_2)}{1 - \frac{\Gamma_4}{2}(\Gamma_1 + \Gamma_2)} \quad (2.14)$$

Assume that the detector is matched and operates in the square-law region. Thus, $\Gamma_4 = 0$, and the detected output voltage is

$$\begin{aligned} v_D &= K|b_4|^2 \\ &= \frac{K}{4}|a_3|^2(|\Gamma_1|^2 + |\Gamma_2|^2 - \Gamma_1\Gamma_2^* - \Gamma_1^*\Gamma_2) \end{aligned} \quad (2.15)$$

where K is a constant depending on the diode characteristic, and Γ_1^* and

Γ_2^* are the complex conjugates of Γ_1 and Γ_2 , respectively. In general,

Γ_1 and Γ_2 are complex and can be expressed as

$$\Gamma_1 = |\Gamma_1| e^{j\theta_1} \quad (2.16)$$

$$\Gamma_2 = |\Gamma_2| e^{j\theta_2} \quad (2.17)$$

If the attenuator in arm 2 is adjusted such that $|\Gamma_1| = |\Gamma_2| = |\Gamma|$, then (2.15) simplifies to

$$v_D = \frac{K}{2} |a_3|^2 |\Gamma|^2 [1 - \cos(\theta_1 - \theta_2)] \quad (2.18)$$

Assuming the amplifier gain to be G , the output voltage of the interferometer is

$$v_o = A(1 - \cos\theta_o) \quad (2.19)$$

where

$$A = \frac{KC}{2} |a_3|^2 |\Gamma|^2 \quad (2.20)$$

and

$$\theta_o = \theta_1 - \theta_2 \quad (2.21)$$

Figure 2.4 illustrates v_o varying as a function of θ_o for a constant amplitude A . This is the characteristic curve of the interferometer when the bridge element (hybrid junction) is a magic tee with port 4 terminated by a matched square-law detector. The response of the bridge to small changes in θ_o is fairly linear for θ_o in the vicinity of $(2n + 1)\frac{\pi}{2}$, where $n = 0, 1, 2, \dots$. It is in these pseudo-linear response regions that the sensitivity of the interferometer is maximum. Suppose the time-varying phase change of Γ_1 is given by

$$\theta(t) = \theta_m \sin \omega_m t \quad (2.22)$$

where θ_m and ω_m are the amplitude and frequency of the phase change, respectively. With the short circuit in arm 2 of the bridge positioned such that

$$\theta_{o, \text{average}} = (2n+1)\frac{\pi}{2},$$

Figure 2.4. Characteristic curve of the interferometric bridge.

(2.19) becomes

$$\begin{aligned}
 v_o(t) &= A[1 - \cos\{(2n+1)\frac{\pi}{2} + \theta(t)\}] \\
 &= A[1 \pm \sin\theta(t)]
 \end{aligned}
 \quad (2.23)$$

The plus sign implies that the response is at a positive-slope linear region of the characteristic curve and vice versa. For a small θ_m , (2.23) can be approximated by

$$v_o(t) \approx A \pm A\theta(t) \quad (2.24)$$

The difference between values of AC component obtained from (2.24) and

(2.23) is within 5% of that obtained from (2.24) for $\theta_m \leq 30^\circ$. Changes within tissue regions are generally small and result in small phase changes of return signals, and, hence, that of the reflected wave in arm 1 of the magic tee. Thus, the approximation of $v_o(t)$ given in (2.24) is sufficient for this application.

The output signal from the interferometer is, therefore, directly proportional to the instantaneous phase change of the reflection coefficient and, hence, to the net change within the tissue region being monitored. The proportionality factor depends upon the sensitivity of the receiver, KG, the power input at port 3, $|a_3|^2$, and the power reflected from the antenna, $|\Gamma|^2$. The last factor is related to the return signals from the tissue region being monitored. The procedure of measuring the instantaneous phase change of the biological impedance can be summarized as follows:

1. Place the matched antenna in contact with the biological tissue region to be monitored.
2. Adjust the attenuator and the short circuit on arm 2 of the magic tee until the average output voltage is minimum. When this condition is met, $|\Gamma_1| = |\Gamma_2|$ and $\theta_1 = \theta_2 + 2n\pi$, where $n = 0, 1, 2, \dots$
3. Readjust the short circuit until the average output voltage is half of its maximum and minimum values. When this condition is met, the interferometer operates on a linear region of its characteristic curve. At this point, the AC component of the output voltage is directly proportional to the time-varying phase change of the reflection coefficient in arm 1.

2.4 Modified Microwave Interferometer System

The receiver of the interferometer system described in the previous section has a square-law diode detector at the input. This is

not a highly sensitive receiver because of the flicker noise at lower IF frequencies [57]. Flicker noise occurs in semiconductor devices such as diodes and the noise power produced by this effect varies inversely with frequency ($1/f$). With zero IF frequency, the flicker noise is very large and reduces the sensitivity of the interferometer. If the reflected power from the observed tissue region is small, which is usually the case because of high attenuation, the detection of small changes within that region by this simple monitoring system becomes difficult. This is evident from (2.19) and Figure 2.4 as $|\Gamma|^2$ becomes smaller. To improve the detection capability, a low noise receiver is needed. This may be achieved either by using an RF amplifier or by replacing the diode detector by a superheterodyne system. For the same performance, the latter seems to have the advantage because of its lower cost, particularly at higher microwave frequencies (X-band). The superheterodyne system is therefore chosen. It consists of a mixer, a local oscillator and an IF amplifier. This input stage is followed by a second detector and an audio amplifier forming a new receiver of the interferometer (Figure 2.5) - a configuration which is well known in radar applications [57]. The mixer and local oscillator (LO) convert the RF signal from the E-arm of the magic tee to an intermediate frequency (IF). The mixer is usually a balanced or double-balanced type so that the noise accompanying the LO signal can be reduced [57].

If the mixer input is matched, then, (2.14) becomes

$$b_4 = \frac{a_3}{2} (\Gamma_1 - \Gamma_2) \quad (2.25)$$

Figure 2.5 Schematic diagram of a superheterodyne receiver for the interferometer.

The signals from the RF source and LO may be expressed as

$$a_3 = A_s \exp[j(\omega_o t + \theta_T)] \quad (2.26)$$

$$v_L(t) = A_L \exp[j(\omega_L t + \theta_L)] \quad (2.27)$$

where A_s , ω_o and θ_T are the amplitude, frequency and phase of the RF source signal, and A_L , ω_L and θ_L are the amplitude, frequency and phase of the LO signal, respectively. If the center frequency of the IF amplifier is ω , then, the LO should be tuned so that $\omega_L = \omega_o \pm \omega$. Using (2.16), (2.17),

and (2.26), equation (2.25) can be written as

$$b_4 = \frac{A}{2} [|\Gamma_1| \exp\{j(\omega_o t + \theta_T + \theta_1)\} - |\Gamma_2| \exp\{j(\omega_o t + \theta_T + \theta_2)\}] \quad (2.28)$$

The signals b_4 and $v_L(t)$ are heterodyned in the mixer. Normally, the signal b_4 is much smaller than $v_L(t)$. The output signal of the mixer includes both input signals, a DC component, and a number of harmonics [66], [69]. The frequency of the output signal contains the sum and difference of the LO and RF signal frequencies. The mixer output signal is fed into the IF amplifier and all components, other than the one whose frequency is $|\omega_L - \omega_o|$, are filtered out. If the $|\omega_L - \omega_o|$ component is passed through, the output signal of the IF amplifier is

$$v_{IF}(t) = K_1 A_s [|\Gamma_1| \exp\{j(\omega t + \theta_L - \theta_T - \theta_1)\} - |\Gamma_2| \exp\{j(\omega t + \theta_L - \theta_T - \theta_2)\}] \quad (2.29)$$

where K_1 is a constant which depends upon the mixer diode characteristic, the LO signal amplitude, and the gain and load impedance of the IF amplifier. Thus, the superheterodyne stage acts as a frequency converter which converts the RF signal of frequency ω_o to the IF signal of frequency ω retaining the phase information.

