Hybrid Model of Ultrasonic Transducer for better Medical Imaging

Apurva Y. Gandole[1], Y. B. Gandole[2]

[2]Adarsha Science J. B. Arts and Birla Commerce Mahavidyalaya Dhamangaon Rly-444709

Abstract: ***Recently, ultrasound becomes prominent in several applications especially in medical field to improve the health services either for diagnostic. The emergence of ultrasound applications raises the need of reliable transducer to comply with that purpose. As polymer material being popular in medical ultrasound, there are chances to combine it with former piezoelectric ceramic material in designing diagnostic transducer to get hybrid characteristics required for multi-frequency application. In this work, SPICE model of ceramic-polymer piezoelectric has been described. With signal conditioner circuit, complete analog system for ultrasound has also been developed.***

Keywords: Ultrasonic, Transducer, Medical imaging, Modeling.

1. Introduction

Recently, ultrasound becomes prominent in several applications especially in medical field to improve the health services either for diagnostic or therapy purpose. For many years, ultrasound has provided clinicians with an affordable and effective imaging tool for applications ranging from cardiology to obstetrics. Development of microbubble contrast agents over the past several decades has enabled ultrasound to distinguish between blood flow and surrounding tissue. Fundamentally, ultrasound images are visual representations of the interaction between sound waves and the medium of wave propagation. In ultrasound imaging, an acoustic pulse is transmitted into the field using a transducer capable of producing a temporally short mechanical wave in response to a voltage applied to the transducer. As the incident wave travels into tissue, some of the wave's energy is reflected back toward the transducer by scatterers in the tissue having different acoustic properties than the background medium. These backscattered acoustic waves are received by the same transducer, which converts mechanical waves into time-varying voltages. These signals are then amplified, digitized, and processed into an image by the ultrasound imaging system. In the most common mode of operation, called "B-mode" ultrasound, grayscale images are formed in which pixel values are proportional to the brightness of scattered acoustic waves. In other system modes, B-mode images are overlaid with colorized maps of blood velocity or integrated energy from moving scatterers. An ultrasonic transducer employed for both transmitting ultrasonic acoustic energy into an immersion medium and for detecting acoustic energy reflected from an object under examination is provided, the transducer having a hybrid transmitter and receiver in which a ceramic piezoelectric material is used to construct a first piezoelectric element for transmitting the acoustic energy, and a polymer piezoelectric material is used to fabricate a second piezoelectric element for receiving the reflected acoustic energy. The hybrid ultrasonic transducer provides improved performance over prior transducers using only a single ceramic piezoelectric element, in that the good transmitting properties of the ceramic are preserved, while the better receiving properties of the polymer piezoelectric are used to improve the sensitivity of the transducer. The polymer piezoelectric has the further advantage of providing a closer match of acoustic impedance to the immersion fluid used in the evaluation of objects. A simulation of transducer's model is useful in order to verify the preliminary design. Hence, SPICE implementation of hybrid multi-frequency transducer has been developed in this paper.

2. Model of Piezoelectric Transducer

The model developed for piezoelectric transducer is shown in fig.1. The block T_1 represents the transmission line. Independent sources V_1 and V_2 are zero value sources, which are used as ammeters in the circuit. F_1 and F_2 are dependent current sources. The value of F_1 is given by $F_1 = h\ C_0 \times I(V_1)$, where $I(V_1)$ is the current through V_1. The voltage across the dependent voltage source E_1 is given by $E_1 = V(4)$, where $V(4)$ is the voltage at node *4* i.e. the voltage across C_1. The dependent current source F_2 which charges C_1 is given by $F_2 = h \times I(V_2)$, where $I(V_2)$ is the current through V_2 and h is the ratio of the piezoelectric stress constant in the direction of propagation and the permittivity with constant strain . Resistor R_1 is included to prevent node 4 from being a floating node. From the mechanical side (i.e. transmission line T1), the difference between the velocity of each surface normal to the propagation path, represented by the currents u_1 and u_2 controls the current source F_1. The node labels E, B and F, respectively, denote the electrical, back, and front ports.

