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Pressure Pulse Distortion by Needle and Fiber Optic Hydrophones due to Nonuniform Sensitivity

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Abstract

Needle and fiber optic hydrophones have frequency-dependent sensitivity, which can result in substantial distortion of nonlinear or broadband pressure pulses. A rigid cylinder model for needle and fiber optic hydrophones was used to predict this distortion. The model was compared with measurements of complex sensitivity for a fiber optic hydrophone and 3 needle hydrophones with sensitive element sizes (d) of 100, 200, 400, and 600 μm . Theoretical and experimental sensitivities agreed to within $12 \pm 3\%$ (RMS normalized magnitude ratio) and 8 ± 3 degrees (RMS phase difference) for the four hydrophones over the range from 1 – 10 MHz. The model predicts that distortions in peak positive pressure can exceed 20% when $d/\lambda_0 < 0.5$ and $\text{SI} > 7\%$ and can exceed 40% when $d/\lambda_0 < 0.5$ and $\text{SI} > 14\%$, where λ_0 is the wavelength of the fundamental component and SI (spectral index) is the fraction of power spectral density contained in harmonics. The model predicts that distortions in peak negative pressure can exceed 15% when $d/\lambda_0 < 1$. Measurements of pulse distortion using a 2.25 MHz source and needle hydrophones with $d = 200, 400$ and $600 \mu\text{m}$ agreed with the model to within a few percent on the average for SI values up to 14%. This work 1) identifies conditions for which needle and fiber optic hydrophones produce substantial distortions in acoustic pressure pulse measurements and 2) offers a practical deconvolution method to suppress these distortions.

Keywords

hydrophone; needle; fiber-optic; acoustic output measurement

I. Introduction

Accurate measurement of pressure fields is essential to ensure safety and effectiveness for diagnostic and therapeutic ultrasound devices [1]. Hydrophones are the most common devices for measurement of acoustic pressure fields. Membrane hydrophones are often preferred because they tend to exhibit less pressure pulse distortion than needle and fiber-optic hydrophones due to their relatively uniform (frequency-independent) sensitivity. However, membrane hydrophones are bulky and therefore impractical for many applications. In addition, membrane hydrophones, with their big planar surfaces, can create reflections that interfere with measurements.

Needle hydrophones are often more economical than membrane hydrophones. Needle and fiber-optic hydrophones are compact and can be inserted into small spaces inaccessible to

membrane hydrophones. Compactness makes needle hydrophones preferable for *in vitro* testing of transcranial systems [2–4]. In addition, some needle [5] and fiber-optic [6–14] hydrophones are more robust for high-intensity therapeutic ultrasound (HITU) measurements than most membrane hydrophones, although robust membrane hydrophone designs also have been proposed [15, 16]. Fiber-optic hydrophones have been used extensively for calibration of HITU [17–23] and lithotripsy [24] systems. Some fiber optic hydrophones have exceptional spatial resolution on the order of 100 μm or less, far superior to most membrane hydrophones. One challenge with needle and fiber optic hydrophones is that they exhibit non-uniform (frequency-dependent) sensitivity at low frequencies. However, pressure pulse distortion due to nonuniform sensitivity can be suppressed in post processing by applying deconvolution to hydrophone measurements [25–30].

Many needle and fiber-optic hydrophones may be modeled as rigid, semi-infinite rods. This model is straightforward for needle hydrophones. This model also applies to many fiber-optic hydrophones for the following reasons. A common fiber-optic hydrophone design involves propagating an optical beam along a fiber core and measuring the optical reflection from the fiber tip / fluid interface. Pressure variations in the fluid induce density variations near the fiber tip / fluid interface. These density variations result in variations in the fluid index of refraction, which result in variations in the reflection coefficient encountered by the optical beam travelling down the fiber core and reflecting back from the tip. Since the fiber-optic hydrophone core typically has far higher characteristic impedance than the surrounding fluid, it may be accurately modeled as a rigid disk [31]. This model applies to many fiber-optic hydrophones, such as those commonly used in HITU [17, 18, 21–23] and lithotripsy [24]. Other fiber-optic hydrophones, such as those based on multi-layer [32], distributed Bragg reflector [33], Fabry-Perot interferometer [34–36] and other [37] designs, have a more complex response than the rigid, semi-infinite rods model predicts.

