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Characterization of harmonic pressure field: Application to medical ultrasound imaging

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Abstract*. Over the last few decades medical ultrasound has become an established diagnostic and therapeutic tool since it produces detailed and high resolution images of tissues in human body. Harmonic imaging is among the recent developments which has brought further improvements to the image quality. Harmonic imaging for tissue or with contrast agent induced a rapid evolution of this modality to diverse clinical uses, among which myocardial perfusion determination seems to be the most important application. This brought the need to understand the physical processes involved in the propagation of finite amplitude sound beams, and the issues for redesigning and optimizing the transducers with higher performances for both tissue imaging and contrast imaging. Concerning tissue harmonic imaging, the advantage of the harmonic beam generated at two times the transmit frequency are translated by reduced reverberations , greater depth of penetration at higher frequencies and improved resolution. In order to characterize the harmonic beam, a time domain solution of the parabolic nonlinear wave equation is used. This equation is traditionally applied in a propagation direction along the central transducer axis, and has been shown to model the pulse propagation satisfactorily. In this work, the characteristics and performances of the second harmonic acoustic beam from a focused piston aperture are described and the physical principles behind tissue harmonic imaging are computed. The field properties are then discussed regarding image quality. Special attention is given to the transmitted and received bandwidths variation and the reception of the pure echo signal.

Key words—**Harmonic Imaging, KZK Equation, Medical Ultrasound Imaging, Nonlinear Acoustic Propagation.**

I. INTRODUCTION

The echocardiographic community has seen wide technical advances over the last two decades. This increasing ascendancy is explained by the advent seen, such as high resolution images (high frequency transducers), blood flow measurements, real time three-dimensional imaging and contrast imaging. In addition to its diagnostic applications, ultrasound has advantage for use in therapy where it is used as a surgical tool when higher intensities are transmitted. Major recent improvements have been achieved in medical ultrasound imaging by exploiting the characteristics of nonlinear fields [1]-[3]. Harmonic generation has been used to create images offering improvements over conventional B-mode images in spatial resolution and, more significantly, in the suppression of acoustic clutter and side-lobe artifacts [4]. The Khokhlov-Zabolotskaya-Kuznetsov (KZK) equation is often used to study nonlinear characteristics of medical ultrasound beams [5]. This equation is applied in a propagation direction along the central transducer axis towards the human body for imaging. In second harmonic imaging mode, a transducer transmits at one frequency and receives at double this frequency. The major advantage of imaging at the second harmonic frequency is a significant reduction of reverberations and artifacts from off axis scatterers in addition to an improved resolution. In this work, we present the advantage of second harmonic over the fundamental in ultrasound imaging, namely the problem of the near field which is a source of artifacts in the medical image, particularly cardiac imaging among obese patients or by reflexion on the ribs, and side-lobes level which are also the origin of artifacts. Improvements in the axial and radial resolutions and surmount of undesirable field parameters. A comparison between the fundamental and harmonic pressure fields is carried out at various excitation parameters to understand the influence of these parameters on the harmonic generation. In order to effectively employ the information, comprised in the second harmonic of the received signal, this information should be properly extracted.

II. BASIC THEORETICAL MODEL

All finite amplitude ultrasonic waves undergo a degree of nonlinear distortion when traveling through real media. The distortion is due to slight nonlinearities in sound propagation that gradually deform the shape of the propagating wave, and result in the development of additional harmonic frequencies that were not present in the initial transmitted wave. More precisely, the reason of the distortion of the wave shape is that the medium (tissue) is not a completely incompressible medium. At the positives cycles of the acoustic pressure wave (compression) the temperature increases and, the density will increase proportionally while during the negatives cycles of the acoustic pressure wave (expansion) the temperature decreases and also the density of the medium [6]-[8]. This change in medium density influences the local propagation speed of sound. Indeed, the positive part of the wave propagates a bit faster than the negative part, leading to a slight deformation in the shape of the wave. This deformation accumulates in depth with propagation distance and is more significant for high acoustic pressure intensities. The distortion manifests itself in the frequency domain by the appearance of additional harmonic signals at integer multiples of the original excitation frequency. The Khokhlov-Zabolotskaya-Kuznetsov (KZK) equation is usually used to describe nonlinear wave propagation [9]. We consider a pressure field generated from each point of the transducer surface propagating in a thermoviscous fluid in the axial direction oz. The KZK equation which describes the combined effects of diffraction, losses and nonlinearity is based on a parabolic approximation and is given by the expression [6]:

$$\frac{\partial^2 p}{\partial z \partial \tau} = \frac{c_0}{2}\nabla_\perp^2 p + \frac{\delta}{2c_0^3}\frac{\partial^3 p}{\partial \tau^3} + \frac{\beta}{2\rho_0 c_0^3}\frac{\partial^2 p}{\partial \tau^2} \quad (1)$$

Where p is the sound pressure, z is the axis of propagation, $\tau = t - z/c_0$ is the retarded time, t is the time, c_0 is the speed of sound, δ is the sound diffusivity corresponding to thermoviscous absorption, ρ_0 is the ambient density of the medium, and β is the coefficient of nonlinearity of the medium (β = 1+B/A where B/A is the nonlinearity parameter). The operator $\nabla_\perp^2$ is the Laplacian in the plane perpendicular to the axis of the propagation. For circular sources (axisymmetric beams), the Laplacian is given by: $\nabla_\perp^2 = \frac{\partial^2}{\partial r^2} + \frac{1}{r}\frac{\partial}{\partial r}$. The first term on the right hand side of equation (1) accounts for diffraction, the second term accounts for absorption (attenuation) and the third term accounts for nonlinearity. The KZK equation describes accurately the propagation of sound fields produced by directive sound sources [6]. However, since this equation is based on a parabolic approximation, its accuracy is limited to distances beyond a few source radii and in a region close to the axis of the transducer [10].

III. COMPUTATIONAL MODEL

The numerical algorithm employed solves the equation in time domain and is based on finite differences method with a stepping in the axis of propagation direction. This algorithm follows similar lines as the algorithm described by Lee and Hamilton [11]. For circular sources, the equation is solved in the half space r≥0 and the boundary condition (∂p/∂r) = 0 is applied when r = 0. The numerical algorithm employed has been previously validated [16]. The simulation is able to determinate the ultrasound field at any point in space and gives information about localized maximun and minimum energy produced by the transducer. Simulations were carried out considering a circular transducer of 20mm of diameter and 60mm of focal distance.

In harmonic imaging, the presence of nonlinear frequency components in the received echo signal is caused either by nonlinear propagation effects of the medium or by the presence of a scatterers that are capable of reflecting (or reradiating) the transmitted energy in nonlinear manner, e.g., contrast agents [17]. As a consequence, several imaging modalities have been developed to acquire the pure harmonic information. These methods require data processing sequences more or less complex. In this paper, the KZK equation is used to characterize the harmonic pressure field and to define scanning parameters that provide optimal harmonic signals. Among the limitations encountered in harmonic imaging is the trade off between resolution and sensitivity. In order to effectively employ the information, contained in the higher harmonics (second harmonic) of the received signals, this information should be properly extracted. On account of the limited transducers bandwidth, the transmitted ultrasound waves must have narrow bandwidths for reduce overlap between fundamental and second harmonic bandwidths. Harmonic received band must not contain components from transmit band and, its components must sufficiently separable from fundamental spectral component for better filtering around second harmonic. Therefore the transmitted fundamental bandwidth should be large enough to ensure resolution while overlapping with harmonic frequencies should be minimized (Fig. 1).

