

Visualization of the interaction of solutions  
through a microfluidic chip using Terahertz  
technology

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Graduate School of  
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DOCTORAL THESIS

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Visualization of the interaction of  
solutions through a microfluidic  
chip using Terahertz technology

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*in the*

Graduate School of Natural Science and Technology

February 16, 2021



# TO WHOM IT MAY CONCERN

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We hereby certify that this is a typical copy of the  
original Doctoral dissertation of

Feroz AHMED

Thesis Title:

Visualization of the interaction of solutions through a  
microfluidic chip using Terahertz technology

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*Signature of Supervisor*

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Professor Toshihiko KIWA

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# Declaration of Authorship

This dissertation and the work presented here for doctoral studies were conducted under the supervision of Professor Toshihiko Kiwa. I, Feroz AHMED, declare that this thesis titled, “Visualization of the interaction of solutions through a microfluidic chip using Terahertz technology” and the work presented in it are my own. I confirm that:

Signed:                      Feroz AHMED                      Student number: 51429651

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Date: February 12, 2021

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# Abstract

Feroz AHMED

*Visualization of the interaction of solutions through a microfluidic chip using Terahertz technology*

Terahertz (THz) waves are electromagnetic waves with frequency components between 100 GHz and 10 THz, shows exclusive spectral features such as a non-ionizing nature, high spatial resolution, good penetration capabilities etc. that make this technology to exhibit its impact on many applications of spectroscopy, sensing, and imaging processes. Therefore, THz wave radiation shows its sensitivity to the chemical solutions, cells, and tissues with different water contents. In addition, THz imaging technique can reveal the orientation of a hydrogen bond owing to its intermolecular bonding. THz radiation gradually decreases when it travels through any medium with a high water concentration. As a result, THz wave can distinguish even small fluctuations in the water contents of chemical and biological solutions for identifying and capturing the interaction of different chemical and biological solutions like pH solutions, DNA, enzyme, tissue and blood samples etc. In the commercial market, a charge-coupled device (CCD) camera is available to capture the images of biological and chemical reactions. However, this CCD camera does not be able to capture the images of the interaction of molecules inside the chemical and biological solutions satisfactorily. Whereas, THz image sensing technology is the recent advancement in the lab-on-chip technology to visualize and capture the images of chemical interactions of pH buffer solutions, blood, tissues, enzyme, DNA and so on in detail

without the use of any fluorescent dye marker which makes it reliable label-free detection and visualization technique. Even microfluidic channels made from different materials are opaque in the visible and near-infrared domains, THz technology is functioning effectively.

In this research, a sensing plate was fabricated as a part of using a sensor. The sensing plate consists of  $SiO_2$  and  $Si$  film on a sapphire substrate. Terahertz waves were generated from the  $Si$  layer in the sensing plate when the femtosecond laser is irradiated to the bottom side of the sensing plate through the sapphire substrate. Electrons jumped into the conduction band from their excitation because they absorbed photons of generated femtosecond laser pulses whose bandgap energies are higher than the bandgap energies of silicon layer. The magnitude of the depletion field depended on the electric potential at the surface of the sensing plate. And, the amplitude of the radiated terahertz waves depended on the magnitude of the depletion field. Thus, the interaction of different concentrations of pH buffer solutions through microfluidic channels was connected to the electric potential on the sensing plate and such interaction of concentrations of different pH solutions was able to visualize by measuring the amplitude of radiated THz waves.

Viscoelastic material PolyDiMethylSiloxane (PDMS) can be one of promising candidates for the fabrication of microfluidic chip (MC) in micro scale for a wide range of applications in biological and medical research. Therefore, PDMS can be used for fabricating cost effective microfluidic chip with different microfluidic channels owing to its excellent properties of optical transparency, easy and swift fabrication process, chemical inertness and low shrinkage rate with the replication of the mold into the micro-scale environment. Moreover, without requiring expensive clean room facilities with time consuming fact, CAD designed 3D printed structure fabricated the replica mold of the microfluidic chip using PDMS. For preparing the PDMS microfluidic chip, cross-linked two agents

of Sylgard 184 Silicone elastomer base agent and Sylgard 184 Silicone elastomer curing agent as an adhesive were employed. Two inlet wells and one outlet well based a 3D structure of a microfluidic chip was drafted using a student version Solid work software, translated it into STL format and then printed the structure from the high definition 3D printer. PDMS sample made microfluidic chip was developed by cross-linking both the base and curing agents at a mass ratio of 10:1. Then, 3D printer printed the designed structure of the microfluidic chip and the cross-linked solution was poured into that printed structure for preparing replica mold of PDMS chip. PDMS was then put into the oven for 24 hours in a controlled 40 degrees Celsius. When it became hard enough, PDMS was used to design the microfluidic chip for the experiment.

The nature of internal fluid flow parameters through a microfluidic chip can be revealed using Terahertz (THz) image sensing technology and ANSYS fluent simulation to evaluate the chemical and biological samples in the micro scale level for the advancement of next healthy generation. Therefore, the trace of internal microfluidic reactions of chemical and biological solutions through a microfluidic chip from both the experimental and simulation-based analyses is a much expected trend to a fluid flow design and estimation engineer in the research and development field for serving the society with the proposed THz technology. In this doctoral dissertation, two inlet wells and one outlet well based, capillary concept a microfluidic chip was designed, and evaluated in terms of its internal performance of microfluidic parameters from fluid dynamics (FD) and its response to the experimental and simulation works. As a part of applied samples for the testing and analysis purpose, two pH=4 and 10 buffer solutions (BS) were injected through the internal micro flow channels (MFC) of the designed microfluidic chip in the above mentioned cross-disciplinary areas. Then, THz laser started to scan the interaction of the mixed solutions for the trace of internal microfluidic reactions by capturing THz images which was

directed by the THz chemical microscope (TCM) of our laboratory. Hence, this fabricated microfluidic chip can detect any imbalanced condition of pH level of human bodies out of metabolism, respiration processes, pH control of liquid foods of food industries for the development of biomedical applications in a wide range.

This dissertation also includes the proposed design of an adaptive digital filter in LABView simulation to reduce the internal noises during the optical signal to electrical signal conversion procedure of the project. The outcome of this dissertation was to improve the contrast and resolution of the THz images without losing the original speed of laser scanned THz images. This filter helped to scan THz images without losing the original speed of TCM with the reduction of total measurement time of the lock-in-amplifier which was connected with the entire THz signalling and imaging system. Thus, the integration of the above mentioned filter with the TCM system can provide the guideline of faster captured images technique with the faster scanning of samples like movie clips, quick occurred chemical reactions etc. in future.

In this dissertation, from the proposed capillary designed microfluidic chip with different dimensional paths of cylindrical shaped channels along with the traditional rectangular channels, it was calculated all microfluidic resistances through the micro flow channels and revealed the variation of internal fluidic parameters to evaluate the performance of the microfluidic chip experimentally. Microfluidic internal parameters like pressure drop, velocity, fluid flow rates of the mixed two pH 4 and 10 buffer solutions were measured and calculated to find out the characteristics of all microfluidic parameters. In addition, ANSYS fluent 2D simulation was also performed for the trace of internal interactions of pH buffer solutions through the designed microfluidic chip. Then, experimental results were validated with the obtained simulation results to emphasize the results in terms of fluid dynamics.

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# Contents

<b>Declaration of Authorship</b>	<b>v</b>
<b>Abstract</b>	<b>vii</b>
<b>Acknowledgements</b>	<b>xi</b>
<b>Contents</b>	<b>xv</b>
<b>List of Figures</b>	<b>xix</b>
<b>List of Tables</b>	<b>xxiii</b>
<b>List of Notations and Symbols</b>	<b>xxv</b>
<b>Research Activities</b>	<b>xxix</b>
<b>1 Introduction</b>	<b>1</b>
1.1 Use of THz Chemical Microscope (TCM) to utilize THz Technology in this research . . . . .	3
1.1.1 Reasons of the use of THz Technology over other technologies . . . . .	4
1.1.2 Noises Involved in THz Signals and Images Processing . . . . .	5
1.1.3 Applied Digital Filter in THz Technology . . . . .	5
1.2 Fluid Dynamics in Microfluidics . . . . .	6
1.2.1 The Speciality of the Designed Microfluidic Chip over other Conventional Microfluidic Chips . . . . .	7
1.3 Outline of this Thesis . . . . .	8
1.4 Motivation . . . . .	10
<b>2 Mechanism of THz Imaging</b>	<b>11</b>
2.1 THz generation principle . . . . .	12
2.2 THz detection principle . . . . .	13
2.3 Experimental set up of TCM . . . . .	15

2.4	Characteristic Curve Analysis from the Experiment . . . . .	16
2.5	Summary . . . . .	18
<b>3</b>	<b>PDMS designed Microfluidic Chip</b>	<b>19</b>
3.1	Preparation of PDMS . . . . .	20
3.1.1	Top and bottom plates design to squeeze PDMS . . . . .	22
3.1.2	Cross-sectional view of designed microfluidic chip and isometric view of central main structure of microfluidic chip . . . . .	23
3.1.3	Advantages and key features of PDMS . . . . .	23
3.1.4	Summary . . . . .	24
<b>4</b>	<b>Microfluidic Parameters Measurement, Calculation and Statistical Analysis</b>	<b>29</b>
4.1	Introduction . . . . .	29
4.2	Mathematics . . . . .	29
4.3	Calculation of Microfluidic Resistances of Microfluidic Chip . . . . .	31
4.4	Pressure Drop Measurement and Velocity Profile Estimation from Fluid Dynamics . . . . .	36
4.5	Summary . . . . .	42
<b>5</b>	<b>Simulation to Emphasize the Experimental Result</b>	<b>43</b>
5.1	Introduction . . . . .	43
5.2	Simulation Algorithm . . . . .	44
5.3	Mesh dependency and validation of simulation in CFD environment . . . . .	51
5.3.1	Mesh dependency and validation of simulation . . . . .	51
5.4	Summary . . . . .	53
<b>6</b>	<b>Digital Filter Design and Testing</b>	<b>55</b>
6.1	Introduction and Motivation . . . . .	55
6.2	Working Principle of an Adaptive Digital Filter . . . . .	55
6.3	Experimental Set up for ADF's Proper Functioning Check . . . . .	57
6.4	Statistical Analysis of Characteristic Curve of ADF . . . . .	59
6.5	Determination of Contrast of THz Images . . . . .	61
6.6	Qualitative Analysis of Contrast with and without Filter . . . . .	62
6.7	Contour of THz Images with and without Filter . . . . .	63
6.7.1	Data Acquisition of THz Images . . . . .	64
6.8	Advantages of an Adaptive Digital Filter . . . . .	65
6.9	Summary . . . . .	65
<b>7</b>	<b>Results and Discussions</b>	<b>67</b>

7.1	Introduction . . . . .	67
7.1.1	THz Imaging results . . . . .	69
7.1.2	2D ANSYS Fluent simulation results . . . . .	70
7.2	Analysis of Velocity Profiles obtained from ANSYS Fluent . . . . .	72
7.3	Summary . . . . .	73
<b>8</b>	<b>Conclusion and Future Works</b>	<b>75</b>
	<b>Bibliography</b>	<b>79</b>
	<b>Biography</b>	<b>95</b>



# List of Figures

2.1	The sensing plate for the TCM system with three different layers fabricated sensing chip . . . . .	11
2.2	THz electromagnetic waves generation with the irradiation of femto second laser pulses onto the layers of the sensing chip . . . . .	12
2.3	THz electromagnetic wave's detection to capture the pH buffer solutions interaction . . . . .	13
2.4	Optical part arrangement of the TCM system for this project . . . . .	14
2.5	Whole THz signaling system to convert the optical signal to electrical signal with a two stage controlling mechanism . . . . .	15
2.6	By applying THz electromagnetic wave, buffer solutions at pH values of 4 and 10, injected through two inlet wells via syringe-connected tubes and flushed out through the outlet well . . . . .	17
2.7	Characteristic curve of THz response with respect to different pH concentration values . . . . .	18
3.1	Different dimensions of micro flow channels' 3D lay out in Solid work. . . . .	19
3.2	3D-printed structure of the microfluidic chip with dimensions of 20 mm × 20 mm scaling in Solid work . . . . .	20
3.3	practical 3D printed structure for preparing replica of this structure . . . . .	21
3.4	Hard replica mold within 3D printed structure . . . . .	22
3.5	Manual cut microfluidic chip attached with Si: sapphire glass substrate (clipped with bottom 3D printed plate) . . . . .	23
3.6	Bottom plate designed in CAD software (In the middle, Si:sapphire glass substrate of 10 mm × 10 mm was clipped and PDMS-made MC was attached with glass substrate . . . . .	24
3.7	Top plate designed in CAD to squeeze the microfluidic chip with glass substrate and bottom sensing plate . . . . .	25
3.8	Top view of internal micro fluidic channels formation in Solid works using different internal designed dimensions . . . . .	25

3.9	Microfluidic chip that attached with Si:sapphire glass substrate, screwed and tilted in a balanced way to squeeze it with top and bottom plate . . . . .	26
3.10	cross-sectional view of designed 3D structure . . . . .	26
3.11	Scaling in solid work of main microfluidic chip's lay out along with micro flow channels . . . . .	27
4.1	Digital pressure machine (DPM) controlled fluids flow through a capillary design based microfluidic chip . . . . .	32
4.2	Internal rectangular, cylindrical resistances calculation of the designed microfluidic chip . . . . .	33
4.3	Different internal dimensional microfluidic wells of the microfluidic chip . . . . .	34
4.4	Equivalent MFR calculation to find out resultant fluid flow rate and pressure drop of entire microfluidic chip . . . . .	35
4.5	Estimation of average pressure drop across the points of two inlet wells using set values of fluid flow rates. . . . .	39
4.6	Quantitative data analysis of average pressure drop across the point of only outlet well using experimental measured values of fluid flow rates. . . . .	40
4.7	Statistical data analysis of average velocity versus fluid flow rate through the outlet well . . . . .	41
4.8	Statistical data analysis of average velocity versus fluid flow rate through the two inlet wells of same designed diameter in dimension . . . . .	42
5.1	Exact dimensional 3D structure formed in ANSYS simulator to run the program with different microfluidic parameters of fluid dynamics . . . . .	44
5.2	Validation of numerical model with experimental data for pressure drops at inlets and outlet . . . . .	48
5.3	Flowchart of 2D ANSYS Fluent simulation using fluid dynamics . . . . .	50
5.4	Mesh dependency for grid-independent study of validation . . . . .	52
6.1	Functionality of an ADF during the computational operation . . . . .	56
6.2	Sensing plate arrangement for the testing of effectiveness of the ADF . . . . .	57
6.3	Conceptual diagram of four wells plate with the application of THz wave . . . . .	58
6.4	Optical set up of TCM system to start measurement for this project . . . . .	59

6.5	Standard deviation versus step size curve for evaluating the characteristics of 100 ms and 30 ms TC of lock-in-amplifier with the application of ADF . . . . .	60
6.6	Calculation and estimation of standard deviation for several pixels of data of THz images inside the upper of the well and outside region of the Well onto sensing plate . . . . .	62
6.7	Contour of THz Image with filter for the contrast of 0.73, TC 100 ms . . . . .	63
6.8	Contour of THz Image without filter for the contrast of 0.32, TC 100 ms . . . . .	63
6.9	Contour of THz Image with filter for the contrast of 0.74, TC 30 ms . . . . .	64
6.10	Contour of THz Image without filter for the contrast of 0.34, TC 30 ms . . . . .	64
7.1	THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.07 mL/min fluid flow rate . . . . .	68
7.2	THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.15 mL/min fluid flow rate . . . . .	68
7.3	THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.25 mL/min fluid flow rate . . . . .	69
7.4	THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.4 mL/min fluid flow rate . . . . .	69
7.5	THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.7 mL/min fluid flow rate . . . . .	69
7.6	THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.8 mL/min fluid flow rate . . . . .	69
7.7	Same set of input velocity applied to each of two inlet wells with respect to different increased volume of fraction; same input velocity, 0.064 m/s applied at A and B points for two pH solutions.	70
7.8	Same set of input velocity applied to each of two inlet wells with respect to different increased volume of fraction; same input velocity, 0.096 m/s applied at A and B points for two pH solutions.	71





# List of Tables

4.1	Pressure drops through pH=4 inlet well for the certain ranges of set values of fluid flow rates from DPM . . . . .	36
4.2	Average Pressure drop through pH=10 inlet well for the certain ranges of set values of fluid flow rates from DPM . . . . .	37
4.3	Average pressure drop through outlet well from the experimental measured fluid flow rates . . . . .	37
4.4	Average velocity through outlet well from the experimental measured fluid flow rates . . . . .	38
4.5	Average velocity measurement through inlet wells using "set values" of the fluid flow rates of DPM . . . . .	38
5.1	Measured Different Fluidic Parameters Given As Input Data And Calculated Input, Output Data. . . . .	45
5.2	Validation Of Experimental And Simulation Based Fluidic Parameters. . . . .	47
5.3	Setup Model and Solver conditions . . . . .	49
5.4	Solution conditions . . . . .	49
5.5	Calculation conditions . . . . .	49
5.6	Mesh Dependency For Grid-Independent Study Of Validation Using Linear Regression Method . . . . .	51
6.1	Estimation of Standard Deviation and Contrast of THz Images .	63



# List of Notations and Symbols

Notation	Description
$\mu$	viscosity of the fluid.
$\rho$	Density of the fluid.
$R_e$	Reynold Number.
$p_{st}$	Static pressure of the fluid.
$\nabla$	Vector differential operator.
$F_{css}$	Surface tension force.
$\alpha_1$	Thermal Diffusivity.
$h$	Internal energy.
$\lambda$	Thermal conductivity.
$\mu m$	Wavelength of the light.
$THz$	High frequency Terahertz signal.
$Q_n$	Fluid flow rate.
$\Delta P_n$	Pressure drop of the fluid.
$R_f$	Microfluidic resistance.
$R_H$	Hydrodynamic resistance.
$D_{hh}$	Characteristic length of a fluid.
$E_q$	For each phase, specific heat.
$T$	Temperature.



*Dedicated to all the persons who I owe the most. To my parents for whom I am in this world, my wife and younger brother who sacrificed the most during my journey in Doctoral research.*



# Research Activities

## Peer-Reviewed Journal Papers (First author)

1. **Feroz Ahmed**, Atsuya Mahana, Katsuya Taniizumi, Jin Wang, Kenji Sakai, and Toshihiko Kiwa “Terahertz imaging technique for monitoring the flow of buffer solutions at different pH values through a microfluidic chip”. In: *Japanese Journal of Applied Physics*, (2021), DOI: 10.35848.

## Peer-Reviewed International Conference Papers (First author)

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2. **Feroz Ahmed**, Tatsuki Kamiya; Yuki Maeno; Toshihiko Kiwa; Kenji Sakai; Keiji Tsukada. “Improvement of contrast of Terahertz Images of a Terahertz Chemical Microscopy using Adaptive Digital Filter”. In: *2019 International Conference on Computer, Communication, Chemical, Materials and Electronic Engineering (IC4ME2)*, Bangladesh, 11-12 July 2019, pp. 1-4. DOI: 47184.2019.9036485.

### Springer Published:

3. **Feroz Ahmed**, Partha Chakraborty, Md. Monirul Kabir, Md. Shahjahan, and Kazuyuki Murase. “An Automatic Speaker Recognition System”. In: *International Conference on Neural Information Processing (ICONIP) 2007*, Japan, 2007, pp 517-526. DOI:10.1007.





# Chapter 1

## Introduction

When God said "Let there be light", He surely must have meant perfectly coherent light and photons. – Charles Townes

Terahertz (THz) - the electromagnetic wave with the frequency components between 100 GHz and 10 THz, possesses extra ordinary spectral features of its radiation, namely non-ionizing [Yan+16], [Dan+19], non-invasive [Yan+16], [Dan+19], phase-sensitive to polar substances [Dan+19], [PMW09], penetration capabilities [Yan+16] through non-polar molecules, high spatial resolution [Dan+19],[NP+17] etc. which show the scenario of this technology with higher significance in case of applications in spectroscopy [PZ15], [He2] sensing [Zho+19] and imaging [Sta+20] processes. Therefore, chemical solutions, cells and tissues with different water contents exhibit with the distinctive responses to the radiation of THz wave. Moreover, the trace of the orientation of hydrogen bond due to its intermolecular bonding can be revealed by THz imaging technology. THz radiation is gradually diminished as one of its important characteristics when it travels through a medium with an increased water concentrations. Therefore, it can distinguish even small fluctuations in water contents of chemical and biological substances like pH buffer solutions, tissues [He2], blood [Jeo+13] etc. which make THz vital technology for revealing and capturing the interaction, chemical reactions of different chemical

solutions and abnormal cell divisions inside the tissues to detect cancer cells and so on. Therefore, it is possible to capture the internal microfluidic reactions of the microfluidic chip using THz laser scanning in the form of THz images with the detailed maps of chemical and biological reactions.

Microfluidics system [SW03; BMW02] is one of emerging solutions for the miniaturization of biological [SW03; Bur02; MH02] and chemical assays since its enrollment in the field of biomedical engineering. Its feature of small scale integration enables it to capture the targets like detection of proteins [BMW02; Ver02; Slo+05], cell sorting and analysis [CK07; BLL06; Che+08; Sal+10], nano-particles [Par+17; Kel+08], cells [GYF04], DNA detection and amplification [Dor+05; Eas+06], germs [Wan+18], chemical ions and so on within the short internal structural distances to make electric or magnetic fields more useful for sensing and detecting the above mentioned all objects. The structures of modern microfluidics are capable of miniaturizing or integrating traditional laboratory practices by making lab-on-chips devices for the sake of reducing time and saving cost [Sai+11a].

