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Electrical impedance matching networks based on filter structures for high frequency ultrasound transducers

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ABSTRACT

An electrical impedance matching (EIM) network is proposed to achieve the broad bandwidth of a high frequency ultrasound transducer and to improve the signal-to-noise ratio (SNR) of an ultrasound image. The proposed EIM network is based on a general filter structure, i.e., either low-pass filter (LPF) or high-pass filter (HPF) structure composing a capacitor and an inductor. Either filter type can be selected depending on the spectral characteristics of a transducer. The prominent advantage of the proposed method is the capability to broaden the spectrum of a transducer by means of modifying its original spectrum. The values of the electronic components are computed with the measured electrical impedances of a transducer and an imaging system. In this paper, the EIM between a custom-made 50 MHz ultrasound needle transducer and a commercial pulser/receiver system was conducted to verify the characteristics and performance of the proposed method. The experimental results demonstrated that the proposed EIM network is capable of achieving the best performance of the transducer, i.e., broad bandwidth and high SNR that are desirable for the high quality of high frequency ultrasound images.

1. Introduction

Both center frequency and spectral bandwidth of ultrasound signals are closely related to the spatial resolution of ultrasound images; the higher center frequency and the broader spectral bandwidth, the higher spatial resolution [1]. In addition, ultrasound signal strength is a key factor to determine a signal-to-noise ratio (SNR) and a contrast-to-noise ratio (CNR) of ultrasound images [2]. For a high image quality, therefore, an ultrasound transducer should have capability to generate a high- strength signal with short pulse length (i.e., broad bandwidth). However, it is demanding to achieve the beneficial factors at the same time.

Piezoelectric materials play a basic role for design and fabrication of the desired transducer; ideal piezoelectric material properties for high quality ultrasound images are a high piezoelectric voltage constant, g_{33}, a high piezoelectric strain constant, d_{33}, an electromechanical coupling coefficient, k_t, close to unity, an acoustic impedance close to 1.5 MRayls (i.e., the acoustic impedance of the human tissue), and low electrical and mechanical losses [3,4]. However, the most piezoelectric materials used for ultrasound imaging transducers have an acoustic impedance much higher than 1.5 MRayls. This hampers increasing the spectral bandwidth of ultrasound. Therefore, acoustic impedance matching is conducted to solve the acoustic impedance mismatching between a piezoelectric material and the human tissue, thus increasing the spectral bandwidth [5–7].

On the other hand, a medical ultrasound imaging system considers a transducer as a capacitive device; the system delivers electrical voltage pulses to a transducer to generate ultrasound signals and detects the voltage produced by the transducer after receiving echo signals from tissue boundaries. The area, thickness, and dielectric constant of a bulk piezoelectric material determine its electrical impedance. The final electrical impedance of a transducer is varied from the original value after acoustic matching layers, a backing block, and an acoustic lens stack up on the piezoelectric material. An electrical cable to connect a transducer to an imaging system also affects the final electrical impedance. For high efficiency of the transmission and reception of ultrasound, therefore, electrical impedance matching between a transducer and an imaging system should be conducted. By doing so, the signal strength and/or the spectral bandwidth of the transducer can be increased [7,8].

High frequency ultrasound transducers have been developed to obtain a high spatial resolution on the order of several tens

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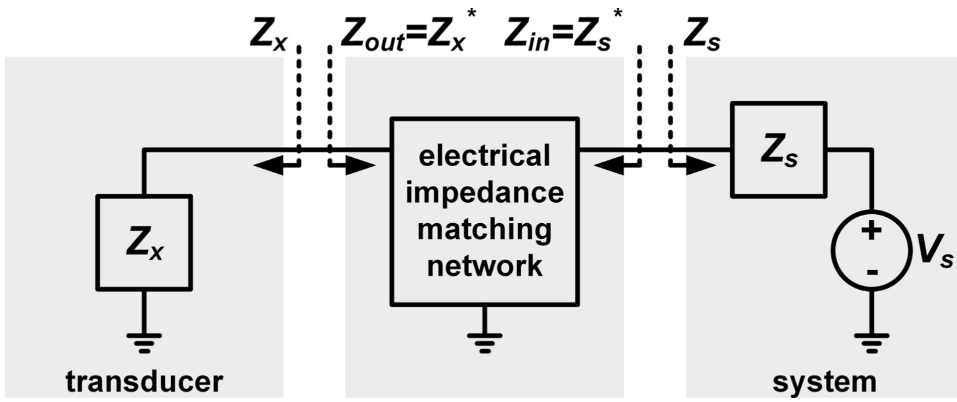

Fig. 1. Simplified configuration of an ultrasound scanner including an imaging system, an electrical impedance matching network, and a transducer.

of micrometers [9–13]. Increase in the center frequency of ultrasound requires reducing the thickness of a piezoelectric material. For single element high frequency transducers, there is a limit to increasing aperture at a given thickness of a piezoelectric material to obtain high signal strength because it causes the electrical impedance of the transducer to approach to zero, thus resulting in a severe electrical impedance mismatch to an imaging system. Note that this problem can be partially solved using piezoelectric materials with a relatively low dielectric constant $\left(\varepsilon^s/\varepsilon_0 < 40\right)$ such as lithium niobate ($LiNbO_3$) [9]. Also, it is challenging to achieve a broad bandwidth, i.e., −6 dB fractional bandwidth larger than 70% mainly due to the lack of piezoelectric materials with a high electromechanical coupling coefficient and passive materials for perfect acoustic matching. As a result, the electrical impedance matching is more crucial for high frequency ultrasound transducers.