Changes within the tissue region result in changes of θ_1 as a function of time. By letting

$$\phi_1(t) = \theta_L - \theta_T - \theta_1(t) = \phi_o + \theta(t) \quad (2.30)$$

$$\text{and} \quad \phi_2 = \theta_L - \theta_T - \theta_2, \quad (2.31)$$

The real part of (2.29) may be rewritten in a trigonometric form as

$$v_{IF}(t) = K_1 A_s [|\Gamma_1| \cos(\omega t + \phi_1(t)) - |\Gamma_2| \cos(\omega t + \phi_2)], \quad (2.32)$$

where ϕ_o is a constant and $\theta(t)$ is the time-varying phase change given by (2.22). Once again, if the attenuator in arm 2 of the magic tee is

adjusted such that $|\Gamma_1| = |\Gamma_2| = |\Gamma|$, then (2.32) is simplified to

$$v_{IF}(t) = K_1 A_s |\Gamma| [\cos(\omega t + \phi_1(t)) - \cos(\omega t + \phi_2(t))]. \quad (2.33)$$

Furthermore, with the short circuit in arm 2 positioned such that

$$\phi_2 = 2n\pi + \phi_0, \quad (2.34)$$

where $n = 0, 1, 2, \dots$, $v_{IF}(t)$ decreases to minimum and is zero for $\theta(t) = 0$. However, if the short circuit is readjusted to

$$\phi_2 = (2n+1)\pi + \phi_0, \quad (2.35)$$

the signal $v_{IF}(t)$ reaches maximum and can be approximately expressed as

$$v_{IF}(t) = 2KA_s |\Gamma| \cos(\omega t + \phi_1(t)). \quad (2.36)$$

The IF signal given by (2.36) is fed into the second detector where the phase change $\theta(t)$ is detected. Details of the second detector are given in Chapter 4. The audio amplifier should have its bandwidth wide enough to cover significant harmonics of the detected signal. The amplifier gain should be adjustable and high enough so that the output signal can be displayed and recorded.

The procedure of monitoring the phase change of the biological impedance using the modified microwave interferometer is fairly straightforward. After placing the antenna in contact with the desired tissue region, the attenuator and short circuit in arm 2 of the magic tee are adjusted to obtain the minimum IF signal. Under this condition, $|\Gamma_1| = |\Gamma_2|$. Then, the short circuit is readjusted until the output signal of the IF amplifier is maximum. When this is done, the second detector is ready to be adjusted to obtain the output signal which is linearly proportional to the phase change of the biological impedance.

CHAPTER 3MICROWAVE DOPPLER RADAR FOR MONITORINGBIOLOGICAL IMPEDANCES

The potential application of a low power microwave Doppler radar for monitoring changes within a human body has been demonstrated experimentally with a specially designed simulator and on normal human subjects. Because of simplicity of design and availability of components, the developed experimental system employed a Microwave Associate MA86656A Doppler transceiver with a Gunn diode oscillator, a ferrite circulator and a Schottky diode mixer integrated in a simple waveguide package. The transceiver operates at a frequency of 10.5 GHz with approximately 10 mW output power. It was used with a dielectric-loaded cylindrical waveguide antenna. The antenna was designed to be matched to the human skin.

3.1 A Simple Simulation of a Biological Tissue Volume:Instrumentation and Calibration

The objectives of this experiment were the following:

- 1) To study how the output signal from the Doppler radar varied with the antenna-artery distance,
- 2) To determine the characteristic curve of the Doppler radar which was used as a converter in this application, and
- 3) To estimate the power reflection coefficient of the artery when it is imbedded in a high water-content tissue.

The simulation of a dynamic phantom model was done in a plastic

box, containing distilled water, which has electromagnetic properties only slightly different from human muscle or blood at 10.5 GHz (Figures 1.1 and 1.2). A section of a bovine artery of approximately 6 mm diameter was inserted between two plastic tubes which were fixed perpendicularly to the two opposite side walls of the box (Figures 3.1 and 3.2). The other ends

Figure 3.1 Dynamic phantom model of the artery

of both plastic tubes were connected to a reciprocating pump by plastic tubings, forming a closed system which was filled with distilled water. The pump caused the artery to expand in a pulsating manner. The amplitude of pulsations can be adjusted by changing the stroke length of the pump.

Figure 3.2 Experimental set-up showing a section of cow artery installed in a plastic container.

Figure 3.3 An engraved 1-mm glass scale, placed close to the artery between a light source and a microscope. The scale is used for measuring the displacement of the artery.

The arterial pulsating amplitude was measured using an optical system. A piece of glass slide with 1-mm engraved scale (0.01 mm per division) was placed in front of the artery. A narrow light beam was transmitted through the artery and the slide to a microscope (Olympus 200043) which was set up outside the box (Figures 3.3 and 3.4). The scale on the micrometer eyepiece lens (OSM202711) of the microscope was then calibrated by focusing the microscope on the slide. With the pump operated at full speed, 60 RPM (close to the normal heart rate of human), the arterial pulsating amplitude was detected on the eyepiece lens. In this experiment, the pulsating amplitude of the artery was adjusted to 0.3 mm which is a typical value of that in human peripheral arteries [33].

The experiment with the Doppler radar was then performed. The Doppler transceiver fitted with the cylindrical waveguide antenna was installed on a platform above the box (Figures 3.1 and 3.5). The transceiver and the antenna can be raised or lowered by using a small jack mounted on the platform. The distance from the antenna to the artery was monitored by a dial depth indicator (Mitutoyo 7212). The signal processing arrangement shown in Figure 3.6 consists of an amplifier, an oscilloscope, a true rms voltmeter and a strip chart recorder. The antenna was immersed in water whose level was about 2 cm above the artery. At a certain distance (z) between the antenna and the artery, the box, which was placed on a moveable platform, was positioned until the detected signal was maximum as indicated on the oscilloscope and the true rms voltmeter. The antenna was then positioned vertically, starting from $z = 0$ to $z = 8$ mm in 0.0635 mm (0.0025 in.) increments, and the detected signal was monitored and recorded by the true rms voltmeter and the strip chart recorder. The $z = 0$ distance here is defined as the position at which the

Figure 3.4 Experimental arrangement for calibrating arterial wall movement with an optical system.

Figure 3.5 Experimental set-up showing a Doppler transceiver with the antenna installed above the artery.

Figure 3.6 Experimental arrangement for the measurement of the detected signal from the Doppler radar.

antenna is just in touch with the artery surface when it was fully expanded. In this experiment, the Doppler transceiver detected the changes within the phantom model which were caused only by the pulsations of the artery.

Figure 3.7 illustrates the variation of the detected signal from the Doppler radar as a function of the antenna-artery distance. Data were obtained from the true rms voltmeter and the dial gage. Examples of detected signals are shown in Figure 3.8 for different values of the antenna-artery distance. These signals correspond to points A, B and C in Figure 3.7, respectively. The signal A and the signal C were obtained when the Doppler radar operated on the upper and lower nonlinear regions of the characteristic curve while the signal B was obtained from a linear mapping with the real pulsations of the artery (Figure 3.9) but having a 180° phase change. This is in agreement with the predicted response (2.4) provided that the constant phase θ_0 and the amplitude V in (2.4) vary linearly with the antenna-artery distance.

The decreasing amplitude of the curve shown in Figure 3.7 is due to the attenuation of the wave in water. The value of the wave attenuation was found to be 43.6 dB/cm which is very close to that shown in Figure 1.1 (43 dB/cm) for the same frequency (10.5 GHz) and temperature (25°C).