At the microscale level, microfluidic devices [YII18] utilize the physical and chemical properties of liquids and gases. They come into being with several benefits over the others existing conventional sized systems. Moreover, microfluidic devices permit the analysis and use of less volume of samples, chemicals and reagents to decrease the global fees of applications . Many operations can be implemented at a time with the excellent features of their compact sizes, shortening the time during the experiment. In addition, their features contain an excellent data quality and substantial parameter control which permit the process of automation while preserving the performances of them. Moreover, they have the capability to both process and analyze the samples with minor sample handling. Microfluidic chips are expanded so that the incorporated automation permits the users to generate multi-step reactions with the requirements

of a low level of expertise and multi-functionalities. The eye-catching services like faster reaction time, enhanced analytical sensitivity, enhanced temperature control, portability, easier automation and the integration of lab routines in the lab-on-a-chip devices make microfluidics versatile for the diverse uses in chemical and biological research. The manufacturing materials of microfluidics are cheap as they do not involve with the use of various costly equipments.

## 1.1 Use of THz Chemical Microscope (TCM) to utilize THz Technology in this research

In the method of THz Time-Domain Spectroscopy (TDS) , the amplitude and phase of short-pulsed incident THz (i.e.,  $10^{12}Hz$ ) radiation show the high sensitivity to the properties of materials of a sample. Therefore, some particular imaging systems which utilize THz TDS, are suitable for the applications [Ton07] of all novel nondestructive testings (NDT) [Kiw+03] . However, the spatial resolution of such systems which is about  $300\mu m$  at 1THz, is so far limited by the wavelength of the THz radiation.

By considering the above mentioned fact, a laser-THz emission microscope (LTME) was proposed and developed for NDT of semiconductor devices like in the large-scale integrated circuits and solar panels [Yam+05; Kiw+03]. One of important features in terms of the spatial resolution of the laser-THz emission technique is that its spatial resolution is not limited by wavelength of the THz radiation, only limited by the wavelength of the femtosecond (i.e.,  $10^{-15}s$ ) laser pulses that is utilized to generate the THz radiation. As a result, comparing with a conventional THz imaging system, higher spatial resolutions can be achieved using an LTME [KST13]. From this concept of sharp spatial resolution criteria and advantages of THz technology, my laboratory developed a

THz chemical microscope (TCM) which is used to visualize, capture and monitor the chemical, biological and/or electrical potential shifts caused by chemical reactions in detail. In my PhD research, I used such TCM based THz technology to utilize its impact on my designed microfluidic chip with the explanation from fluid dynamics. Besides, a digital filter was integrated with the THz controlling system to check the acceptable visibility of the brightness of contrast of THz images using the above mentioned TCM.

### **1.1.1 Reasons of the use of THz Technology over other technologies**

The traditional image captured cameras do not have the full satisfactory features to visualize the internal chemical and biological reactions inside the microfluidic chips [Ahm+21] convincingly. Comparing with the application of charge coupled device (CCD) cameras available in commercial markets for capturing the images of the interaction of chemical and biological reactions inside the microfluidic chips, the use of THz image sensing technology is the latest version as one of important value added services in lab-on-chip technology for visualizing and capturing of the image maps of chemical and biological reactions. Thus, the breakthrough of the above mentioned novel THz imaging technique is evident with the visualization, detection and capturing of chemical reactions of different pH buffer solutions without using any fluorescent dye marker. Therefore, such technique is regarded as a reliable label free detection and visualization technique. THz technology can be effective though the microfluidic channels fabricated from different materials which are opaque in visible and near-infrared domain. That means; they do not show the transparency in light. In the long run, from my designed microfluidic chip, for the fabrication of more moderate versions of the microfluidic chips, choice of materials or extension of properties of materials can be possible to add by experimenting the degree of effectiveness

of THz technology in a wide scale range for the sake of the progressive reports on chemical and medical research.

### 1.1.2 Noises Involved in THz Signals and Images Processing

In THz image sensing technique, there are two types of components of noises [Han+06] existed. One is signal noise and another one is imaging noise. The signal noise is responsible for the deformation of THz pulses in some unwanted ways, as opposed to the noise which an artifact of the process of imaging. On the contrary, imaging noise is associated with the noise for example; the incoming analogue signal was sampled by the Analog-to-Digital Converter (ADC) via the lock-in-amplifier where definitely the generation of noises in a process which invariably resulted in a digital approximation of the original analogue signal in spite of the presence of a low pass filter inside the lock-in-amplifier. Moreover, the alignment of each of optical instruments specially the alignment of pumping and trigger beams of THz signal were done by the human made controlling which in turn have the possibility of not maintaining the alignment always in a perfection mood. As a result, imaging noise might be occurred. To overcome different types of regular and irregular noises associated with the optical to electrical converted signal, it was necessary to design a compatible digital filter to handle such noises and reduced the noises to ensure the higher resolution and contrast based processing of THz images.

### 1.1.3 Applied Digital Filter in THz Technology

From the experiences of the existence of different noises involved in THz pulsed imaging technique, an Adaptive Digital Filter (ADF) was designed [Ahm+19] and integrated with the central THz image sensing system. There was also another important objective hidden inside such type of designing filter. The

objective was the visualization of the chemical and biological reactions with higher THz imaging quality and higher scanning speed. Therefore, it was required to improve the contrast of the THz images with higher quality of images. I applied an ADF to the TCM system and evaluated the quality of images qualitatively. The outcome was effective to improve the imaging quality without losing the scanning speed of the TCM and improved the total measurement time of each of samples from THz scanning laser as well. This filter design and its integration to the TCM system have already opened the door to capture the imaging maps of faster spreading out of chemical, biological solutions and/or movie clip. Thus, a breakthrough of faster scanning procedure for the faster movement of chemical and biological solutions would be effective and the latest edition in our novel cutting edge research of THz image sensing technology.

## 1.2 Fluid Dynamics in Microfluidics

The growing demands of microfluidics systems create the opportunity in the research and development field vastly to analyze and visualize [HSR13] the internal fluid flow parameters inside the micro flow channels (MFC) of a microfluidic chip (MC) which is the essential phenomenon to ensure good quality based chemical and biological products for the betterment of next healthy generation. As a result, it is a huge expected trend to a fluid flow design and estimation engineer to characterize, optimize and evaluate the performance of the interaction of chemical and biological solutions through the MFC for the quality assurance in the microfluidic platform. In this research work, I designed and fabricated two inlet wells and one outlet well based a capillary [Ola+18] design concept based MC from 3D printing and polydimethylsiloxane (PDMS)-made mold. Two pH 4 and 10 Buffer Solutions (BS) were injected into the two inlet wells and flushed out through the only outlet well of the capillary designed MC to analyze internal fluidic parameters of mixed two viscous BS.

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It was built up a microfluidic circuit network [Oh+12] from different dimensions of MFC and fluidic parameters used for the designed MC. From the basic concept of Fluid Dynamics (FD), the pressure drop profiles, velocity profile of mixed two pH buffer solutions through two inlet wells and one outlet well were calculated, plotted and analyzed to evaluate the performance of the capillary designed MC. At the end, the experimental results were validated by the 2D simulation applied to the above mentioned MC using ANSYS Fluent simulator and from simulated data, velocity in each portion of the MFC was analyzed.

### **1.2.1 The Speciality of the Designed Microfluidic Chip over other Conventional Microfluidic Chips**

Owing to the structures of smaller dimensions of the different internal MFC of MC, the flow of fluids through MFC exhibit the laminar [KCD16c] nature of fluids' flow. However, due to the complexities of the states of the mixture of fluids, when two or more different viscous fluids mix with each other through the MFC, it is obvious to simulate, visualize and analyze the flow patterns of fluids at each point of the internal flow channels of a MC for the characterization and optimization of the nature of flow of fluids and different internal dimensions of MFC. Therefore, we need to generate 2D simulation using ANSYS fluent simulator [BGP08; APA19] with the output of simple and precise able fluid flow patterned data influenced by the friction due to the viscosities of the fluids within the internal layers of MFC [UH+18] with the variation of different speeds of different fluids. Thus, the performance evaluation and optimization of different dimensions of internal MFC can be achieved by the analyses of characteristics of internal microfluidic parameters like pressure drops, velocities, viscosities, microfluidic resistances etc.

Compared to the increased demands of varieties of complex internal architectures' formed microfluidic chips, the conventional method for the control of

the flow profiles through the internal straight forward rectangular paths based microfluidic chips [Sai+11b] due to the linear behavior of hydrodynamic equations at low Reynolds number, is experiencing tremendous challenge. Moreover, many chemical and biological researches based on lab on chips require precise ideas of the samples of the fluids flow through different dimensional and patterned (rectangular paths, circular shaped cylindrical paths, common junction circular paths and, so on) areas with the sharp analyses based on both the experimental and simulation works. Therefore, we designed and fabricated a Y-shaped MC with different dimensional rectangular paths, cylindrical shaped circular wells, a common junction based circular path [GGBY16; BL17] to combine the flow of two fluids. As fluid flow design and estimation engineers, experimental and simulation works gave us the scope of the optimization of different dimensional shaped paths of different flow of fluids based on fundamentals of fluid dynamics [Lin+16]. As a result, we achieved the goal to optimize a symmetric two arms and a common mixed fluids provided MFC contained a Y-shaped MC due to its good response of handling the flow of fluids. Further, for the injected two pH 4 and 10 BS through the MC, experimental measurement data were used to calculate all hydrodynamic parameters via internal MFC using FD.

### 1.3 Outline of this Thesis

The remainder of this dissertation has been planned and structured as follows:

**Chapter 2** describes and defines the fundamental concept of THz image sensing technology, the higher spatial resolution of that technology in comparison with conventional laser excited system, experimental optical set up of TCM system to use for the experiment with the microfluidic chip, THz signal flow graph of the TCM system, the mechanism of THz generation, detection principles.



**Chapter 3** introduces the capillary designed concept based an MC, its fabrication and final replica mold formation for making experimental shaped polydimethylsiloxane (PDMS) based an MC, advantages of using such PDMS made MC.

**Chapter 4** contains the mathematical expressions used for the calculation of internal microfluidic parameters of the MC, using all hydrodynamic parameters, establishment of a microfluidic circuit network and calculation of parameters of such network with the determination of the nature of flow of different fluids through the MC.

**Chapter 5** includes the characteristics curves of pressure drop profiles, velocity profile according to FD from the experimental result of the designed MC. Then it was simulated for the trace of fluidic patterns of two different pH BS through the MC in ANSYS Fluent simulator. This chapter also concludes the very important validation of experimental results from the analysis of simulation based results. This chapter also explains the results of the obtained data of THz scanned lasers. The main contribution of this chapter is to find the THz imaging data which were recorded for the different fluid flow rates (FFR) of two pH BS by evaluating the average peak amplitude data of THz with respect to different concentrations of different pH BS.

**Chapter 6** presents the basic principle of ADF and its algorithm following least mean square (LMS) of digital signal processing technique. It also introduces the optical set up of TCM system to check the efficiency of ADF which was integrated with the central THz image processing system for monitoring and visualization.

**Chapter 7** concludes the qualitative analysis of the designed ADF for the improvement of the contrast of THz images. Here for the evaluation of the performance of the ADF, mili-Q water was used as sample to determine the different parameters of ADF.

Finally, **Chapter 8** includes this dissertation along with an outline of the future works.

## 1.4 Motivation

The research trend on different patterned MFC are the concerned matter for the sharp analysis of blood, cell samples, chemical solutions for the early medical diagnosis of different patients in comparison with the conventional straight forward rectangular path MFC [Sai+11a] which actually motivated me to fabricate capillary design concept based different patterned channels like cylindrical or circular path channels, common junction circular channel and rectangular paths within an MC.

Moreover, for the reduction of noises from THz signal data, it was required to implement an universal digital filter in spite of a low pass filter integration with our optical to electrical conversion part of lock-in-amplifier. Thus, the central TCM system connected with analog-to-digital converter (ADC) and lock-in amplifier were required to deliver almost noises free THz images which motivated me to design and test an ADF with our central THz image controlling system.

## Chapter 2

# Mechanism of THz Imaging

This chapter recalls the mechanism of THz generation, detection principles, experimental optical set up of TCM system used to operate the designed microfluidic chip, THz signal flow graph of the TCM system. The details presented in this chapter, will be found in [Kiw+19; Kiw+07], [Ahm+19; Ahm+21].

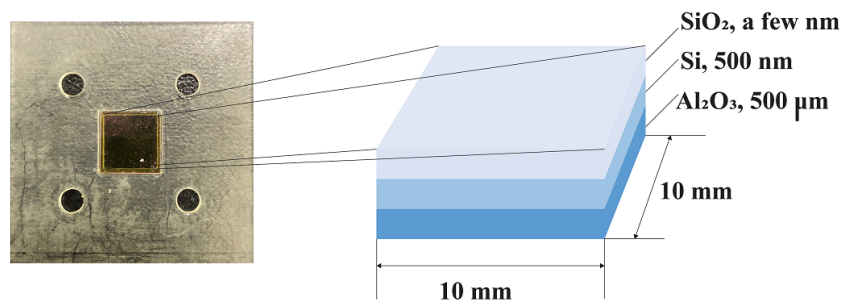


FIGURE 2.1: The sensing plate for the TCM system with three different layers fabricated sensing chip

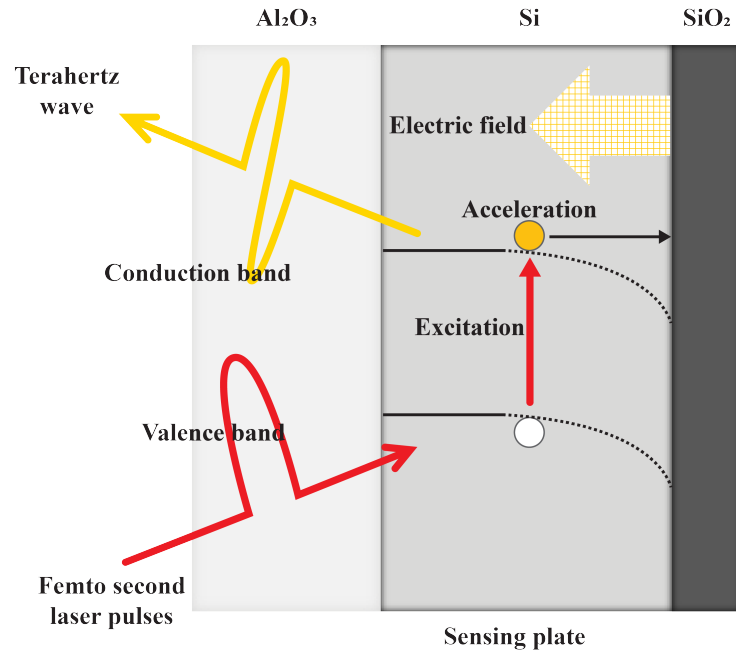


FIGURE 2.2: THz electromagnetic waves generation with the irradiation of femto second laser pulses onto the layers of the sensing chip

## 2.1 THz generation principle

To establish full set up of the TCM system, we used a sensing chip as a sensor, as shown in Fig. 2.1. It comprises of three layers of thin films of  $\text{SiO}_2$  and  $\text{Si}$  deposited on a sapphire substrate. When the silicon layer was illuminated by femtosecond laser pulses of 800 nm wavelength from the sapphire substrate side, the electrons in the valence band was excited to jump to the conduction band, as shown in Fig. 2.2. An electron was excited to jump into the conduction band because it absorbed a photon of the femtosecond laser pulse, whose photon energy was higher than the bandgap energy of silicon. The dependency of the magnitude of the formed depletion region was on the electrical potential at the surface of the sensing plate. As a result, we can determine radiated THz wave's amplitude, which depended on the magnitude of the depletion field. When pH buffer solutions of different pH values like 1.68, 4, 6.86 and 10, flowing through the micro flow channels interacted on the sensing plate, the electric potential at the surface of the sensing plate and the amplitude of the radiated THz waves

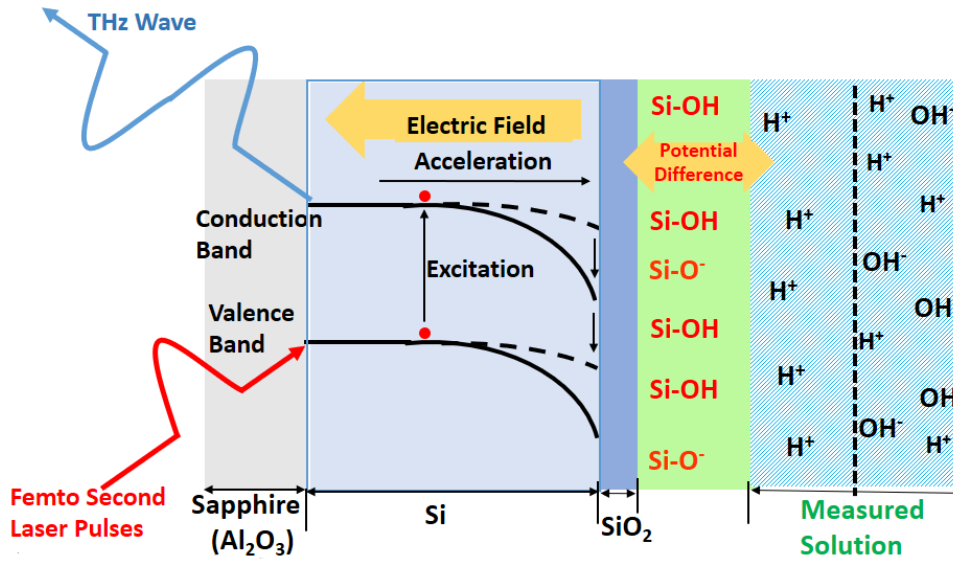


FIGURE 2.3: THz electromagnetic wave's detection to capture the pH buffer solutions interaction

were varied. As a result, a THz electromagnetic wave is proportional to the differential of the current density by the carrier acceleration that was generated according to 2.1.

$$E_{THz} \propto \frac{dJ}{dt} = e \frac{dn}{dt} \mu E_d S + en \mu \frac{dE_d}{dt} S \quad (2.1)$$

In 2.1,  $E_{THz}$  denotes the THz electromagnetic wave,  $J$  indicates the current density,  $e$  means the electric charge,  $n$  defines the photo-excited carrier density,  $S$  refers to the irradiated area of the laser spot,  $E_d$  symbolizes the surface electric field, and  $\mu$  denotes the carrier mobility. Figure 2.2 depicts the generation principle of the terahertz waves that was originated from the sensing plate.

## 2.2 THz detection principle

In the project, THz laser was irradiated to the bottom side of the sensing plate through the sapphire substrate. Depletion field's magnitude depended on the electric potential at the surface of the sensing plate. And, the radiated terahertz

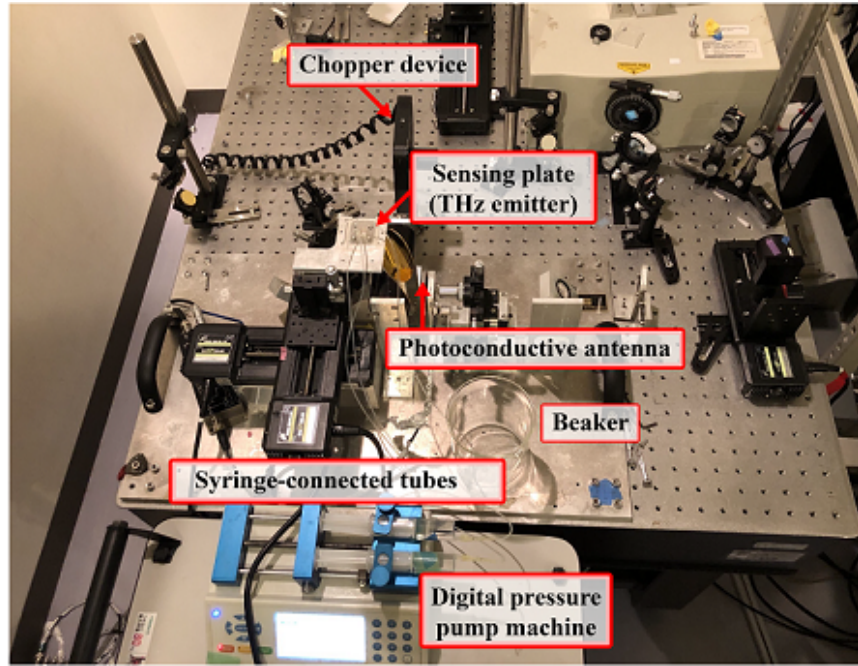


FIGURE 2.4: Optical part arrangement of the TCM system for this project

waves' amplitude depended on the magnitude of the depletion field. The chemical reactions of different concentration based pH buffer solutions occurred onto the sensing plate that led to the change in the electric potential at the surface of the sensing plate by detecting the interaction of the continuous flow of  $H^+$  and  $OH^-$  ions based different pH buffer solutions. As a result, the concentration of  $H^+$  and  $OH^-$  ions inside pH buffer solutions was able to be measured by measuring the amplitude of the radiated terahertz waves. TCM system measured the amplitude of THz waves that radiated from the sensing plate.

The  $H^+$  ion concentration is higher than the  $OH^-$  ion concentration for acidic nature based pH solutions (4, 1.68, and 6.86). On the contrary, the  $OH^-$  ion concentration is higher than the  $H^+$  ion concentration for a pH=10 buffer solution. In both cases, the  $H^+$  and  $OH^-$  ions interacted with each other to reach chemical equilibrium that led to a potential shift on the surface of the sensing plate. This operation led to changes in the acceleration of the carriers. Therefore, generated terahertz wave's intensity also changed, and it was measured the interaction between different  $H^+$  and  $OH^-$  ion concentrations from the change

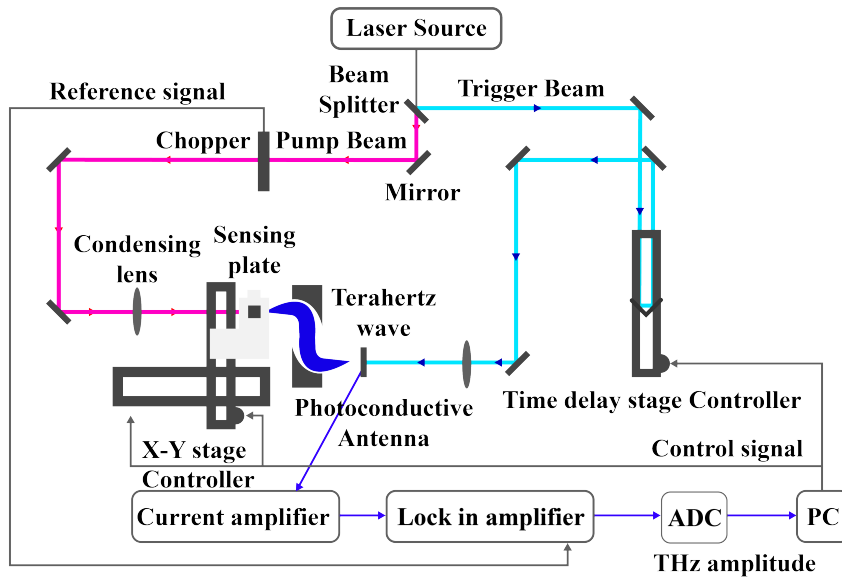


FIGURE 2.5: Whole THz signaling system to convert the optical signal to electrical signal with a two stage controlling mechanism

of the intensity of the THz amplitude. Figure 2.3 depicts the principle of THz detection .

