FEM SIMULAION FOR ACOUSTIC CHARACTERIZATION OF MEDICAL ULTRASOUND TRANSDUCERS USING COMSOL MULTIPHYSICS

PHÂN TÍCH ĐẶC TÍNH ÂM HỌC CỦA ĐẦU DÒ SIÊU ÂM BẰNG MÔ HÌNH PHẦN TỬ HỮU HẠN SỬ DỤNG PHẦN MỀM COMSOL MULTIPHYSICS

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Abstract - The design of a high-performance ultrasound transducer ultimately plays an important role in determining the quality of the image. The simulation based COMSOL Multiphysics (Finite Element Model -FEM) is the most powerful and accurate tool available for this application. The electrical impedance, the transducer center frequency and relative bandwidth of the ultrasound transducer is simulated by using FEM model, and then compared to XTRANS. The key point of the study is that the results between simulation 1D analysis with XTRANS and 2D analysis with FEM model need to be well in agreement to get more reliable, repeatable, and consistent results. Moreover, matching layer thickness of transducer will be set up and changed to survey its influence on the electrical impedance, the center frequency and relative bandwidth of the ultrasound transducer.

Key words - FEM; XTRANS; ultrasound transducer; COMSOL; piezoelectric.

Tóm tắt - Thiết kế một đầu dò siêu âm có công suất lớn đóng một vai trò cực kỳ quan trọng trong xác định chất lượng hình ảnh siêu âm trong lĩnh vực y học. Phần mềm COMSOL Multiphysics (Finite Element Model-FEM) là một công cụ mạnh và cho kết quả chính xác trong mô phỏng đầu dò siêu âm. Trở kháng điện, tần số trung tâm và băng thông hoạt động của đầu dò siêu âm được mô phỏng bằng mô hình phần tử hữu hạn FEM (2D), sau đó so sánh kết quả với mô phỏng bằng phần mềm XTRANS (1D). Điểm nổi bật của bài báo này là kết quả mô phỏng từ hai phần mềm trên có sai số rất nhỏ. Thêm vào đó, độ dày lớp ghép trong cấu trúc của đầu dò siêu âm sẽ được thay đổi để khảo sát sự ảnh hưởng của chúng lên các thông số kỹ thuật của đầu dò như trở kháng điện, tần số trung tâm và băng thông hoạt động cũng được nghiên cứu và trình bày.

Từ khóa - FEM; XTRANS; đầu dò siêu âm; COMSOL; áp điện.

1. Introduction

Medical ultrasound imaging application is well developed due to the improvement of its resolution and low price compared with computed tomography and magnetic resonance imaging [1]. The differences in acoustic properties of different types of tissue allow the scanner to generate an image of a part of the body, based on echo signals. Lead Zirconate Titanate (PZT) is commonly used as active material for ultrasound technology due to its high electromechanical coupling factor and sensitivity [2]. Therefore, PZT material is used as piezoelectric active layer in this design of ultrasound transducer for medical applications. The center frequency is designed to be around 5.2 MHz. In order to have a good sensitivity, the design is added by backing and matching layer.

Indeed, ultrasonic transducers are difficult to characterize, since they are complex electromechanical devices that convert electrical into acoustical energy and vice versa. The electrical impedance, transducer center frequency, bandwidth and the sensitivity are ones of the key paramaters related to the performance of a medical ultrasound transducer [3]. In this paper, these parameters will be studied by using COMSOL Multiphysics (Finite Element Model -FEM) and XTRANS program. Moreover, Transducer with two matching layers and without matching layer will be also applied to the model to survey the change in the centre frequency as well as operating bandwidth. All simulations are the most powerful and accurate tool available to characterize and optimize these parameters of Ultrasound Transducers.

2. Device structure and simulation method

2.1. Design of ultrasound transducer structure

The structure of the design is shown in Figure 1. It consists of a layer's piezoelectric material which is embedded between two electrodes layers. Two matching layers are used in the model to have further broad bandwidth of the transducer. The transducer works in the environment which is represented as load in the model. A light backing layer is used to absorb the energy so that there will be no reflection wave. The area of transducer is set to be $75 mm^2$.