Using the informations from Figures 3.7 through 3.9 and from other recorded signals, the characteristic curve of the Doppler transceiver may be sketched as shown in Figure 3.10. In this experiment, the phantom model contained only one object whose pulsation was sinusoidal since it was derived from a known source (the pump). Therefore, if the Doppler transceiver was forced to operate on a linear region of the characteristic curve, a sinusoidal response would always be obtained. To achieve this,

Figure 3.7 Detected signal (RMS value) as a function of the antenna-artery distance.

SIGNAL AT POINT A

$$z = 1.97 \text{ mm}$$

SIGNAL AT POINT B

$$z = 2.35 \text{ mm}$$

SIGNAL AT POINT C

$$z = 2.79 \text{ mm}$$

Figure 3.8 Output signals from the Doppler radar for different antenna-artery distances (z). Chart speed = 10 mm/s.

one could insert a variable phase shifter between the transceiver and the antenna to compensate for the unwanted phase. However, both magnitude and

Figure 3.9 Mapping arterial pulsations for different operating points on the characteristic curve of the Doppler radar.

phase of the reflection coefficient at the input of the antenna are nonlinear functions of the antenna-object distance, z (Figure 4.9). The first

Figure 3:10 Characteristic curve of the Doppler radar which operates as a converter.
 λ_w is the wavelength in water.

nonlinear region occurs when the antenna is approximately $\lambda_m/4$ above the object, where λ_m is the wavelength in the medium of wave propagation.

Since this inherent characteristic of the antenna can not be changed, employing the phase shifter for this purpose will not improve the performance of the transceiver.

It is interesting to ascertain how much power is reflected from the artery as compared with that from a known-reflecting object in a lossy medium. This information can be used to estimate the capability of the Doppler radar for detection of changes within an artery-contained lossy medium.

The measurement of the power reflection coefficient of the artery was performed in the same simulator. The antenna-artery distance was set at 3.2 mm where the Doppler transceiver operated on its linear characteristic (Figure 3.10). A thin copper sheet (thickness = 0.1 mm, area = 4.2 cm²) was placed on the upper surface of the artery, using a very light arm which was clamped with a stand. The output signal from the transceiver was measured and compared with the value shown in Figure 3.7 for the same distance (3.2 mm). The square of the ratio of the output signal from the transceiver with the copper sheet removed to that with the copper sheet on the artery is defined as the power reflection coefficient of the artery which was found to be in the order of 27 dB. The arterial wall contains elastic tissue and smooth muscle [6] and its permittivity is very close to that of muscle [70]. It therefore provides a low power reflection coefficient in muscle and hence in water.

3.2 Results for Human Subjects

A number of measurements were made by placing the antennas of the

10.5-GHz and 3-GHz Doppler radars on various parts of a normal human subject, each of which contains a major artery. The results of measurements are shown in Figure 3.11 for the signals from the superficial arteries and in Figure 3.12 for the signals from the abdominal aorta.

The 3-GHz Doppler radar was assembled using standard laboratory components joined together by semi-rigid coaxial lines. The 3-GHz antenna consists of a section of a standard WR62 waveguide filled with a dielectric material ($\epsilon'_r = 30$). The length of the antenna and the length of the coupling probe were selected and adjusted for matching to the skin at the operating frequency. Due to the high attenuation in the tissue, reliable measurements were not possible for the abdominal aorta using the 10.5-GHz Doppler radar. The 3-GHz system provides significantly larger penetration in tissues, but it was found to have less sensitivity as compared to the 10.5-GHz unit.

The pulsating nature of the arteries is clearly seen. Each recorded signal indicates the changes within the tissue region under measurement. These changes result from the pulsations of the underlying arteries. Even though the Doppler radar could be adjusted to operate on a linear region of its characteristic curve, it is very difficult to say when this condition is met since the sources (real physical changes) are unknown. It is therefore difficult to tell whether the detected signal is a real mapping of the physical changes. One way to circumvent this problem is to use the microwave interferometer system as a monitoring device in which a test signal can be introduced to monitor its detection performance. This is presented in Chapter 4.

a. RADIAL (RIGHT)

b. FEMORAL (RIGHT)

c. POSTERIOR TIBIAL (RIGHT)

d. DORSALIS PEDIS (RIGHT)

Figure 3.11 Recorded signals from various parts of a normal male, each of which contains a major superficial artery (10.5-GHz Doppler radar).

A) AT THE LOWER END OF
THE BREASTBONE.

B) AT 3 CM BELOW THE
BREASTBONE.

C) AT 6 CM BELOW THE
BREASTBONE.

Figure 3.12 Signals from the abdominal aorta of a normal human subject (3-GHz Doppler radar). Chart speed = 25 mm/s.

CHAPTER 4
MICROWAVE INTERFEROMETER SYSTEMS
FOR MONITORING BIOLOGICAL IMPEDANCES

This chapter discusses the details of the implementation of the interferometer system proposed in Chapter 2. The system implementation was made at two frequencies, 9.3 GHz and 915 MHz, which have been selected due to the availability of components. The experimental arrangement for testing of both systems as well as the experimental results are presented. The last section discusses the resolution and stability of each system in detecting of small phase changes of the reflected signal.

4.1 Transmission Characteristics of Microwave Interferometers

An X-band waveguide magic tee and an SMA-connector 4-port hybrid tee (Merrimac HJM-0.75K) were used as central bridge elements of the 9.3-GHz and the 915-MHz interferometers, respectively. To evaluate the performance of these elements, it is quite important to know their transmission coefficients under selected conditions. To determine the transmission coefficient, the scattering parameters and the reflection coefficient at the input of the receiver front-end, (Γ_4), must be known. A waveguide balanced mixer based on a sidewall 3-dB directional coupler and an SMA-connector double balanced mixer (Merrimac DMM-6-1500) were used as the receiver front-ends in the 9.3-GHz and the 915-MHz set-ups, respectively. Measurements of the scattering parameters of both hybrid tees and the Γ_4 -parameters of both mixers were performed using a network analyzer (Hewlett-Packard 8410A). Transmission coefficients were

calculated from (2.12) for different values of L_R , where L_R is the return loss in the collinear arms of the hybrid tee. Calculated results are plotted as a function of the phase difference of the reflected signals at both collinear arms in Figure 4.1 for the 9.3-GHz magic tee and in Figure 4.2 for the 915-MHz hybrid tee. The transmission characteristic curves of the two devices are very similar except for L_R higher than 40 dB for which those of the 915-MHz one tend to approach the straight line faster. Due to the imperfections of the magic tee characteristic, the curves in Figures 4.1 and 4.2 are shifted in the direction of larger phase differences for higher values of L_R .

Microwave interferometers reported in the literature have their operating points located in the regions of minimum transmission coefficient and the detection of phase change is based on amplitude sensitivity [60], [62], [63]. This method is appropriate only when the signal attenuation in the collinear arms is small. As the signal attenuation increases, the linear regions on the transmission characteristic curve become small, and, hence, the sensitivity of the detection decreases. To obtain a better performance, the operation of the interferometer should be set in the vicinity of the maximum of the transmission characteristics and the method of phase sensitive detection should be employed. At the maximum of the transmission characteristic curve, the magnitude of the transmission coefficient remains almost constant for a small phase change even when the signal attenuation increases. This feature limits the amplitude-modulation (AM) noise generated in the phase detection process. The AM noise remains almost constant for different signal attenuations corresponding to the same phase change. For example, for the phase change of less than 50° , the transmission coefficient variations are

Figure 4.1 Transmission coefficient of the 9.3-GHz magic tee vs phase difference of reflected signals at collinear arms. L_R is the return loss in the collinear arms.