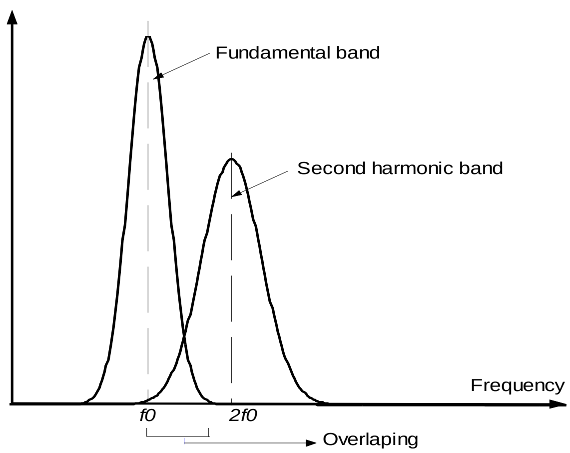
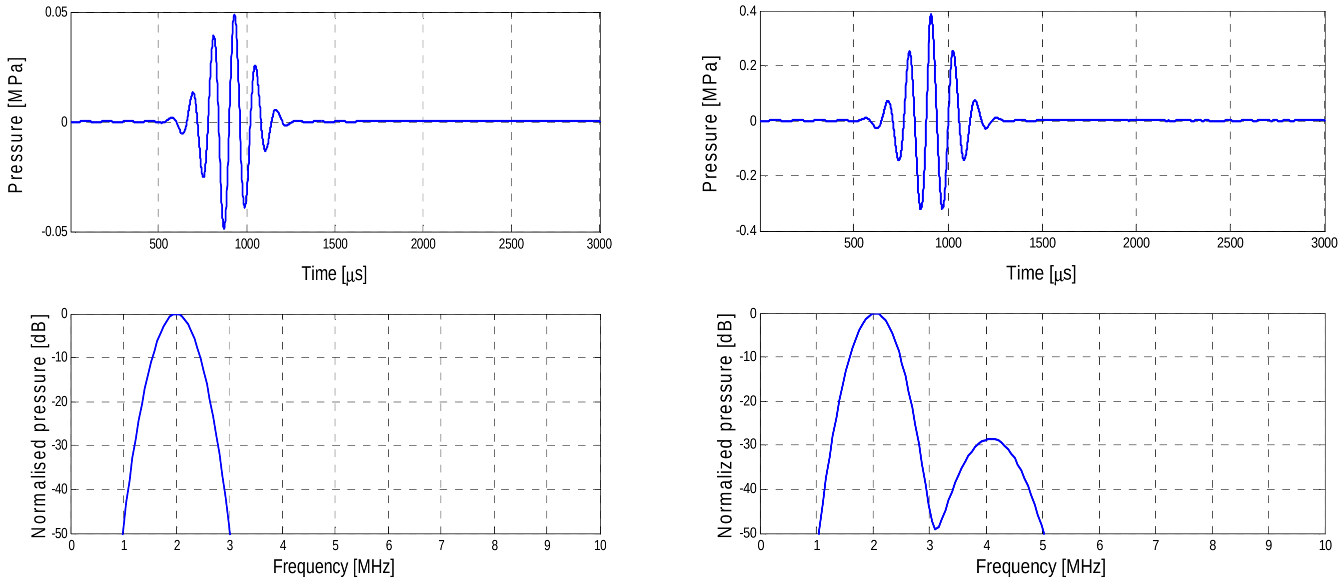


Figure 1: Overlap between the fundamental and second harmonic frequency bands.