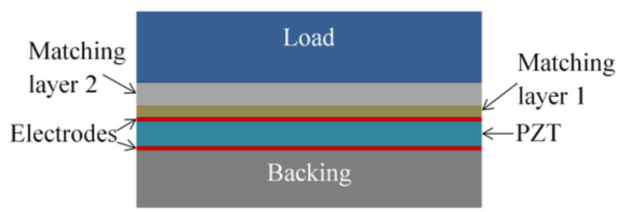


Figure 1. *Structure of Transducer.*

The used materials for the model are shown in Table 1. Gold is commonly used as electrode in MEMS technology due to their high electric conductivity and their chemical inertia. Acoustic backing and matching layers are essential for high performance ultrasound transducers owing to the large impedance mismatch between the piezoelectric element and tissue [4]. In order to understand the effect of backing and matching layers of a transducer, a schema of transmit pulse is shown in Figure 2. With a PZT transducer with no backing and matching layers, the generated pulse transmit has a long ring down time [5]. The design provides a good sensitivity but the solution is poor. By adding a backing layer to the transducer, the ringing decreases and the sensitivity of this design also decreases. The introducing matching layers improve the energy transfer efficiency in the forward direction. As a result, the ring down time is short and the sensitivity of beam is good. Widening the bandwidth of transducers by introducing quarter wavelength matching layers to the piezoelectric ceramic may result in a reduction in sensitivity [6]. It is shown that constructive reinforcement of acoustic waves will occur when the thickness of the matching layer is λ/4 [7].

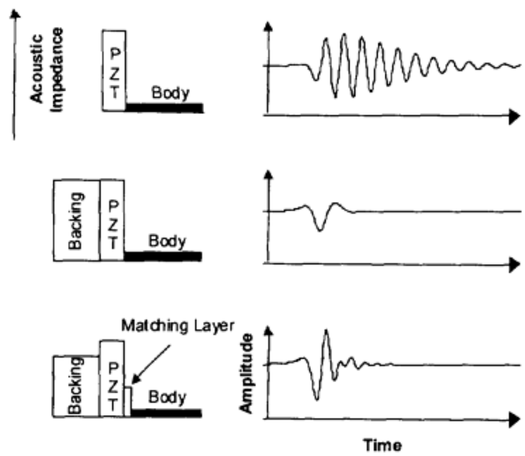


Figure 2. *Schematics show the effect of the backing and matching layer impedance on transmit pulse sensitivity and the ring down time [5]*

Table 1. *Materials used in the model [1], [3]*

Layer	Thickness [mm]	Materials	Impedance [MRayls]	Wave velocity [m/s]	Q factor
Loading	-	Water	1.5	1500[1]	1000
Matching 1	0.14844	Sliver epoxy-pp	7.3	1900	31.3
Matching 2	0.21046	Poly Propylene	2.41	2740	62.6
Electrode	0.002	Au	62.5	3240	1000
Piezoelectric	0.368	PZT5A [2]	33.7	4350	75
Baking	-	Esolder 3022	5.92	1850	4

For short impulse response and broad bandwidth, ultrasonic transducers are designed with one or two matching layers in the front of PZT to transfer a maximum power from a generator to a load. The basic physics are borrowed from transmission line theory, where *n* stages of transmission line sections of varying impedances, each quarter wavelength long, are used to match the generator to a load. In the acoustic analog, the acoustic impedance is used, and the transmission line sections become matching layers of intermediate impedance values between PZT and water. The thickness dimension is nominally one-quarter of the wavelength at the centre.

This design uses two different matching layers. The choice of matching layers needs to be considered based on acoustic impedance of piezoelectric material and the impedance of the load medium. The ideal acoustic impedances of the first Z_{m1} and second Z_{m2} are calculated in the following equations as shown in Table 2:

Table 2. *Acoustic impedance calculation of matching layers*

Layer	Equation	Impedance [MRayls]
Matching Layer 1	$Z_p^{4/7}Z_l^{3/7}$	8.879
Matching Layer 2	$Z_p^{1/7}Z_l^{6/7}$	2.339

where Z_p and Z_l are the acoustic impedance of the piezoelectric material and the load medium. From the calculation, the silver epoxy-pp with impedance 7.3 MRayls and Poly Propylene with impedance 2.41 MRayls are chosen as the matching layer 1 and matching layer 2. The solder 3022 is chosen as a backing material. For medical application, the acoustic properties of load medium are considered as water while the objective is to excite and detect ultrasound in human or animal tissues [10].