Jones studied the frequency response of a rigid, semi-infinite rod to a normally-incident plane wave and showed that the spatial-average (over the rod surface) pressure amplitude has a maximum response at $f_{\text{max}} = 1.2 c / (\pi a)$, where f = frequency, c = sound speed and a = cylinder radius, that is approximately double the response in the limit $f \ll f_{\text{max}}$ [38]. An equivalent approach that uses vibrating planar piston theory [31] is discussed in Sec. II. The model does not simply average the free (hydrophone absent) field over a surface corresponding to the hydrophone sensitive element but also accounts for the fact that the presence of the hydrophone perturbs the field by imposing a boundary condition of zero normal component of particle velocity on the hydrophone sensitive element surface (as a consequence of the assumption of rigidity). This frequency-dependent response can introduce substantial pressure pulse distortion for broadband signals measured with needle and fiber-optic hydrophones. In addition, Jones obtained the phase response, which is also frequency-dependent and therefore contributes to pulse distortion. Measured frequency responses for typical commercial needle and fiber-optic hydrophones are consistent with this model at low frequencies, as will be shown in this paper. Some hydrophones incorporate features such as pre-amplification that cause them to deviate from the model. However, if the frequency dependence of these features is known, then the model can be adapted accordingly.

Figure 1 illustrates a potential problem of nonuniform response in the context of medical ultrasound signals. The top panel shows the spectrum of a nonlinear signal (blue) and the hydrophone frequency-dependent sensitivity (red). The bottom panel shows an unfiltered waveform (blue) and a waveform distorted by the hydrophone (red dash). The hydrophone boosts harmonics relative to the fundamental, resulting in a waveform with exaggerated high-frequency content, which is manifested as sharper, taller, compressional peaks.

In this paper, the model is validated with measurements of complex sensitivity performed using time delay spectrometry (TDS) for a fiber-optic hydrophone and 3 needle hydrophones with sensitive element sizes (d) of 100, 200, 400, and 600 μm . A simulation based on the model is then validated with experimental measurements and used to predict distortions in measurements of peak compressional pressure (p_+) peak rarefactional pressure (p_-), and pulse intensity integral (pii) due to non-uniform sensitivity over a wide range of experimental parameters (sensitive element size, fundamental frequency, bandwidth, and nonlinearity). The likelihood for mechanical bioeffects is thought to be related to p_- while the likelihood for thermal bioeffects is thought to be related to PII [39]. In histotripsy applications, in which mechanical fractionation of the tissue is accomplished using repetitive high intensity ultrasound pulses, p_+ is important for distinguishing regimes of cavitation histotripsy and boiling histotripsy [20].

There are many sources of uncertainty in hydrophone measurements, including calibration uncertainty, spatial averaging, variation of hydrophone sensitivity with temperature, and hydrophone sensitivity drift over time. These affect all hydrophones, not just needle and fiber-optic hydrophones. The goal of the present study was to isolate the effects of nonuniform sensitivity of needle and fiber-optic hydrophones on estimates of p_+ , p_- , and pii .

II. Model

A. Frequency-dependent Hydrophone Response

Needle and fiber optic response was modeled by spatially averaging acoustic pressure over the end of a rigid, semi-infinite, cylindrical rod [31, 38, 40]. One approach involves a solution for scattering of a plane harmonic scalar wave by a semi-infinite circular cylindrical rod of diameter $2a$ when the boundary condition is $u=0$ or $u/\nu=0$, where u is the scalar field and ν is the normal to the rod [38]. A related approach models a fiber optic hydrophone as a vibrating planar piston (fiber core) with radius a_2 in an infinite rigid baffle (fiber cladding) with inner radius a_2 and outer radius a_1 [31]. The two approaches should give similar results when $a = a_1 = a_2$. The transfer function for the fiber optic hydrophone model is [31]

$$H(f) = 2 - \frac{1}{\pi a_2^2} \int_0^{2\pi} \int_0^{a_2^2} e^{-j2\pi f b(r, \theta)/c} r dr d\theta$$

where f = frequency, c = sound speed, and

$$b(r, \theta) = \sqrt{r^2 \cos^2 \theta + \left(a_1^2 - r^2\right)} - r \cos \theta.$$