## 2.3 Experimental set up of TCM

We selected a Ti: sapphire laser as a light source to develop a THz signaling system. 780 nm with a pulse width of 100 fs at the full width at half maximum was the center wavelength of the laser pulse. 790 W with a repetition rate of 82 MHz was the output of the laser system. We selected a digital pressure machine (DPM) connected to the two inlet wells of the microfluidic chip via two syringe-connected tubes for allowing the flow of different pH buffer solutions. The mixed pH solutions like pH 4, 1.68, 6.86, and 10, were flushed out spontaneously and stored into a beaker using another tube of 1 m from the outlet well. Figure 2.4 illustrates the experimental setup .

A THz detector detected the amplitude of the THz pulses. A bowtie-type photoconductive antenna made by low-temperature-grown GaAs thin film, was selected as a detector. The peak amplitude of the THz wave was determined from the optical delay of the trigger pulses to the detector, that was fixed. A lock-in technique was employed for improving the signal-to-noise ratio (SNR) of the THz signal. To reduce the noise from the THz signal data, We selected a chopper (To synchronize the THz data, the frequency of the chopper was 2 kHz ) in the lock-in technique. Furthermore, an adaptive digital filter (ADF) and an analog-to-digital converter (ADC) were programmed by a two-stage time delay controlling mechanism in LabVIEW® for improving the contrast of the THz images and sampling the THz signal data in a compatible manner based on the Nyquist sampling theorem. The THz signaling system in the form of a signal flow graph (SFG) shows the sequential architecture of its operation, as drawn in Fig. 2.5.

## 2.4 Characteristic Curve Analysis from the Experiment

During the experiment (Fig. 2.4), THz data of the interaction of two pH = 4 and 10 buffer solutions were captured and plotted as the contour of the THz images with respect to different fluid flow rates (viz: 0.07, 0.15, 0.25, 0.4, and 0.6 mL/min etc.) of digital pressure machine that applied to each of the two inlet wells of the designed microfluidic chip, driven by a THz laser-scanned TCM system.

The averaged data of the THz amplitudes with respect to different concentrations of the pH value are plotted in Figs. (2.7 and 2.6) using the programmed average [Ahm+19] raw data of the contour images of THz from the experimental measurement. To measure and evaluate such raw THz data, a total area



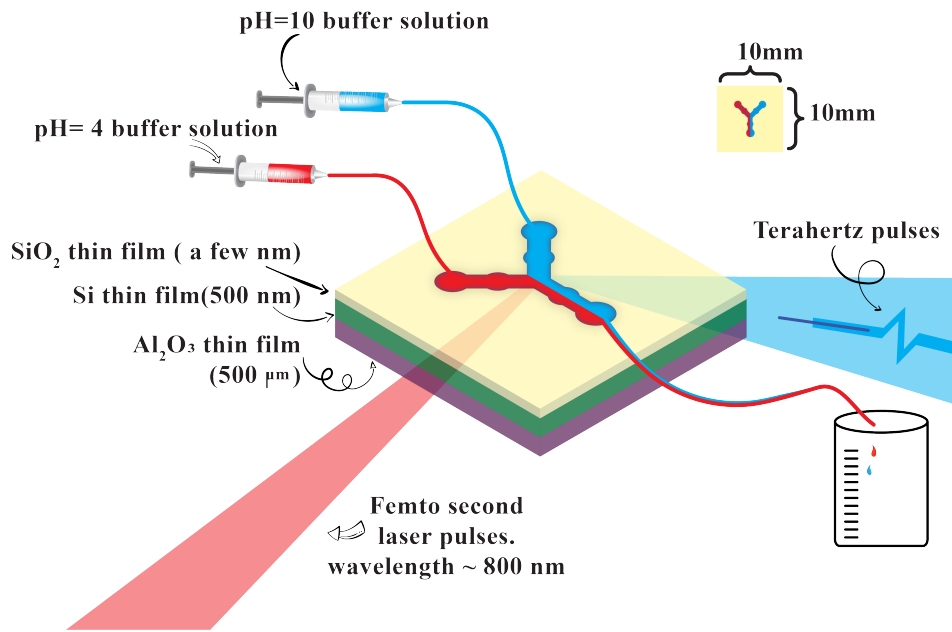


FIGURE 2.6: By applying THz electromagnetic wave, buffer solutions at pH values of 4 and 10, injected through two inlet wells via syringe-connected tubes and flushed out through the outlet well

of laser scanned of 10 mm × 10 mm was programmed in LabVIEW® by averaging the lock-in-amplifier data over the total measurement time [Ahm+19]. The characteristic curve denotes the criteria of the peak values of the THz data, that increased monotonically with the increased values of the pH concentration. Meanwhile, from the THz generation and detection principles (Figs. 2.2 and 2.3), the negative value of pH = 1.68 in Fig. 2.7 can be realized significantly, where a downward or reverse shift in the depletion layer indicated the potential shift in the depletion region, which led to a negative peak amplitude of the THz data. This sensitiveness of the TCM to pH values proved the quantitative measurements of the pH values via the micro flow channels of the designed microfluidic chip. In this way, even a small variation in the pH level can indicate the sensitiveness of THz amplitudes data to different pH levels data with the interaction of the pH values through the microfluidic chip, where the applications of pH buffer solutions is of great interest.

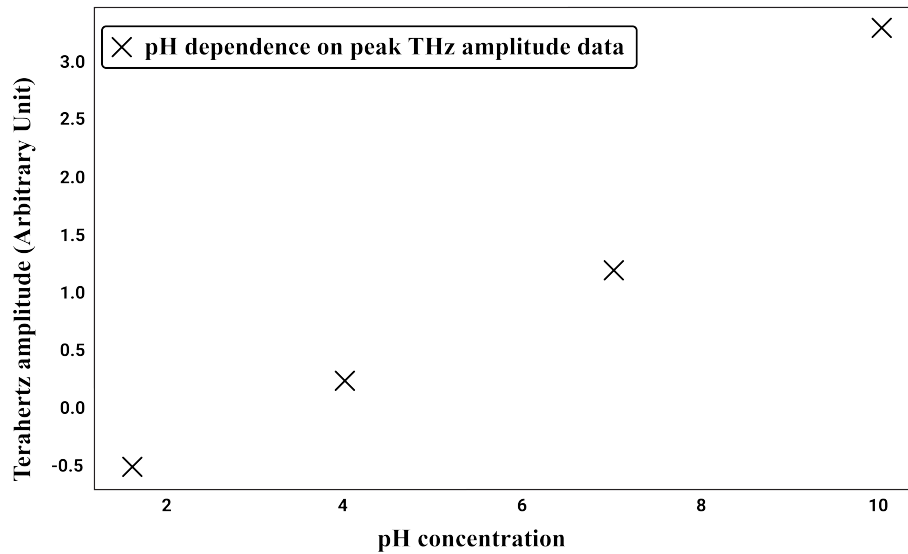


FIGURE 2.7: Characteristic curve of THz response with respect to different pH concentration values

## 2.5 Summary

This chapter defined the fundamentals of TCM system system's hardware set up, its mechanism followed by working principles of advanced physics and engineering. Moreover, the measurement and experimental data obtained from THz image sensing technology revealed the effectiveness of THz laser scanning in order to capture the internal microfluidic reactions by showing variations in different buffer solutions of different pH values.

## Chapter 3

# PDMS designed Microfluidic Chip

**Chapter 3** introduces the capillary designed concept based a microfluidic chip, its fabrication and final replica mold shape for making PDMS based a microfluidic chip, advantages of using such PDMS-made microfluidic chip.

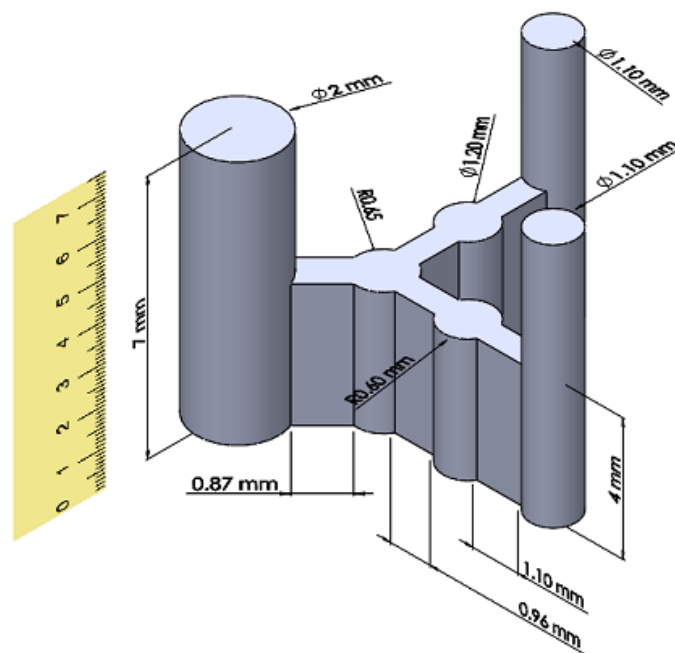


FIGURE 3.1: Different dimensions of micro flow channels' 3D layout in Solid work.

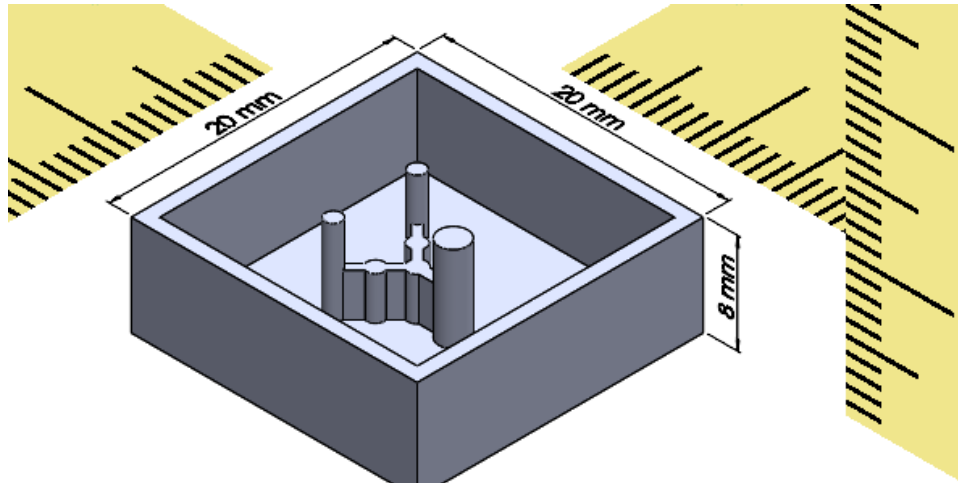


FIGURE 3.2: 3D-printed structure of the microfluidic chip with dimensions of 20 mm  $\times$  20 mm scaling in Solid work

### 3.1 Preparation of PDMS

PDMS [Bor+18] is frequently used to fabricate micro total analysis systems [Cul+14; ZH19] and lab-on-a-chips [MT17]. One base and one curing agent named, a sylgard® 184 silicone elastomer base agent and a sylgard ®184 silicone elastomer curing agent as an adhesive were chosen to fabricate PDMS replica mold that are manufactured by Dow Corning (Midland, Maryland, USA). A PDMS-made microfluidic structure was developed followed by an irreversible assembly process using the above mentioned base and curing agents. Two inlet wells and one outlet well based (Fig. 3.1) a 3D structure of a microfluidic chip was drafted using a student version Solid work software, translated it into STL format (Fig. 3.2) and then printed the structure (Fig. 3.3) from the high definition 3D printer (Agilista-3100; serial no.: 96M14458) for this research work. PDMS sample made microfluidic chip was developed by cross-linking both the base and curing agents at a mass ratio of 10:1 [Ver+14; Ahm+21].

In the next step, for the fabrication of a replica mold of the 3D structure, the two cross-linked base and curing agents were poured into this structure of the dimensions of 20 mm  $\times$  20 mm and height of 7 mm from the upper part of base surface (Fig 3.3 and Fig. 3.10). The poured solutions were then placed

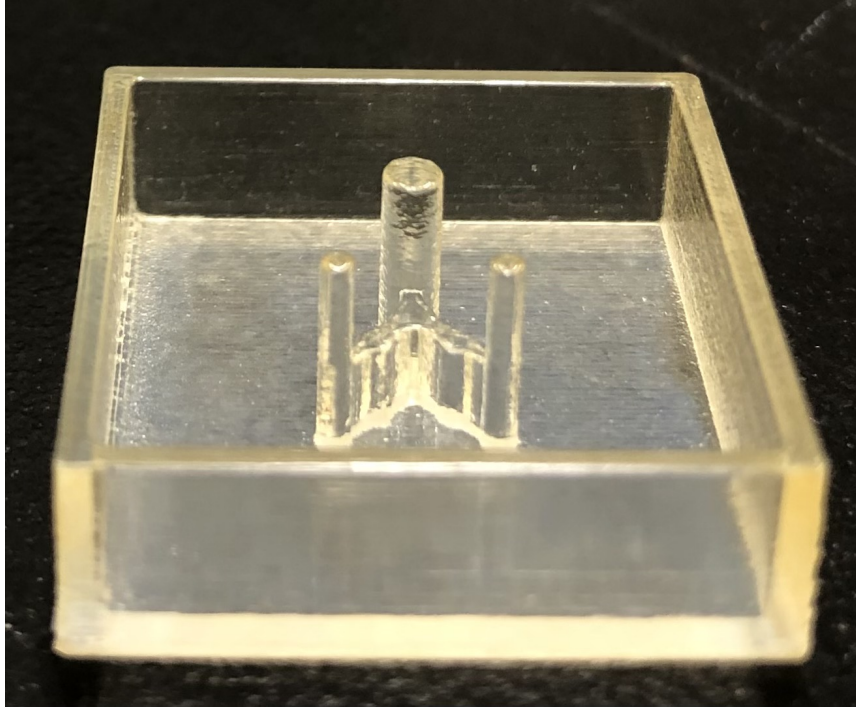


FIGURE 3.3: practical 3D printed structure for preparing replica of this structure

into an oven at a controlled temperature of  $40^{\circ}\text{C}$  for 24 h for the confirmation of sufficient hard shape of the replica mold (Fig. 3.4). The main part of central micro flow channels were designed at the height of 7 mm for each of two inlets and the only outlet from the upper part of base surface and at the height of 4 mm of internal micro flow channels from the top part of the base surface of the 3D printed structure (Fig.3.1). When the replica mold within the structure is hard enough, the replica mold was cut manually from the dimensions of 20 mm  $\times$  20 mm structure to the dimensions of 10 mm  $\times$  10 mm (Fig. 3.5). Thus, a PDMS made microfluidic chip was formed for doing the experiment. The bottom part of the microfluidic chip was attached with a Si: sapphire glass substrate to ensure the flow of fluids without any interruption or the leakage inside the micro flow channels of the microfluidic chip [Ahm+21].

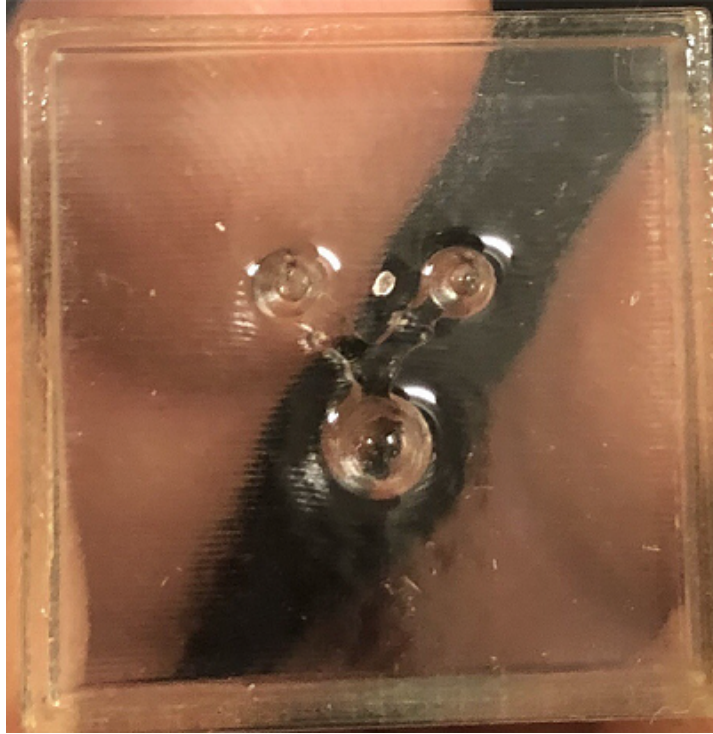


FIGURE 3.4: Hard replica mold within 3D printed structure

### 3.1.1 Top and bottom plates design to squeeze PDMS

It was difficult to ensure intensive flow of fluids through the micro flow channels without any upper external support to squeeze the PDMS-made microfluidic chip. Therefore, a similar top plate (Fig. 3.7) of  $40\text{ mm} \times 40\text{ mm}$  was designed, screwed and tilted mechanically with the bottom sensing plate (Fig. 3.6) by squeezing the microfluidic chip (Fig. 3.9) with the Si: sapphire glass substrate. As a result, without any risk of the leakage of the solutions outside the micro flow channels, the spontaneous flow of butter solutions was ensured. As a precautionary step, in order to avoid any chance of the leakage of the fluidic solutions outside the micro flow channels, the screwing and tilting should be balanced enough for the microfluidic chip during the experiment time [Ahm+21].

The whole procedure for the experimental microfluidic chip with such arrangement is shown in Fig. 3.9 below.

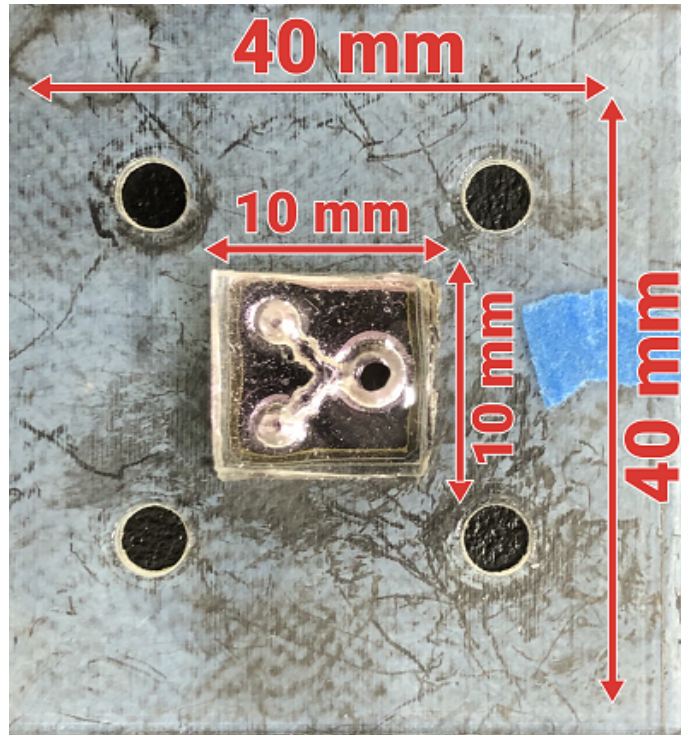


FIGURE 3.5: Manual cut microfluidic chip attached with Si: sapphire glass substrate (clipped with bottom 3D printed plate)

### 3.1.2 Cross-sectional view of designed microfluidic chip and isometric view of central main structure of microfluidic chip

The top isometric view of the designed chip (Fig. 3.11), side cross sectional view (Fig. 3.10) and top close dimensional view (Fig. 3.8) were drawn in solid works software [SSW15; Mat13] to understand the design criteria.

### 3.1.3 Advantages and key features of PDMS

The primary advantages [Bor+18] of it over other substrate materials are the economic feature with low cost, easy and fast fabrication, high gas permeability and excellent optical transparency through the visible wavelengths into the near ultra-violet wavelengths and chemical inertness. Also, the interesting features of low shrinkage rate and the capability to replicate micro-scale environment make PDMS to suit nicely with soft lithography and 3D structure printing processes.

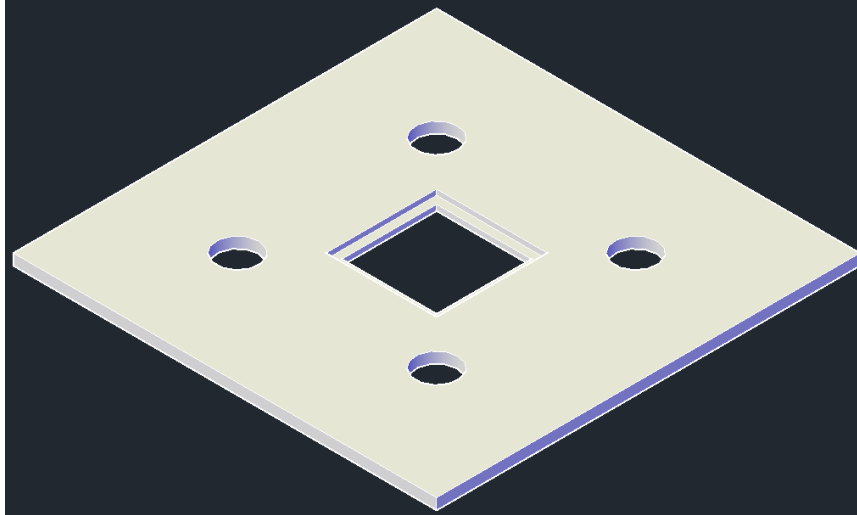


FIGURE 3.6: Bottom plate designed in CAD software (In the middle, Si:sapphire glass substrate of  $10\text{ mm} \times 10\text{ mm}$  was clipped and PDMS-made MC was attached with glass substrate)

### 3.1.4 Summary

PDMS is so popular material used for the fabrication of microfluidic chip in the clean room and laboratories using different photolithographic process and 3D printing technique. Here in this dissertation, PDMS-made microfluidic chip was used efficiently to test chemical solutions for the control quality control of pH controlled foods.