Figure 4.2 Transmission coefficient of the 915-MHz hybrid tee vs phase difference of reflected signals at collinear arms. L_R is the return loss in the collinear arms.

less than 0.3 dB. The method of setting the operating point of the interferometer was described in Chapter 2.

4.2 Implementation of the Microwave Interferometer System.

In order to detect the phase change of the output signal from the IF amplifier, a signal of the same frequency is needed as a reference. The IF signal given by (2.36) may be written as

$$v_{IF}(t) = A_1 \cos(\omega t + \phi_0 + \theta(t)), \quad (4.1)$$

where

$$A_1 = 2KA_s |\Gamma|.$$

The reference signal may be expressed as

$$v_{REF}(t) = A_2 \cos(\omega t + \psi_0), \quad (4.2)$$

where ψ_0 is a constant phase.

An output signal proportional to the phase change $\theta(t)$ can be obtained by feeding $v_{IF}(t)$ and $v_{REF}(t)$ into the RF and LO ports of a double-balanced mixer (DBM) (Mini-Circuits RPD-1) wherein their phases are compared. This DBM was chosen as a phase detector because of its low cost, high isolation between all three ports (50 dB) and relatively high sensitivity (8 mV/°). The high isolation minimizes the DC offset and therefore the error in phase detection. The performance characteristic of the DBM as a phase detector has been very well described in the literature [66]. For the best performance, the levels of $v_{IF}(t)$ and $v_{REF}(t)$ should be within the specified limit so that the DBM can operate in a saturated mode. For the RPD-1 mixer, the $v_{IF}(t)$ level at the RF port should be at least 1 dBm but not greater than 10 dBm while the $v_{REF}(t)$ level is fixed at 7 dBm. When both signal levels are at 7 dBm, the phase detector

provides the maximum output voltage. Another important requirement is that the phase difference between $v_{IF}(t)$ and $v_{REF}(t)$ should be in the vicinity of $\frac{\pi}{2}$ so that the phase detector can operate on the linear region of its response curve. Therefore, an IF phase shifter is needed at the input of the phase detector to vary the phase of $v_{IF}(t)$ until $\psi_0 - \phi_0 = \frac{\pi}{2}$ which is indicated by a zero DC component in the output signal.

A block diagram of the complete phase detection unit of the microwave interferometer system is shown in Figure 4.3. This is the second detector which was mentioned in Section 2.4. The IF phase shifter can be electrically controlled and has its center frequency at 30 MHz. This frequency was chosen because of the availability of an IF amplifier (GR 1216-A). A step attenuator and a preamplifier are used to adjust the drive level of $v_{IF}(t)$ to the required 7 dBm. The output signal from the phase detector is amplified and lowpass-filtered for display and recording.

Since both $v_{IF}(t)$ and $v_{REF}(t)$ must have the same frequency (30 MHz) for phase detection, a reference signal oscillator is needed. Also, the local oscillator should be phase-locked with the transmitter so that its frequency always differs from that of the RF source by 30 MHz. The phase-locked circuit for the voltage-controlled oscillator (VCO) or local oscillator is shown in Figure 4.4. The circuit is known as a frequency translation loop for shifting an input frequency to a desired output frequency [71]. The operation of this circuit requires an initial adjustment of the VCO so that its uncontrolled frequency is close to the desired output ($\omega_0 + \omega$). A portion of the VCO signal

Figure 4.3 Phase detection unit (second detector) in the microwave interferometer system.

Figure 4.4 Phase-locked circuit for the local oscillator.

is then heterodyned with the incoming RF signal of frequency ω_0 and the resulting signal whose frequency is close to ω is produced. This resulting signal is compared with the signal from the reference oscillator whose frequency is exactly ω (30 MHz), and the loop is closed back to the VCO so that the RF mixer beat-note is locked to the reference oscillator. The loop is also capable of locking to $\omega_0 - \omega$ by a proper adjustment of the VCO.

A complete schematic diagram of the microwave interferometer system for monitoring changes of biological impedances is shown in Figure 4.5. An isolator is used here to absorb the reflected power from the H-port of the magic tee when the operating point of the interferometer

Figure 4.5 Schematic diagram of the interferometer system for monitoring biological impedances.

is being set. The output from the reference signal source (CTS Knights JKTCXO-16A crystal-controlled oscillator) is fed into a power splitter (Mini-Circuits ZSC-2-1W) in which it is divided equally into two signals of equal phase to supply the phase-locked circuit and the phase detection unit. A test signal may be injected into arm 2 using a phase modulator to monitor the performance of the interferometer system. The phase change of biological impedance and the phase change due to the test signal can be separated after the phase detection process by appropriate filtering. When the latter is found to be the same as its original, measurements of the first one can start. However, this always occurs when the operating point of the interferometer is on the maximum of its transmission characteristics. The phase modulator in the 9.3-GHz set-up is a waveguide-mounted varactor which is biased to obtain a linear phase change with the test signal. The modulator is terminated by a moveable short circuit whose position was adjusted to obtain maximum sensitivity. In the 915-MHz set-up, the RF phase shifter (Merrimac PSAM-3-915) also acts as a phase modulator. A bias voltage on its control terminal provides a constant phase shift while the test signal riding on this voltage gives a corresponding phase change. It should be underlined that when the interferometer system is properly adjusted, there is no DC component at the output. This feature eliminates the need of using AC coupling elements at the input of the audio-frequency amplifier.

4.3 Testing the Interferometer System with an Ideal Simulator

Because of the availability of X-band components, the 9.3-GHz interferometer system was assembled and tested first. The purpose of the test was to check the performance of the interferometer system in

detecting changes of biological impedances. For simplicity of experimental arrangement, the biological impedance was simulated by a variable attenuator (FXR X164A) and a moveable short circuit (Hewlett-Packard X923A). The simulator was connected to arm 1 of the magic tee. The phase of the reflected signal from arm 1 was varied by moving the short circuit. Changes of the impedance were monitored by measuring displacements of the short circuit for various values of the attenuation setting. The power input to the H-port of the magic tee was kept constant at 10 mW from a Gunn diode oscillator (GE 2120C). Another Gunn oscillator with varactor tuning was used as a VCO.

The static calibration of the displacement of the short circuit was performed using a dial-gage indicator (Mitutoyo 2902-08). A linear variable differential transformer (LVDT) (Hewlett-Packard 7DCDT-100) was used for the dynamic calibration. During the dynamic calibration and the displacement measurements of the object, the short circuit was driven by an electric motor (Figure 4.6). The output signals from the LVDT and the interferometer system were displayed, recorded by the strip chart recorder and measured by the true RMS voltmeter. Typical output signal spectra are shown in Figure 4.7. The amplitude spectra were obtained using a digital computer (PDP 11/34) and the fast Fourier-transform (FFT) algorithm. It is obvious that both output signals are sinusoidal waves with no detectable nonlinear distortions. Therefore, in the measurement of small displacements or changes, the interferometer system can provide an output signal of the same waveform as that obtained from the commonly used device (LVDT).

The measured values of the interferometer output voltage vs the short circuit displacement for different values of attenuation setting on

Figure 4.6 A dial gage indicator for static calibration and a LVDT for dynamic calibration in the displacement measurements of the moveable short circuit.

arm 1 are shown in Figure 4.8. It is evident that the characteristic remains linear even for the return loss in arm 1 as high as 60 dB (30 dB attenuation setting). The sensitivity of the instrument depends on the return loss. Typical values for an X-band (9.3 GHz) interferometer are 0.5 mV/mm for 20 dB of the return loss and 15 μ V/mm for 60 dB. The noise level is 3.6 μ V. The uncertainty in voltage measurements depends upon the uncertainty of the RMS voltmeter and was estimated to be within 3% of the measured value. The uncertainty in the measurements of the object displacement by the dial gage indicator is ± 0.01 mm.

