Model of multipath propagation of ultrasonic pulses generated in soft tissue by linear arrays

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Arrays of ultrasonic transducers used in medical diagnosis for safe and non-invasive visualization of the inside of a human body usually consist of many elementary piezoceramic transducers. In such an electromechanical construction of the array, while activating individual transducers, creation of crosstalk can be observed in the form of propagation of electric voltage and mechanical vibration on neighboring elements, which leads to distortion of the generated acoustical field, and in consequence reduces the quality of reconstructed medical images. Complexity of the problem rises rapidly in the case of arrays forming ultrasonic beams. In this paper, authors developed a numerical amplitude-phase model of multipath propagation of ultrasonic pulses generated in a soft tissue-like medium by such arrays. The model allowed simulation of acoustic field distributions, and to examine the influence of beam focusing in transmission mode on these distributions, taking into account electrical and mechanical crosstalk.

Keywords: model of multipath propagation of ultrasonic pulses, linear arrays of ultrasonic transducers, crosstalk, acoustical field, water, tissue.

1. Introduction

Generating ultrasonic wave beams using linear arrays of piezoelectric transducers is a complex process in which both the array structure and medium properties have a direct impact on the imaging quality. Resolution is one of the most important parameters determining the quality of ultrasound images. In order to improve the imaging of specific areas of studied tissue, the methods of focusing the signal energy through interference of the ultrasonic waves in the selected area are used; such as focusing or changing the direction of energy propagation (wavefront deflection) [1]. The principle of focusing the wavefront by delaying the signals transmitted from successive elements of linear array transducers has been used in this paper (Fig.1).

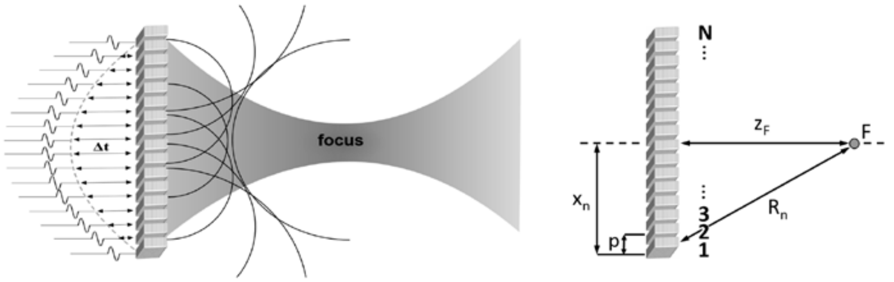


Fig. 1. Focusing the wavefront by delaying the transmitted signals.

The propagation paths of pulses from successive transducers in the array to the focus increase with the increasing distance from the center of the transducer (lying on the axis of the focus), in a given transmitting section. Therefore, the transmitting transducer located closer to the axis of the focus, must be excited with the increasing delay [2]:

$$\Delta t_n = \frac{1}{c}\left(z_F - \sqrt{x_n^2 + z_F^2}\right), \qquad (1)$$

where c – the speed of propagation of ultrasound in the medium, x_n – distance from the center of the peripheral transducer in the array to the center of the array, z_F – distance from the center of the array to the focus (focal length).

The array of ultrasonic transducers is an electromechanical structure in which, when the individual transducers are powered, the mechanical vibrations are transmitted by the structural components to the neighboring piezoelectric elements, exciting them [3,4]. An insufficient mechanical isolation in the array structure is the reason for this phenomenon. In addition, when activating the transducer by the transmitted signal, electric voltages are transferred to adjacent elements, which powers and excites them. Some of the reasons for this phenomenon are non-zero electrical capacitances between the piezoelectric elements, the leads of electrodes and socket pins, as well as electromagnetic induction [3,5,6]. As a result, the transducers which should not operate at that moment, generate unwanted ultrasonic waves in the medium with uncontrolled amplitudes and delays in the form of the so-called “acoustic crosstalk”. This causes interference with the transmitted useful wave pulses in different phases and locations in the medium, which consequently affects the size and position of the focus created in the process of forming the ultrasonic wave beam, and the total pressure wave amplitude in the focal point (Fig.2).

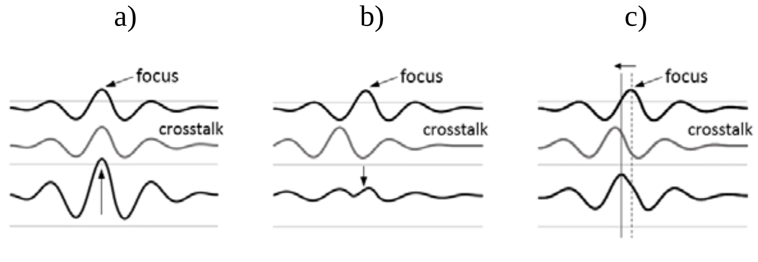


Fig. 2. The illustration of possible effects of crosstalk interference and the useful signal in the focus: a) amplitude amplifying, b) amplitude fading, c) displacing the focus.