In the present paper, both models [31, 38] were tested by comparison with our experimental measurements of hydrophone sensitivity based on time delay spectrometry (TDS). Pulse distortions for a wide variety of pulse shapes were estimated by subjecting simulated nonlinear diagnostic pulses to theoretical frequency-dependent complex sensitivities. Another model for needle hydrophone sensitivity accounts not only for diffraction but also for multiple reflections within the layered structure and includes properties of polyvinylidene fluoride (PVDF) such as attenuation [41]. This model, however, requires detailed knowledge of the hydrophone structural characteristics, which is not always available.

B. Nonlinear Pressure Pulse

The pressure pulses used in simulations were based on the model of Ayme and Carstensen for nonlinear medical ultrasound signals [42]. Their formula represents the signal as a sum of n sinusoids, each with frequency $n f_c$ (where f_c is the center frequency for the fundamental component) and a $\pi/4$ phase shift, with amplitudes proportional to $1/n$. Zeqiri and Bond [43] modified this formula by changing the harmonic amplitude, $1/n$, to the form derived by Blackstock for plane periodic waves in a semi-infinite fluid, $B_n(\sigma)$ [44]. With this modification, the acoustic pressure pulse, $p_i(t)$, has a continuously adjustable shock parameter, σ , to modulate the level of nonlinearity.

$$p_i(t) = m(t) \left[\sum_{n=1}^{100} B_n(\sigma) \sin\left(2\pi n f_c t + \frac{\pi}{4}\right) \right]$$

where t is time and

$$B_n(\sigma) = \frac{2}{n\pi} V_b + \frac{2}{n\pi\sigma} \int_{\Phi_{\min}}^{\pi} \cos n(\Phi - \sigma \sin \Phi) d\Phi$$

For $0 < \sigma < 1$, the shock amplitude $V_b = 0$, $\Phi_{\min} = 0$, and $B_n(\sigma) = (2/n\sigma) J_n(n\sigma)$ where J_n is the ordinary Bessel function of order n . For $\sigma > 1$, $\Phi_{\min} = \sigma V_b$ and V_b is the first nonzero root to the transcendental equation $V_b = \sin(\sigma V_b)$ [43]. As pointed out by Zeqiri and Bond, Bacon has proposed a method for estimating σ from experimentally measured quantities [45, 46].

For the present study, an envelope function, $m(t)$, was included to simulate the pulsed nature of medical ultrasound signals. $m(t)$ had three phases: a rising portion proportional to $[1 - \exp(-t/t_1)]$ followed by a constant middle portion followed by a decaying portion proportional to $\exp(-t/t_2)$. The durations of the three phases of the envelope could be adjusted in order to control the fractional bandwidth of the simulated signals. One hundred harmonics were summed in order to result in nonlinear properties observed in medical

ultrasound: compressional peaking, rarefactional rounding, and shock front formation. Signals were sampled at 2 GHz using Matlab (Natick, MA).

The degree of nonlinearity in the signal was quantified using the “spectral index,” SI , originally proposed by Duck [47].

$$SI = \frac{\int_{f_a}^{\infty} P(f)df}{\int_0^{\infty} P(f)df}$$

where $P(f)$ is the power spectrum and f_a is a spectral boundary, chosen here to be equal to 1.5 times the center frequency of the fundamental spectral component in order to fall half way between the fundamental and the first harmonic. Another alternative would be to set f_a to the frequency at which the spectral value first falls to a level 6 dB below its value at the center frequency [47], but this approach can be problematic for highly nonlinear, very wideband signals for which the -6 dB point can lie far outside the fundamental lobe. The spectral index reflects the fraction of power contained in frequencies above the fundamental.