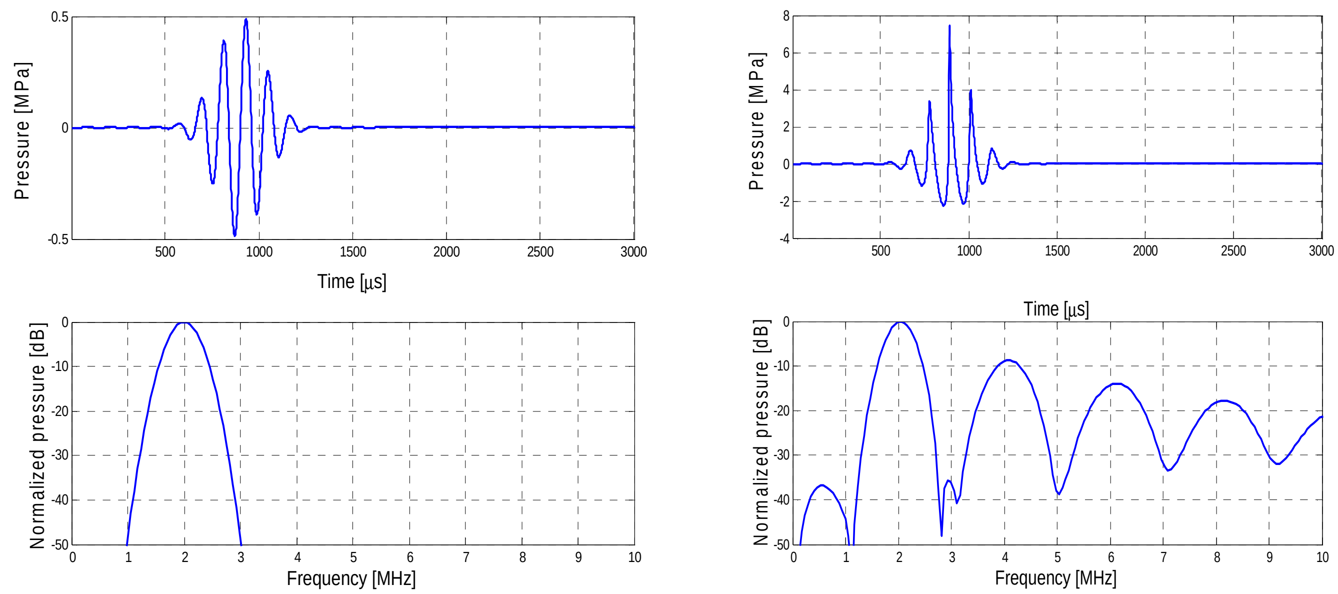
IV. RESULTS AND DISCUSSION

Figures 2 and 3 show examples of pressure waveforms at the source and at the focal distance of the piston transducer. Figure 2 shows the results at 50 KPa, 2 MHz and 3 cycles transmit waveform while Fig. 3 shows the results for a 500 kPa acoustic pressure. The distortion of the waveform after propagation is more significant for the high transmitted pressure (fig. 3B), where we appreciate a larger number of harmonic components located at multiple frequencies of the excitation frequency.



A. **B.**

Figure 2: 50 KPa, 2 MHz and 3 cycles transducer excitation. **A.** Top: waveform at piston plane; bottom: corresponding spectrum. **B.** Top: distortion of the waveform in focal depth; bottom: corresponding spectrum.



A. **B.**

Figure 3: 500 KPa, 2 MHz and 3 cycles transducer excitation. **A.** Top: waveform at piston plane; bottom: corresponding spectrum. **B.** Top: distortion of the waveform in focal depth; bottom: corresponding spectrum.

The improvement in image quality for harmonic imaging is related to three principal characteristics of the harmonic sound field. The harmonic components are generated gradually according to the distance of propagation. Thus, near to the probe, the second harmonic signal level is very low, and as a consequence, all the interferences and reverberations resulting from obstacles of near field such as skin, ribs or fatty tissue will be strongly reduced. These reverberations are responsible indeed for many artifacts in the ultrasound image. Figures 4 and 5 show respectively the variations of the second harmonic near field pressure level and the variations of the second harmonic side lobes level with the duration of the excitation. Both the near field level and side lobe levels increase when the length of the transmitted signal is increased. Thus, decreasing undesirable near field and side lobe levels obliges excitations of narrow durations.

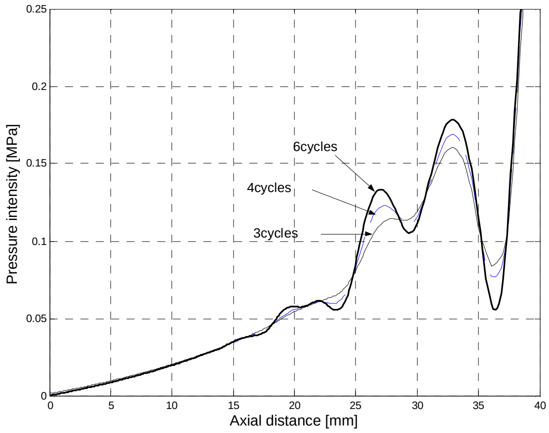


Figure 4: Second harmonic axial near field variation with the number of cycle in excitation. Case of 2MHz frequency excitation.