2. Overview of available models

The propagation of ultrasound in soft tissue can be simulated numerically using a mathematical model describing the propagation of a three-dimensional acoustic wave of finite amplitude, based on the modification of a non-linear KZK equation [7,8]. Such a model takes into account the effect of linear and nonlinear phenomena. However, it requires the use of sophisticated and specialized computational software tools.

Numerical modeling of the propagation of ultrasonic pulses in the time domain in the structure of the biological medium using the Finite Differences Method (FDM) can be another, somewhat less complicated, way to do this [9]. Individual layers of tissue on parallel bordering planes are considered as homogeneous media with a frequency-dependent attenuation of waves. Reflection and transmission coefficients of ultrasound at the boundaries of layers are determined analytically for given angles of incidence. Reflection and transmission characteristics at layer boundaries are determined by numerical simulation of the propagation of broadband ultrasonic pulses, calculated based on peak amplitudes of pulses which are incident, reflected, and penetrating through the layers [9]. Thus, multiple reflections also are taken into account.

The distribution of the acoustic field generated in the biological medium by an ultrasonic linear array with a given geometry, considered as a series of small, adjacent sources can be determined from the Huygens principle: however, in the case of an elementary transducer of finite size (the set of point sources), this method is inefficient (time-consuming), and it is difficult to execute a simulation with high resolution in this way. This method sums complex values of sound pressure generated by each source point at each point of the medium by the formula [10]:

$$p(R,t) = j\frac{\rho \cdot c \cdot k}{2\pi \cdot R} \cdot V_a \cdot S \cdot e^{j(\omega t - kR)} \ , \tag{2}$$

where ρ – density of the medium in which the source is located, c – the speed of propagation of the ultrasonic wave in the medium, k – wavenumber, ω – source pulsation, S – source area, V_a – the amplitude of the acoustic velocity on the source surface.

A versatile, fast, algorithm for the numerical summation of fields of elementary transducers, shown in detail in [11,12,13,14], can also be used for the calculation of the acoustic field distributions in any two-dimensional array of ultrasonic transducers. This algorithm uses a formula for the instantaneous sound pressure generated in the far field by a flat ultrasonic transducer placed in a spherical coordinates system [15]. The value of sound pressure $p(x,y,z)$ generated by all active elements of the array is calculated as the sum of pressures p_{mn} generated by one element in many points of the medium, by virtually moving the transducer with a given value of acoustic velocity $V_a(m,n)$ [13,14].

In the present paper, due to the requirement to include multiple crosstalk, while exciting multiple elementary transducers of the linear array powered by specifically delayed pulses (electronic focusing in ultrasound B-mode scanners), we have decided to develop our own amplitude and phase model; based on the analysis of multipath propagation of ultrasonic pulses with a predetermined shape, in such a way that it enables the visualization and evaluation of the impact of crosstalk on the distribution of the focused acoustic field, in tissue-like media.

3. Model of multipath propagation of ultrasonic pulses

The developed model can generate pulses of ultrasonic waves with any given shape, based on the envelope function:

$$f(x) = 1 - |x|^m \quad . \tag{3}$$

Using this function, it is possible to develop a universal formula to give a model of *burst* signals with a given envelope (depending on the value of the exponent *m*), delay Δt, filled with a sinusoidal signal with a number of cycles m_c and frequency f_o, generated in a quasi-homogeneous medium [16]:

$$p = p_o\left(1 - \left|\frac{2(t-\Delta t)\,f_o}{m_c} - 1\right|^m\right) \cdot \sin\left(\omega_o(t-\Delta t) - \varphi_o\right) \ , \tag{4}$$

where $\omega_o = 2\pi f_o$, p – acoustic pressure, p_o – the amplitude of the acoustic pressure, t – time, Δt – pulse delay, φ_o – initial phase.

In order to obtain a good approximation of actual pulses generated in soft tissue, or in an aqueous medium, by the transducers in the ultrasonic array [16], two pieces of signals created based on the formula (4) were assembled in such a way that it was possible to adequately model the pulse rise and fall times (Fig. 3).

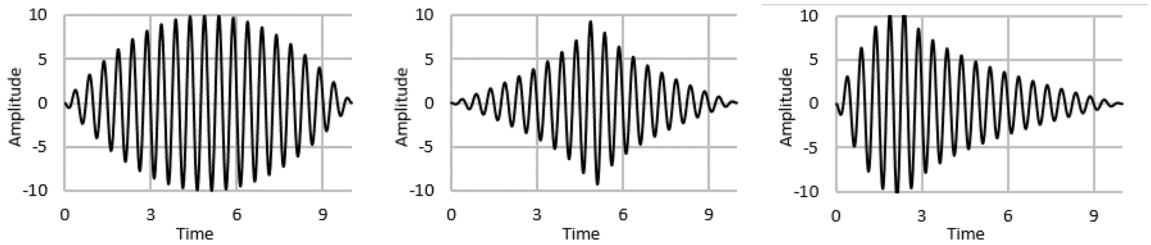


Fig. 3. Exemplary pulse shapes obtained by the formula (4): presented pulses are much longer than in reality in order to better reflect its envelopes – used scanning pulses have a length from one to several periods.