Figure 2 shows the 16 signals tested in the simulation. Nonlinearity (SI) increases from left to right, and fractional bandwidth (-6 dB) of the fundamental lobe (BW_0) increases from top to bottom. The center frequencies of the fundamental spectral lobes were chosen to avoid a simple monotonic relationship with either SI or BW_0 . It can be seen that as nonlinearity (SI) increases, compressional peaks become sharper while rarefactional peaks become more rounded. Figure 3 shows the corresponding spectra.

Although the relative amplitudes of the harmonics in the simulations were based on theory for plane waves [44] and the constant phase shift of $\pi/4$ for all harmonics in the simulations [42] may seem like an oversimplification, Figure 2 suggests that the signal model proposed here produces signals that are consistent with signals produced by focused transducers encountered in medical ultrasound [48].

Several authors have explored many metrics for quantification of nonlinearity [47, 49–52]. Some metrics are convenient because they may be computed directly from simulated or measured signals, without knowledge of the source transducer or beam properties. These include the following. As mentioned above, the *spectral index*, SI , is the fraction of power contained in frequencies above the fundamental. The *second harmonic ratio* is the ratio of the amplitudes of the second to first harmonics in the spectrum. The *asymmetric ratio*, p_+ / p_- , is the ratio of the peak compressional pressure to the peak rarefactional pressure. The *Ostrovskii / Sutin propagation parameter*, σ_s , is the ratio of the difference between p_+ and p_- to the sum of p_+ and p_- . The *acoustic pulse crest factor* is defined as p_+ / p_{rms} , where p_{rms} is the root-mean-square value of the pressure. Other proposed metrics such as the *acoustic propagation parameter*, σ_m , and the *local distortion parameter*, σ_q , have merits but require empirical measurements of system properties such as focal gain and local area factor (square root of the ratio of the source aperture to the beam area) [52].

The relationships among nonlinearity metrics depend on the detailed structure of the acoustic beam, including relative amplitudes and phases of the harmonics in the spectrum. For example, the time domain metrics are thought to be particularly sensitive to diffraction (that is, interference among harmonics) [52]. In addition, time-domain metrics would be expected to exhibit more dependence on hydrophone phase response (which is often unknown) than SI . Factors such as these lead to the concern that predictions of the model could depend on the particular choice of nonlinearity metric (in this case, SI). In order to investigate this possibility, SI , second harmonic ratio, p_+ / p_- , σ_s , and p_+/p_{rms} were computed and compared for all simulated and experimental signals.

III. Experimental Methods

A. Sensitivity Measurements

Theoretical models were compared with measurements of complex sensitivities for four hydrophones without preamplifiers (see Table 1). The complex sensitivity of the fiber optic hydrophone was obtained from the Fourier transform of the impulse response provided by the manufacturer. Complex sensitivities for the three needle hydrophones were measured using a digital time-delay spectrometry (TDS) system previously described [27, 53]. (There are many methods for measuring sensitivity magnitude but fewer methods for measuring sensitivity phase [53–57], with some TDS methods extending to frequencies as high as 40 MHz [27, 56] and a nonlinear method providing phase measurements at harmonic frequencies up to 100 MHz [55].) Some methods have shown that common hydrophone measurement systems can be accurately modeled as minimum phase systems [27, 53, 56].) A PVDF membrane hydrophone (HMB-200, Onda Corp., Sunnyvale, CA) was used for a reference in substitution experiments. The complex sensitivity of the reference membrane hydrophone was measured from 1 to 40 MHz by an independent laboratory (National Physical Laboratory, U.K.). TDS sensitivity measurements were performed using four different broadband source transducers (approximate center frequencies: 2 MHz, 5 MHz, 30 MHz, and 50 MHz) in order to obtain hydrophone magnitude and phase response over four frequencies bands (roughly 1–3 MHz, 3–6 MHz, 6–25 MHz, 25–40 MHz) that collectively spanned the range from 1 to 40 MHz. The transducers included (in order of ascending center frequency) 1) a custom bi-concave transducer [58], 2) a 1.27 cm diameter, 5 MHz transducer (IS0504HR, Valpey Fisher, Hopkinton, MA), 3) a 0.635 cm, 30 MHz transducer (ZF3001, Panametrics, Waltham, MA), and 4) a 0.635 cm, 50 MHz transducer (P150-2-R2.00, Olympus, Waltham, MA). For more detail, see [27].