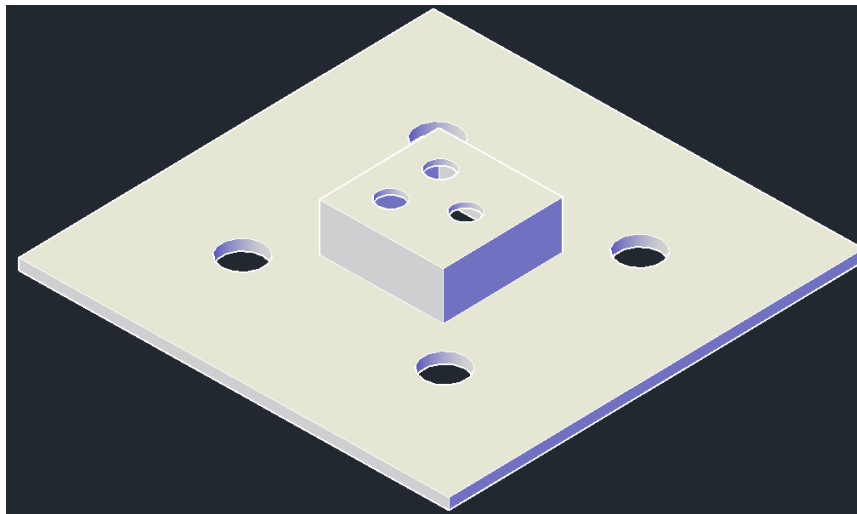


FIGURE 3.7: Top plate designed in CAD to squeeze the microfluidic chip with glass substrate and bottom sensing plate

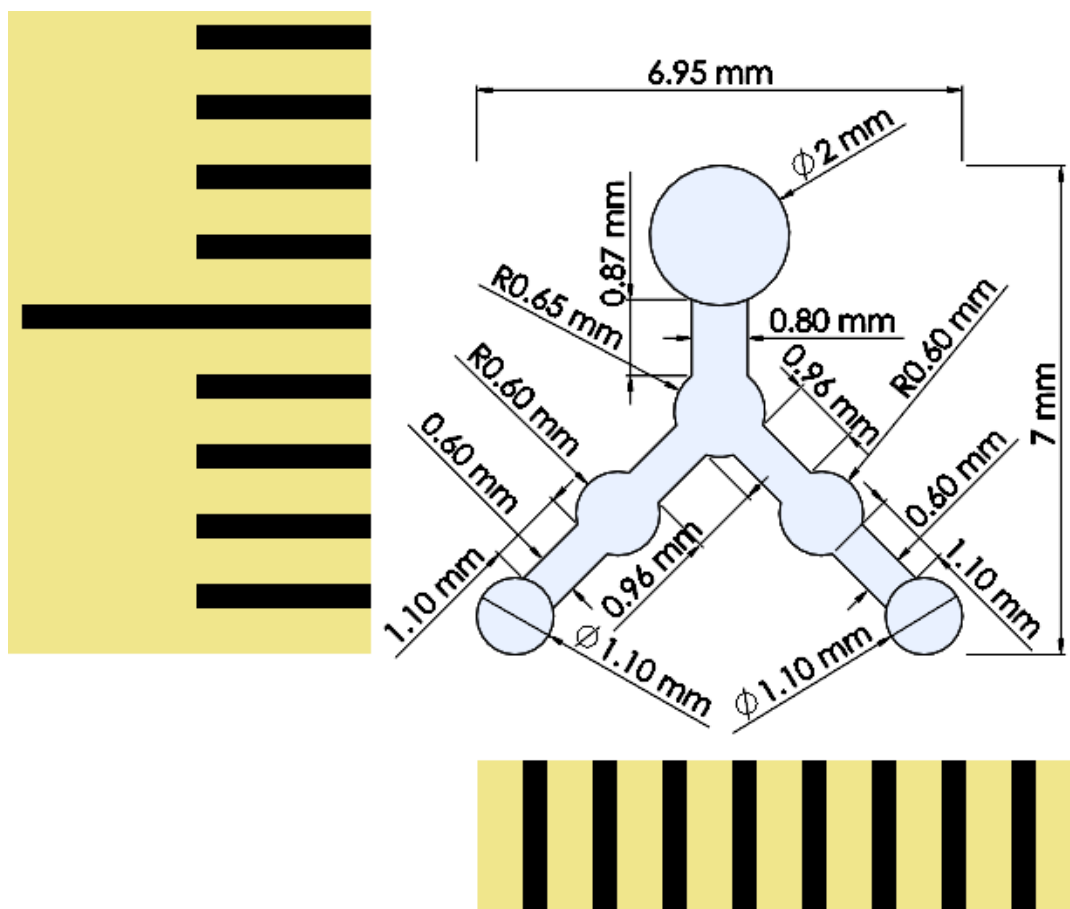


FIGURE 3.8: Top view of internal micro fluidic channels formation in Solid works using different internal designed dimensions

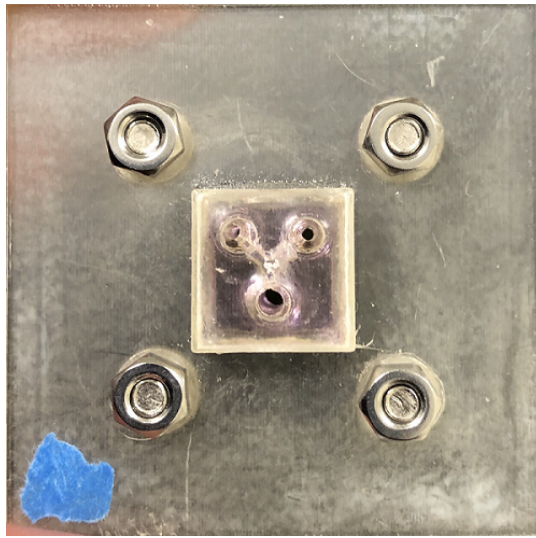


FIGURE 3.9: Microfluidic chip that attached with Si:sapphire glass substrate, screwed and tilted in a balanced way to squeeze it with top and bottom plate

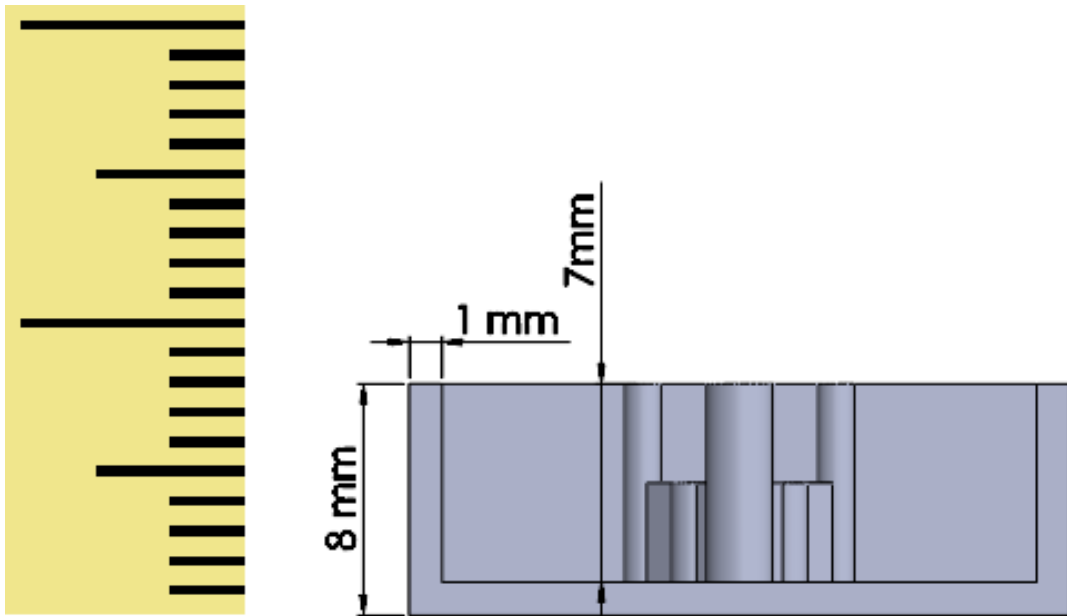


FIGURE 3.10: cross-sectional view of designed 3D structure

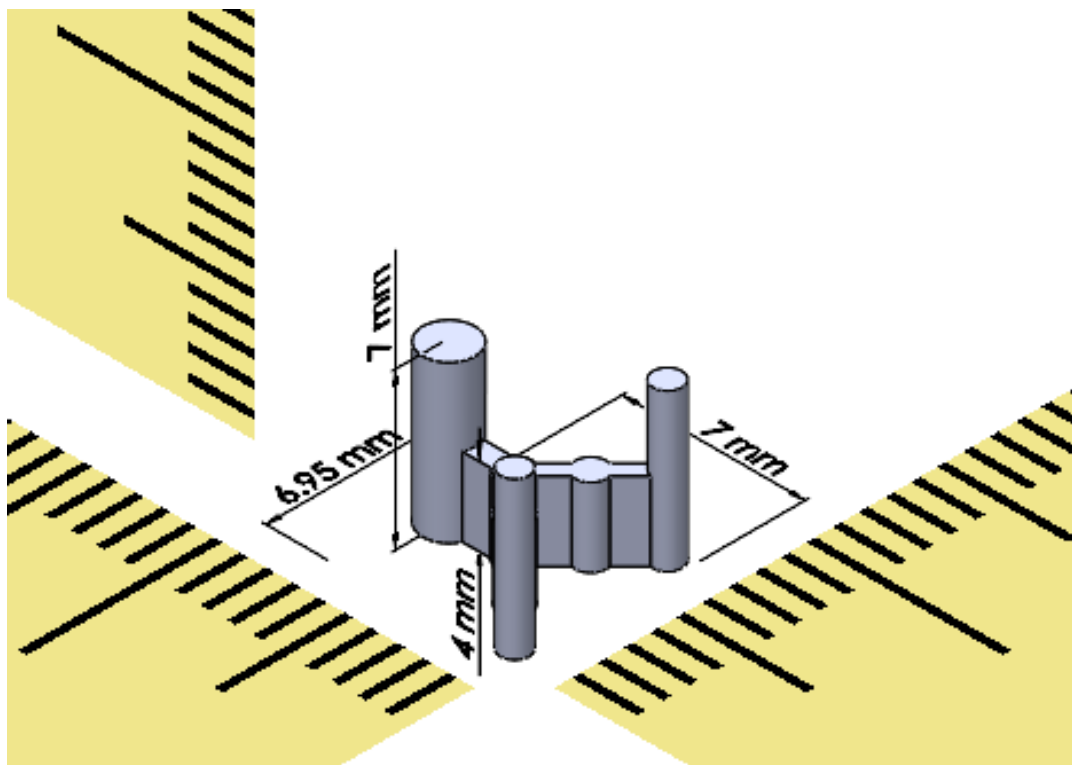


FIGURE 3.11: Scaling in solid work of main microfluidic chip's lay out along with micro flow channels



## Chapter 4

# Microfluidic Parameters

## Measurement, Calculation and Statistical Analysis

### 4.1 Introduction

#### Background and Motivation

Evaluation of microfluidic parameters for the analysis of the performance of a microfluidic chip is important because the calm and quiet nature of flow of fluids solutions can give the suitable micro-scale environment for the analysis of chemical and biological samples like pH controller liquid food samples, blood samples etc.

### 4.2 Mathematics

On the basis of nature of the geometry of the micro flow channels , the way it is combined the resistances of electrical circuit components, the similar way micro fluidic resistances (MFR) participate to calculate the total resistance of a microfluidic platform. According to the concept of the electrical circuit model,

to deduce fluid flow rate, MFR values and given pressure drop applied to a microfluidic chip followed by the Hagen- Poiseuille's equation [PPUn a] is used:

$$\Delta P_n = Q_n R_f \quad (4.1)$$

where  $R_f$  is the MFR of micro flow channels,  $Q_n$  is the volumetric different fluid flow rates and  $\Delta P_n$  is different pressure drops across the micro flow channels correspond to different fluid flow rates respectively.  $n = 1, 2, 3 \dots N$ . In case of laminar nature of an isothermal, incompressible and isotropic Newtonian fluid flow through a tube or cylindrical shaped inlets and one outlet with a circular cross section and no-slip boundary conditions on sidewalls of microfluidic structure, the MFR can be modeled mathematically [GGBY16] below:

$$R_{Tubing} = R_{Circular/Cylinder} = \frac{128\mu L}{\pi d^4} \quad (4.2)$$

where  $\mu$  =viscosity of a fluid,  $L$  =length of the given tube or cylindrical shaped well,  $d$  = diameter of the given tube or cylindrical shaped well.  $\frac{1}{d^4}$  term signifies that even a small change in  $d$ , there will be a vast change in values of fluidic resistances. For the rectangular cross section, fluid flow rate,  $Q$  can be formulized in terms of pressure drop  $\Delta P$  over the micro flow channels' length,  $L$  according to equation 4.1 and can be deduced as:

$$Q = \Delta P \frac{w_1 h^3}{12\mu L} \quad (4.3)$$

The pressure drop  $\Delta P$  can be expressed to show a relation in between hydrodynamic resistance [GGBY16; BL17] in the channel,  $R_H$ , also called chip resistance,  $R_{chip}$  and the FFR similar to the voltage drops in electrical current.

$$\Delta P = QR_H = Q \frac{12\mu L}{w_1 h^3} \quad (4.4)$$

where

$$R_H = R_{chip} = \frac{12\mu L}{w_1 h^3} \quad (4.5)$$

$w_1$  is in between width and height of rectangular channels of smaller dimension and  $h$  is the height of larger dimension. Equation 4.5 is to calculate the hydrodynamic or chip resistance . The continuity equation [PPUn b] is,

$$Q = A_1 \bar{v} \quad (4.6)$$

where  $A_1$  = area of either a circular or a rectangular path and  $\bar{v}$  is the average velocity of a given fluid. Area of inlets/outlet,  $A_1 = 2\pi r(r+h)$  where  $r$  =radius of cylindrical shaped inlets/outlet and  $h$ =height of them. Reynolds number  $R_e$  is a dimensionless number to measure a fluid in motion qualitatively that evaluates the ratio of convective forces to viscous forces. It is expressed as:

$$R_e = \rho D_{hh} \frac{\bar{v}}{\mu} \quad (4.7)$$

where  $\bar{v}$  is the average fluid velocity and  $D_{hh}$  is the characteristic length of a fluid.  $R_e \leq 2300$  means the flow is laminar and.  $R_e > 2300$  means the flow is turbulent [PPUn a].

### 4.3 Calculation of Microfluidic Resistances of Microfluidic Chip

From the experiment in the laboratory, it was measured the viscosities of pH=4 and 10 buffer solutions of 0.5 mPa·s and 0.59 mPa·s respectively. The viscosity of pH=4 and 10 mixed solution is 0.58 mPa·s. The given length of each of two syringes connected two tubes was 2 m length and 2 mm diameter. The designed microfluidic chip of two inlet wells and one outlet well based micro flow channels consisted of the intermediate wells, a common junction well (CJW) and

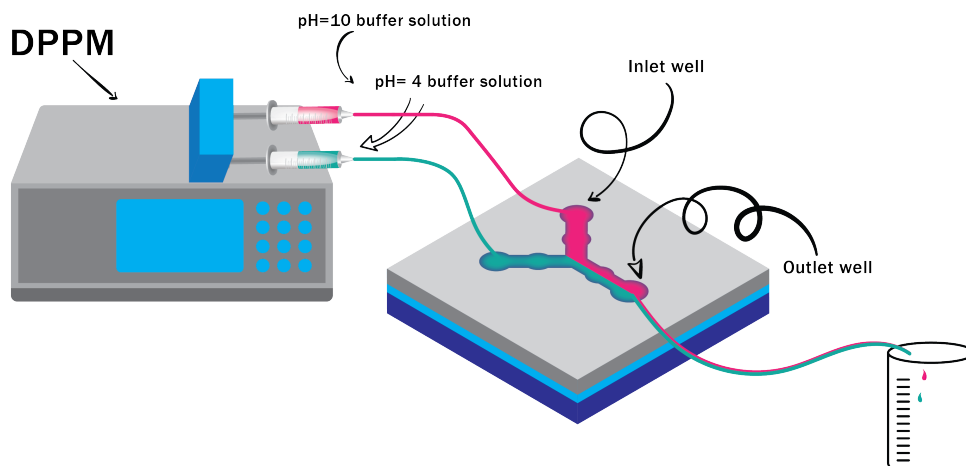


FIGURE 4.1: Digital pressure machine (DPM) controlled fluids flow through a capillary design based microfluidic chip

rectangular paths which was drawn in Fig. 4.4.

The mixed accumulated buffer solutions into the common junction well were flushed out through the outlet well into a beaker using a tube of 1 m length and diameter of 2 mm. The height of each of two inlet wells and one outlet well were 7 mm from the fabricated base surface of the MC structure and the height of micro flow channels was 4 mm from that of the base surface. Therefore, using all of the above fluid flow parameters for different dimensions of microfluidic chip, we calculated tubing resistances [GGBY16] for the syringes connected tubes, chip resistances [BL17] for the rectangular paths and cylindrical resistances [GGBY16] for the circular (Fig. 3.1) shaped cylindrical fluid flow paths with the other hydrodynamic properties of the injected pH=4 and 10 buffer solutions using equations 4.2 and 4.5 that sketched in Fig. 3.1. In the Fig. 4.2, from the sides of pH 4 and 10 buffer solutions, if we move towards the final common path of these mixed two solutions, we can find that the values of all types of microfluidic resistances are decreasing gradually. It means that from the injected flow paths of both pH 4 and 10 buffer solutions, the fluid flow rates are increasing according to the equation 4.1 at the final destination of the flushed out of the mixed two buffer solutions via a beaker.



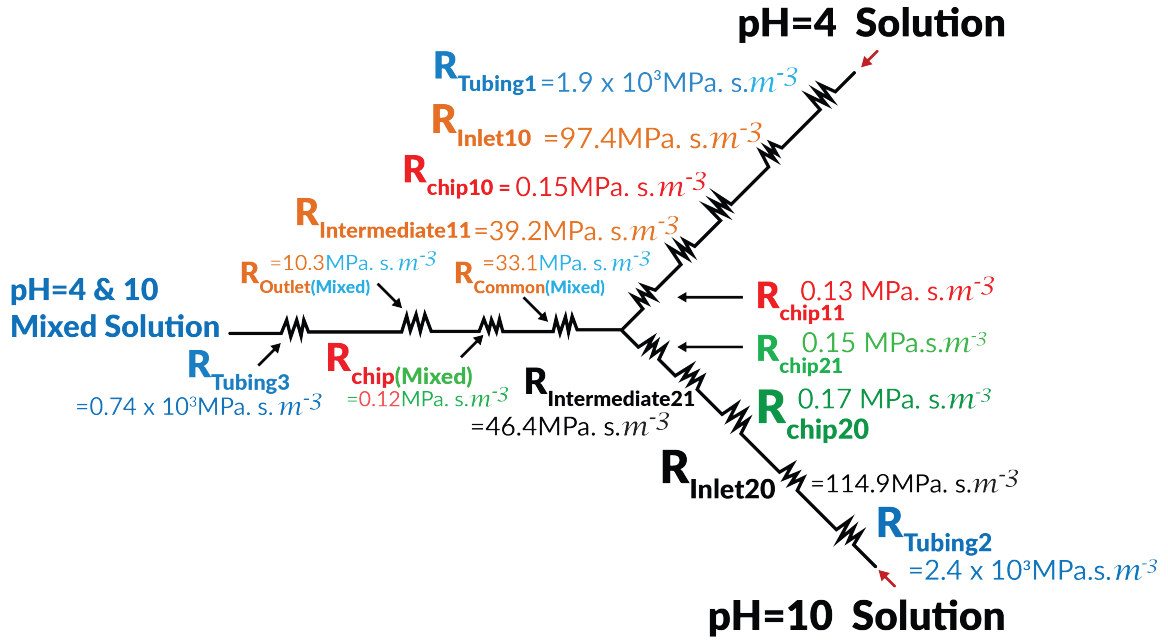


FIGURE 4.2: Internal rectangular, cylindrical resistances calculation of the designed microfluidic chip

During the design of the microfluidic chip, the diameter of the outlet well and the dimensions of all different patterned common paths towards the outlet well were made larger in comparison with the diameters of each of two inlets and other different dimensions with different patterned (rectangular paths, circular shaped cylindrical paths) fluid flow paths. Because after the mixture of two fluids, heat transfer, temperature change and other internal parameters inside the micro flow channels might affect the nature of mixed fluids widely. For pH=4 solution's injected fluid flow path, if the number of MFR arranged in series, an equivalent single microfluidic resistance has a hydraulic resistance equal to the sum of four hydraulic resistances [Oh+12].