The elaborated model takes into account the drop in pressure caused by the attenuation of the amplitude of ultrasonic pulses during propagation in the medium, according to the formula:

$$p(z - z_o) = p_{z_o} \cdot e^{-\alpha\cdot(z-z_o)} \ , \tag{5}$$

where p – acoustic pressure, α – pressure attenuation coefficient, z – wave penetration depth, z_o – the distance near the transducer surface. The speed of ultrasound c = 1540 m/s, and attenuation coefficient $\alpha(f)$ = 0.5 dB/cm/MHz were assumed, which corresponds to the attenuation in soft tissue (average) α = 1 dB/cm for the frequency of 2 MHz [1,2,14]. The model simulates the transmission mode of operation of the linear array, and does not account for the phenomena of refraction, deflection, diffraction, reflection, and scattering of ultrasound waves, as well as heterogeneity, nonlinearity, and geometrical limitations of the medium.

It was also assumed that each excited transducer generates an ultrasonic wave beam of rays, with the directivity characteristics according to its size and operating frequency. This research studied rectangular transducers, for which the following formula for directivity characteristics has been used [15]:

$$K(\theta,\varphi)=K(\theta)\cdot K(\varphi)=\left|\frac{\sin\left(\dfrac{\pi\cdot a\cdot\sin\theta}{\lambda}\right)}{\left(\dfrac{\pi\cdot a\cdot\sin\theta}{\lambda}\right)}\cdot\frac{\sin\left(\dfrac{\pi\cdot b\cdot\sin\varphi}{\lambda}\right)}{\left(\dfrac{\pi\cdot b\cdot\sin\varphi}{\lambda}\right)}\right|, \tag{6}$$

where a, b – surface dimensions of the elementary transducer in the array, λ – wavelength; θ, φ – wave beam deflection angles, horizontally and vertically, respectively. Rays of the generated ultrasonic wave beam are considered in the model as pulses, with the same predetermined shape and decreasing amplitude, as a result of attenuation in the medium and the directivity of the transducer.

Crosstalk signals in the developed model are generated in the same way as the transmitted signal, by adjusting the amplitudes to a predetermined size. This calculation was based on the measurements of crosstalk carried out previously (Fig.4) [3,4].

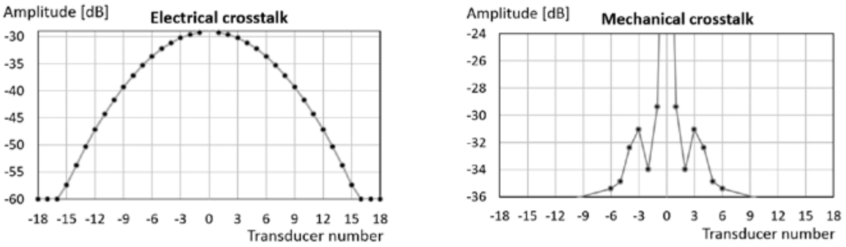


Fig. 4. Crosstalk amplitudes assumed for calculations (in relation to the assumed amplitude of the useful signal in the medium = 10 kPa).

Crosstalk signals are then summed in the medium with useful signals while taking into account the delays due to the time of wave transition from individual transducers to the established focus. In the case of mechanical crosstalk, the delay due to the propagation time of crosstalk by the array structure is additionally considered [3]. Thus, simulated electrical and mechanical crosstalk has the same frequency and shape as transmitted signal, but they differ in amplitude and delays. The number of signals generated in the developed model is calculated according to the formula:

$$n(1+(n-1)+(n-1))=2n^2-n\ , \tag{7}$$

where n – the number of transducers in the array. For example, due to the need to analyse the mutual crosstalk between the excited transducers in the linear array, as many as 7875 signals are generated for the model of a 63-element array that interfere with one another in the medium. In this model, due to the symmetry about the axis of the central array element, it has been assumed that the number of transducers is odd.

Pulses generated in the described way (formula (4), after adjusting for delays, attenuation (formula (5)) and characteristics of directivity (formula (6)) are added in the focus; as well as at any point which is a pixel of the "screen" in the acoustic field distribution imaging plane (XY) at the focal distance z_F (Fig.5), for the moment in time in which the pulses generated by all active transducers in the linear array achieve the maximum positive amplitude at this point. In the same way, crosstalk signals with given amplitudes and delays resulting from the distance of the source of the crosstalk from the point of the screen, and the time of pulse propagation by the array structure (in the case of mechanical crosstalk) are added. Transducer sizes and distances between them in the array that have been used in the calculations are shown in Fig. 5.