Figure 4.7 Amplitude spectra of the LVDT output signal (above) and the interferometer output signal (below).
Object displacement = 1 mm.
Attenuation setting on arm 1 = 20 dB.

Figure 4.8 Detection characteristic of the experimental 9.3 GHz interferometer system. The noise level is $3.6 \mu\text{V}$.

4.4 Testing the Interferometer System with a Simple Simulator

It has been shown in the last section that the microwave interferometer system provides an output voltage proportional to the object displacement in the ideal simulator. It is interesting to know how the interferometer output voltage varies with the distance between the antenna and the scattering object in a tissue medium. Such information can be used to estimate the capability of the interferometer system to detect displacements of an object imbedded deep in the lossy medium (tissue). This information may be obtained by testing the instrument with a simple simulator, consisting of a perfect reflector being displaced in a liquid muscle phantom.

Before testing the instrument, the dielectric-loaded cylindrical waveguide antenna designed for the 9.3-GHz interferometer system was tested using a network analyzer (Hewlett-Packard 8410A). The purpose of the test was to measure the input reflection coefficient of the antenna facing the muscle phantom which contained the perfect reflector. A mixture of water and glycerol was used as a phantom solution which has dielectric properties close to that of muscle at 9.3 GHz. The proportion of the two materials was 30% of glycerol to 70% of water (by volume). The solution was contained in a plastic box with a piece of brass (8 cm x 8 cm) placed at the bottom. The aperture of the antenna was covered with a plastic film to prevent the solution from leaking into it. The box was placed on a jack so that the distance between the antenna and the reflector could be adjusted. This distance was monitored by a dial depth indicator (Mitutoyo 2902-08). The results from the measurements of the input reflection coefficient of the antenna are shown in Figure 4.9. The phase change is defined as the change

Figure 4.9 Input reflection coefficient versus the antenna-reflector spacing for the 9.3-GHz antenna (dielectric loaded cylindrical waveguide) facing the muscle phantom.

in phase of the reflection coefficient as the distance between the object and the antenna increases. The phase change is fairly linear up to the antenna-reflector spacing of approximately $\lambda_m/2$, where λ_m is the wavelength in the muscle phantom. The magnitude of the reflection coefficient becomes nonlinear when the spacing is slightly greater than $\lambda_m/4$. The nonlinear characteristics of the magnitude and phase of the input reflection coefficient are caused by the reflected signals (relatively small) at the input of the antenna and the multiple reflections within the waveguide section. As the spacing increases, these reflections become comparable with the signal reflected from the object and modify the input reflection coefficient significantly. The multiple reflections may be minimized by shortening the antenna. However, by doing so, the matching characteristic of the antenna will be degraded since its length is very critical and has been selected for matching purposes only.

An effort was made to develop an antenna whose characteristic does not depend on its length. It was found experimentally that the flanged rectangular waveguide fitted with a capacitive screw and an inductive diaphragm at one end provided return loss greater than 23 dB when facing the muscle phantom alone. The diaphragm of 3 mm width was used to cover the waveguide flange and the screw was mounted at 2 mm from the aperture. The depth of the screw was adjusted until a good match was obtained. Measurements of the input reflection coefficient of this antenna with the reflector placed in the muscle phantom were conducted and the experimental results are shown in Figure 4.10. It is evident that the characteristic of this antenna is better than that of the cylindrical waveguide (Figure 4.9), particularly the magnitude of the reflection coefficient. However, the nonlinear phase change is still observed when

Figure 4.10 Input reflection coefficient versus the antenna-reflector spacing for the antenna (flanged rectangular waveguide mounted with a capacitive screw and an inductive diaphragm) facing the muscle phantom.

the antenna-reflector spacing is greater than $\lambda_m/2$. The $\lambda_m/2$ distance therefore limits the detection range of the microwave interferometer system in the measurement of object displacements in an electromagnetically semi-transparent medium. At 9.3 GHz, the $\lambda_m/2$ distance is 2.56 mm and is not sufficient for the interferometer system to be used for deep-lying arteries. It is therefore necessary to lower the frequency of operation of the system in order to extend the detection range. Because of the availability of elements [72], [73], 915-MHz frequency was chosen as a second frequency of operation. The input reflection coefficient of the 915-MHz slot antenna was measured and the results are illustrated in Figure 4.11. The linear range of phase change has been extended to 20 mm. In the 915-MHz measurements, a mixture of 1 molal solution of aqueous sodium chloride and glycerol was used as a muscle phantom. The proportion of the two materials was 46.5% to 53.5% (by volume), respectively. The relative permittivity of the mixture was measured and found to be $53-j28$ which is very close to that of muscle at 915 MHz.

The interferometer system at 915 MHz was tested as follows. A thin copper plate (8 cm x 8 cm) was displaced with a low speed (6 times/min) in the phantom using an electric motor. Several tiny holes were made on the plate so that the liquid could pass through during the displacement. The slow speed and holes are needed to minimize the drag and make the object displacement uniform. Two varactor-tuned transistor oscillators (Solid State Tech. SSV-0109) were used as an RF source and a VCO in the interferometer system. The power input to the H-port of the magic tee was kept constant at 10 mW. The interferometer output signal was displayed on an oscilloscope (Tektronix R 5103N) and was recorded on a strip chart recorder (Watanabe WA3001).

Figure 4.11 Input reflection coefficient versus the antenna-reflector spacing for the 915-MHz antenna (microstrip slot radiator) facing the muscle phantom.

It was mentioned in Section 4.1 that in order to obtain the maximum performance and hence maximum output signal, the operation of the interferometer should be set in the vicinity of the maximum of the transmission characteristics. Thus, the interferometer should be adjusted for every measurement. This seems to be impractical in medical applications whenever measurements from tissue region to tissue region and from subject to subject have to be performed. For practical use, the interferometer may be set (adjusted) only once throughout the measurements. This is possible since, on each transmission characteristic curve in Figure 4.2, the transmission coefficient decreases only a few dBs from its maximum value for 180° phase difference (corresponding to $\lambda_m/2$ distance in the medium) around the center setting point. This means that only the magnitude of the output signal will change when the interferometer is detuned from its center setting point, while the shape of the signal waveform will remain the same. To verify this postulate, the central setting point of the interferometer was selected at 10.8 mm of the antenna-reflector spacing which is about half of the linear region of the phase change curve (Figure 4.11). Zero spacing was taken as the position where the antenna was just in touch with the reflector when it was up to the top. The displacement of the reflector was set at 0.5 mm. Peak-to-peak values of the interferometer output voltage as read from the oscilloscope are plotted as a function of the spacing in Figure 4.12. The distortion of the output signal was observed when the spacing was greater than 21 mm. Each value of the output voltage is proportional to the phase change and, hence, the reflector displacement around a corresponding point on the phase characteristic curve of the reflection coefficient (Figure 4.11). The proportionality factor depends upon the magnitude of the reflection coefficient, the detuning

Figure 4.12 Interferometer output voltage versus the antenna-reflector spacing. Reflector displacement = 0.5 mm. Amplifier gain = 20.

distance from the setting point, the sensitivity of the phase detector, and the gain of the amplifier. When the spacing decreases from 10.8 mm, the signal attenuation in arm 1 of the magic tee decreases and the operating point of the interferometer is shifted to be on the upper transmission coefficient curve (Figure 4.2), resulting in a higher value of the output voltage. As the spacing decreases further and is less than 8 mm, the output voltage starts to drop since the operating point falls along the transmission characteristic curve, resulting in a smaller value of the transmission coefficient. When the spacing is greater than 10.8 mm, the output voltage decreases due to higher signal attenuation in arm 1 and smaller transmission coefficient. The uncertainty in voltage measurements depends upon the uncertainty of readings from the oscilloscope and was estimated to be within 3% of the measured value. Experimental results illustrated in Figure 4.12 show that, by adjusting the interferometer only once, reasonable output signal can still be obtained throughout the detection range. For metal displacement objects, the maximum detection range that the interferometer system can provide a linear output signal is approximately $\lambda_m/2$, where λ_m is the wavelength in the medium containing the object. For other kinds of displacement objects, the maximum detection range will be less since they provide smaller reflected signals.