Figure 6 shows fundamental (F) and, second harmonic (2F) ultrasound field in direction of propagation from a circular transducer of 20 mm of diameter. The demonstrates that the build up of the second harmonic field is cumulative as explained by very low field intensity near the source and then increases with the propagation distance. In the case of high excitation pressure, the second harmonic ultrasound field level in focal point is much better in its level than in the case of low pressure, leading to a better contrast and consequently to improved image quality. As it was mentioned earlier, the second harmonic component provides a better penetration. Figure 7 illustrates the difference in focal depth between the fundamental and the second harmonic fields. In figure 8 is shown the difference in focal depth between the fundamental and the second harmonic axial field as a function of the excitation frequency. This difference in depth decreases when the frequency of excitation increases. This phenomenon is due primarily to the attenuation. The second main acoustic property of the second harmonic field is the low level of the side lobes and grating lobes.

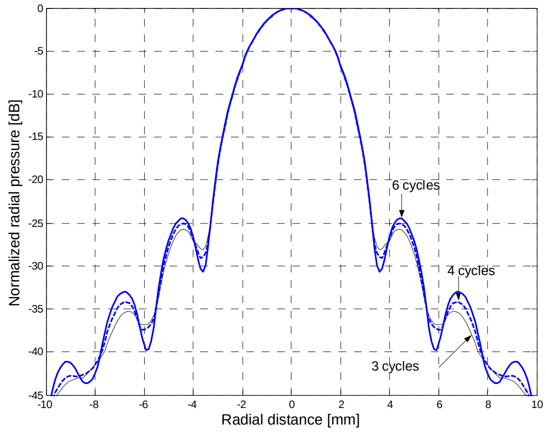


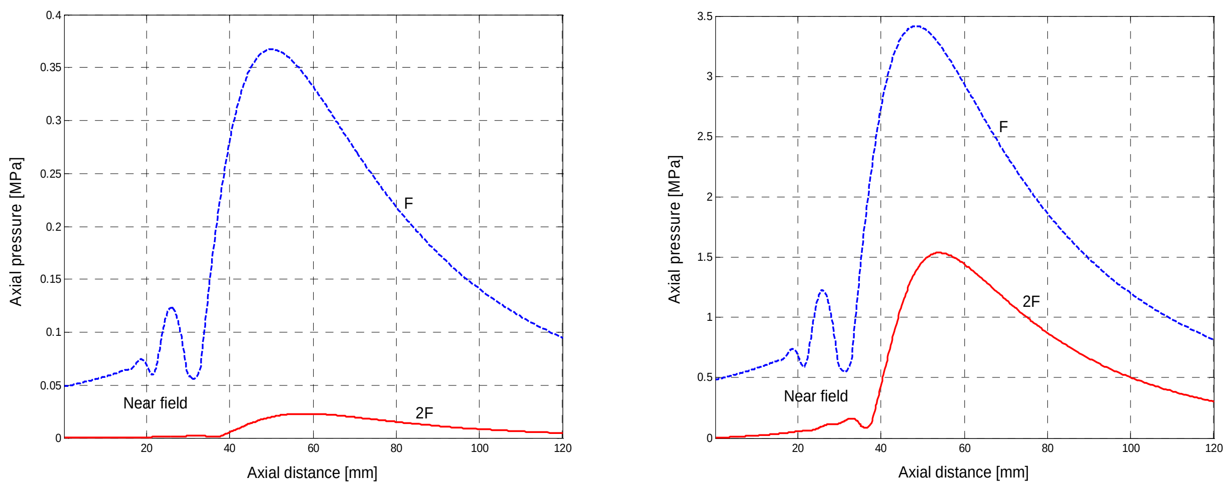
Figure 5: Second harmonic side lobes level variation with the number of cycle in excitation. Case of 2MHz frequency excitation.