$$R_{1eq.} = R_{inlet10} + R_{chip10} + R_{intermediate11} + R_{chip11} \quad (4.8)$$

$$R_{1eq.} = 169.85 MPa.s.m^{-3} \quad (4.9)$$

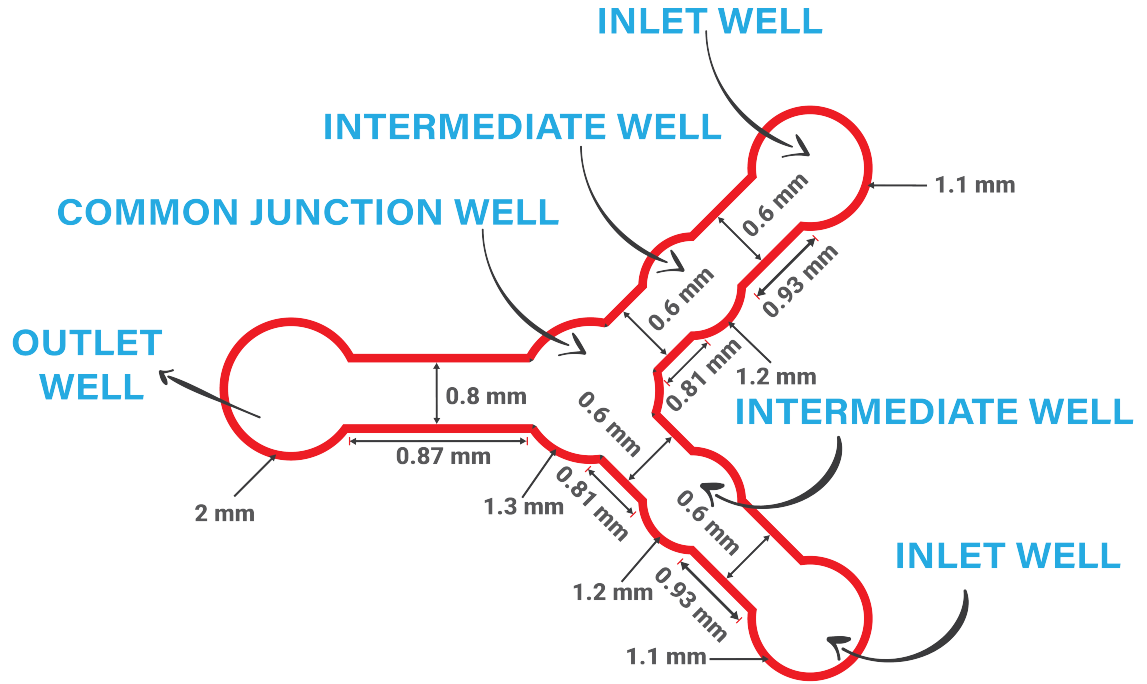


FIGURE 4.3: Different internal dimensional microfluidic wells of the microfluidic chip

Similarly, for pH=10 Solution, the injected Fluid Flow Path (FFP),

$$R_{2eq.} = R_{inlet20} + R_{chip20} + R_{intermediate21} + R_{chip21} \quad (4.10)$$

$$R_{2eq.} = 161.62MPa.s.m^{-3} \quad (4.11)$$

And, for Mixed common FFP,

$$R_{3eq.} = R_{com.mix} + R_{chip_{mix}} + R_{out.mix} \quad (4.12)$$

$$R_{3eq.} = 43.52MPa.s.m^{-3} \quad (4.13)$$

As an external pressure source driven MC, we denoted our used syringe pump conveying fluids flow through two tubes as independent constant fluid flow

sources  $Q_{S1}, Q_{S2}$ . Practically, our MC were connected with digital pressure machine via syringe-connected tubes where two inlets and one outlet normally opened at atmospheric pressure to flush out the mixture of two pH BS. From the electrical circuit analogy [Oh+12] compared with the microfluidic network analogy, the outlet is regarded as a ground as it was opened at the atmospheric pressure. The equivalent MFR of entire MC was drawn in Fig. 4.4. It was

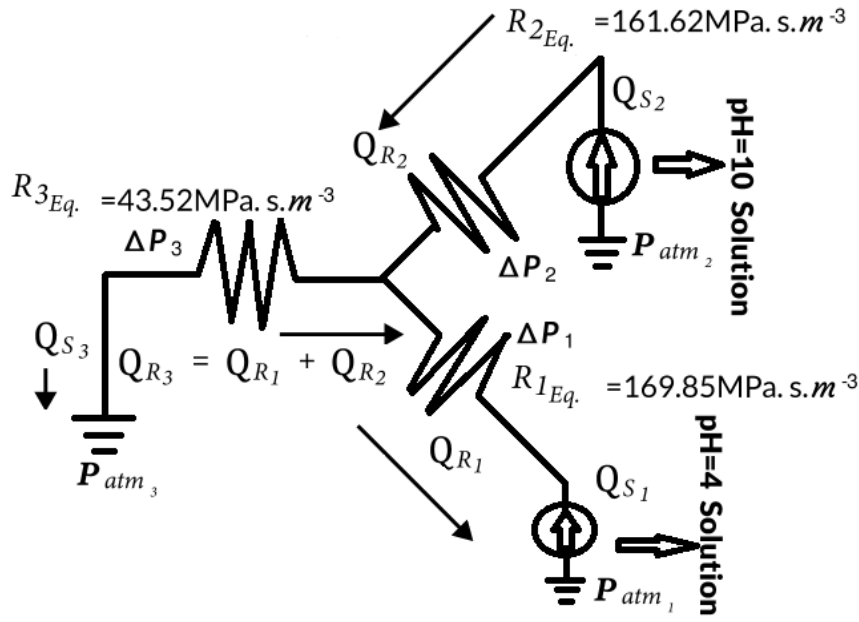


FIGURE 4.4: Equivalent MFR calculation to find out resultant fluid flow rate and pressure drop of entire microfluidic chip

found out  $R_e$  at outlet well of microfluidic chip by calculating average velocity from the experimental measured values of different fluid flow rates. For outlet well, mixed solutions' density is 0.9901 g/mL. Average velocity at outlet well according to the experimental measured different fluid flow rates is calculated below:

$$v = \frac{(0.031 + 0.057 + 0.09 + 0.14 + 0.3 + 0.15 + 0.39 + 0.481)}{7} \quad (4.14)$$

$$v = 0.21ms^{-1} \quad (4.15)$$

Therefore, from (7), we got  $R_e = 2.51$  .

TABLE 4.1: Pressure drops through pH=4 inlet well for the certain ranges of set values of fluid flow rates from DPM

Inlet Flow rate, $Q_n, mL/min$	$\Delta P_1, Pa$	$\Delta P_2, Pa$	$\Delta P_3, Pa$	$\Delta P_4, Pa$	$\Delta P_5, Pa$	Average Pressure drop, $Pa$
0.03 to 0.07	0.049	0.065	0.081	0.097	0.114	0.0812
0.09 to 0.13	0.146	0.162	0.179	0.195	0.211	0.1786
0.14 to 0.18	0.227	0.244	0.26	0.276	0.292	0.2598
0.23 to 0.27	0.373	0.39	0.406	0.422	0.438	0.4058
0.38 to 0.42	0.617	0.633	0.649	0.666	0.682	0.6494
0.58 to 0.62	0.942	0.958	0.974	0.99	1.007	0.9742
0.78 to 0.82	1.266	1.283	1.299	1.315	1.331	1.2988
0.97 to 1.01	1.575	1.591	1.607	1.624	1.64	1.6074

#### 4.4 Pressure Drop Measurement and Velocity Profile Estimation from Fluid Dynamics

During the experiment, the digital pressure machine was set at different fluid flow rates by pumping two syringe-connected tubes automatically to allow the flow of pH=4 and 10 buffer solutions through the designed microfluidic chip. Different pressure drops corresponding to different five “set values” of fluid flow rates from digital pressure machine were calculated for two inlet wells from equation 4.1 by putting the values of cylindrical resistances,  $R_{inlet10}$  and  $R_{inlet20}$  of two inlet wells from Fig. 4.2. Then, by choosing certain intervals (viz. 0.3 to 07 mL/min, 0.09 to 0.13 mL/min etc.) for different fluid flow rates, the average pressure drop was calculated (Table 4.1 and Table 4.2 ) for each of fluid flow rate’s interval at two inlet wells and data were plotted that increased monotonically (Fig.4.5) with the increased fluid flow rates in case of two pH buffer solutions.

In Fig. 4.6, with the increased values of fluid flow rate through the microfluidic chip, the average pressure drop increased linearly to satisfy the equation 4.1 of Hagen-Poiseuille’s law of fluid dynamics. During the experimental measurement, each reading of pressure drop was taken five times repeatedly with

TABLE 4.2: Average Pressure drop through pH=10 inlet well for the certain ranges of set values of fluid flow rates from DPM

Inlet Flow rate, $Q_n, mL/min$	$\Delta P_1, Pa$	$\Delta P_2, Pa$	$\Delta P_3, Pa$	$\Delta P_4, Pa$	$\Delta P_5, Pa$	Average Pressure drop, $Pa$
0.03 to 0.07	0.057	0.077	0.096	0.115	0.134	0.0958
0.09 to 0.13	0.172	0.192	0.211	0.23	0.245	0.21
0.14 to 0.18	0.268	0.287	0.306	0.326	0.345	0.3064
0.23 to 0.27	0.441	0.46	0.48	0.5	0.517	0.4796
0.38 to 0.42	0.728	0.747	0.766	0.785	0.804	0.766
0.58 to 0.62	1.111	1.13	1.149	1.168	1.188	1.1492
0.78 to 0.82	1.494	1.513	1.532	1.551	1.571	1.5322
0.97 to 1.01	1.896	1.915	1.935	1.877	1.858	1.8962

TABLE 4.3: Average pressure drop through outlet well from the experimental measured fluid flow rates

In each Inlet, $Q_{in}, mL/min$	Outlet Flow rate, $Q_0, mL/min$	$\Delta P_1, Pa$	$\Delta P_2, Pa$	$\Delta P_3, Pa$	$\Delta P_4, Pa$	$\Delta P_5, Pa$	Average Pressure drop, $Pa$
0.07	0.093	0.0159	0.0158	0.0161	0.162	0.016	0.016
0.1	0.17	0.029	0.031	0.03	0.028	0.28	0.0796
0.15	0.27	0.047	0.046	0.047	0.047	0.045	0.0464
0.25	0.41	0.07	0.07	0.073	0.076	0.071	0.072
0.4	0.59	0.102	0.103	0.1	0.101	0.1	0.1012
0.6	0.9	0.151	0.152	0.16	0.155	0.156	0.1548
0.8	1.16	0.2	0.201	0.2	0.202	0.2	0.2006
1	1.45	0.25	0.25	0.252	0.25	0.251	0.2506

respect to the same fluid flow rate using trial and error method. Then, each average pressure drop data was calculated with respect to each fluid flow rate and found out the range of standard deviation for each average pressure drop value. From the graph, all average pressure drop data followed linear nature except 0.15 mL/min data that was deviated from the normal criteria because of error in trial and error method during the measurement time. The microfluidic resistance  $R_{outletmixed}$  at the outlet well was taken from the calculated resistance values of Fig. 4.2.

The average pressure drop (refer to Table 4.1 and Table 4.2) in each interval

TABLE 4.4: Average velocity through outlet well from the experimental measured fluid flow rates

Outlet Flow rate, $Q_0, mL/min$	$v_1, mm/s$	$v_2, mm/s$	$v_3, mm/s$	$v_4, mm/s$	$v_5, mm/s$	$v_{Avg}, mm/s$
0.093	0.031	0.03	0.031	0.031	0.03	0.0306
0.17	0.056	0.059	0.058	0.055	0.055	0.0566
0.27	0.091	0.088	0.091	0.09	0.087	0.0894
0.41	0.134	0.134	0.14	0.15	0.14	0.1396
0.59	0.292	0.3	0.305	0.3	0.302	0.2998
0.9	0.151	0.152	0.16	0.155	0.156	0.1548
1.16	0.39	0.39	0.38	0.39	0.39	0.388
1.45	0.474	0.48	0.49	0.481	0.484	0.4818

TABLE 4.5: Average velocity measurement through inlet wells using "set values" of the fluid flow rates of DPM

Inlets Flow rate, $Q_{in}, mL/min$	$v_1, mm/s$	$v_2, mm/s$	$v_3, mm/s$	$v_4, mm/s$	$v_5, mm/s$	$v_{Avg}, mm/s$
0.03 to 0.07	0.019	0.026	0.032	0.038	0.045	0.032
0.09 to 0.13	0.057	0.064	0.07	0.077	0.083	0.0702
0.14 to 0.18	0.089	0.096	0.102	0.109	0.115	0.1022
0.23 to 0.27	0.153	0.255	0.166	0.172	0.147	0.1596
0.38 to 0.42	0.243	0.377	0.249	0.262	0.268	0.2554
0.58 to 0.62	0.37	0.152	0.383	0.39	0.4	0.384
0.78 to 0.82	0.498	0.505	0.511	0.517	0.523	0.5108
0.97 to 1.01	0.62	0.626	0.632	0.639	0.645	0.6324

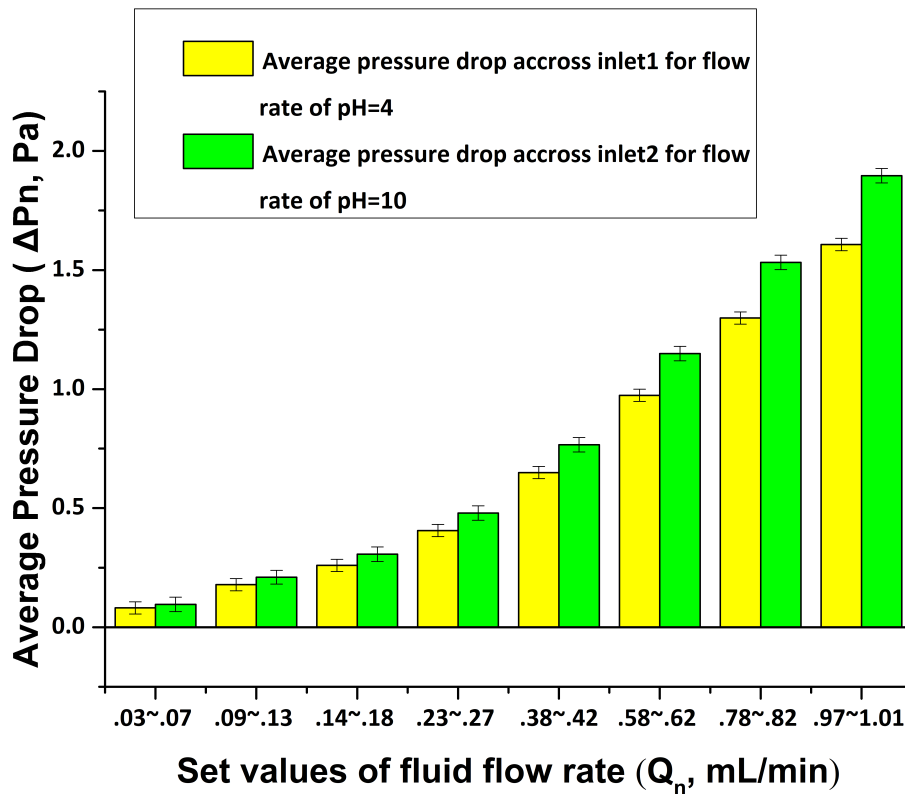


FIGURE 4.5: Estimation of average pressure drop across the points of two inlet wells using set values of fluid flow rates.

of fluid flow rate (Fig. 4.5) is higher than the average pressure drop (Table4.3) in each case at the outlet well of Fig. 4.6. Because syringe-connected tubes experienced huge pressure drop as the air did not be able to fully escape from inside the syringes in reality. To push into the plungers of syringes automatically by digital pressure machine, the air inside the plungers of two syringes were compressed into smaller volume that created much high pressure inside the two syringes. As a consequence, this high pressure automatically applied to the tubes connected with the syringes that affected the effective diameters of flexible structure of two tubes instantaneously to handle the high Pressure drop inside the tubes. Therefore, these two syringe-connected tubes were connected with two inlet wells that created overall higher impact on the pressure drop inside the two inlet wells than that of a tube connected outlet well. There is also another important reason of such type of higher pressure drop across

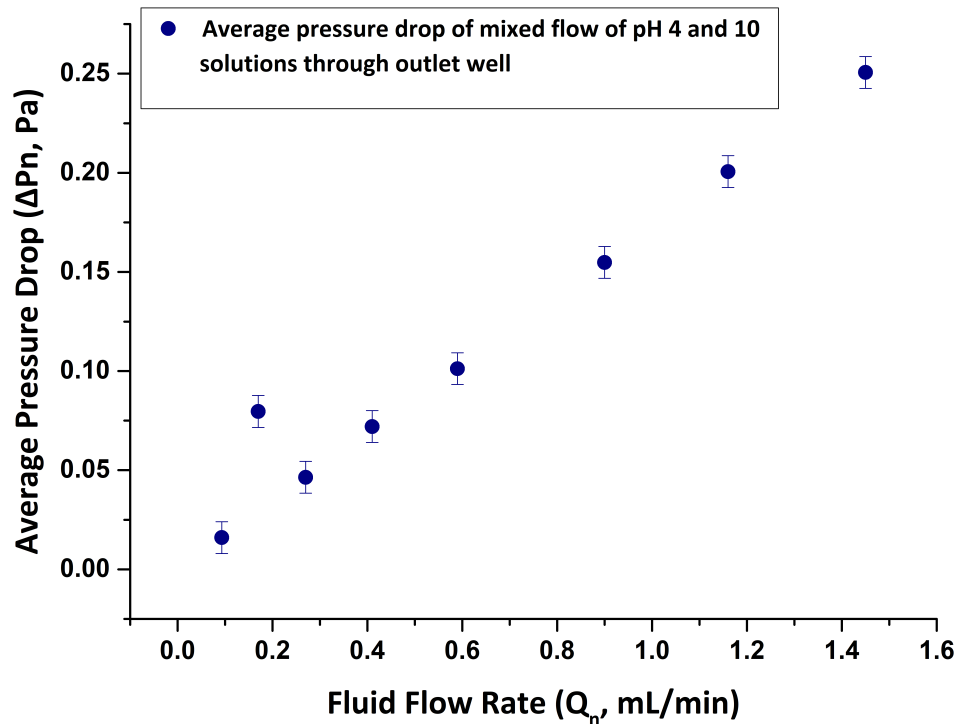


FIGURE 4.6: Quantitative data analysis of average pressure drop across the point of only outlet well using experimental measured values of fluid flow rates.

the two inlet wells. The diameter of each of cylindrical shaped two inlet wells designed (1.1 mm each) was lower than the diameter of outlet well (2 mm). As a result, the pressure drop across two inlet wells experienced higher pressure drop 4.5 compared with that of outlet well 4.6, according to the equation 4.1 where pressure drop is proportional to the fluid flow rate.

In Fig. 4.7, the average velocity at the outlet well was estimated by taking five readings (refer to Table 4.4) of measured value of same fluid flow rate at the outlet well. For doing this measurement, mixed pH 4 and 10 buffer solutions were flushed out automatically into a beaker through a tube-connected with only outlet well of the microfluidic chip. Then, the volume of taken-out mixed fluids calculated from the total occupied volume of the beaker in mL for the certain programmed total measurement time. As a result, each fluid flow rate



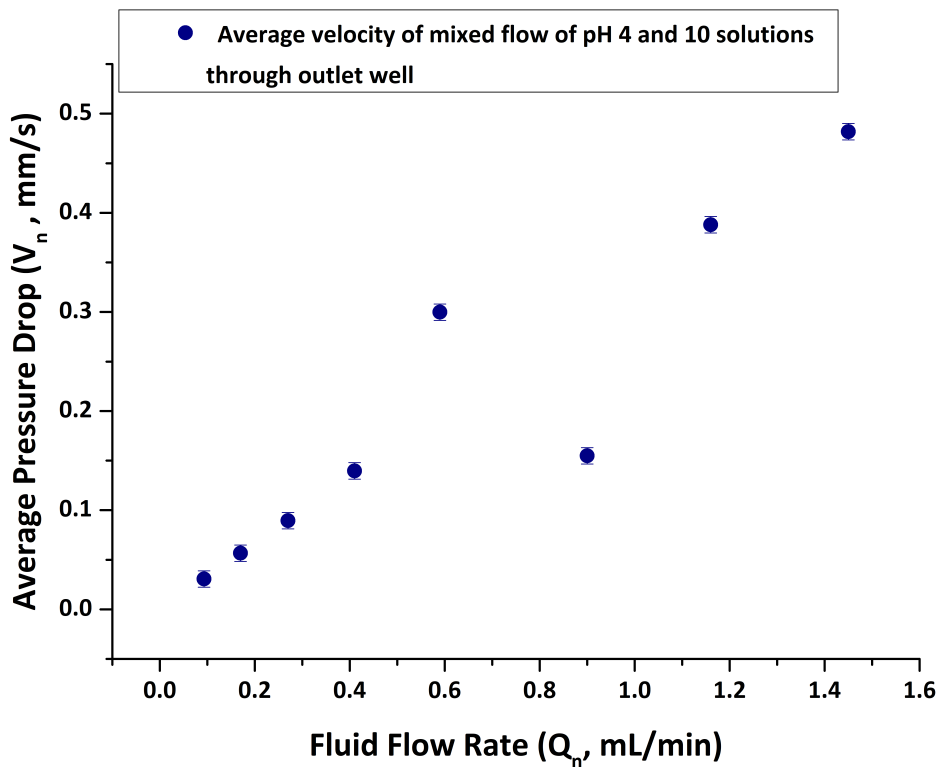


FIGURE 4.7: Statistical data analysis of average velocity versus fluid flow rate through the outlet well

was calculated. From such five readings of each of fluid flow rate, five readings of velocity was calculated in each case and using equation 4.6, average velocity was calculated. The curve was plotted to find the different average velocities at the outlet well for different fluid flow rates. The curve shows linear nature of average increased velocity with respect to different fluid flow rates except the fluid flow rates at 0.6 and 0.7 mL/min because of the error in trial and error method during the experimental measurement time.