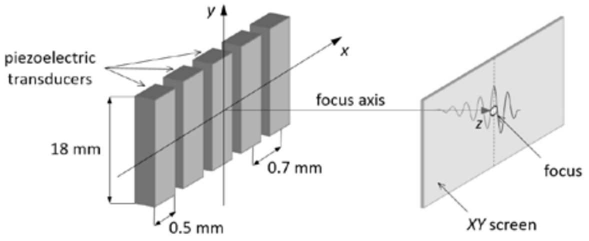


Fig. 5. *XY* calculations plane placed in the focal position ("screen").

4. Results of simulation and analysis

It was assumed in this paper that the ultrasonic wave generated by the central transducer of the modeled array, at the distance of 50 mm on its axis in the tissue-like medium, is a 6-period sinusoidal pulse with a frequency of 2 MHz, and a peak amplitude of 10 kPa, with the rapidly growing wave front ($2T$, where $T = 1/f_o$) and long decay time ($4T$) (Fig.6).

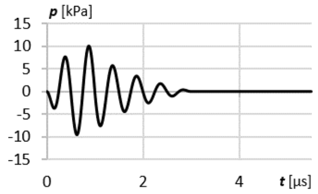


Fig. 6. Ultrasound pulse generated by a central transducer in the array in the tissue-like medium, at a depth of 50 mm.

Ultrasonic pulse focusing simulations with crosstalk (electrical and mechanical), have been carried out for 6 linear arrays with different numbers of transducers: 7, 15, 31, 63, 127, 255. All elements in each variant were excited, and took part in the focusing process. In addition, simulations were carried out for two fixed focal distances: 50 mm and 10 mm. The results shown in Fig.7 – Fig.10 are signals assembled on the *Z* axis (connecting the focus to the middle of the central transducer in the array – Fig.5) at the moment in which the pulses generated by all active linear array transducers reach maximum positive amplitude in the focus.

For arrays consisting of 7, 15, 31, 63 transducers, respectively, and the focal distance of z_F = 50 mm, crosstalk increases the signal amplitude at the focus by adding to the transmitted pulses; and the signal amplitude in the focus, after adding to crosstalk, is a bit lower for the other arrays (127 and 255 elements) (Fig.7, Fig.14). In the case of focal distance z_F = 10 mm, crosstalk increases the signal amplitude at the focus for the 7-element and 15-element array, they slightly reduce it for the array consisting of 31 – 255 transducers (Fig.9, Fig.14a). This is caused by adding up signals with different phases due to different paths of crosstalk signals in the medium. For the same reason, the amplitudes of crosstalk after the assembly on the focal axis do not grow linearly with the increase of the number of excited transducers (Fig.8, Fig.10). For example, the amplitude of the assembled crosstalk signal in the 7-element array for the focal distance z_F = 10 mm is almost 3-fold that of the 15-element array (Fig.10). Calculations show that crosstalk does not significantly affect the distribution of acoustic pressure in the array axis for amplitudes assumed in the model (Fig.7, Fig.9).

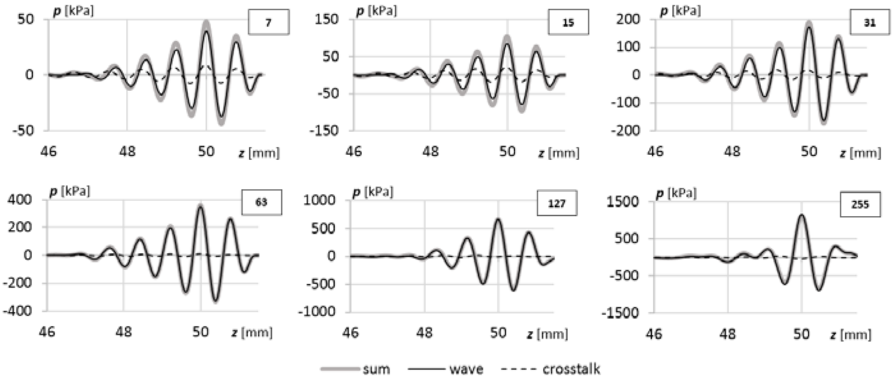


Fig. 7. The comparison of transmission signals, isolated crosstalk and the sum of transmission signals and crosstalk on the *Z* axis, for the linear array consisting of 7, 15, 31, 63, 127, 255 simultaneously excited transducers, after focusing at the distance of $z_F = 50$ mm.