Indeed, considering the nonlinear relationship between fundamental and harmonic components, the second harmonic field presents a significant suppression of side lobes and grating lobes. These lobes are known to be responsible of the artifacts coming from obstacles or reflectors located off-axis. This characteristic is shown in figure 9 in which are displayed the normalized radial ultrasound field profiles at fundamental and second harmonic frequencies. The second harmonic field shows a consequent suppression of the side lobes level expressed here by a reduction of about 5 dB. The third property of second harmonic field responsible for the improvement of image quality is its beam width. Figure 9 shows also that the second harmonic beam width is narrower than the fundamental one with a reduction of the about 60 %. The following table shows that beam widths are definitely better than if we consider linear propagation. This reduction in beam width indicates an improvement in lateral resolution of ultrasound image at this frequency.
In order to ensure that the higher frequencies are due only to harmonics generated by a nonlinear process, the transmitter must be restricted to a band of frequencies around the fundamental frequency. Therefore overlap between fundamental band and second harmonic band should be avoided. Figure 10.A shows the frequency bandwidths of the fundamental and second harmonic components as a function of number of cycles in the transmit pulse as calculated at -10 dB and -20 dB levels.

TABLE I
BEAM WIDTH OF 20mm PISTON DIAMETER AT 2MHz, 3 CYCLES OF EXCITATION.

Transmit pulse intensity	Fundamental beam width at -10dB	Fundamental beam width at -20dB	2^{nd} harmonic beam width at -10dB	2^{nd} harmonic beam width at -20dB
50KPa	3.6mm	5mm	2.2mm	2.9mm
500KPa	3.8mm	5.2mm	2.4mm	3.1mm

The curves demonstrate a similar trend for both fundamental and second harmonic bandwidths. Figure 10.B shows the bandwidth of second harmonic component as a function of transmit fundamental bandwidth at two different applied acoustic pressures (50KPa and 500KPa). The bandwidth of the second harmonic component increases when the transmit bandwidth increases. For example, at 500 KPa transmit pressure, 80% fundamental bandwidth engenders a 50% harmonic bandwidth. This means that the second harmonic component extends from 3 MHz to 5 MHz; while 80% transmit bandwidth extends from 1.2 MHz to 2.8 MHz. At 50 KPa transmit pressure, 80% fundamental bandwidth engenders a 56% harmonic bandwidth and, the second harmonic component extends from 2.88 MHz to 5.12 MHz, either 240KHz of band is gained. At this level, we conclude that the second harmonic component is purely a nonlinear component and has not been transmitted since no overlap occurred between the linear and the nonlinear components.



A. **B.**

Figure 6: Axial field towards axe of propagation focalised at 60 mm. **A.** Low pressure (50 KPa). Dashed line: fundamental field; solid line: 2nd harmonic field. **B.** High pressure (500 KPa). Dashed line: fundamental field; solid line: 2nd harmonic field.

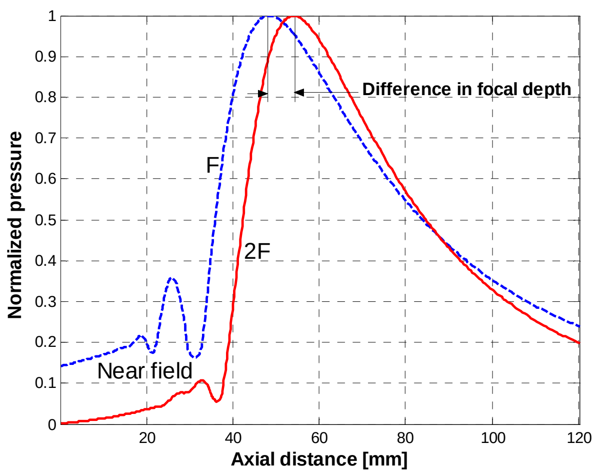


Figure 7: Illustration of the difference in focal depth. Dashed line: fundamental field; solid line: 2nd harmonic field

