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Improvement of hydrophone measurements on diagnostic ultrasound machines using broadband complex-valued calibration data

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Abstract. Non-ideal hydrophone frequency responses may impede correct acoustic output measurements on medical ultrasound equipment, in particular when high frequency or nonlinearly distorted waveforms have to be detected. It is shown that correct pressure waveforms $p(t)$ and the associated standard pulse parameters such as positive peak pressure $p_+$ and rarefractional peak pressure $p_-$ can be obtained by impulse deconvolution if the non-ideal frequency response of the hydrophone $M(f)$ is provided with high frequency resolution both in amplitude and phase in a broad frequency range. The complex-valued calibration data required can be obtained by a secondary hydrophone calibration technique which was recently developed and uses broadband, nonlinearly distorted, focused ultrasound pulses and an optical multilayer hydrophone as reference. The results obtained for a membrane and a needle-type hydrophone are applied to improve exposure measurements on a commercial diagnostic ultrasound machine. This is shown for different operation modes and parameter settings of the diagnostic machine by comparison of the pressure pulse waveforms and pulse parameters obtained by the commonly applied evaluation method using the voltage-to-pressure transfer factor at the acoustic working frequency $f_{awf}$ with those obtained by pulse deconvolution using the complete broadband complex-valued transfer function $M(f)$.

1. Introduction

The acoustic output of diagnostic ultrasound machines is characterized by standard parameters and indices to allow potential risks for the patient to be estimated. The basis of such characterizations is provided by measurements of the ultrasonic pressure fields emitted in water using calibrated hydrophones. International exposure guidelines state requirements and recommendations for the hydrophone bandwidths which should extend up to several (up to eight) times the acoustic working frequency $f_{awf}$ of the pulses to be measured to account for the presence of harmonics due to nonlinear sound propagation [1],[2]. Since commonly available piezoelectric hydrophones can hardly meet these frequency response flatness demands, an adequate correction procedure may be an appropriate alternative.

It is shown that correct pressure waveforms $p(t)$ and the associated standard pulse parameters such as positive peak pressure $p_+$ and rarefractional peak pressure $p_-$ can be obtained by impulse deconvolution if the non-ideal frequency response of the hydrophone $M(f)$ is provided with high frequency resolution both in amplitude and phase in a broad frequency range. The complex-valued
calibration data required can be obtained by substitution calibration using time-delay spectrometry (TDS) [3] or by a secondary hydrophone calibration technique which was recently developed and uses broadband, nonlinearly distorted, focused ultrasound pulses and an optical multilayer hydrophone as reference [4]. The calibration results obtained for membrane and needle-type hydrophones are applied to significantly improve exposure measurements on a commercial diagnostic ultrasound machine. This is shown for various operation modes and parameter settings of the diagnostic machine by comparison of the pulse parameters obtained by the commonly applied evaluation method using the voltage-to-pressure transfer factor at the acoustic working frequency $M(f_{awf})$ with those obtained by pulse deconvolution using the complete broadband complex-valued transfer function $M(f)$.

2. Amplitude and phase calibration of hydrophones

A simple method for amplitude and phase calibration that has previously been used to determine the complex frequency responses of fiber-optic hydrophones [5] is applied to the calibration of piezoelectric ultrasonic hydrophones. An optical multilayer hydrophone [6] is used as a reference receiver in a secondary calibration technique. Due to its flat constant amplitude frequency response in a very broad frequency range from 1 to 75 MHz, the system is expected to provide also a flat phase response. The calibration method is based on the measurement of broadband nonlinearly distorted pulses of a focusing source transducer (Karl Deutsch GmbH, acoustic focusing, frequency range: 3-12 MHz, diameter: 12 mm, nominal focal length: 50 mm) driven by a pulse generator with the reference and, successively, with the hydrophone to be calibrated using the same excitation conditions (figure 1). The frequency spectra of the pulses are calculated by numerical Fourier transformation of the time waveforms. Due to the short excitation voltage pulse and the nonlinear propagation of the sound wave producing higher harmonics, the broad frequency range from 1 up to 70 MHz is covered using this transducer-pulse generator combination. The complex-valued frequency response of the hydrophone under test is obtained by division of the respective frequency spectra with high frequency resolution in a broad frequency range. Due to the dimension and structure of the sound field used, the calibration method described is restricted to hydrophones with 0.2 mm diameter or smaller [4]. The phase data for larger diameter hydrophones may be determined in a form appropriate for impulse deconvolution applications by TDS substitution calibration using the optical multilayer hydrophone as the primary phase standard and a small diameter hydrophone for transfer.