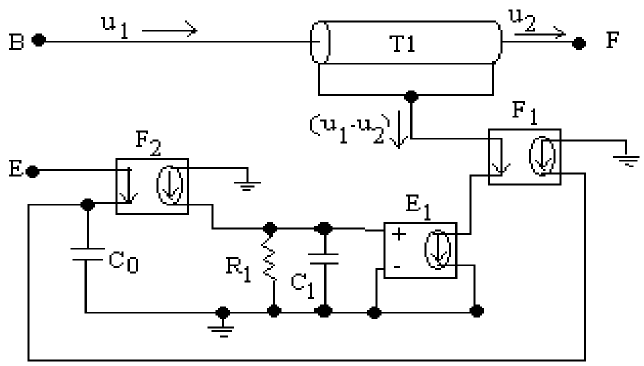

Figure 1: Model of a piezoelectric transducer

3. Model of PZT 5A - PVDF Hybrid Transducer

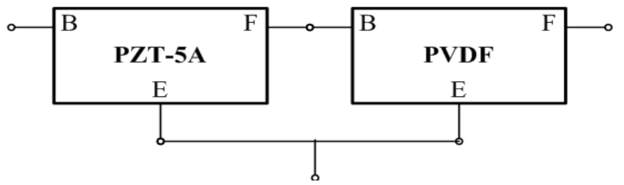

Figure 2: Hybrid Transducer Model

Fig. 2 shows series configuration of two material's equivalent models. Although in this model lossy characteristics (mechanical, dielectric, and electromechanical) of piezoelectric are considered, it must be taken to note that polymer material has complex additional losses than those of ceramic material. The piezoelectric material PZT-5A and PVDF whose material data was obtained from [1], [2], [3], and [6] was chosen and given in table 1

Table 1: Physical properties of Transducers at 25 ^{o} C

S. No	Physical properties at 25^{0}C	PZT-5A	PVDF
1	Density (ρ) (kg/m^3)	7750 [3]	1780 [2]
2	Mechanical Q (Q_m)	75 [3]	19 [6]
3	Sound velocity (c) (m/s)	4350 [3]	2200 [2]
4	Permitivity with constant strain (ε^s) (C^2/Nm$^{2)}$	7.35 ×10^{-9} [3]	55.78 x 10^{-9} [6]
5	Piezoelectric stress constant (e^{33}) (C/m$^{2)}$	15.8 [3]	0.16 [6]
6	Acoustic Impedance (MRayl)	33.7 [3]	2.7 [2]
7	Piezoelectric Constant (10^{-12} C/N)	d_{33} = 374 [1] d_{15} = 584 [1]	d31=23 [6] d32=4 [6] d33=-33 [6]
8	Coupling factor (K_{33})	[1] 0.66 [3]	0.2 [6]

Assisted with the definition of the low loss characteristic impedances equation, following relationships can be obtained

$$L = A \cdot \rho \tag{1}$$

$$C = \frac{1}{A\rho c^2} \tag{2}$$

$$R = 2\rho c A \alpha_v \tag{3}$$

$$G = \frac{2\alpha_{tc}}{\rho c A} \tag{4}$$

Mechanically, a transmission line T of length *len* (m) represents the acoustical layer. The length is selected to achieve the desired center frequency *f* (Hz) of the transducer. With fixed ends, the piezoelectric plate has a fundamental resonant frequency as

$$f = \frac{c(T)}{2 \cdot len} \tag{5}$$

Where *c(T)* is the velocity of sound through it at temperature *T*.

Using equations (1), (1) and the piezoelectric material density ρ, required for transmission line, *L* and *C* values can be calculated. The mechanical factor Q_m describes the shape of the resonance peak in the frequency domain. The relation between angular frequency ω, inductance *L* and the resistance *R* is given as [5]:

$$Q_m(T) = \frac{\omega L}{R} \tag{6}$$

In the electrical section, the static capacitance C_0 (T) at temperature T is calculated as

$$C_0(T) = \frac{\varepsilon^S(T) \cdot A}{len} \tag{7}$$

where ε^s(T) (C^2/Nm^2) is the permittivity with constant strain at temperature T [3]. The latter is related to the permittivity with constant stress (free) ε^T as :

$$\frac{\varepsilon^T(T)}{\varepsilon^S(T)} = \frac{1}{1-k^2(T)} \tag{8}$$

Where *k* (T) is the piezoelectric coupling constant at temperature T.