4.5 Resolution and Stability of Interferometer Systems

In this section, we shall discuss the accuracy of the phase change of the impedance seen at the antenna input which the microwave interferometer system is capable of detecting. Factors which govern the detection capability of the system are the attenuation in the medium,

the power and the frequency drift of the RF oscillator, the receiver noise, and the mechanical stability of the system. The effect of the receiver noise can be analyzed by considering the schematic diagrams shown in Figures 2.5 and 4.3 which, after simplifying, may be redrawn as illustrated in Figure 4.13.

Figure 4.13 Simplified diagram of the receiver.

The simplified diagram of the receiver consists of three networks in cascade which have different noise figures, available gain and noise bandwidths. The receiver bandwidth is determined by the bandwidth of the audio amplifier as the remaining networks are wide-band components. Let F_j, G_j and B_j be the noise figure, available gain and noise bandwidth, respectively, of the j^{th} network. The overall noise figure F of the three circuits in cascade is given by [57].

$$F = F_1 + \frac{B_2(F_2-1)}{B_1 G_1} + \frac{B_3(F_3-1)}{B_1 G_1 G_2} \quad (4.3)$$

Because of the high gain of the IF amplifier in the front-end

($G = 72$ dB), the overall noise figure is approximately to the noise figure of the first network, i.e., $F \approx F_1$. Using a noise figure meter (Hewlett-Packard 432A), values of F_1 of the receivers of the 9.3-GHz and 915-MHz interferometer systems were measured and found to be 7.6 dB and 11.8 dB, respectively. The noise figure of the receiver may also be written as [57]

$$F = \frac{S_i/N_i}{S_o/N_o} \quad (4.4)$$

where S_i is the input signal power, N_i is the input noise power, S_o is the output signal power, and N_o is the output noise power. The input noise power is the available-thermal noise power which is given by

$$N_i = kTB \quad (4.5)$$

where k = Boltzmann's constant = 1.38×10^{-23} J/K, T is the temperature in Kelvin, and B is the receiver bandwidth in Hz. At room temperature, taken as $T = T_o = 290$ K, $kT_o = 4 \times 10^{-21}$ W/Hz. Using (4.3) - (4.5) and taking the approximation of the noise figure into account, the output signal-to-noise ratio can be expressed as

$$\frac{S_o}{N_o} = \frac{S_i}{kT_o BF_1} \quad (4.6)$$

or

$$\frac{E_o^2}{E_n^2} = \frac{E_o^2}{E_n^2} = \frac{S_i}{kT_o BF_1} \quad (4.7)$$

where E_o is the time-average signal amplitude and E_n is the time-average noise amplitude. When the phase detector is adjusted to operate in the linear region of its response curve, the output voltage is

$$E_o = K_D \theta \quad (4.8)$$

where K_D is the sensitivity of the phase detector and θ is the phase change. Small incremental changes in phase, $\Delta\theta$, can be related to output changes by

$$\Delta E_o = K_D \Delta\theta \quad (4.9)$$

To determine the apparent phase change caused by the receiver noise, let $\Delta E_o = E_n$, then from (4.7) to (4.9),

$$\frac{\Delta\theta^2}{\theta^2} = \frac{\Delta\theta^2}{\theta^2} = \frac{kT_o BF_1}{S_1} \quad (4.10)$$

or

$$\frac{\sqrt{\Delta\theta^2}}{\theta} = \sqrt{\frac{kT_o BF_1}{S_1}}$$

Figure 4.14 shows the plot of $\sqrt{\Delta\theta^2}/\theta$ as a function of the return loss in the collinear arms of the magic tee, L_R , for different values of input power at the H-arm and for $B = 10$ Hz. The variation of $\sqrt{\Delta\theta^2}/\theta$ as a function of B for different values of L_R and for input power = 10 mW is illustrated in Figure 4.15. As observed, the 9.3-GHz interferometer system offers a better phase resolution, $\Delta\theta$, than the 915-MHz one since it has a smaller receiver noise figure. Besides the receiver noise figure, it is clear that the resolution capability of microwave interferometer is limited by the receiver bandwidth, input power at the H-port, and signal attenuation in the medium.

Figure 4.14 Relative phase resolution versus the return loss in the collinear arms for different values of input power at the H-port. The receiver bandwidth = 10 Hz.

Figure 4.15 Relative phase resolution versus the receiver bandwidth for different values of the return loss in the collinear arms. Input power at the H-arm = 10 mW.

The receiver bandwidth is controlled by the detection bandwidth of the audio amplifier, B_3 , which was usually set at 10 Hz. It is evident that the apparent phase change due to the receiver noise is much smaller than that caused by the variations within the medium of measurements. For example, for a 3 mm short circuit displacement in the ideal simulator (Figure 4.6) and a 10 dB attenuation setting, the incremental phase change due to the receiver noise is of the order of 3.9×10^{-8} rad or approximately 2.2×10^7 times smaller than the phase change caused by the displacement.

The effect of the frequency drift of the RF oscillator in the phase detection is obvious because the phase of the impedance seen at the antenna is proportional to the frequency. Typical frequency stabilities are 35 ppm/ $^{\circ}$ C for the Gunn oscillator (GE C-2120C) and 200 ppm/ $^{\circ}$ C for the transistor oscillator (Solid State Tech. SSV-0109). These frequency drifts result in the change of the interferometer output voltage by the same fractions. As an example, for a 3 mm short circuit displacement in the ideal simulator (Figure 4.6) and a 10 dB attenuation setting, the detected output voltage is 1450 μ V (Figure 4.9) and the change due to the frequency drift is estimated to be 0.05 μ V/ $^{\circ}$ C. This change is less than the equivalent input noise of the audio amplifier (PAR 113) and, therefore, it does not produce any change in the amplifier output signal. In practice, the effect of the receiver noise and the RF oscillator frequency drift is generally smaller than that caused by the mechanical instability of the instrument.

CHAPTER 5

MICROWAVE INTERFEROMETERS IN

MEDICAL APPLICATIONS

It has been demonstrated in Chapter 4 that the microwave interferometer can be used to monitor the object displacement in an electromagnetically semi-transparent medium (muscle phantom). In this Chapter, the potential of using the microwave interferometer for monitoring changes within a tissue region will be demonstrated by testing the instrument using a human phantom. This is followed by the experimental results of the tests with normal human subjects and the discussion on certain aspects regarding to the potential of using this instrument in medical applications.

5.1 Monitoring Changes within a Phantom by the Microwave Interferometer

To demonstrate the potential of using the microwave interferometer system for monitoring changes within a tissue region, a simulated model of the middle part of a human thigh was constructed. This part of the human body was chosen for the simulation because its tissue structure is less complex as compared with that of other parts [74], [75]. The cross section of the mid-thigh, looking from outside to the bone and in the vicinity to the major blood vessels, consists of the skin layer (thickness ≈ 3 mm), the fat layer (thickness ≈ 4 mm), femoral artery and vein which are surrounded with fat and muscle, respectively, as shown in Figure 5.1. There are also various small blood vessels and nerves spreading in this section, but these result in only minor changes and can be neglected in the simulation.