Generally electrical impedance matching (EIM) is conducted after measuring the electrical impedance of a transducer as a function of frequency. As a simple way, either a shunt inductor or a series inductor can be used to adjust the electrical impedance of a transducer to that of an imaging system at a specific frequency [7]. However, this simple method is not optimal for broad bandwidth. Recently, more sophisticated matching circuits have been proposed for maximum acoustic power delivery at a specific frequency [14,15] and broadband electrical impedance matching in very low frequency range (less than 1 MHz) [16]. These methods utilize a model-based calculation or a computerized Smith chart to find the optimal values of the electronic components in an EIM network.

In this paper, we propose a new EIM network for the broad bandwidth of high frequency ultrasound transducers such as an intravascular ultrasound (IVUS) transducer. The proposed EIM network has a broadband filter structure of which configuration and electronic component values determine the spectral response of a transducer. The filter-structured EIM network consists of a capacitor and an inductor. Another novelty of the proposed method is a way to select the values of the electronic components used in the EIM network; based on the electrical impedances measured from a transducer and an imaging system, the values are computed at first and fine tuning is carried out to achieve the best performance, i.e., broad bandwidth and high SNR.

2. Materials and methods

2.1. EIM network based on filter structure

A general ultrasound scanner can be simplified as shown in Fig. 1; an EIM network lies in between an imaging system and a transducer. The best efficiency of ultrasound transmission can be achieved when the input and output impedances of an EIM network (Z_{in} and Z_{out}) are equal to the complex conjugates of the electrical impedance of a system (Z_s) and of a transducer (Z_x) at a certain frequency, respectively. Each electrical impedance consists of resistive and reactive components, i.e., $Z_s(\omega) = R_s(\omega) + jX_s(\omega)$ and $Z_x(\omega) = R_x(\omega) + jX_x(\omega)$, where j represents the imaginary unit and ω is an angular frequency. In the proposed method, $Z_s(\omega)$ and $Z_x(\omega)$ are measured using an impedance analyzer at first. A relation between the magnitudes of $Z_s(\omega)$ and $Z_x(\omega)$ at a frequency of interest determines the configuration of the proposed EIM network as shown in Figs. 2 and 3.

2.1.1. Case 1 $|Z_S| > |Z_x|$

When the electrical impedance magnitude of a system is larger than that of a transducer, two reactive components of the proposed EIM network (A(ω) and B(ω) in Fig. 2(a)) are subsequently connected to the transducer in series and in parallel. For an optimal EIM, the input and output electrical impedances of the EIM network should be, respectively,

$$\boldsymbol{Z_{in}(\omega) = Z_s^*(\omega) = R_s(\omega) - jX_s(\omega)}, \tag{1}$$

$$\boldsymbol{Z_{out}(\omega) = Z_x^*(\omega) = R_x(\omega) - jX_x(\omega)}. \tag{2}$$

Under these conditions, A(ω) and B(ω) can be found by the analysis of the circuit in Fig. 2(a) and these are represented with the resistive and reactive components measured from the system and the transducer as follows:

$$\boldsymbol{A(\omega) = \frac{-R_x(\omega)X_s(\omega) \pm \sqrt{R_x(\omega)R_s(\omega)\left[R_x(\omega)^2 + X_x(\omega)^2 - R_x(\omega)R_s(\omega)\right]}}{R_x(\omega)}}, \tag{3}$$

$$\boldsymbol{B(\omega) = \frac{-X_x(\omega)R_s(\omega) \pm \sqrt{R_x(\omega)R_s(\omega)\left[R_x(\omega)^2 + X_x(\omega)^2 - R_x(\omega)R_s(\omega)\right]}}{R_s}}. \tag{4}$$

The detailed derivation is presented in Appendix. Depending on $Z_s(\omega)$ and $Z_x(\omega)$, A (ω) and B(ω) can be either negative or positive value; the negative value corresponds to capacitance and the positive value to inductance. In the proposed method, the combination of either positive A(ω) and negative B (ω) or negative A(ω) and positive B(ω) is considered as an EIM network. The former combination has the structure of a low-pass filter and the latter combination is a high-pass filter (see Fig. 2(b)). These are summarized in Table 1.

2.1.2. Case 2 $|Z_S| < |Z_x|$

If the electrical impedance magnitude of a system is smaller than that of a transducer, a shunt reactive component, i.e., B(ω) should be