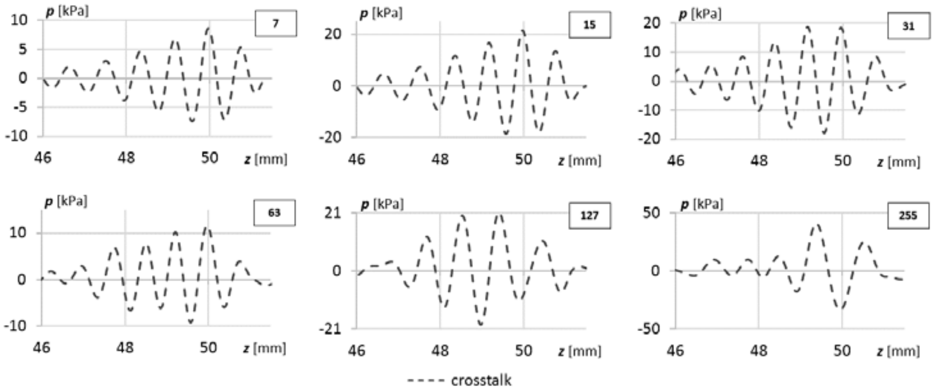


Fig. 8. The comparison of isolated crosstalk on the *Z* axis, for the linear array consisting of 7, 15, 31, 63, 127, 255 simultaneously excited transducers, after focusing at the distance of $z_F = 50$ mm.

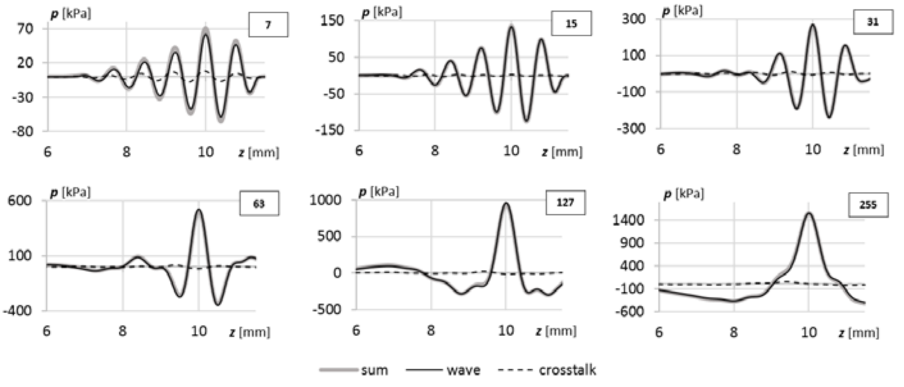


Fig. 9. The comparison of transmission signals, isolated crosstalk and the sum of transmission signals and crosstalk on the *Z* axis, for the linear array consisting of 7, 15, 31, 63, 127, 255 simultaneously excited transducers, after focusing at the distance of $z_F = 10$ mm.

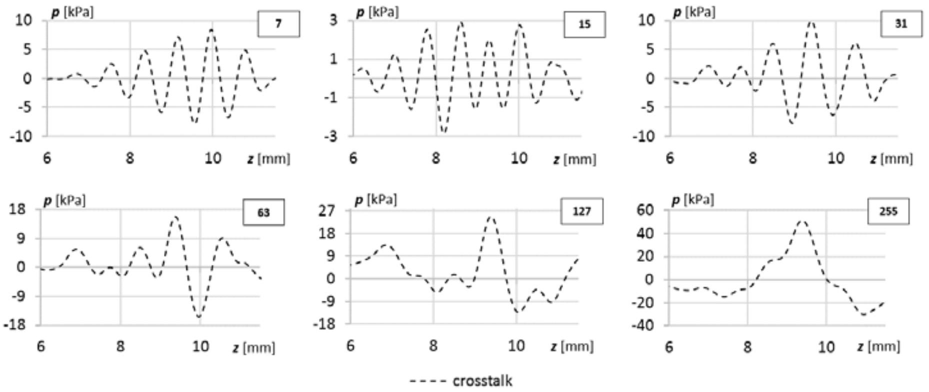


Fig. 10. The comparison of isolated crosstalk on the *Z* axis, for the linear array consisting of 7, 15, 31, 63, 127, 255 simultaneously excited transducers, after focusing at the distance of z_F = 10 mm.

The analysis of the focus displacement in the *Z* axis as a result of crosstalk is shown in Fig.11. This displacement has been referenced to the wavelength in the soft tissue λ = 1540 ms / 2 MHz = 0.77 mm. The interference of transmitted pulses with crosstalk results in the focus displacement along the *Z* axis, in the direction towards the transducer array, or, in the opposite direction (Fig.11). In the studied conditions, the displacement is relatively small (0.1 % – 0.5 % of the wavelength λ) and increases for a small number of excited transducers (Fig.11).