![Figure 1. Hydrophone substitution calibration procedure; I. measurement of a broadband focused ultrasound pulse with the optical reference hydrophone, II. measurement with the hydrophone to be calibrated.](image)

The calibration result obtained for a bilaminar membrane hydrophone with a nominal diameter of 0.2 mm and a PVDF thickness of $2 \times 15 \mu m$ is depicted in figure 2 (left). The amplitude response increases monotonously up to the thickness mode resonance at $\sim 31$ MHz and decreases at higher frequencies. Very good agreement is found with results of a primary interferometric calibration [7] at
discrete frequencies in the range from 1 to 40 MHz. The second calibration example is for a needle-type PVDF hydrophone with a nominal diameter of the sensing element of 0.2 mm, an outer diameter of the needle of 0.5 mm, and a foil thickness of 9 µm. This sample shows very strong variations in the amplitude response in the frequency range from 1 to 20 MHz and the thickness mode resonance peak can be observed at \( \sim 38 \) MHz (figure 2, right). However, this hydrophone is expected to provide limited results for exposure measurements, but here it confirms the high frequency resolution and broadband capabilities of the calibration method investigated. Figure 2 also shows the phase responses of both the membrane and the needle-type hydrophone. For the membrane hydrophone the phase response is flat up to \( \sim 22 \) MHz and shows some variation at higher frequencies. Similar to the respective amplitude response variations, the phase response for the needle-type hydrophone shows strong variations in the frequency range from 1 to 20 MHz and smoother changes at higher frequencies.

![Figure 2. Complex-valued frequency responses of a PVDF bilaminar membrane hydrophone with 15 µm layer thickness (left) and of a PVDF needle-type hydrophone with 9 µm layer thickness (right) obtained by broadband pulse calibration (uncertainties displayed relate to 95% confidence level); comparison of membrane hydrophone amplitude response with results from primary interferometric calibration (left).](image)

3. Hydrophone measurements using pulse deconvolution

Exemplar exposure measurements on a commercial diagnostic ultrasound machine were performed using both hydrophones calibrated. Pulse measurements were made in degassed water at distances \( z = 30 \) mm and \( z = 100 \) mm from the linear array transducer for two different non-scanning beam modes to cover both very short broadband pulses (M-mode) and more narrow-band bursts (pulsedoppler mode). For each measurement the focus and the flow measurement window (for pulsedoppler mode) were set to the distance \( z \), and the hydrophone was adjusted to the lateral position with maximum pulse integral calculated from the voltage-time signal. Simultaneous B-mode imaging was deactivated during the pulse measurements. The analog bandwidth of the oscilloscope was set to 250 MHz, and 2500 data points spanning \( T = 10 \) µs were acquired for each pulse. The time and frequency sampling increments were the same as in the hydrophone calibration data sets. Pressure-time waveforms were obtained from the voltage data by both methods for comparison: A) common conversion using the factor \( M(f_{aw}) \), where \( f_{aw} \) was determined as the mean value of the two -3dB points in the impulse amplitude spectrum already corrected for the hydrophone frequency response, and B) deconvolution of the voltage-time signals \( u(t) \) with the hydrophone impulse response.
\( m(t) = \mathcal{F}^{-1}(M(f)) \) (\( \mathcal{F} \): Fourier transform) performed in the frequency domain using the complex-valued broadband transfer function \( M(f) \):

\[ p(t) = \mathcal{F}^{-1}(\mathcal{F}(u(t))/M(f)). \]  