The Q factor in this model is calculated from the loss of materials by the equation [4]:

$$Q = \frac{8.7\pi \times f}{\alpha_{dB} \times c}$$

where f is the excited frequency, α_{dB} is the loss of material at frequency f and c is the wave velocity of the material.

2.2. Characterization of transducer analysis with FEM simulation

In simulation, the electrical impedance, operated bandwidth, and the center frequency of two Ultrasound Transducers, consisting of with and without matching layers are simulated and compared by using two software programs, XTRANS and COMSOL Multiphysics [11]. The method in designing the transducer follows the Mason model of an acoustic wave propagation. The graphic user interface XTRANS toolbox is taken as a design and characterisation toolbox for 1D single element transducer. Then, 2D finite element models of the transducer with the condition of voltage and charge driving have been made using COMSOL software. The analytical model with Xtrans is well established in transducer modelling. However, it is limited by being a 1D model. Consequently, the effect of a finite width of transducer structure is not included. To overcome the limitation of 1D simulation, the Finite Element Method (FEM) simulation is done by using COMSOL Multiphysics. The key point is that the results between simulation 1D analysis with XTRANS and 2D analysis with COMSOL Multiphysics need to be well in agreement to get more reliable, repeatable, and consistent results. COMSOL Multiphysics is a core tool in this study to perform numerical simulations. It exists in several versions. In this paper, version 4.4 is applied. One of the key features in this program is the chance to combine difference physical interfaces. The Acoustic-Piezoelectric Interaction interface is applied; this couples the acoustic wave equation with the governing partial differential equation (PDEs) in structural mechanics.

All domains in the model are meshed as free tetrahedral, with adequate minimum element dimensions which are displayed in Table 3. This permits a good compromise between computational time and accurate results.

Table 3. *Mesh used in the model*

Layer	Mesh
Water	water_lambda/10
Matching	ML1_lambda/20
Electrodes	electrode_lambda/20
Piezoelectric	piezo_lambda/20
Backing	back_lambda/20

Table 4 that lists all expressions is used to calculate the impedance of transducer in COMSOL Multiphysics, where intop1 means it is integrating magnetic field strength over the boundary of upper PZT surface.

Table 4. *Electrical impendence expressions*

Name	Expression	Unit
Ipiezo	2*pi*intop1(r*acpz.nJ)	A
Zepiezo	Piezo/Ipiezo	Ω
Zepiezo_r	real(Vpiezo/Ipiezo)	Ω
Zepiezo_i	imag(Vpiezo/Ipiezo)	Ω
Zepiezo_mag	abs(Zepiezo)	Ω
Zepiezo_phase	arg(Zepiezo)*(180/pi)	Rad

3. Results and discussion

Here the results obtained from the FEM models and XTRANS are presented. The time for complete simulation is approximately only 10min (on a Dell V5470 workstation, 16GB).

3.1. Electrical Impedance of Transducers

The results of the 2D FEM simulations for the electrical impedance of Transducer are shown in Figure 3. The results of the analytical 1D are plotted in the same graph for comparison. It is shown that there is a good agreement between the analytical model and the FEM simulation.