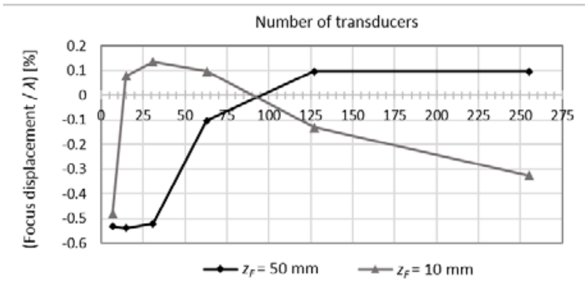
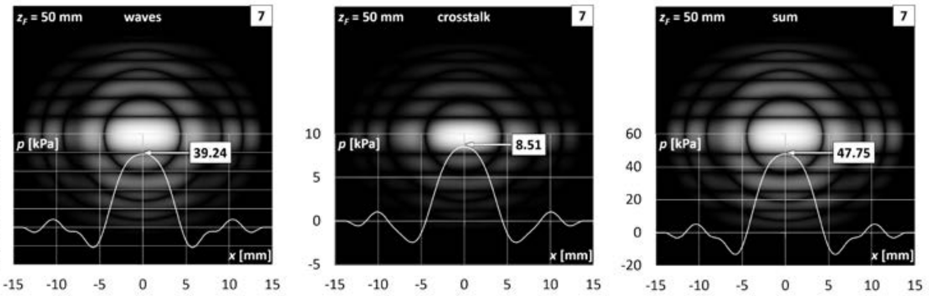


Fig. 11. Focus displacement with respect to wavelength λ (positive number means further from the matrix, negative number means closer to the matrix) caused by crosstalk.

Acoustic field distribution, simulated using a developed model shown in Fig.12 and Fig.13, present the formation of the acoustic field for different lengths of linear arrays of ultrasonic transducers in the transmission mode with focusing. The second column in the drawings contains the distribution of the acoustic field generated only by crosstalk, while the last column shows the effect of crosstalk on the creation of the acoustic field in each of the considered cases.