(1)

An example of the pressure waveforms for M-mode operation measured with the membrane hydrophone at \( z = 30 \) mm is shown in figure 3 where a typical property of the membrane hydrophone can be observed. Due to the transfer function increasing with increasing frequency up to the thickness mode resonance, the positive peak pressure value \( p_+ = \max(p(t)) \) is strongly overestimated and the rarefactional peak pressure \( p_- = \min(p(t)) \) slightly underestimated when using conversion method A) in comparison with the results obtained by conversion method B). Here the oscillations in the decreasing edges of the positive voltage peaks are corrected very well, and a much more reasonable pressure waveform is obtained because the weighting effect of the hydrophone frequency response is numerically compensated. Similar effects were observed with the membrane hydrophone for a larger distance and also for the pulse-Doppler mode at two different possible working frequencies \( f_{aw} \). The results for \( p_+ \), \( p_- \), and the pulse intensity integral:

\[ PII = \int_{t_1}^{t_2} \rho c p(t)^2 \, dt, \]

with \( \rho \): density, \( c \): sound velocity in water, and the complete pressure-time waveform \( p(t) \neq 0 \) lying in the time interval \([t_1, t_2]\), are listed in table 1 for the different parameter settings. In these exemplar measurements, \( p_+ \) is overestimated by up to \(-50\%\), \( p_- \) is underestimated by up to \(-11\%\), and \( PII \) is overestimated by up to \(-28\%\) when using conversion method A) in comparison with the broadband evaluation method B). For the measurements performed, maximum \( PII \) occurs in pulse-Doppler mode, for the lower frequency setting, and at \( z = 30 \) mm which is close to the fixed elevational focus of the transducer. It should be noted that \( p_+ \) is of less importance than \( p_- \) and \( PII \) for the output characterization of diagnostic ultrasound equipment according to current standards [8].

**Table 1.** Pulse parameters derived from measurements with a membrane hydrophone using A) transfer factor \( M(f_{aw}) \) and B) broadband complex-valued transfer function \( M(f) \) for different settings of a diagnostic ultrasound machine; numbers in percent: deviations of results after conversion method A) from results after B).

| Mode | \( z \) (mm) | \( f_{aw} \) (MHz) | \( M(f_{aw}) \) (mV/MPa) | Method A) \( M(f_{aw}) \) | Method B) \( M(f) \) |
|------|-------------|-------------------|----------------|-----------------|-----------------|
|      |             |                   |                | \( p_+ \) (MPa) | \( p_- \) (MPa) | \( PII \) (J m\(^{-2}\)) | \( p_+ \) (MPa) | \( p_- \) (MPa) | \( PII \) (J m\(^{-2}\)) |
| M    | 30          | 5.5               | 14.37          | 6.27            | 1.76            | 0.562           | 4.19            | 1.93            | 0.438           |
|      |             |                   |                | +50\%           | -9\%            | +28\%           |                  |                  |                  |
| M    | 100         | 4.8               | 14.49          | 1.79            | 0.57            | 0.064           | 1.23            | 0.64            | 0.053           |
|      |             |                   |                | +46\%           | -11\%           | +21\%           |                  |                  |                  |
| pD   | 30          | 5.3               | 14.38          | 3.53            | 1.27            | 1.858           | 2.62            | 1.32            | 1.674           |
|      |             |                   |                | +35\%           | -4\%            | +11\%           |                  |                  |                  |
| pD   | 100         | 5.2               | 14.39          | 1.71            | 0.40            | 0.326           | 1.19            | 0.43            | 0.266           |
|      |             |                   |                | +44\%           | -7\%            | +23\%           |                  |                  |                  |
| pD   | 30          | 6.9               | 14.58          | 4.58            | 1.59            | 1.783           | 3.11            | 1.76            | 1.476           |
|      |             |                   |                | +47\%           | -10\%           | +18\%           |                  |                  |                  |
An example of the pressure-time waveforms for pulse-Doppler mode operation \((z = 30 \text{ mm}, f = 5.3 \text{ MHz})\) obtained by the needle-type hydrophone using both conversion methods A) and B) and comparison with the respective result of the membrane hydrophone measurement using conversion method B) is depicted in figure 4. Conversion method A) leads to very large deviations, since the needle-type hydrophone frequency response shows very strong variations in the frequency range of interest (cf. figure 2) leading to much stronger waveform distortions than with the membrane hydrophone. The pulse parameters obtained with the needle-type hydrophone for the same parameter settings of the diagnostic ultrasound machine as given in table 1 showed an overestimation by up to 150\% for \(p_+\), an overestimation by up to 274\% for \(p_-\), and an overestimation by up to 767\% for \(\text{PII}\) in comparison with the membrane hydrophone results obtained by impulse deconvolution. In general, the results seemed to be less predictable than for the membrane hydrophone due to the much stronger variations of the frequency response. But even for this most non-ideal hydrophone, the results could be improved so as to arrive at more reasonable values using the broadband conversion method B). In this case, the deviations from the respective membrane hydrophone measurements using method B) ranged from -5\% to +22\% for \(p_+\), from +3\% to +30\% for \(p_-\), and from +8\% to +28\% for \(\text{PII}\).