In the Fig.4.8, using equation 4.6, for different "set values" of fluid flow rates from digital pressure machine, average values of certain interval of fluid flow rates (viz. 0.3mL/min to 0.07mL/min, 0.09mL/min to 0.13mL/min etc.) were calculated (refer to Table 4.5) and plotted that increased monotonically the increased intervals of fluid flow rates. During the design process, the diameter of each inlet was designed at the same value of 1.1 mm. As a result, using

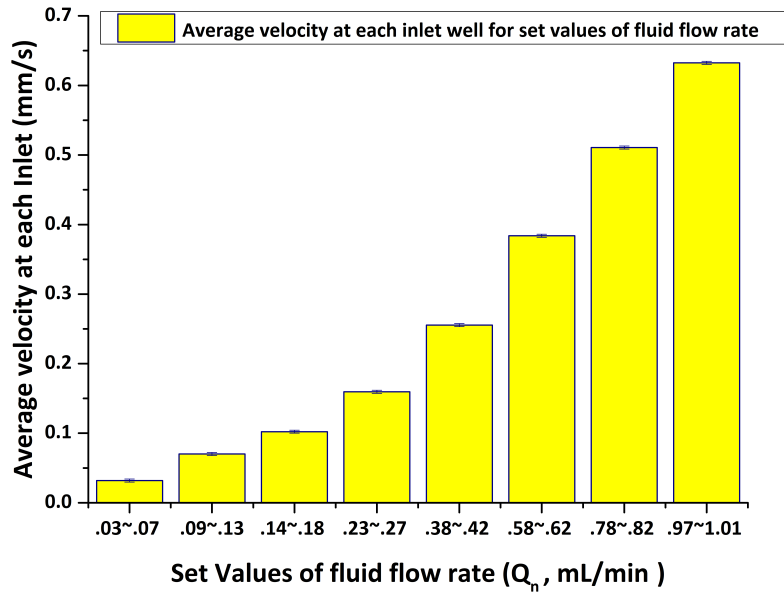


FIGURE 4.8: Statistical data analysis of average velocity versus fluid flow rate through the two inlet wells of same designed diameter in dimension

equation 4.6, same area was calculated in each case to get a gross idea of velocity profiles in each of two inlet wells of the microfluidic chip. Compared to the Fig. 4.7, average velocity is higher at the two inlet wells of Fig. 4.8 because of the diameter of outlet is larger (2 mm) than each of two inlet wells that tend to make the lower average velocity experienced by the only outlet well according to equation 4.6.

## 4.5 Summary

The internal microfluidic parameters of the designed microfluidic chip were statistically plotted, analyzed and evaluated. At the beginning, the microfluidic resistances were calculated for the internal different patterned micro flow channels. The outcome of this experimental results is used for the emphasis with the simulation results.

## Chapter 5

# Simulation to Emphasize the Experimental Result

### 5.1 Introduction

Use of ANSYS Fluent software for computational fluid dynamics (CFD) simulation [SLL14] is specifically useful in case of the analysis, study of the behaviors of microfluidic flow that ranges from fundamental research to design of devices [B81], [HES13],[Kim+08]. This computational method shows its merits, including the design of robust devices and its capability to simulate the complex and coupled physics swiftly at very low cost [Gre+09]. A lot of fluid mechanics researchers used CFD simulator for studying the mechanism of hydrodynamics behavior of fluids and cell mechanical environments [CWL06], [YA12], [Liu+13], [Sun+13], [Xu+13]. The numerical simulation of internal flow of biological and chemical solutions through the microfluidic chip is necessary to compare with the experimental data as the performance of the microfluidic chip can be evaluated in terms of the behavior of the microfluidic parameters with the simulated and experimental data of the real world.

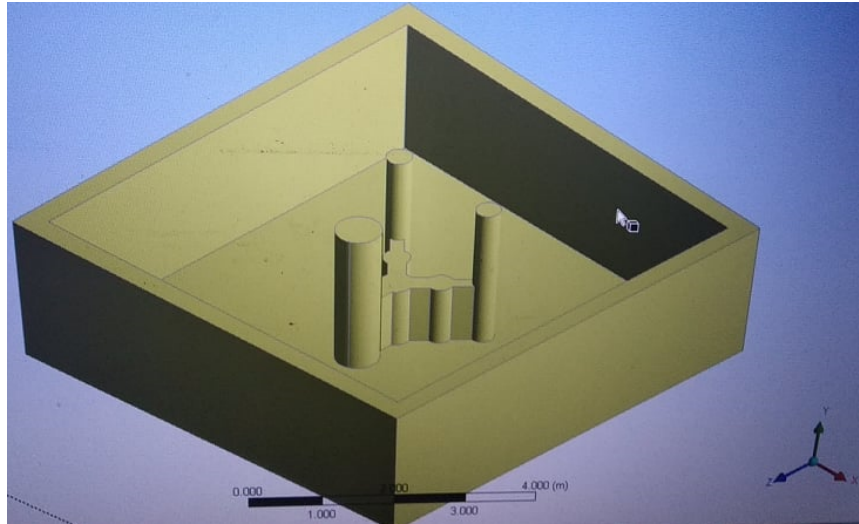


FIGURE 5.1: Exact dimensional 3D structure formed in ANSYS simulator to run the program with different microfluidic parameters of fluid dynamics

## 5.2 Simulation Algorithm

### Physical Model and CAD Modelling

During the experiment, flow of two different pH buffer solutions have been realized through the micro flow channels through the microfluidic chip. The geometric structure narrated already in the chapter 4 with the reference of Fig. 3.1.

The flow of internal chemical buffer solutions through the channels were numerically solved with ANSYS academic version of Fluent workbench [Flu+11] when the experiment was performed with the designed microfluidic chip. The solid works model was implemented in the design modeler with suitable geometric structure (Fig.5.1) exactly same as the dimensions of the fabrication of the micro flow channels of the microfluidic chip. For selecting better convergence criteria and reducing the computational cost, the model was precisely reduced to the axial symmetry of the domain into a planar two-dimensional section. The viscosities and densities of different pH buffer solutions were considered with other properties of water listed in Table 5.1 using numerical model. Using the

TABLE 5.1: Measured Different Fluidic Parameters Given As Input Data And Calculated Input, Output Data.

pH Buffer Solutions	Viscosity, $mPasm^{-3}$	Density, ( $g/mL$ )	Reynold Number	Microfluidic Resistances, $MPasm^{-3}$
pH=4, Inlet 1	0.5	0.982	3.71	$1.9 \times 10^3$
pH=10, Inlet 2	0.59	0.992	3.18	$2.4 \times 10^3$
Mixed pH 4 and 10, Outlet	0.58	0.99	2.51	10.3

experimental measured data, all microfluidic parameters were calculated.

### Governing equations

During the experiment, the mixing and others properties of fluid dynamics were examined in the outlet section. In order to track between the interface of mixing pH solutions, multi-phase volume of fluid (VOF) [BGP08; HN81; Eri05b; Kat19] with two Eulerian phases had been considered. For the dynamic behavior of these different pH fluids' coalescence in stagnant environment, the continuity, energy and momentum governing equations [KCD16a] of fluid dynamics were activated in this numerical approach too. They were approached by following the Euler form.

equation of continuity:

$$\nabla \cdot v = 0 \quad (5.1)$$

$$\frac{\partial \alpha_1}{\partial t} + v \cdot \nabla \alpha_1 = 0 \quad (5.2)$$

equation of momentum:

$$\frac{\partial(\rho v)}{\partial t} + \nabla \cdot (\rho v v) = -\nabla p_{st} + \nabla \cdot [\mu(\nabla v + \nabla v^T)] + F_{css} \quad (5.3)$$

equation of energy:

$$\frac{\partial(\rho h)}{\partial t} + \rho v \cdot \nabla h = \nabla(\lambda \nabla T) \quad (5.4)$$

where  $v$  is the fluid velocity ( $m/s$ ),  $\rho$  is the fluid density ( $kg/m^3$ ),  $p_{st}$  is the pressure ( $Pa$ ),  $\mu$  is the dynamic viscosity ( $\rho Pa.s$ ),  $\alpha_1$  is the thermal diffusivity,  $T$  is the temperature,  $\lambda$  thermal conductivity,  $h$  is the internal energy and  $F_{css}$  is the surface tension.

For the 2 phases, in each of control volume, density, viscosity, thermal conductivity are given by the following equations.

$$\rho = \alpha_1 \rho_w + (1 - \alpha_1) \rho_0 \quad (5.5)$$

$$\mu = \alpha_1 \mu_w + (1 - \alpha_1) \mu_0 \quad (5.6)$$

$$\lambda = \alpha_1 \lambda_w + (1 - \alpha_1) \lambda_0 \quad (5.7)$$

In general form, for a system of n-phase, all other properties in the volume-fraction-averaged will take on the following form:

$$x = \sum \alpha_q x_q \quad (5.8)$$

The VOF model will treat the energy,  $E_e$ , and the temperature,  $T$ , as mass-averaged variables:

$$E_e = \frac{\sum_{q=1}^n \alpha_q \rho_q E_q}{\sum_{q=1}^n \alpha_q \rho_q} \quad (5.9)$$

where  $E_q$  for each phase is on the basis of the specific heat of that phase and the shared temperature. In this model, continuum surface force (CSF) model

TABLE 5.2: Validation Of Experimental And Simulation Based Fluidic Parameters.

$Q_I, mL/min$	$Q_O, mL/min$	pH4, $\Delta P_{I_1}$	pH10, $\Delta P_{I_2}, Pa$	$\Delta P_{O_{Exp.}}^*$	$\Delta P_{O_{Sim.}} Pa$	$E., \Delta P_{O_{Sim.}} (\%)$
0.07	0.093	0.0812	0.0958	0.0012*	0.001	16.67
0.1	0.17	0.1786	0.21	0.0796	0.054	32.16
0.15	0.27	0.2598	0.3064	0.026*	0.017	34.62
0.25	0.41	0.4058	0.4796	0.045*	0.031	31.11
0.4	0.59	0.6494	0.766	0.1012	0.086	15.02
0.6	0.9	0.9742	1.1492	0.087*	0.091	-4.6
0.8	1.16	1.2988	1.5322	0.1106*	0.094	15.01
1	1.45	1.6074	1.8962	0.1606*	0.114	29.02

Values of asterik sign (\*) indicate the application of regression method to select the nearest values compared with the simulation results to minimize the percentages of error, E.

[DKC15] was used to calculate surface tension force with non-conservative formulation.

For time-dependent VOF calculation, all governing equations narrated above, were solved using an explicit time-matching scheme. ANSYS Fluent automatically refined the time step for the integration of the volume of fractions' equations. However, it was influenced this time step calculation by modifying the "Courant number". Anyone can choose to update the volume fraction once for each time step, or once for each iteration within each time step. For the simulation purpose, very small time step was considered in our process and  $1e^{-3}$  converging criteria was taken as the residual for all the equations. All the notations of above equations denote usual meaning used in fluid dynamics.

### Flow Chart

When the capillary tube based microfluidic model was designed completely in ANSYS Fluent environment for the two pH buffer solutions mixture, number of time steps was selected at first and then iteration limit was given. When the solution got converged for one time step, the solution got updated. After the above process to be converged, the next time step began. In this way, every

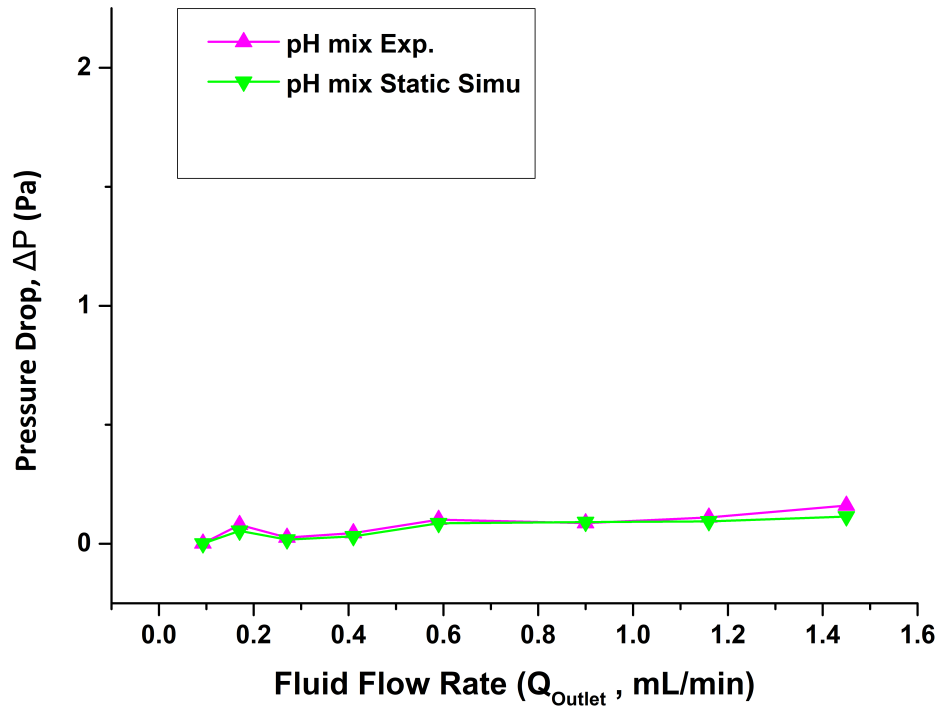


FIGURE 5.2: Validation of numerical model with experimental data for pressure drops at inlets and outlet

time when the consecutive solutions got converged, the next time step began and so on. The flow chart of general procedure of simulation ( Fig.5.3) has been stated below:

The important values and conditions used from simulator are described in the following tables (Tables 5.3, 5.4 and 5.5):

### Numerical settings

Using powerful commercial computational fluid dynamics (CFD) package of ANSYS Fluent [BGP08; APA19; KCD16b], the numerical simulation was carried out with the 0.001 residual convergence conditions for pressure, density, momentum and  $10^{-8}$  for energy. In general, pressure based solver with absolute velocity for transient time dependent model was used . The fluid was considered to flow in the laminar region. The boundary walls were imposed with no slip



TABLE 5.3: Setup Model and Solver conditions

Solver Type	Pressure Based
Solver velocity formation	Absolute
Time	Transient
2D space	Planar
Models	VOF multiphase model, Energy, Laminar viscous
Materials	Water(liquid) with experimental pH value of substances
Boundary Conditions	Velocity inlet, Pressure outlet, No-slip stationary wall function
Reference Zone	Surface body

TABLE 5.4: Solution conditions

Pressure-velocity Scheme	Coupling	Simple Type
Spatial Discretization Gradient		Least squares cell based
Spatial Discretization pressure		PRESTO!
Spatial Discretization momentum		Second order upwind
Spatial Discretization volume fraction		Compressive
Spatial Discretization Energy		Second order upwind
Transient Formulation		First order Implicit
Under-Relaxation factors		Pressure-0.3, Density-1, Body force-1, Momentum-0.7, Vaporization Mass-1, Volume fraction-0.5, Energy-1

TABLE 5.5: Calculation conditions

Initialization type	Hybrid initialization
Time advancement type	Fixed
Time advancement method	User-specified
Number of time steps	20,30,50,100 (as CPU capacity)
Time steps Size(s)	0.001
Max iteration/Time steps	20
Reporting Interval	1
Profile update interval	1

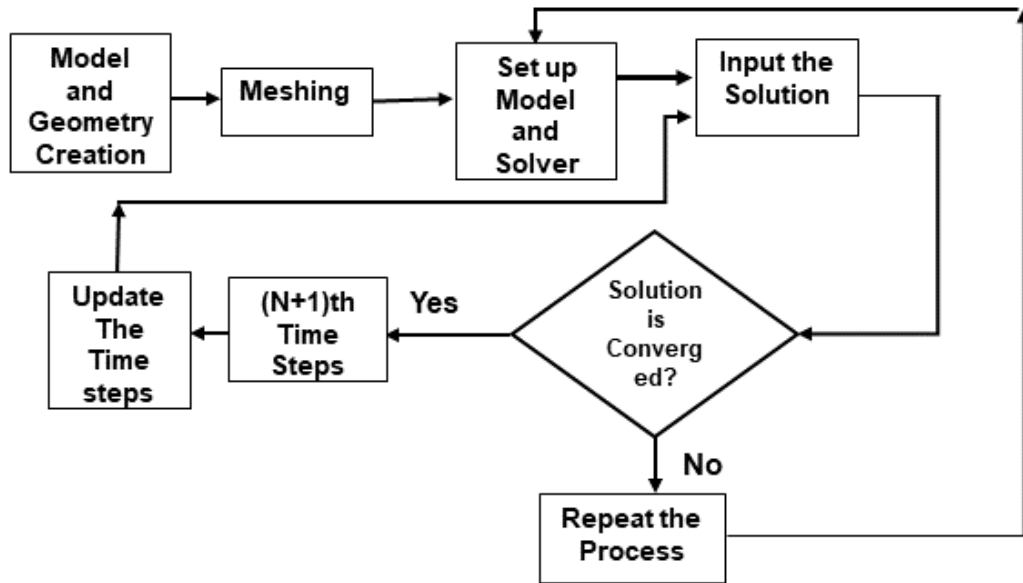


FIGURE 5.3: Flowchart of 2D ANSYS Fluent simulation using fluid dynamics

condition and constant 300 K temperature. The input applied boundary conditions were velocity in inlets for both pH 4 and pH 10 solutions with same given velocity applied at each of two inlets and different densities, viscosities etc. from Table 5.1. The properties of both pH fluids were assigned to a new material section in fluent database. The outlet of mixing pH buffer solution was considered in pressure outlet. Simple scheme for pressure velocity coupling algorithm was used in solution methods. For spatial discretization least square cell-based gradient with PRESTO, pressure and second order upwind momentum for first order implicit were used also in solution methods. The under-relaxation factors in solution controls were taken as a half from default to improved convergence rate. Hybrid initialization was used with 0.001s time steps. Fixed type time advancement with 20 number of time steps for maximum 100 iterations were assigned to calculate. However, the solution was converged within 450 to 800 iterations when the residual reached the limiting point.

TABLE 5.6: Mesh Dependency For Grid-Independent Study Of Validation Using Linear Regression Method

Element Size, $mm$	Node	Elements	$\Delta P_{mixexp.}(Pa)$ *	$\Delta P_{mixsimu.}(Pa)$
$4.5 \times 10^{-2}$	12323	12020	0.087	0.01295
$5 \times 10^{-2}$	10458	10185	0.087	0.086
$5.5 \times 10^{-2}$	9145	8898	0.087	0.0404
$6 \times 10^{-2}$	8030	7805	0.087	0.0117
$7 \times 10^{-2}$	6474	6278	0.087	0.0136
$8 \times 10^{-2}$	5496	5323	0.087	0.0145

Values of asterik sign (\*) indicate the application of the regression method

## 5.3 Mesh dependency and validation of simulation in CFD environment

### 5.3.1 Mesh dependency and validation of simulation

The simulation is validated with experimental result of pressure drops in our designed micro fluidic channels of two inlets and one outlet with the deviation range from  $\pm 34\%$  to  $\pm 4.6\%$ . Using  $4.5 \times 10^{-5}$ ,  $5 \times 10^{-5}$ ,  $5.5 \times 10^{-5}$ ,  $6 \times 10^{-5}$ ,  $7 \times 10^{-5}$ ,  $8 \times 10^{-5}$  m element sizes with 10 inflation layers in boundary, the model is the mesh properly orthogonal rectangular structure. All the simulation then carried out for  $5 \times 10^{-5}$  elements size with 10185 elements as it gave the lowest deviation from the trials. The data estimation using regression method is described in Table 5.2 and Table 5.6. Fig. 5.2 and Fig. 5.4 were drawn graphically for the model validation and mesh dependency [SLL14] respectively.

When two pH buffer fluids via MFC were flushed out through the outlet, outlet itself was opened at the same atmospheric pressure of the outside temperature. With the velocity of two mixed pH fluids via inlets, it was found net mixed product of these fluids through the open space of outlet. Therefore, in reality, final fluidic parameters at the outlet were considered from the mixture of two separate pH fluids and three dimensional diameter based shape of the outlet

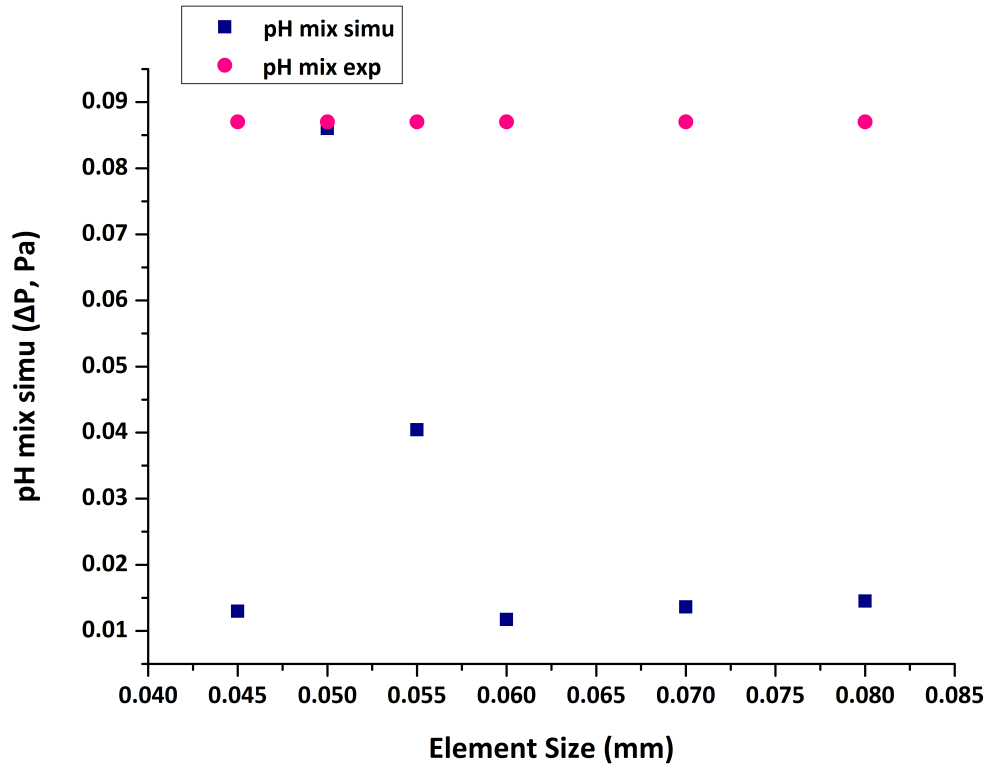


FIGURE 5.4: Mesh dependency for grid-independent study of validation

which was higher than that of each of two inlets. However, in two dimensional simulations, we considered only the fluids flow through the planar surface without taking into consideration the effect of fluids via three dimensional micro flow channels. That is why, the deviations are seemed to be varied from -4.6 % to 34.62 % in the simulation process. The percentage of error of change of pressure drop in between experimental and simulation result is calculated as:

$$(\%)Error = \frac{\Delta P_{outExp.} - \Delta P_{outSimu.}}{\Delta P_{outExp.}} \quad (5.10)$$

On the other hand, for example, from the first measurement of experimental data of Table 4.3, for each of two inlets, total of  $(0.1+0.1) = 0.2$  mL/min FFR, we got corresponding net value of 0.17 mL/min FFR at the outlet. This is because of the loss of a significant amount of fluids inside the surface walls of

---

micro flow channels due to the consideration of a certain measurement time by stopping the flow of fluids. Therefore, to overcome such difficulty in practical case, it was necessary to do linear regression by taking averages of pressure drops at the outlet. Considering the lowest deviation with the experiment, mesh dependency test was applied for the highest 10185 elements carried out by the velocity ranges from 0.26 mm/s to 0.38 mm/s as it gave lowest deviation during the testing process. At these ranges, the deviation from the experimental results showed minimum error to consider for the validation of simulation results.