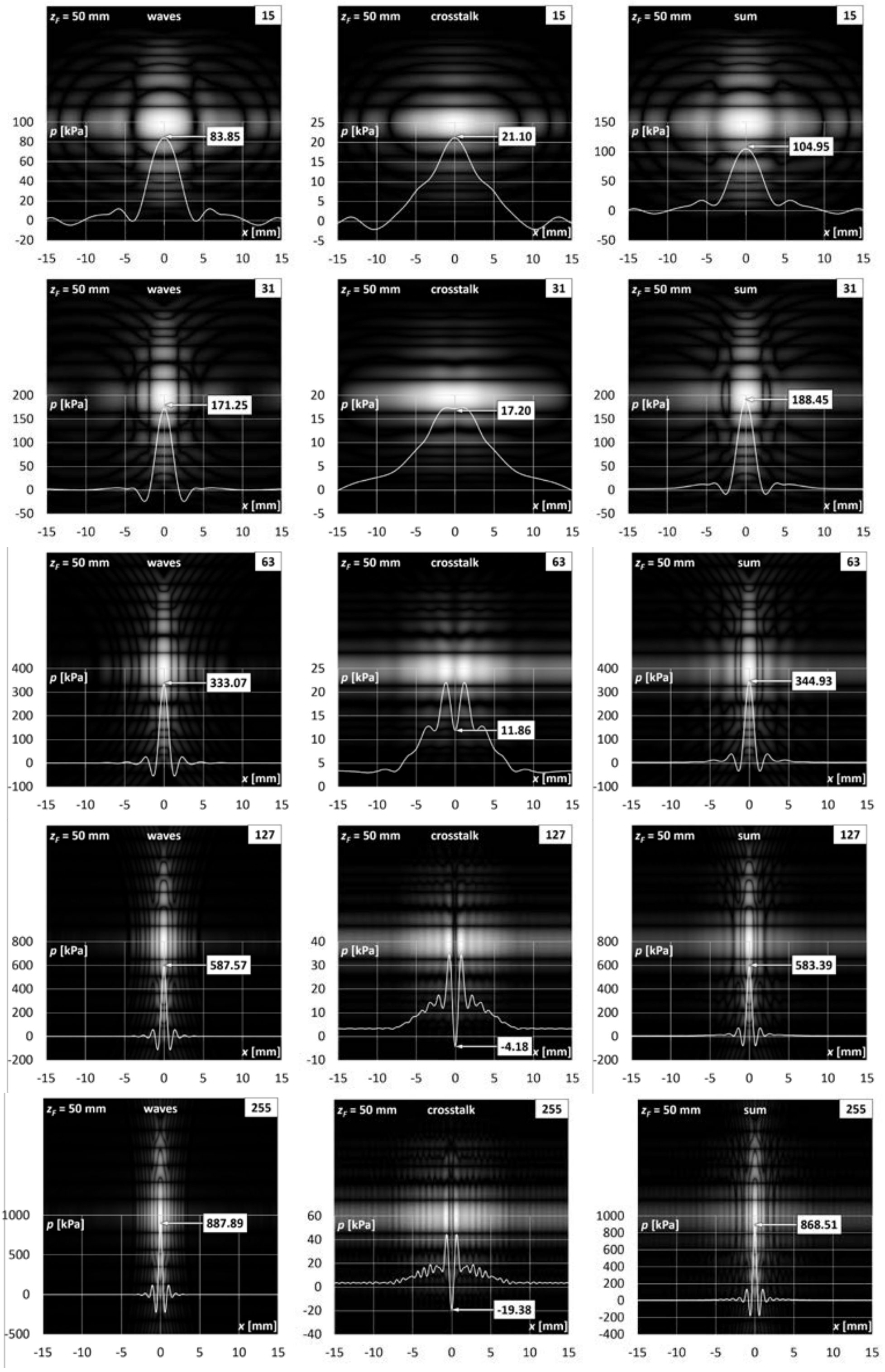
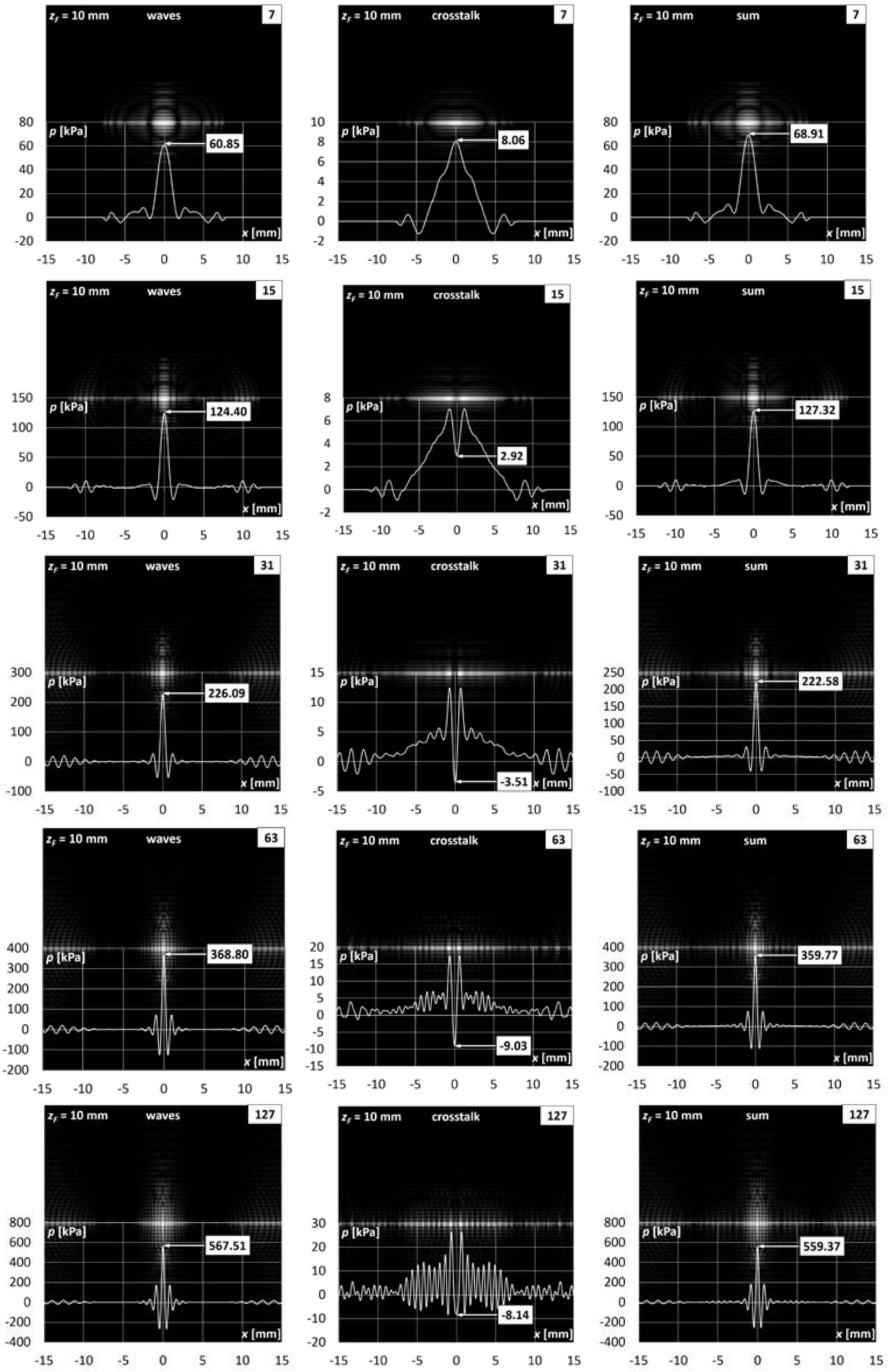


Fig. 12. Distributions of the acoustic field (the assembly of pulses without crosstalk, superposition of crosstalk only, cumulative distribution – presented in columns) at the distance of z_F = 50 mm from the linear array made of 7, 15, 31, 63, 127, 255 transducers – presented in rows.