Testing the interferometer system with the simulated model was

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Figure 5.1 Cross-sectional view of the middle part of human thigh (right side) [74]. Scale 3:4.

initially planned to be performed at both 915 MHz and 9.3 GHz. However, due to high attenuation in biological tissues at 9.3 GHz (Figure 1.1), a reliable detected signal may not be easily obtained. As shown in Figure 5.2, the phase change of the input reflection coefficient of the antenna is linear for the antenna-artery distance within approximately $\lambda_m/4$, where λ_m is the wavelength in muscle phantom. At 9.3 GHz, $\lambda_m/4 = 1.28$ mm which is still less than human skin thickness (see Figure 5.1). Major changes within the tissue region occur beyond this distance and the system might end up operating on a nonlinear region of the phase change characteristic which results in less sensitivity and distortions. Due to this limitation, testing of the 9.3-GHz system with the simulated model was not performed. The experiment was therefore left to be conducted only for the 915-MHz system. As illustrated in Figure 5.3, the phase change of the input reflection coefficient of the 915-MHz antenna is fairly linear within the antenna-artery distance up to 6.1 mm. This range is sufficient to detect changes resulting from the artery in the simulated model.

To construct the simulated model of a human thigh at 915 MHz, several phantom materials are required to simulate skin, fat, blood vessels, blood, muscle and bone. One could use animal samples (except fat) for this purpose, but they may contaminate the experimental set-up and cause an inconvenience in the measurements. Since bone and fat have almost the same dielectric properties [40], only one phantom material is needed for the simulation. The same principle can also be applied to skin and muscle. The composition and permittivity of phantom materials for these tissues are available in literature [40]. The muscle phantom is composed of water, sodium chloride, polyethylene powder and a gelling

PHASE CHANGE OF THE REFLECTION COEFFICIENT (%)

ANTENNA-ARTERY SPACING (mm)

Figure 5.2 Input reflection coefficient versus the antenna-artery spacing for the cylindrical waveguide antenna (9.3 GHz) facing the muscle phantom.

Figure 5.3 Input reflection coefficient versus the antenna-artery spacing for the antenna (915 MHz microstrip slot radiator) facing the muscle phantom.

plastic super stuff (Whamo Manufacturing) with the proportion by weight of 75.44%, 0.907%, 15.2%, and 8.45%, respectively. Fat phantom material is available only in solid form and cannot be used for this simulation. Animal fats and vegetable oils have their relative permittivity, ϵ_r^* , in the order of 2.6-j0.2 at 1 GHz [76]. These materials still cannot be used for the simulation because their permittivity is much lower than that of human fat ($\epsilon_r' = 5.3-7.5$, $\epsilon_r'' = 1.5-2.7$) at the same frequency, where ϵ_r' is the relative dielectric constant and ϵ_r'' is the relative loss factor. Some liquid chemicals, for example cyclohexanol ($\epsilon_r^* = 4.8-j2.24$ at 915 MHz), have their permittivities very close to that of fat. However, they are toxic and volatile and are not suitable to be used as fat phantom materials for modeling experiments.

Several attempts were made in an effort to develop a phantom material for human fat in such a form that it should not greatly disturb the arterial wall motions in the simulated model. It was found empirically that a mixture of cold cream (Chesebrough Pond's) and corn oil (Mazola) with a composite percentage by volume of 72.4% to 27.6% give $\epsilon_r^* = 7-j1.14$ (at 915 MHz) which is very close to that of human fat. The nonvolatile and nonpoisonous characteristics make the developed material appropriate to be used as a fat phantom for modeling experiments. The muscle equivalent liquid described in Chapter 4 was used as simulated blood. Two sections of bovine artery, each of which has its length and diameter of approximately 6 cm and 1 cm, were used as femoral artery and vein.

The simulated model around the femoral artery at the middle part of a human thigh is illustrated in Figure 5.4. A plastic tube of 2.5 cm diameter containing the fat phantom (cold cream and corn oil) was used to simulate the bone. The tube was surrounded by the muscle phantom

Figure 5.4 Tissue phantom around the femoral artery at the middle part of human thigh.

which was contained in a plastic box covered with a thin sheet of mylar at the top. Two bundles of the muscle phantom wrapped up in plastic films were placed on the top of the box and beside the two bovine blood vessels. The locations of the blood vessels with respect to the simulated bone were scaled up from those shown in Figure 5.1. Both blood vessels contained the liquid muscle phantom. The vein was inserted between two plastic tubes (diameter ≈ 1 cm) which were fixed on the supporters at the two opposite sides of the box (not shown in Figure 5.4). Another end of each plastic tube was closed with a screw. The human skin was simulated by a slab of the muscle phantom (thickness ≈ 3 mm) which was wrapped up in a plastic film. This simulated skin was attached to the antenna when testing the interferometer system. The space between the simulated skin and the muscle phantom was filled with the fat phantom.

The whole simulated tissue model was contained in another plastic box wherein the artery was connected to another pair of plastic tubes (diameter ≈ 1 cm) as shown in Figure 5.5. Both plastic tubes were fixed perpendicularly to the two opposite side walls and were extended to the outside of the box. After filling these tubes with the liquid muscle phantom, one end of the tube at the outside was closed while the other was connected to a brass cylinder which also contained the muscle phantom. The volume of the cylinder was varied by changing the position of the piston. One turn of the handle resulted in the volume change by 0.09 cm³. This change was consequently transmitted into the simulated model in terms of a pressure which caused a change in the artery diameter.

Testing the interferometer system with the simulated tissue model was then performed. The experimental arrangement for the measurement of changes within the simulated model is shown in Figure 5.5. The antenna

Figure 5.5 Experimental arrangement for the measurement of changes within the simulated tissue model resulting from varying the volume of the cylinder.

22
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was attached to a small platform which can be raised or lowered using a small jack. The distance between the antenna and the artery was monitored by a dial depth indicator (Mitutoyo 7212). The output of the interferometer system was connected to an X-Y recorder (Watanabe WX4401). The X-axis sweep speed of the recorder was set at 0.5 cm/s which was slow enough to complete 5 turns of twisting the piston handle when recording the interferometer signal. Each measurement was made by turning the piston handle in the clockwise direction and marking the recorded signal when each turn was completed. The piston handle was then turned backwards to the original position before taking the next measurement. The gain of the amplifier (PAR 113) (Figure 4.3) in the interferometer system was kept constant at 100 throughout the measurements. The measured values of the interferometer output voltage versus the increase in volume of the artery in the simulated model for different values of the fat-layer thickness are shown in Figure 5.6. It is evident that the interferometer output voltage is linearly proportional to the change within the simulated model which results from the increase in volume of the artery. The sensitivity of the instrument depends on the thickness of fat between the skin and the artery. For an amplifier gain equal to 100, typical values for the 915-MHz interferometer are 8.0 V/cm^3 for 2.5-mm thickness of fat and 1.6 V/cm^3 for 11.4-mm. The uncertainty in the voltage measurements depends upon the uncertainty of the X-Y recorder and was estimated to be within $\pm 10 \text{ mV}$.

The experimental results shown in Figure 5.6 were obtained based on the measurements of changes in the static pressure applied to the artery. In human subjects, changes in the applied pressure occur in cycles (60-80 times/min for normal adults) due to the motion (contraction and

Figure 5.6 Output voltage from the 915-MHz interferometer system versus the volume increase of the artery in the simulated biological tissue model for different values of the fat-layer thickness.

relaxation) of the heart as it pumps the blood. Based on the experimental results illustrated in Figure 4.9, it can be expected that for the dynamic changes within a tissue region the interferometer system maintains its linear characteristic.