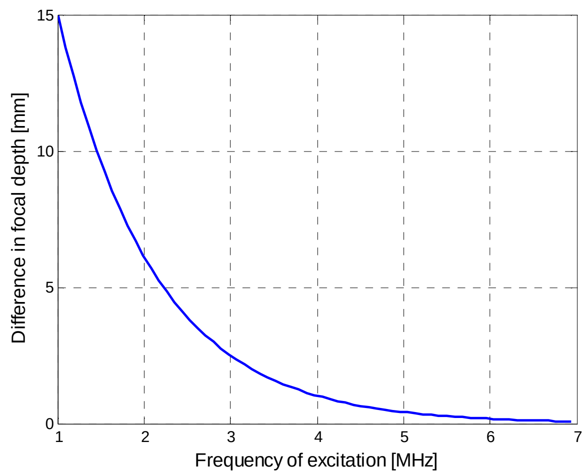


Figure 8 : Difference in focal depth between the fundamental and second harmonic as function of the frequency.

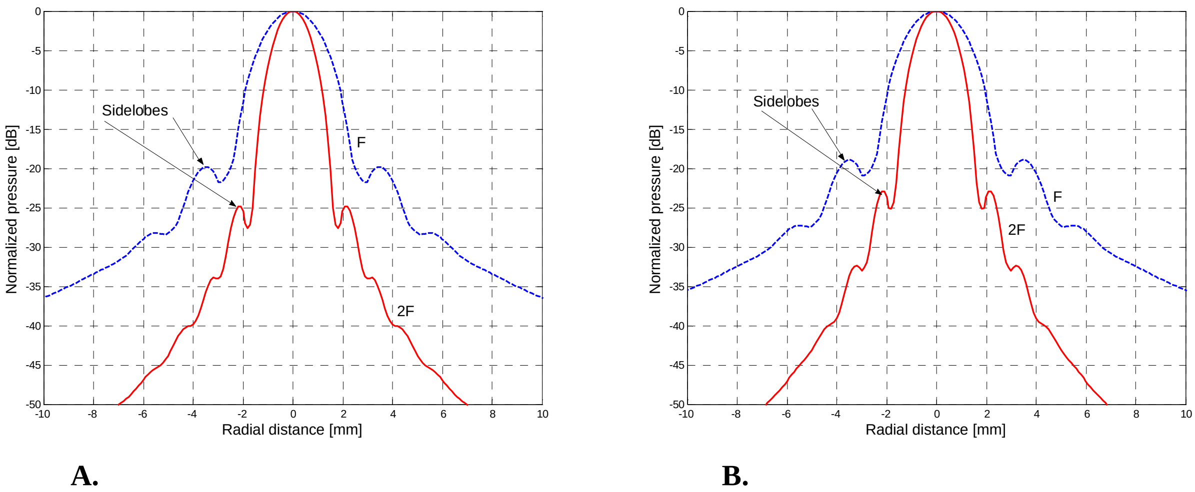


Figure 9: Radial field. **A.** Low pressure (50 KPa). Dashed line: fundamental field; solid line: 2^{nd} harmonic field. **B.** High pressure (500 KPa). Dashed line: fundamental field; solid line: 2^{nd} harmonic field.