## 5.4 Summary

This chapter describes the process of simulation of different viscous pH buffer solutions using CFD based intensive environment to analyze the behavior of hydrodynamic parameters of microfluidic chip from fluid dynamics. Moreover, the simulation was done to compare with the experimental result so that the performance of the microfluidic chip can be evaluated by flowing two pH buffer fluids.



## Chapter 6

# Digital Filter Design and Testing

### 6.1 Introduction and Motivation

This chapter introduces with the requirement of doing fast scanning of higher contrast of THz images. Here, for this purpose, it was necessary to use filter in THz imaging process. The reasons of the selection of adaptive digital filter (ADF) [GB15], [Bar+07], [JCM12] over any other filter, are of its computational simplicity, robustness and popular use in biomedical, electrical and communication engineering specially.

### 6.2 Working Principle of an Adaptive Digital Filter

An ADF adjusts the filter coefficients, followed by the method of an adaptive least mean square (LMS) algorithm. While as a computational device, ADF attempts to model the relationship between two signals in real time in an iterative manner. A sample from a digital input signal  $x(n)$  is fed into a device, called an adaptive filter, that computes a corresponding output signal sample  $y(n)$  at time  $n$ . The output signal is compared to a second signal  $d(n)$ , called the desired response signal, by subtracting the two samples at time  $n$ . This difference

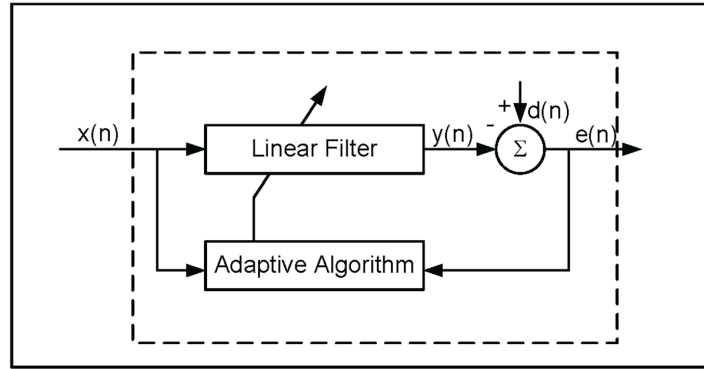


FIGURE 6.1: Functionality of an ADF during the computational operation

signal, given by  $e(n) = d(n) - y(n)$  is known as the error signal. The error signal is fed into a procedure which alters or adapts the parameters of the filter from time  $n$  to time  $(n + 1)$  in a well-defined manner. This proposed LMS algorithm adjusts the coefficients of the linear filter in iterative process to minimize the power of  $e(n)$ . The LMS does the following arithmetical operations to update the coefficients of an adaptive finite impulse response (FIR) filter [Hay08]:

Step 1:

$$y(n) = \vec{u}^T(n) \cdot \vec{w}_1(n) \quad (6.1)$$

where,  $\vec{u}(n)$  is the filter input vector and

$$\vec{u}(n) = x(n)x(n-1)\dots\dots\dots x(n-N+1) \quad (6.2)$$

where,  $\vec{w}_1(n)$  is the filter coefficients vector.

Step 2: To calculate the error signal  $e(n)$  from the equation;

$$e(n) = d(n) - y(n) \quad (6.3)$$

Step 3: To update the filter coefficients from the equation;

$$\vec{w}_1(n+1) = (1 - \mu)\vec{w}_1(n) + \mu \cdot e(n) \cdot \vec{u}(n) \quad (6.4)$$



where  $\mu$  denotes the step size (SS) as a gain constant parameter of the ADF. The whole process of adaptation is represented by the oblique arrow that pierces the adaptive filter block in the Fig. 6.1.

### 6.3 Experimental Set up for ADF's Proper Functioning Check

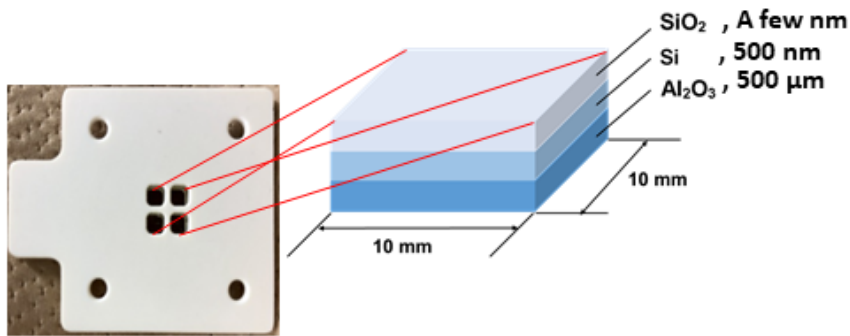


FIGURE 6.2: Sensing plate arrangement for the testing of effectiveness of the ADF

For the integration of the ADF with the TCM system, LMS algorithm using digital signal processing technique has been programmed and implemented in laboratory virtual instrument engineering workbench (LabVIEW®) by analyzing the improvement for capturing higher contrast of THz images. Four wells sensing plate in Fig.6.2 was used to test the proper functioning of the ADF by pouring Milli-Q water solutions onto the sensing plate of each of four wells. The conceptual diagram of such four wells plate with the application of THz electromagnetic wave using the TCM system is shown in Fig.6.3.

The same working principle of section 2.3 and set up of TCM system were used here for this testing except the digital pressure machine, microfluidic chip and samples of chapter 2. From femtosecond laser pulses, two types of laser beams were generated; the pumping and trigger beams. First of all, the alignment of pumping and trigger beams of THz laser was finished for each of the measurement of TCM system. Then, the position of laser was moved by starting the

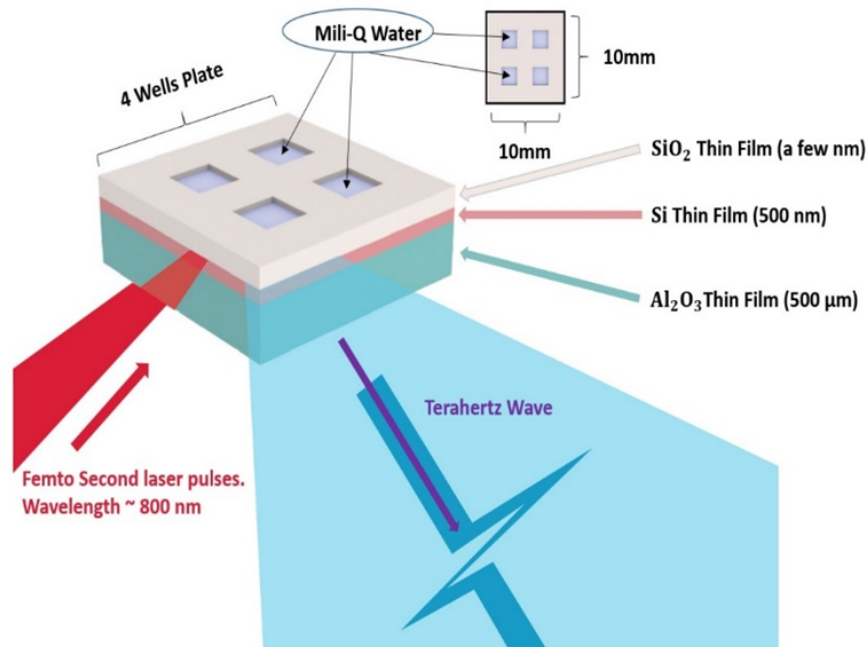


FIGURE 6.3: Conceptual diagram of four wells plate with the application of THz wave

scan of the mili-Q water samples on the sensing plate. The optical signal was detected by the photo conductive antenna, used as a detector. For the identification of the position of peak amplitude of THz data, the delay stage of chopper was used. For choosing the correct sampling frequency, number of samples of signal data to be sampled according to Nyquist sampling theorem, the analog to digital converter (ADC) was programmed in LabVIEW® (section 2.3). Then, using ADF, sampled data were filtered by averaging the raw THz data from the simulation of LabVIEW®. THz imaging data were captured and plotted using personal computer (PC). At the end, the average filtered data were plotted in the form of contour of THz images by measuring points by points of imaging step.

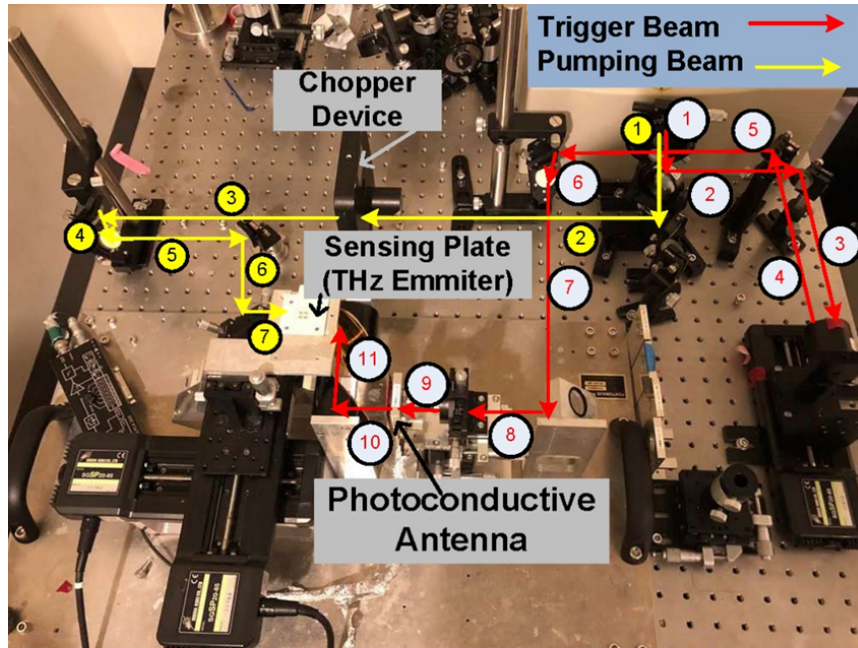


FIGURE 6.4: Optical set up of TCM system to start measurement for this project

## 6.4 Statistical Analysis of Characteristic Curve of ADF

The quantitative analysis of the characteristic curve of ADF was plotted from the variation of standard deviation (SD) with respect to step size (SS) with the optimization of step size as a gain constant-one of vital parameters of the filter [Ahm+19]. From the algorithm of ADF, it can be said that convergence speed is used for adjusting the minimum error with the main input signal. The aim of this project is to fix the optimum condition level of convergence speed of LMS algorithm with respect to step size from which the total measurement time of the THz imaging was reduced without losing the original scanning speed of TCM by internally applied ADF. Fig.6.5 shows standard deviation versus step size curve where the time constants (TC) of lock-in-amplifier were set as 30 ms and 100 ms.

From the computational analysis of LMS algorithm, 0.001, 0.01, 0.05, 0.5, 1,1.5,

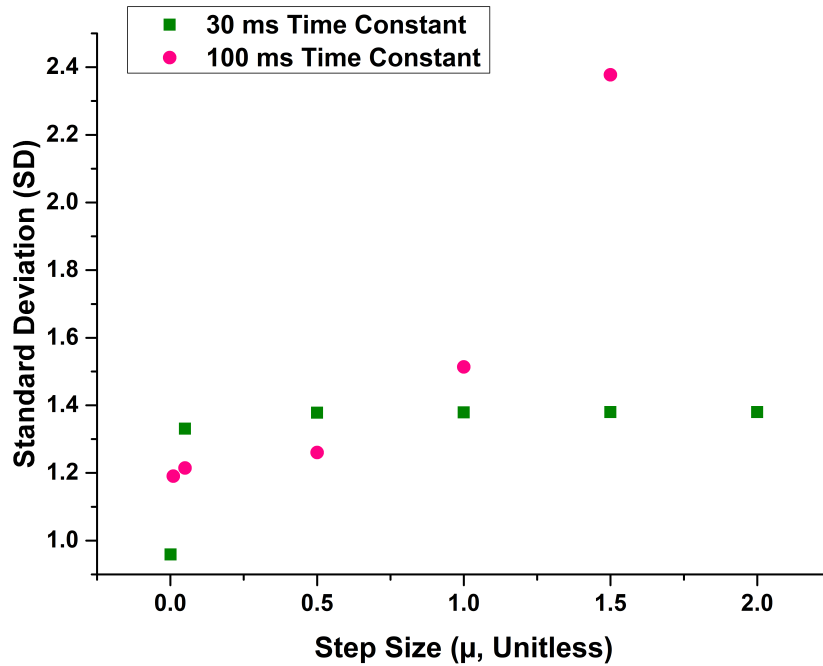


FIGURE 6.5: Standard deviation versus step size curve for evaluating the characteristics of 100 ms and 30 ms TC of lock-in-amplifier with the application of ADF

and 2 step size values were given as input data to calculate corresponding standard deviation values so that optimum value of standard deviation would be fixed for the high precision based performance of the ADF with the higher quality of contrast of THz images along with the unchanged original scanning speed of the TCM system.

From the quantitative analysis, step size's value of 0.001 was chosen as an optimum value for getting higher contrast of THz images by analyzing parameters' variation of the filter. Therefore, the performance of the ADF was evaluated in terms of the standard deviations of images as a function of step sizes of the filter and plotted in Fig.6.5.

In Fig.6.5, for two different family of curves, the standard deviations of data of the signal were not saturated for 100 ms rather than increased monotonically with larger values of step sizes. On the contrary, the curve for 30 ms exhibited

with the criteria of early saturation even at standard deviation value of 0.01. Therefore, during the analysis of two qualitative results, for the step size of 0.001 of ADF, higher contrast was achieved with the reduction of total measurement time of TCM by taking into the consideration of the optimum level issue of standard deviation. By considering this fact, the data captured with ADF were (a) contrast of 0.74 for 30ms time constant of lock-in-amplifier corresponding to the standard deviation value of 0.96 and, (b) contrast of 0.73 for 100ms time constant of lock-in-amplifier corresponding to the standard deviation value of 1.19. From the basic principle of lock-in-amplifier, time constant is inversely proportional to the bandwidth of low pass filter (LPF) integrated with the lock-in-amplifier. Therefore, with the lower value of the time constant, the bandwidth of LPF became wider with lower signal-to-noise-ratio (SNR) and faster measurement. Higher value of time constant subjects to narrower bandwidth of low pass filter with higher signal-to-noise ratio and slower measurement fact. The designed adaptive digital filter worked on 30 ms time constant brought early saturation, compared with 100 ms time constant. Because in the case of 30 ms, ADF got much wider bandwidth to collect more signal information by adapting with the input THz signal to reduce the degree of standard deviation and minimize error according to its working principle followed by LMS algorithm [Ahm+19].

## 6.5 Determination of Contrast of THz Images

For the performance evaluation in terms of the quality of images using ADF, The following process was used to estimate the standard deviation and contrast of the THz images:

Firstly, the values of pixels inside and outside of the well were taken, that was marked by the red-yellow dashed squares region in the graphic of Fig.6.6.

Secondly, calculation for the determination of standard deviations of the pixels were performed in the both inside and outside of wells

Thirdly, to evaluate the contrast of the images, the values of pixels inside and outside of the wells  $I_{max}$  and  $I_{min}$  were taken respectively and estimated by:

$$contrast = \frac{I_{max.} - I_{min.}}{I_{max.} + I_{min.}} \quad (6.5)$$

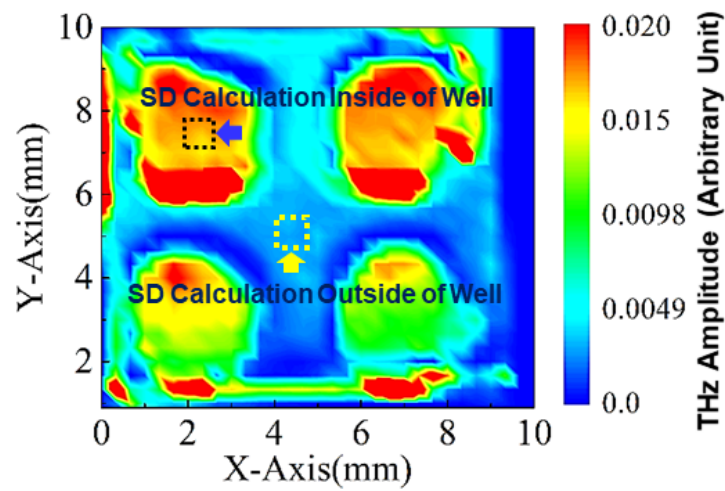


FIGURE 6.6: Calculation and estimation of standard deviation for several pixels of data of THz images inside the upper of the well and outside region of the Well onto sensing plate

## 6.6 Qualitative Analysis of Contrast with and without Filter

The quality of contrast without and with filter are estimated in this section. In Table I, the values of contrast with filter are more than double than that of without filter. Therefore, to capture higher contrast of THz images with higher scanning speed, the primarily chosen contrast values are 0.74 and 0.73 for 30 ms and 100 ms time constant respectively. However, the aim was to keep standard deviation values as lower as possible with respect to the step sizes of the ADF in parallel with the target of higher quality of contrast of THz images. Therefore,

TABLE 6.1: Estimation of Standard Deviation and Contrast of THz Images

With or Without Filter	100ms Time Constant			30ms Time Constant		
	SS	SD	Contrast	SS	SD	Contrast
Without Filter	N/A	0.01	0.32	N/A	0.07	0.34
With Filter	0.001	1.19	0.73	0.001	0.96	0.74

the recommendation here is the lowest value of standard deviation of 0.96 with respect to step size of 0.001 of the filter whose contrast is of 0.74 for 30ms time constant of the lock-in-amplifier. As a result, contrast of THz images was improved notably. Moreover, from the view of the time constant, the contrast did not compromise by decreasing the time constant from 100 ms to 30 ms that suggested the reduction of the total measurement time of THz data.

The statistical analysis and calculation without and with filter for the standard deviations and the contrast of images are shown in the Table 6.1.

## 6.7 Contour of THz Images with and without Filter

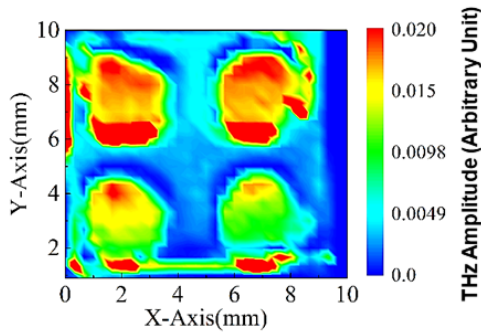


FIGURE 6.7: Contour of THz Image with filter for the contrast of 0.73, TC 100 ms

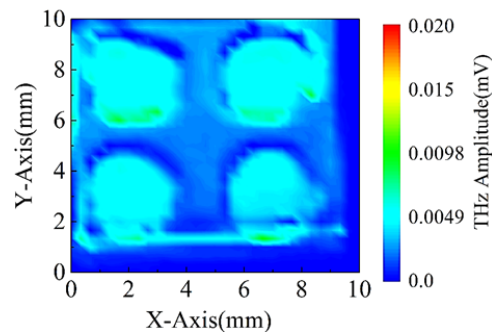


FIGURE 6.8: Contour of THz Image without filter for the contrast of 0.32, TC 100 ms

The outcome of THz images with and without filter for 30 ms time constant of lock-in-amplifier are plotted in Fig.6.7 and Fig.6.8 respectively. Moreover, the contour of THz images for 100 ms time constant of lock-in-amplifier are

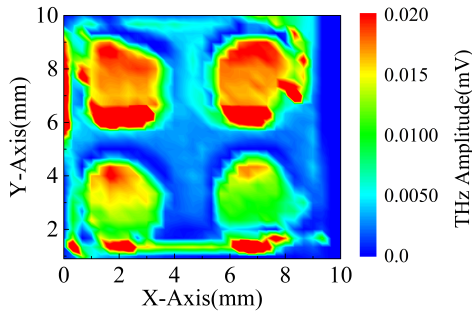


FIGURE 6.9: Contour of THz Image with filter for the contrast of 0.74, TC 30 ms

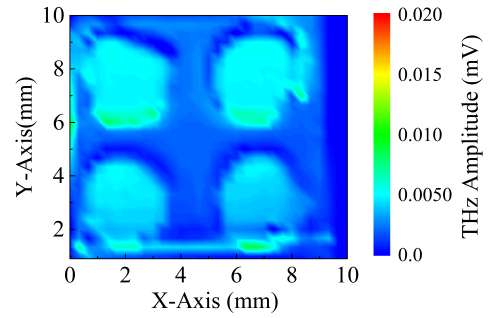


FIGURE 6.10: Contour of THz Image without filter for the contrast of 0.34, TC 30 ms

plotted in Fig.6.9 and Fig.6.10 respectively. During the experiment, two upper wells of the sensing plate of above mentioned all figures, were poured with 5  $\mu\text{L}$  mili-Q water solutions, while lower two wells were poured with 1  $\mu\text{L}$  mili-Q water solutions.