The mechanical and electrical sections interact with two current controlled sources. From the mechanical side, the deformation itself is not measurable, but the current representing the rate of deformation is the difference between the velocity of each surface normal to the propagation path, represented by the currents u_1 and u_2, is the rate of deformation. This current (u_1 - u_2) controls the current source F_1. It has a gain equal to the product of the transmitting constant *h* (*N/C*), and the capacitance C_0. *h* is the ratio of the piezoelectric stress constant e^{33} *(C/m*^{2}) in the direction of propagation and the permittivity with zero or constant strain ε^S. In the thickness mode it is [3].

$$h(T) = \frac{e^{33}(T)}{\varepsilon^S(T)} \tag{9}$$

This source's output is in parallel with the capacitor C_0(T). The result is a potential difference across the capacitor that is proportional to the deformation. In the electrical section, the current through the capacitor C_0(T) controls the current source F_2. The gain for this second current source is *h*(T). Its output needs to be integrated to obtain the total charge on the electrodes that proportionally deforms the transducer. The integration is performed by the capacitor C_1. The voltage controlled voltage source E_1 with unity gain is a one-way isolation for the integrator.

To evaluate the model, the model parameters of PZT-5A and PVDF transducers were calculated using equations (1, 2, 3, 5, 6, 7, 9) and given in table 2.

Table 2: Model parameters of Transducers

S.No.	Model parameters	PZT-5A	PVDF
	Physical parameters		
1	Diameter (mm)	12.7	12.5
2	Cross sectional area (A) (m^2)	0.0001267	0.0001227
3	Center frequency (MHz)	5MHz	5MHz
	Equivalent lossy transmission line parameters (Mechanical section)		
4	C	53.8nF	945.8nF
5	R	411kΩ	361.18 kΩ
6	L	981mH	218 mH
7	G	0	0
8	len	435µm	220 µm
	Electrical section parameter		
9	Static capacitance C_0	2.14nF	31.14nF
	Controlled sources parameter		
10	Transmitting constant (h) (N/C)	2.15×10^9	2.87×10^6
11	Current source gain (F_2)	2.15×10^9	2.87×10^6
12	Dependant current source gain (F_1)	4.60	0.09
13	Voltage control voltage source gain (E_1)	1	1
14	R_1	1 KΩ	1 KΩ
15	C_1	1F	1F

4. Simulation Setup for Ultrasonic System

The analogous simulation schematic setup is described in figure 3, with the transducer sub circuit shown in figure 1. In this schematic an ultrasonic probe with acoustic matching layer is symbolized by the two three-port blocks X1 and X2, which involve established PSPICE piezoelectric model. The measuring cell is modeled using lossy transmission line.

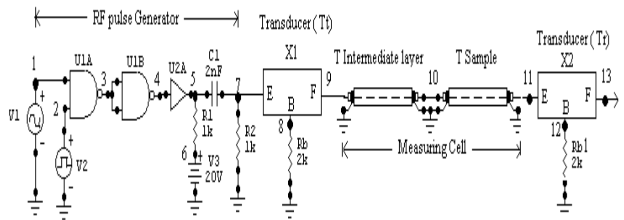

Figure 3: Simulation setup Schematic for ultrasonic test system.

5. Result and Discussion

Transient analysis of the transducer model was done with the configuration shown in figure 4. An oscillation was observed after excitation of the piezoelectric crystal. The received signal was compared in the time domain. (Fig. 5, 6, 7)

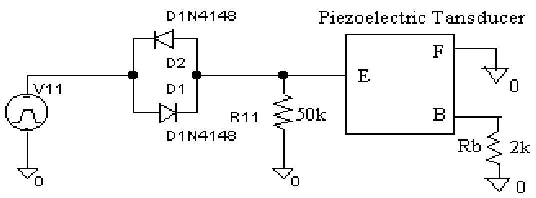

Figure 4: Simulation setup for analysis of transient behavior of transducers.

Certain polymer characteristic losses are neglected to simplify the preliminary design at this stage. Fig. 6 shows the transient response of series configuration model.

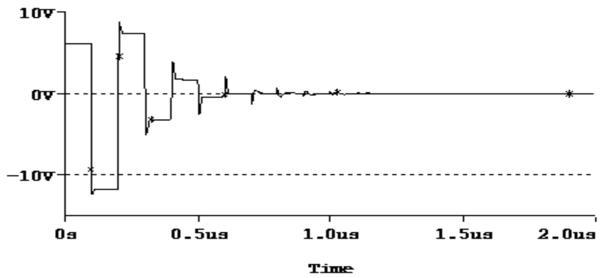

Figure 5: Transient response of PZT-5A piezoelectric Transducer.