**Figure 3.** M-mode pressure-time waveform produced by a commercial diagnostic ultrasound machine \((z = 30 \text{ mm});\) membrane hydrophone measurement; voltage-to-pressure conversion as commonly applied using \(M(f_{\text{act}})\) (method A)), and using the broadband complex-valued frequency response \(M(f)\) as determined by pulse calibration (method B)).

**Figure 4.** Pulse-Doppler mode pressure-time waveform produced by a commercial diagnostic ultrasound machine \((z = 30 \text{ mm}, f = 5.3 \text{ MHz});\) needle-type hydrophone measurement; voltage-to-pressure conversion using \(M(f_{\text{act}})\) (method A)), and using the broadband complex-valued frequency response \(M(f)\) as determined by pulse calibration (method B)); comparison with membrane hydrophone measurement; voltage-to-pressure conversion method B).
4. Conclusions
A secondary hydrophone calibration technique using an optical multilayer hydrophone as the reference receiver was applied to calibrate both a piezoelectric membrane and a needle-type hydrophone in amplitude and phase. The complex-valued calibration data obtained can be applied to significantly improve hydrophone measurement results. This was demonstrated by exemplar exposure measurements on a typical diagnostic ultrasound machine in two different operation modes and at various parameter settings. The pulse parameters obtained by the evaluation method commonly applied using the voltage-to-pressure transfer factor at the acoustic working frequency \( M(f_{\text{awf}}) \) were compared with those obtained by pulse deconvolution using the broadband complex-valued transfer function \( M(f) \). The membrane hydrophone was shown to systematically overestimate the positive peak pressure \( p_+ \) by up to \( \sim 50\% \), underestimate the rarefractional peak pressure \( p_- \) by up to \( \sim 11\% \), and overestimate the pulse intensity integral \( PII \) by up to \( \sim 28\% \) if no broadband conversion is applied.

Using the most non-ideal needle-type hydrophone all pulse parameters were dramatically overestimated due to the strong variations of the frequency response, but the results could be improved to a large extent using the broadband conversion method.

The broadband voltage-to-pressure conversion method suggested increases the usable bandwidth of real non-ideal hydrophones to a large extent by numerical means. It will not raise the technical expense of exposure measurements very much if the complex-valued transfer function is supplied in an appropriate form, since the calculation of frequency spectra by Fourier transformation should already be implemented for the calculation of the working frequency \( f_{\text{awf}} \) from the pulses measured. Unlike in the present investigation, the broadband voltage-to-pressure conversion should be performed automatically during the exposure measurements by the data acquisition computer program to allow for correct maximum search procedures when scanning the ultrasound fields.

5. References
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