Model functions were fit to experimental data by minimizing the mean square difference between the model and the experimental sensitivity magnitude over a range of frequencies. The adjustable parameters in the fit were a multiplicative constant and sensitive element size d . The upper limit of the frequency range of the fit could be limited by two factors. First, one model only provided predictions up to a maximum frequency corresponding to $ka = 10$ [38] where $k=2\pi/\lambda$ and $\lambda=f/c$. Second, as will be seen in the Results section, two needle hydrophones had design complexities that caused their frequency-dependent sensitivities to deviate from the simple cylinder model at frequencies much higher than the maximum

response frequency, $f_{\max} = 2.4c/(\pi d)$. In these cases, the upper limit was set near the frequency at which the deviations began.

B. Pulse Distortion Measurements

The effects of nonuniform sensitivity on measurements of p_+ and p_- were investigated by comparing measurements of ultrasound fields in a water tank performed using the HMB-200 membrane hydrophone (with a relatively flat frequency response from 1 – 20 MHz) with measurements performed using the needle and fiber optic hydrophones. The source was a KB K12401 Aerotech (Lewistown, PA) single-element transducer with center frequency = 2.25 MHz, diameter = 1.91 cm, and focal distance = 10.2 cm. A Tektronix (Beaverton, OR) AFG 3102 dual channel arbitrary function generator and an Amplifier Research (Souderton, PA) Model 150A 100B 150W power amplifier were used to generate the electric signal that was fed into the source transducer. Voltage outputs from hydrophones were digitized at 1 GHz using a Tektronix 3012C Digital Phosphor Oscilloscope. The transducer was driven by 8-cycle tone bursts. This burst duration was much shorter than durations for HITU pulses but was long enough to achieve an approximately continuous wave condition in the center of the pulse, allowing measurements of pressure distortion relevant for HITU. In addition to driving the transducer at its resonance frequency of 2.25 MHz, measurements were performed at 1.75 MHz and 3 MHz in order to investigate different values for d/λ_0 where λ_0 is the wavelength corresponding to the fundamental frequency, f_c . For each driving frequency and each hydrophone, measurements were performed at 6 different driving levels in order to span a range of signal nonlinearity (SI) but avoiding levels that could produce sufficiently high pressures to damage the hydrophones. The function generator peak-to-peak voltages were 31, 62, 93, 124, 156, 190 mV (1.75 MHz), and 20, 40, 60, 80, 100, 120 mV (2.25 and 3 MHz).

Voltage output signals were analyzed off-line using Matlab. Signals were low-pass filtered with a Gaussian global low-pass filter $\exp(-f^2/2\Sigma^2)$ with $\Sigma = 40$ MHz. Signals were normalized to their spectral magnitude at the driving frequency. This eliminated pressure discrepancies due to uncertainties in sensitivity magnitude, which has been reported to be on the order of ten percent [59], so that pressure discrepancies exclusively due to frequency-dependent sensitivity could be elucidated.

IV. Results

A. Hydrophone Sensitivity Measurements

Figure 4 shows experimental measurements and theoretical fits for sensitivities of the four hydrophones. Since the complex sensitivity of the fiber optic hydrophone was obtained from the Fourier transform of the impulse response provided by the manufacturer, the experimental values in the top row of Figure 4 go all the way down to zero frequency. The experimental values for the three needle hydrophones do not include values below 1 MHz because of experimental signal-to-noise ratio limitations. Similar high-pass filter behavior for low frequencies ($f < f_{\max}$) has been measured for needle [5, 26, 27, 29, 40, 41, 60] and fiber optic [8, 14] hydrophones previously.