### 6.7.1 Data Acquisition of THz Images

THz data obtained from the experiment, were averaged without and with filter using LabVIEW® simulation. By setting lock-in-amplifier time constants of 30 ms and 100 ms, with and without filter, two-stage controlling LabVIEW® simulation was used to adjust ADC's number of samples data = 100, sampling frequency,  $f_s = 1\text{kHz}$ , followed by the Nyquist sampling theorem for the time interval of 1 ms in case of each measurement. The total acquisition time for measurement in both cases = (time constant of Lock-in-amplifier + No. of samples  $\times$  time interval of ADC).

In case of (10 mm x 10 mm) dimensions of sensing plate, 6300points = 1mm was programmed using LabVIEW® with the resolution of 333 $\mu\text{m}$  for each of pixel data in order to scan total images by THz laser. Thus, during the experiment with (10 mm x 10 mm) dimensional sensing plate, THz laser scanned 63000 by 63000 points for each of measurement steps to capture each of THz images



of Fig.6.7 and Fig.6.8. Therefore, ADF was successfully tested the sample of Mili-Q water inside the four wells by recording higher contrast of THz images.

## 6.8 Advantages of an Adaptive Digital Filter

Using least mean square (LMS) algorithm, an ADF was designed and simulated by LabVIEW®. The causes of choosing ADF over any other filter, are for the reasons below: (i) One of ADF's parameters, convergence speed (CS) manages optimal solution with iteration method (ii) To possess miss-adjustment ability for measuring the amount by which the final value of Mean Square Error (MSE), averaged over a gross calculation of filtering, deviated from the mean square error (iii) To have tracking capability by following statistical variations in the environment of a random, non-stationary, time varying signal, (iv) ADF can be able to adjust even small estimation errors from the small disturbances of any source that imply its robustness (v) Compared with other methods of signal processing filters, it has computational simplicity.

## 6.9 Summary

This chapter experimentally observed that the filter would improve the contrast double without compromising existing scanning speed of the TCM system. In addition, total measurement time was reduced from the optimum value of standard deviation corresponds to the step size of ADF.



## Chapter 7

# Results and Discussions

### 7.1 Introduction

Transport of fluids [HSR13] phenomena on the micro scale level have achieved particular importance owing to the progressive demand for more efficient and sustainable processes. Especially, to co-relate the nano- and micro technologies that require a clear understanding of multi-scale coherence. The recent advancement in micro-fluidic and nano-fluidic technologies have been moving forward in parallel ways by the advancement in methods for the direct optical measurement of transport of fluids phenomena on these scales. A lot of methods for micro-scale flow visualization have been employed and examined since the late 1990s for fulfilling the better spatial resolution criteria. However, the sharp spatial resolution criteria is still a hot cake to capture the best images of the internal fluidic reactions of chemical and biological solutions.

Even if the flow conditions of chemical and biological solutions on the microscale level are almost laminar owing to the smaller dimensions and therefore, high accuracy are applicable for the numerical simulations, there are some remarkable causes for the basic necessity of the visualization [HSR13] of the experimental flow of the fluids. In most of the cases, for example, data on chemical media such as diffusion coefficients, reaction rates and the accurate data of geometries

and wall conditions of micro flow channels are not be able to find out. Furthermore, through the micro flow channels, chemical reactions are not calculable to date for the interaction between mass transport and conversion, especially in case of catalytic processes have taken place simultaneously. As a result, the visualization of micro-scale flow is an effective method and tool to understand and optimize the micro flow channels in micro scales. For the visualization of internal microfluidic reactions in micro-scale level convincingly, most measurement techniques make use of a microscope [HSR13] to get suitable spatial resolution. Therefore, optical access is necessary in order to allow a view into the internal micro flow channels without interrupting the conditions of the flow of fluids. Here, THz image sensing technology using THz chemical microscope with  $0.8\mu\text{m}$  wavelength meets up the demand of sharp spatial resolution criteria.

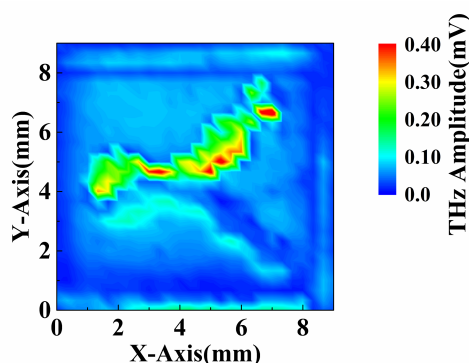


FIGURE 7.1: THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.07 mL/min fluid flow rate

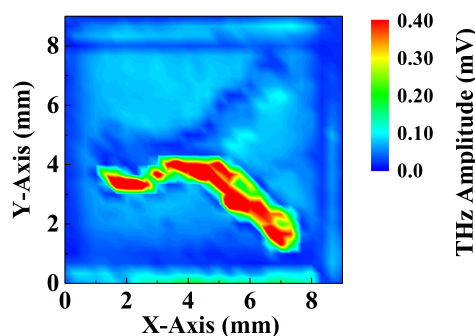


FIGURE 7.2: THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.15 mL/min fluid flow rate

Moreover, in this dissertation, ANSYS Fluent software for computational fluid dynamics (CFD) simulation was used to explain not only the experimental flow phenomena but also predict new flow phenomena, providing a better realization of the chemical and biological micro-environment and fundamentals of microfluidic flows in case of manipulation and analysis. The objective of this ANSYS based CFD simulation is to simulate small-scale fluid flow with the visualization of the interaction of different fluids inside the internal micro-environment

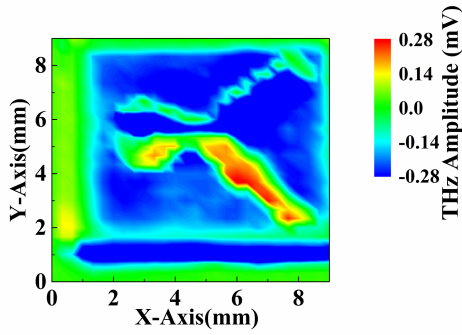


FIGURE 7.3: THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.25 mL/min fluid flow rate

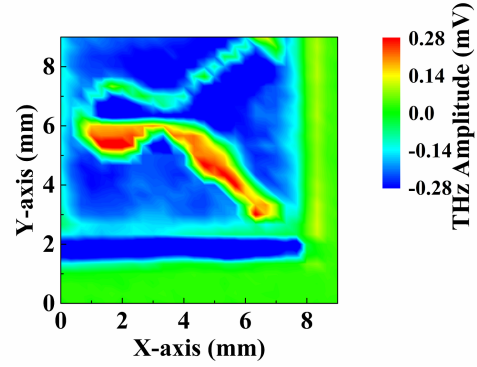


FIGURE 7.4: THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.4 mL/min fluid flow rate

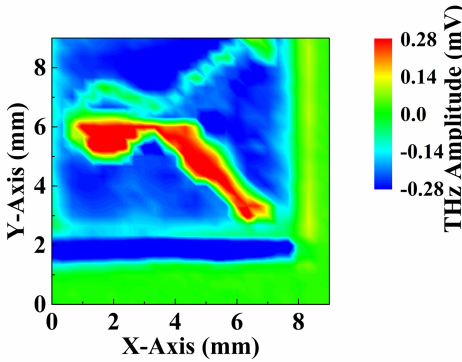


FIGURE 7.5: THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.7 mL/min fluid flow rate

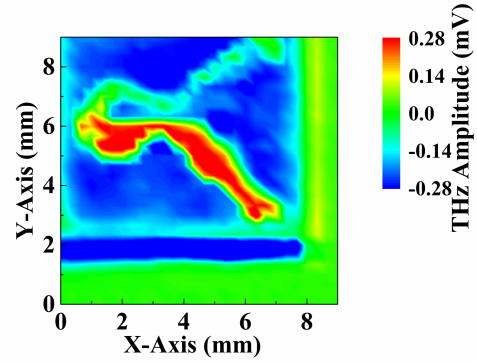


FIGURE 7.6: THz captured data for the interaction of pH 4 and 10 buffer solutions at 0.8 mL/min fluid flow rate

using volume of fluid (VOF) [Eri05a] method, followed by the THz laser scanned visualization method. In addition, from experimental measurement data of different microfluidic parameters, the experimental results were validated with the simulation generated data to investigate optimal chip design, flow fundamentals and possible applications of microfluidic lab-on-a-chip systems.

### 7.1.1 THz Imaging results

During the experiment, THz data were recorded to capture the interaction of two different viscous pH=4 and 10 buffer solutions through the micro flow channels of the designed microfluidic chip. The contour of THz Images driven by the THz laser scanned TCM system were plotted for different fluid flow rates

(viz. 0.1, 0.15, 0.25, 0.4, 0.6 mL/min etc.) that were applied to each of two inlet wells of micro flow channels as input data. For each experimental reading, using syringe-connected two tubes, same two set values of fluid flow rate from digital pressure machine were applied at each of two inlet wells of the microfluidic chip. The captured image maps of internal microfluidic reactions of two above mentioned pH solutions through the fabricated micro flow channels were shown in Fig. 7.1, Fig. 7.2, Fig. 7.3, Fig. 7.4, Fig. 7.5 and Fig. 7.6 in the form of THz images.

### 7.1.2 2D ANSYS Fluent simulation results

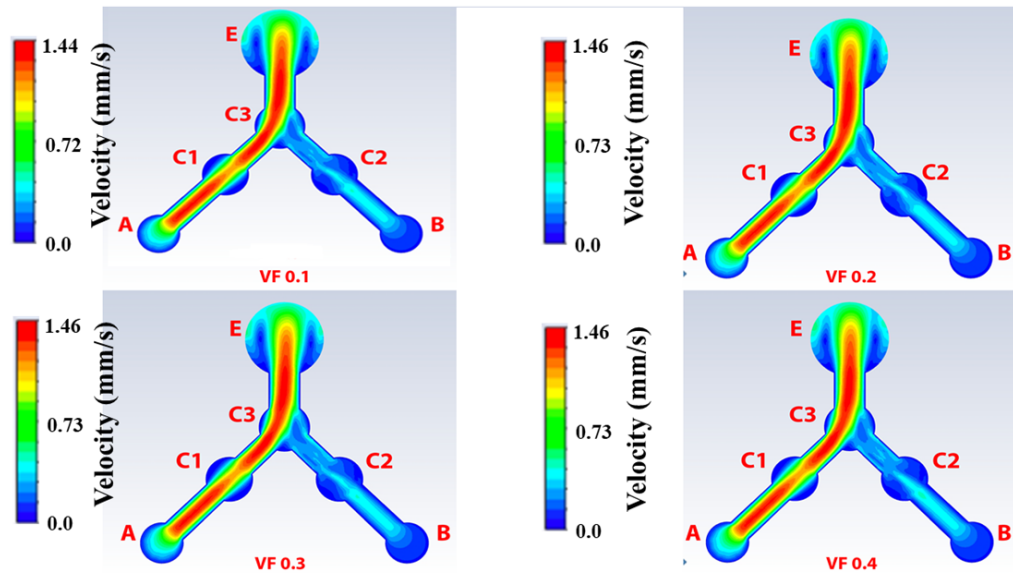


FIGURE 7.7: Same set of input velocity applied to each of two inlet wells with respect to different increased volume of fraction; same input velocity, 0.064 m/s applied at A and B points for two pH solutions.

Here, the change of volume of fractions (VF) using VOF [Eri05a] method is utilized to predict the nature of two immiscible phases of fluids flow by evaluating the influence of such multi-phase modeling approach on the flow of pH 4 and 10 viscous acidic and base fluids.

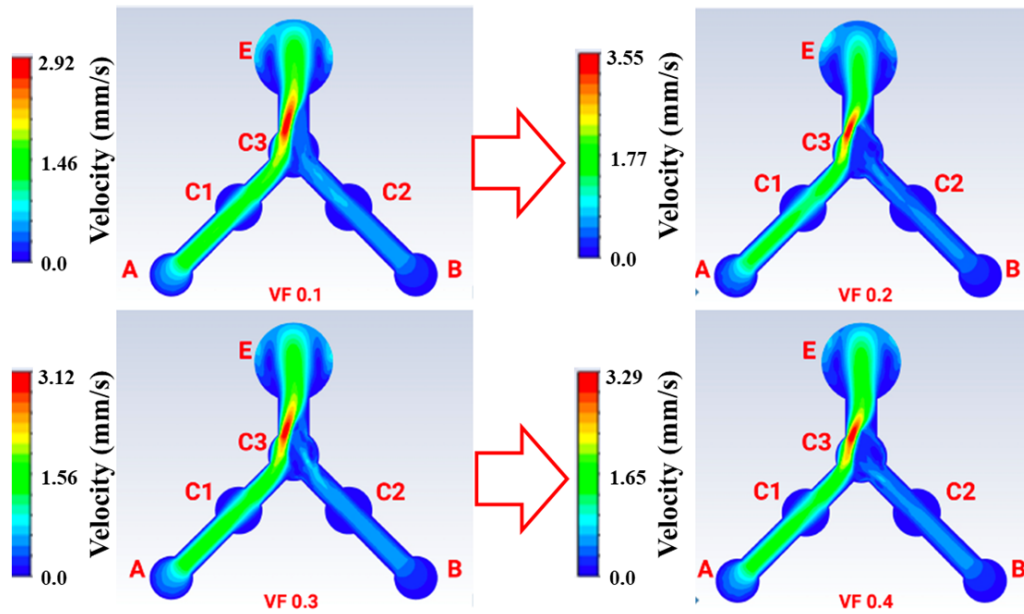


FIGURE 7.8: Same set of input velocity applied to each of two inlet wells with respect to different increased volume of fraction; same input velocity, 0.096 m/s applied at A and B points for two pH solutions.

Using ANSYS Fluent for CFD simulation, with the increased values of VF, different velocity profiles-based flow of solutions were simulated in Fig. 7.7 and Fig. 7.8 from inlet points via intermediate points  $C_1$  and  $C_2$  and common junction well point,  $C_3$  to exit point  $E$ . In Fig. 7.7 and Fig. 7.8, based on the concept of fluid dynamics, two fluid flow solutions didn't mix with each other by flowing through each of two layers separately. Therefore, pH=4 and pH=10, two different viscous buffer solutions exhibited the laminar nature of fluid flow. Two different viscous solutions travelling through the micro flow channels, created friction at the boundaries of micro flow channels that in turn created a bit speed tendency to deflect slightly from the spontaneous flow of fluid flow solutions. The higher the viscosity of a fluid, the higher the tendency of the deflection of the speed of fluid flow was observed, with the increased fluid flow rates as well. Therefore, pH=10 solution denotes the yellowish and red mixed color marked velocity profiles in Fig. 7.7 and Fig. 7.8 and its viscosity 0.58 mPa.s is higher than that of pH=4 solution. pH=4 solution mapped with the surface of its flow

path which is shown in the simulated velocity profiles. Moreover, the pH=10 velocity profiles possess higher dragging effect owing to the frictional force experienced inside the internal layers of micro flow channels than that of pH 4 solution.

In Fig. 7.8, with the increased values of VF, the velocity profile path showed so slight change for the same applied velocity (0.096 m/s corresponding to 0.15 mL/min) at each of A and B points. Similarly, in Fig. 7.7, same input velocity of 0.064 m/s corresponding to 0.1 mL/min fluid flow rate for two different viscous pH solutions was applied at each of “A” and “B” points, with the increased values of VF. However, in Fig. 7.7, the simulated data do not show any appreciable change inside the paths of velocity profiles even though the change of VF values controlled data. It signifies that velocities profiles are nearly same with the increased values of VF. As a result, the simulated data with optimum velocity 0.064 m/s corresponding to 0.1 mL/min of Fig. 7.7 would be countable for the almost ideal fluid flow patterns through the micro flow channels of the simulated microfluidic chip. The reason is that they show consistent behavior of fluids flow with the outcome of almost same velocity profiles of laminar nature in spite of the increased values of VF.

## 7.2 Analysis of Velocity Profiles obtained from ANSYS Fluent

According to the previous section's assigned notation of Fig. 7.7 and Fig. 7.8, pH=4 and 10 viscous acidic and base solutions were inserted into A and B points respectively in order to compare the velocity-based flow patterns inside the microfluidic structure from simulation and experimental measured velocity profiles. The objective of this discussion is to satisfy the simulation results based on continuity equation of fluid dynamics.



Fig. 7.7 and Fig. 7.8 indicate different profiles of velocities inside the micro flow channels at the starting same input velocity of 0.064 m/s and 0.096 m/s respectively. In the design of the microfluidic chip, the dimensions that were used, same all dimensions were used during the simulation process. In Fig. 7.7 and Fig. 7.8, the diameters of different points were  $A_{dia.} = B_{dia.} = 1.1$  mm,  $C_{1dia.} = C_{2dia.} = 1.2$  mm and  $E_{dia.} = 2$  mm. In Fig. 7.7, through A and B inlet points, in each case, initial velocity showed high value and through  $C_1$  and  $C_2$  points, velocity showed low profiles. The reason is that the diameter of each of  $C_1$  and  $C_2$  of that parts of the chip was larger than that of each of A and B. When fluid travelled through the  $C_1$  and  $C_2$  points, velocity of fluid increased again because of the smaller diameter of rectangular path provided channel. At  $C_{3dia.} = 1.2$  mm point, here the larger diameter was designed. Therefore, at this common junction point  $C_3$ , the velocity of pH=4 fluid's flow was dropped. Moreover, the other pH=10 fluid also travelled this common junction point that surely affected the velocity of pH=4 fluid due to the heat transfer, pressure etc. phenomena. After leaving from  $C_3$  point, it was designed here a rectangular path of 0.8 mm diameter that boosted to increase the velocity of mixed two pH fluids again. At the outlet point E, velocity again dropped due to the larger diameter of 2mm than any other dimensional path of full fluid flow channels.

### 7.3 Summary

In this chapter, THz generated laser scanned the internal microfluidic reactions to visualize the chemical reactions with sharp spatial resolution to get the best outcome of the this project. Then, ANSYS Fluent with CFD simulation was executed to analyze the flow patterns of fluid flow solutions on the basis of fluid dynamics.



## Chapter 8

# Conclusion and Future Works

This dissertation proposed a THz imaging technique to characterize and observe the flow of different pH buffer solutions through the designed microfluidic chip where the testing of pH level maintained liquid foods, blood samples analysis to observe any pH variation can be possible in our daily life. From technical point of view, the performance of the designed microfluidic chip was evaluated in terms of the analysis of hydrodynamic parameters of fluid dynamics that actually controls the performance of the microfluidic chip. Moreover, an universal digital filter is proposed in this dissertation to test its effectiveness to capture the imaging maps of biological and chemical solutions shapely. From this dissertation, it was improved the contrast of the THz images using such filter.

**Chapter 1** introduces the recent advancement of this project work, motivation, background, and contributions of the study. **Chapter 2** describes and defines the mechanism of THz generation and detection principles. **Chapter 3** introduces the fabrication, manufacturing process of PDMS designed microfluidic chip. In **Chapter 4**, the hydrodynamic parameters were calculated from fluid dynamics, experimented using the designed microfluidic chip to find out the characteristics of different internal microfluidic parameters. **Chapter 5** contains the ANSYS Fluent software based CFD simulation to evaluate the

experimental result with the simulation result so that the performance of the designed microfluidic chip can be improved in a appreciable manner. **Chapter 6** explains the proposed adaptive digital filter design for the improvement of the contrast of the THz images so that the TCM system can allow faster measurement of chemical and biological samples. **Chapter 7** is very much crucial to find the outcome of this dissertation so that one can realize the achievement of this dissertation fully. Here, the THz imaging maps of interaction of pH buffer solutions were captured and discussed from the technical view of fluid dynamics. Moreover, CFD based simulation results show the visualization of the two pH buffer solutions through the internal micro flow channels of the designed microfluidic chip.

As future works, more complex structure of the microfluidic chip will be designed to test more than two chemical solutions at a time through more than two internal micro flow channels of the designed microfluidic chip. Thus, from the interaction of more than two chemical and biological solutions, it can be possible to visualize the interaction of more than two solutions in the form of THz images. Apart from this, the designed adaptive filter's version will be improved by following different methodologies of digital signal processing techniques so that the faster scanning of biological and chemical solutions using THz laser can be possible. The ultimate target is to use such microfluidic structures for showing the contribution in real life applications of biomedical engineering. Moreover, fluid mechanics design engineers have been searching for higher spatial resolution criteria using so many optical arrangements to capture sharp microfluidic reactions based images. However, still they have achieved maximum of  $6.47 \mu\text{m}$  spatial resolution [Tor+20] by applying optical method for the measurement of heat and flow of fluid solutions. The THz technology used here already achieved  $333 \mu\text{m}$  spatial resolution criteria for each of pixel of THz image to capture the internal microfluidic reactions that is more than 50 times

higher than that of optical measurement by the fluid mechanics engineers. In future, we can achieve more sharp spatial resolution by modifying the optical set up of the TCM system for the faster scanning and movement of THz laser. Moreover, the filter's moderate version will be used for capturing the faster scanning of movie clips, faster chemical reactions made solutions in future.



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# Index

- AD converter, 5, 16
- AD filter, 5
- ANSYS Simulator, 7
- ANSYS simulator, 48
- Average THz data, 16
- average velocity, 40
- CCD camera, 4
- CFD simulator, 43
- Contrast, 5
- Cylindrical path, 8
- DPM, 15
- femtosecond laser, 12
- Imaging noise, 5
- LabVIEW, 16
- laminar, 7, 48
- lock-in-amplifier, 5
- LTME, 3
- microfluidic chip, 6
- microfluidics, 2
- NDT, 3
- PDMS, 6, 20
- Rectangular path, 8
- Regression method, 51
- Reynold number, 8
- SFG, 16
- Solid work, 20
- TDS, 3
- Terahertz, 1
- THz detection, 15
- THz detector, 16
- THz generation, 13
- Ti:Sapphire laser, 15
- VOF, 45

# Biography

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