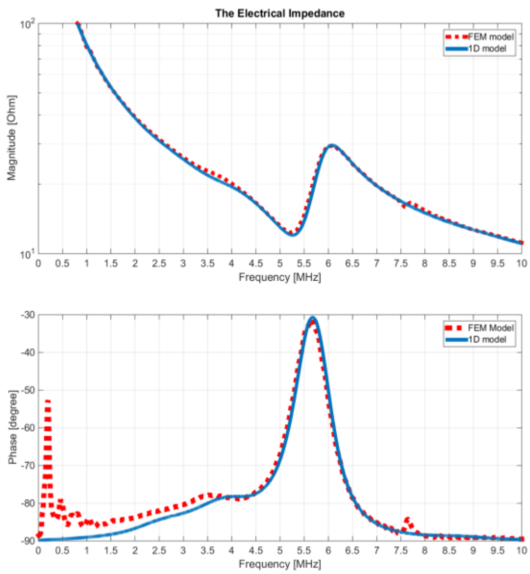


Figure 3. *Comparison of 1D and 2D simulation of the electrical impedance*

A frequency response analysis is performed, from 0 Hz to 10 MHz. The resonance frequency of transducers is around 5.2 MHz. Both first thickness vibration mode (around 5.2 MHz) and radial vibration mode (from 1 KHz up to 2 MHz) can be recognized. For this radial mode, the direction of vibration is oriented orthogonal to the direction of polarization. The diameter in the direction of propagation of this mode is much greater than its thickness. Since the resonance frequency depends on the diameter of transducer, hence a large length implies a low resonant frequency drive. For this thickness mode, the vibration is oriented along the direction of polarization. The resonant frequency depends on the thickness of the device; therefore, a thin plate implies a high frequency drive.

Figure 4 corresponds to the piezoelectric ultrasound transducer with and without two matching layers under the voltage and charge driving condition. The electrical impedance Z of transducer is calculated by Comsol with relation:

$$Z = \frac{V}{I} = \frac{V}{\int J dA}$$

where V, I, A correspond to applied voltage, induced current and area of the electrode.

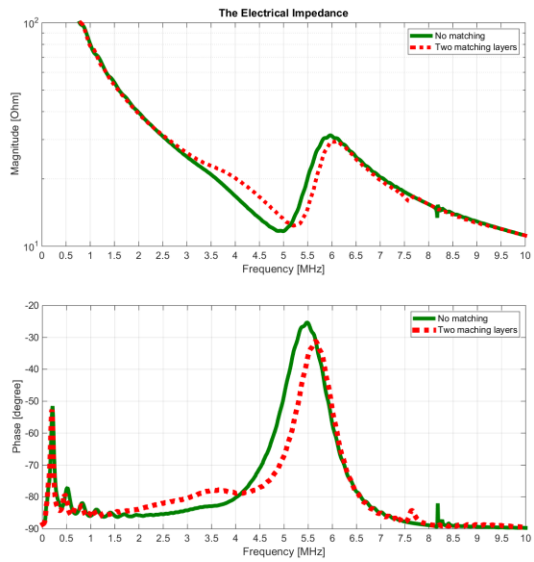


Figure 4. *2D FEM and 1D model response of 5.2 MHz transducer without and with matching layers*

3.2. The center frequency and relative bandwidth of Transducer

Bandwidth is one of the key parameters related to the performance of the medical ultrasonic transducers [3]. The study of electro-acoustic transfer function is done by one-dimensional analysis based on Mason model using the 2D FEM model. After that these results are compared to Matlab-based package Xtrans. The plot of amplitude of transfer function is shown in Figure 5. From the plot of transfer function amplitude, the achieved -3dB bandwidth of the design is from 3.125MHz to 5.9MHz with the center frequency of 5.2MHz which can be said 53% relative bandwidth.

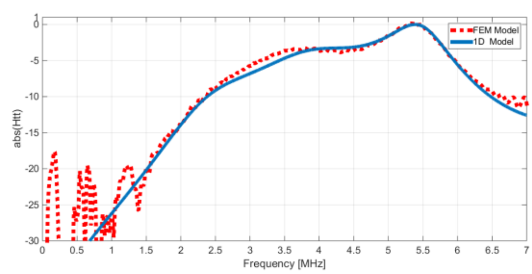


Figure 5. *Comparison of 1D and 2D simulation of the transfer function*

The Figure presents a fairly agreement (shape, center frequency and bandwidth) between the amplitude plots of transfer response obtained from 2D and 1D model. The response exhibits a large number of ripples and even significant peaks and at some frequencies both in and out of the band. These ripples are probably caused by lateral effect of sound waves resulting from the finite lateral dimensions, which may occur either inside an element or cross coupling between elements.