z_F = 10 mm
waves
crosstalk
sum
7
15
31
63
127
p [kPa]
x [mm]
60.85
8.06
68.91
124.40
2.92
127.32
226.09
-3.51
222.58
368.80
-9.03
359.77
567.51
-8.14
559.37
-15
-10
-5
0
5
10
15

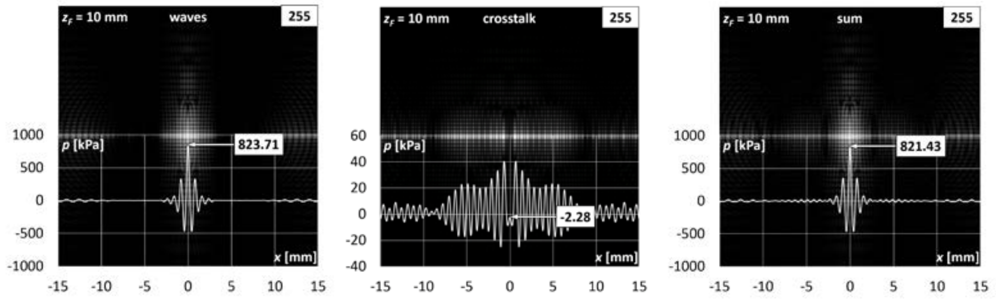


Fig. 13. Distributions of the acoustic field (the assembly of pulses without crosstalk, superposition of crosstalk only, cumulative distribution – presented in columns) at the distance of z_F = 10 mm from the linear array made of 7, 15, 31, 63, 127, 255 transducers – presented in rows.

Figure 14 shows the results of changes in the signal amplitude in the focus (Fig.14a) and a 3-decibel width of the focus (Fig.14b) resulting from crosstalk for the arrays made of 7 – 255 elements.

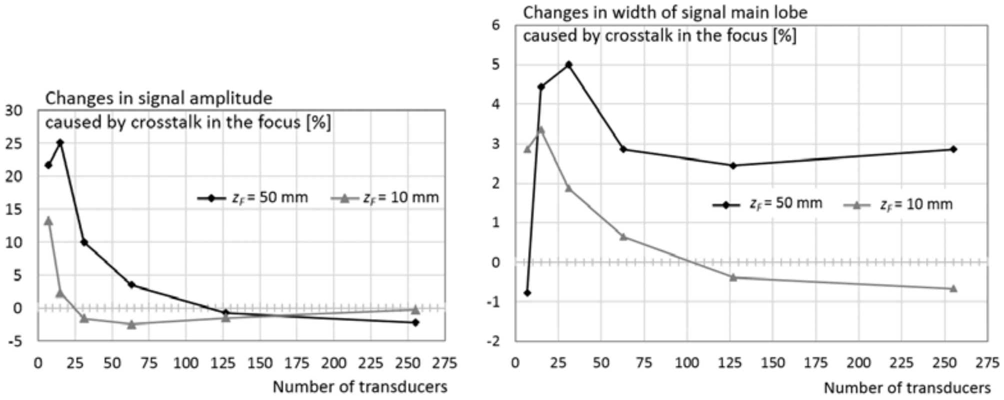

a) b)

Fig. 14. Changes in signal amplitude in the focus (a) and a 3-decibel focus width (b) resulting from crosstalk depending on the number of transducers in a linear array.

5. Conclusions

The developed computer model of multipath propagation of ultrasonic waves, generated in soft tissue by a linear array of ultrasound transducers, allowed generation of the wave assembly on the focal axis. The assembly consists of transmitted pulses generated by the array, isolated crosstalk, and the sum of transmitted pulses with crosstalk for a given number of excited transducers in the array. The model also enabled generation of the acoustic field distribution, including crosstalk, for given focal length values. The analysis of the results obtained enables evaluation of the effect of crosstalk (the total of electrical and mechanical) on the generation of the acoustic field, in the process of focusing in ultrasound linear arrays.

The study shows that the crosstalk affects the distribution of the acoustic field produced by a linear ultrasonic array operating in the focusing mode. This impact is difficult to determine *a priori*, because it depends nonlinearly on the number of excited transducers, focal length, and the distribution of crosstalk amplitudes; however, in all cases, the interference with crosstalk resulted in blurring the focus, which increased with the decreasing effective aperture of the array (Fig.14b). It has also been shown that crosstalk increases the signal amplitude in the focus by adding up to the transmitted pulses for small apertures and slightly reduces it for larger ones (Fig.14b). Focus blur, and changes to the signal amplitude in the focus, they are generally higher the greater is the focal length.

A slight displacement in the focus was also observed (0.1 % – 0.5 % of the wavelength) along the focal axis (Fig.11).

Acoustic field distributions which include crosstalk (Fig.12, Fig.13) simulated using the developed model, show distortions in different places outside the focus that limit the dynamics of imaging. These distortions decrease with the increasing focal length.

One can assume that these phenomena will intensify with the increasing amplitude of crosstalk.

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