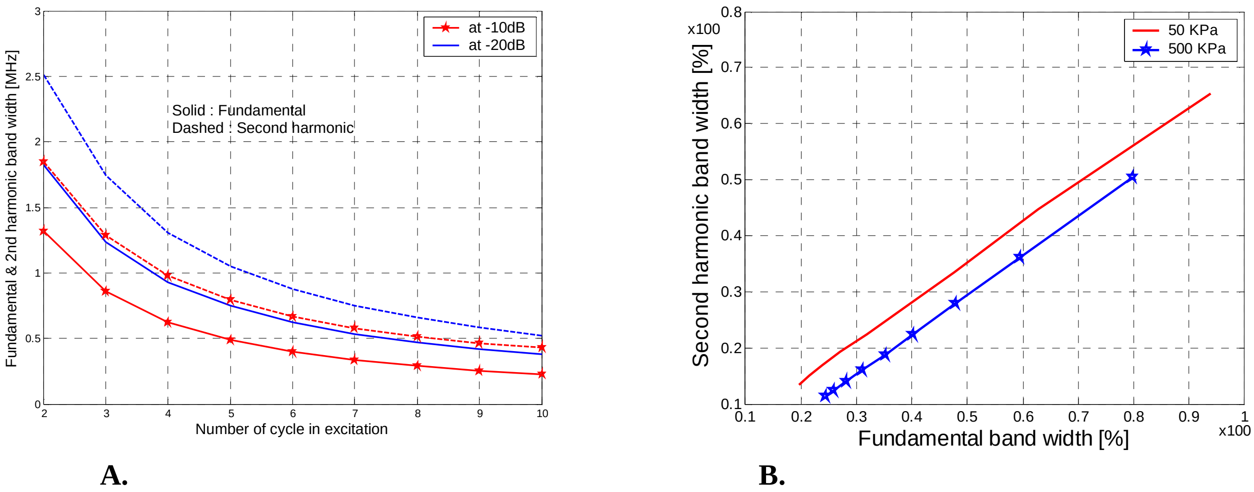


Figure 10: Bandwidth. **A.** Bandwidth as a function of number of cycle in transmit pulse. **B.** Fundamental and second harmonic bandwidth in percentage.

TABLE II
OVERLAPPING BETWEEN FUNDAMENTAL AND SECOND HARMONIC COMPONENT BANDS CATHES AT -40dB.

Transmit pulse [Kpa]	Frequency [MHz]	Number of cycle in excitation	Overlap [MHz]
50	2	1	> 2
		1,5	1,176
		2	0,3654
		2,5	N. O
	3	1	> 3
		1,5	2,0146
		2	0,8626
		2,5	0,1075
		3	N. O
500	2	1	> 2
		1,5	1,4923
		2	0,5058
		2,5	N. O
	3	1	> 3
		1,5	2,3196
		2	0,5964
		2,5	N. O

Table II shows the calculated values of frequency overlapping between fundamental and second harmonic components at two different pressures and frequencies (N. O means no overlapping). Bands were calculated at -40dB to show the overlapping and, in order to acquire the maximum of information contained in the received second harmonic signal as illustrated in figure 11. For a given pressure and frequency, the overlapping decrease if the excitation duration is increased. Overlapping is significant for small durations because the frequency band is inversely proportional with duration thus, we return again to the compromise but in this case between overlapping and axial resolution. According to table II, we can say that the overlapping increases for few hundreds of kilohertz when the excitation pressure is increased by a factor of 10.

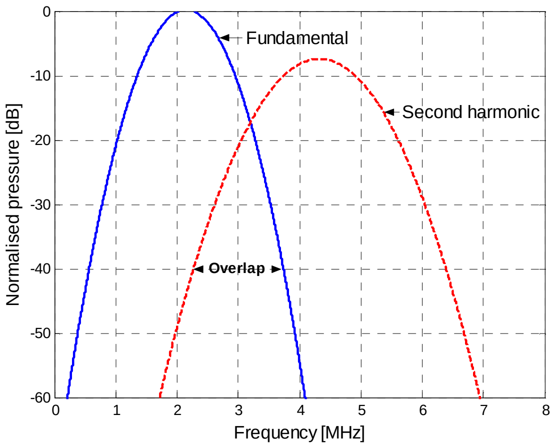


Figure 11: Overlapping illustration. Case of an excitation of 500KPa, 2MHz frequency and 1,5 cycles.

V. CONCLUSION

The quality of the ultrasound image depends strongly on the axial and radial field profiles characteristics. Moreover the level and the frequency bandwidth of transmitted acoustic pressure are important. Restriction of the receive bandwidth degrades the resolution of the resulting image, thus framing again a fundamental compromise in harmonic imaging between contrast and resolution. Also, in a spectrum of a same ultrasound wave the second harmonic band is broad compared to the fundamental one. To receive correctly and without overlap the pure nonlinear component of the second harmonic, a novel method differs from the conventional filtering approaches must be studied (filters are never ideal).

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