It is noted that the transfer function H_{tt} is calculated by Comsol with equation:

$$H_{tt} = \frac{\frac{1}{A}\int v_n dA}{V}$$

where v_n is normal velocity of the acoustic wave at the contact surface between transducer and load.

Figure 6 illustrates the comparison between two models without and with matching layers. The blue line represents no matching and red one two matching layers. Obviously, although the blue and red line have the same magnitude at most of these frequency ranges (from 4 MHz to 7 MHz), the differences are also found at relative bandwidth.

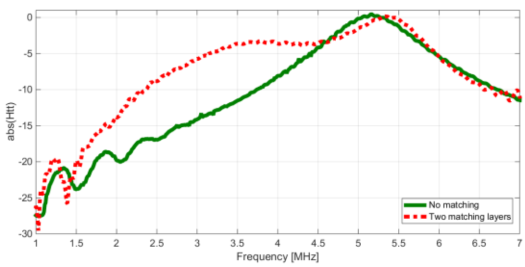
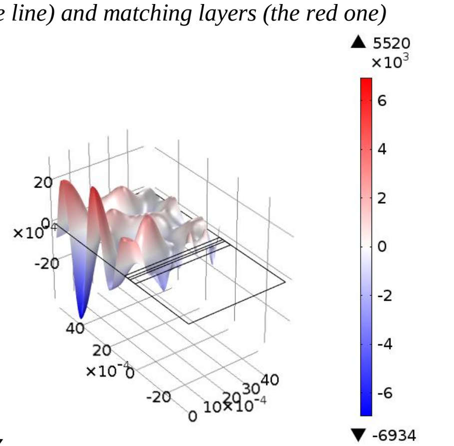
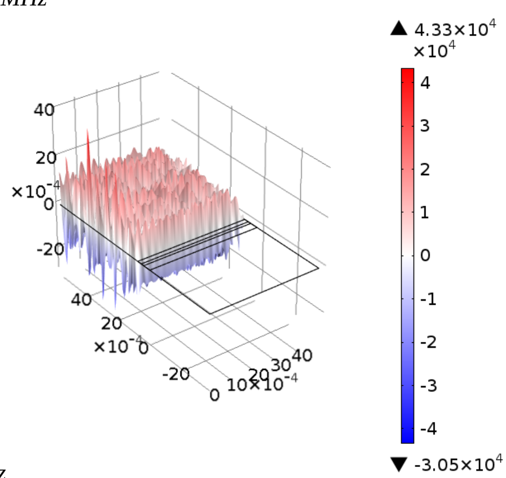


Figure 6. *The transfer function for transducer without matching (the blue line) and matching layers (the red one)*



a. 0.85 MHz



b. 5.2 MHz

Figure 7. *2D FEM cutting of wave propagation in tissue medium at 0.85 MHz and 5.25 MHz by voltage driven Transducer*

From Figure 6, it could be seen that the relative bandwidth of transducer without matching layers is approximately 27% (-3db bandwidth from 4.5 MHz to 5.9 MHz), much lower than one with matching layers.

The shape of acoustic wave propagation at this resonant frequency is shown in Figure 7a. However, while in 1D model, the transfer function seems to be smooth for the whole observed frequency range, the transfer function of 2 D FEM model in charge and voltage control is not only rough but there also exist voltage of some local resonance and anti-resonance points. These unexpected resonant frequency, *i.e.* in case of voltage driving condition is around 0.85 MHz and has a wave shape in Figure 7b.

4. Conclusion

The design and analysis based on FEM model with PZT as the active layer transducer has been presented. The design uses one backing layer and two matching layer to achieve a good sensitivity as well as the relative bandwidth. The simulated values of 1D simulation with Xtrans and 2D simulation with COMSOL Multiphysics agree to most of the values with small error in these parameters. Thus, the models can be validated for further applications. 1D and 2D simulation indicate that the design has a center frequency around 5MHz and the relative bandwidth 53%. Many different designs can be simulated with the material parameters or geometrical design variables to study the change in new designs of ultrasound imaging probes.

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