5.2 Results for Human Subjects

The microwave interferometer systems were also tested on several normal human subjects. Signals from the radial artery were recorded using the 9.3-GHz system. The results are illustrated in Figure 5.7. Similar to the 10.5-GHz Doppler radar, this system can be used only to detect changes within tissue regions of superficial arteries. The signals from the abdominal aorta, detected at various locations below the breastbone using the 915-MHz system, are shown in Figure 5.8. The

Figure 5.7. Interferometer output signal from a radial artery (9.3-GHz system).

recorded signals show better pulsating nature of the abdominal aorta as compared to those recorded by the 3-GHz Doppler system. Small artifacts shown in Figure 5.8(b) and (c) are due to the movement resulting from breathing and perhaps, pulsations of other adjacent arteries. The 915-MHz

A) AT 2 CM BELOW THE
LOWER END OF THE
BREASTBONE.

B) AT 6 CM BELOW THE
BREASTBONE.

C) AT 10 CM BELOW THE
BREASTBONE.

Figure 5.8 Interferometer output signal from the abdominal aorta
of a normal human subject (915-MHz system).
Chart speed = 10 mm/s.

interferometer system was also tested on the middle part of human thigh but reliable signals were not obtained. This is because the system has relatively small sensitivity as derived from the phase change characteristic of the antenna (Figure 5.3). Also, the pulsations of the underlying artery may not be as large as arterial wall displacements produced in the simulated model.

There are two aspects to be considered regarding to the potential use of the microwave interferometer system in medical applications. The first one would be the selection of the frequency of operation which was found to be related to the sensitivity of the system. The frequency should not be too high otherwise the electromagnetic waves will be highly attenuated in the test tissue region (Figure 1.1) and, therefore, reliable scattered signals will not be obtained. On the other hand, if the frequency is too low, the sensitivity of the detection may not be sufficient even though deeper penetration of waves in the test region is possible. High frequencies are therefore suitable for detecting changes within the superficial region. For deep-lying objects, lowering the frequency of operation should be considered and made such that the sensitivity is still sufficient for reliable detection. Also, a rough estimation should be made to ensure that the antenna-object distance is within the linear detection range of the system. The second aspect is concerned with the variability of the human body in structure from person to person. This has caused a big problem in using the instrument as a diagnostic tool because the same antenna cannot be used for every human subject. To solve this problem, a tunable antenna is needed.

CHAPTER 6

CONCLUSIONS AND SUGGESTIONS

6.1 Summary

The development of two methods for continuous monitoring the time-varying biological impedances at microwave frequencies has been described in this thesis. The methods can be used to measure changes within a tissue region arising from arterial pulsations. The first method is based on using a microwave Doppler radar to monitor the phase change of the return signal from the tissue region, whereas the second method is based on measuring the phase change of the input reflection coefficient of the antenna using a microwave interferometric system. The return signal and the reflection coefficient are related, with their phase changes varying proportionally to the net changes within the tissue region.

Both the basic principles and mathematical analysis of each method were first discussed. Experimental arrangements for the testing of the 10.5-GHz Doppler radar with a simulator, consisting of a section of bovine artery installed in a water bath, were then described. These were followed by experimental results from tests using the simulator and from tests on human subjects. It was found that the detection characteristic of the Doppler radar varies sinusoidally as a function of the antenna-artery spacing. The output signal from the Doppler radar was observed to be proportional to the arterial wall displacement only when the device operated on the linear regions of this detection characteristic. It was also found experimentally that, at 10.5 GHz, the signal reflected from an animal artery pulsating in water (or muscle) was of the order of 30 dB lower than that reflected from a metal reflector.

The design and implementation of the microwave interferometric

system were carried out at 9.3 GHz and 915 MHz. These systems employ a hybrid junction (magic tee) as a central element which compares the phase of the signal reflected from the tissue region with that of a reference signal using a phase sensitive detector. The detection performance of the system was checked by applying a test signal into arm 2 of the magic tee using a phase modulator. The best performance was found when the output signal from the E-arm was maximum. The 9.3 GHz system was tested with a simulator, consisting of a variable waveguide attenuator and a waveguide moveable short circuit driven by an electric motor. The interferometer output voltage was found to be linear with the short circuit displacement for the signal attenuation in the collinear arms below 60 dB (30 dB attenuation setting on arm 1). The sensitivity of the interferometric system depends on the signal attenuation. For the 9.3-GHz system, typical values are 0.5 mV/mm for 20 dB of the signal attenuation and 15 μ V/mm for 60 dB. The noise level was 3.6 μ V.

The input reflection coefficients for different antennas used for diagnostic applications were measured using a microwave network analyzer. It was found that the phase of the reflection coefficient varied fairly linearly with the antenna-reflector distance for distances less than $\lambda_m/2$, where λ_m is the wavelength in the muscle phantom. The 915-MHz interferometric system was tested with a simple simulator, consisting of a reflector being displaced sinusoidally in liquid muscle phantom, to verify this distance limitation. The interferometer output signal was observed to be proportional to the reflector displacement for the antenna-object distance less than 20 mm ($\approx \lambda_m/2$). The distance for linear detection is less than $\lambda_m/2$ for those objects which are not perfect reflectors. For animal artery, as an example, the linear detection

distance was found experimentally to be in the order of $\lambda_m/4$.

A simulated model of mid human thigh was built for testing the interferometric system. Due to large signal attenuations in human tissues and small linear detection distance at 9.3 GHz, testing the interferometer system with the simulated model was performed only at 915 MHz. Variations within the model were made by manually changing the volume of a cylinder connected to the artery. The interferometer output voltage, recorded on an X-Y recorder, varied linearly with changes of the cylinder volume.

The sensitivity of the instrument depends upon the thickness of fat layer between the skin and the artery. Typical values are 8 V/cm^3 for 2.5 mm of the fat thickness and 1.6 V/cm^3 for 11.4 mm.

Testing the interferometer systems was also performed on human subjects. Due to the high attenuation in biological tissues, the 9.3-GHz system provides reliable measurements only for superficial arteries. The 915-MHz system was found to be suitable for measurements of deep-lying objects such as the abdominal aorta. However, it was found to be less sensitive than the 9.3 GHz system. The selection of the frequency of operation requires, therefore, a trade-off, between the sensitivity and the penetration depth in the tissue region. At present, any information about the mechanical properties of the arteries can be obtained from the analysis of the shape of the waveform recorded by the interferometer system.

Experimental results have shown a potential for using the microwave Doppler radar and microwave interferometer system as diagnostic tools for monitoring changes within a tissue region. For the microwave Doppler radar, a variable phase shifter would be required to set the operating point in the linear region of the detection characteristic.

However, it is difficult to determine when this condition is met because the exact distance between the antenna and the test region is not known a priori. The interferometric system is more controllable and its operating point can be easily adjusted to obtain maximum sensitivity. The present set-up is complicated and bulky but can be integrated and miniaturized using microwave integrated circuit technology.

6.2 Suggestions for Future Research

Continued research in many areas of the microwave interferometric system presented in this thesis is required to bring the instrument out of the laboratory and into clinical use.

The first suggestion would be the development of an antenna which is adjustable thereby allowing impedance matching once it is in contact with a tissue region. This adjustable antenna would certainly improve the performance and sensitivity of the interferometric system as well as other microwave diagnostic devices. Besides being well-matched when in contact with the tissue, the antenna should also be simple, compact, light and easy to handle, and have minimum leakage.

The initial alignment of the interferometer and specifically the adjustment of the IF phase and power level required for maximum sensitivity has been done manually. This process is inconvenient, time consuming and requires a knowledgeable operator. An automated system could be developed and incorporated into the interferometric system to accomplish this purpose.

Other suggestions include conducting experimental studies at other sections of the human body, for example, the brain and abdomen, where

other diagnostic methods are simply not able to monitor changes within underlying tissues. These studies would require building simulated models to be tested with the interferometric system. Impairment of each artery in the model should be made in order to observe the change at the output signal. Finally, further studies should be undertaken with human subjects to correlate the results with those obtained from the simulated models.

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