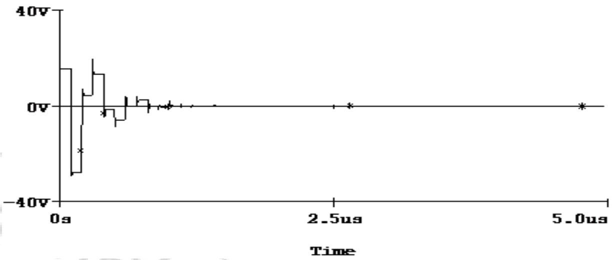

Figure 6: Transient response of PVDF Transducer

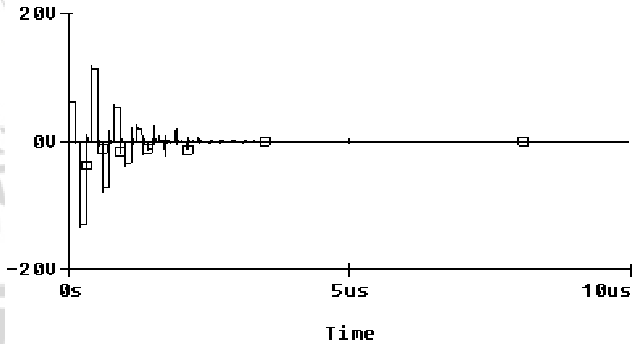

Figure 7: Transient response of ceramic-polymer piezoelectric transducer (PZT-5A+PVDF) for multi-frequency ultrasonic system

Fig. 8 shows frequency response of transducer. AC analysis was conducted to observe frequency behavior from 1 MHz to 10 MHz. There are three peaks of power spectrums: at 2.8 MHz, 5.5 MHz, and 8 MHz. The last spectrum is higher than another, but for overall dB, bandwidth from 2.5 MHz to 8.5 MHz is considered flat.

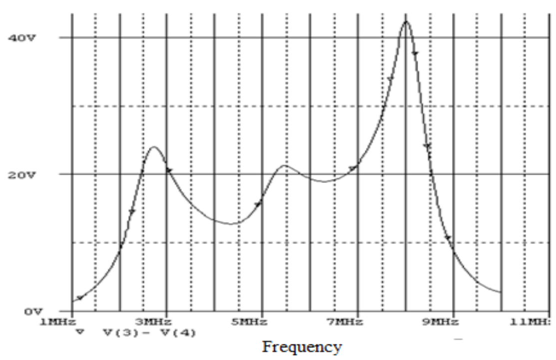

Figure 8: Frequency response of Transducer

Electrical impedance of transducer was observed as in Fig.9. It gives turning point at about 1 MHz.

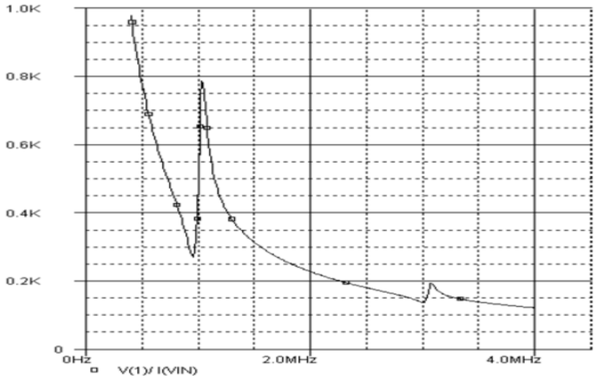

Figure 9: Electrical Impedance of Transducer

Experimental validation for water sample is shown in fig. 10

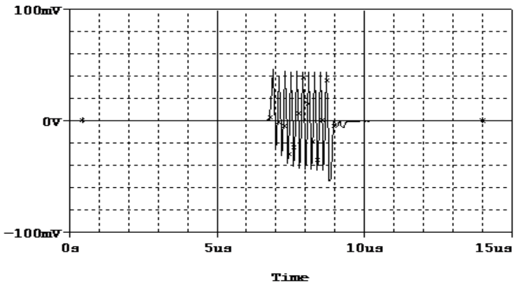

Figure 10: Complete transient received by 5MHz ceramic-polymer piezoelectric transducer (PZT-5A+PVDF) for multi-frequency ultrasonic system at 25°C in distilled water at d= 1 cm.

6. Conclusion

A model of ceramic-polymer piezoelectric has been described. Simulation level shows that by hybridization, characteristics of both materials are providing a satisfying performance for multi-frequency transducer. Initially transducer test for ceramic and polymer model were generated. By filtering and amplifying frequency range from 1 MHz until 10 MHz, the system offers wideband medical ultrasonic acceptance. It gives smooth result of ultrasound signal for medical purposes.

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