The two models, “Model 1” [38] and “Model 2” [31], gave nearly identical results. Theoretical and experimental sensitivities agreed to within $12 \pm 3\%$ (mean $\pm$ standard deviation of RMS magnitude difference) and 8 ± 3 degrees (mean $\pm$ standard deviation of RMS phase difference) for the four hydrophones. The effective sensitive element diameter was an adjustable parameter in the fits and is shown in Table I. The effective sensitive element diameter was always within 28% of the nominal value. The maximum sensitivity is expected at $f_{\max} = 2.4c/(\pi d)$ [38]. The optimal estimates of sensitive element size based on the minimum least-squares model fits to experimental sensitivity gave values of d of 128, 176, 308, and 584 μm for the four hydrophones, which corresponds to $f_{\max} = 8.8, 6.4, 3.7,$ and 1.9 MHz, consistent with Fig. 4. Needle #3 has a peak between 1 and 2 MHz, which is consistent with simulation based on the multilayered structure model [41]. In order to capture the high-pass filter behavior for rigid cylindrical hydrophones with sensitive elements larger than 600 μm , sensitivity should be measured at frequencies below 1 MHz [61].

As can be seen in Fig. 1a, the cylinder model predicts an approximate doubling of sensitivity magnitude between very low frequencies ($f \ll f_{\max}$) and the maximum response frequency, $f_{\max}$, followed by relatively constant sensitivity magnitude for frequencies above $f_{\max}$. Fig. 4 shows that Needle #1 and Needle #2 are consistent with the rigid cylinder model for frequencies up to 10 MHz, but show diminished sensitivity for frequencies above 10 MHz compared with the nearly-constant sensitivity predicted by the cylinder model. This disparity is presumably due to design features that deviate from the simple cylinder model. In these cases, the upper limits of the displayed model values were set near 10 MHz. As can be seen in Fig. 3, most of the spectral energy in the test signals was contained in frequencies below 10 MHz.

Figure 5 shows relationships, obtained from simulation and experiments, between the spectral index, SI , and four other metrics to describe nonlinearity. All metrics appear to obey simple monotonic relationships with SI . Although relationships among nonlinearity metrics in principle depend on the structure of an acoustic beam, including relative amplitudes and phases of the harmonics in the spectrum, the simulated and experimental values seem very consistent with each other, supporting the relevance of the simulation signal model. Although the relative amplitudes of the harmonics in the simulations were based on theory for plane waves [44], the signal model used here seems consistent with experimental data acquired using a focused transducer [48]. Although the constant phase shift of $\pi/4$ for all harmonics in the simulations [42] may seem like an oversimplification, the signal model shows nonlinearity metrics (both time-domain and frequency-domain) similar to experimentally measured values.

B. Acoustic Output Measurements

Figure 6 shows distortions in simulated peak compressional pressure before and after application of the model sensitivity function. The results are shown as functions of d/λ_0 and SI where λ_0 is the wavelength corresponding to the fundamental frequency, f_c . Experimental needle hydrophone data points are plotted in the panel in Figure 6 with SI value (3%, 7%, 14%, or 22%) closest to that obtained from companion membrane

hydrophone measurements, provided the experimental SI value was less than 3% above and no more than 1.5% below the simulated value (3%, 7%, 14%, or 22%). It can be seen that the distortion in peak compressional pressure caused by the non-uniform hydrophone sensitivity is primarily due to two factors: d/λ_0 and SI . Center frequency and fractional bandwidths of the signals were far less influential in distorting peak compressional pressure. Experimental measurements were consistent with the simulations. Data were not acquired at $SI = 22\%$ because the required intensities were feared to potentially damage hydrophones.

Figure 7 shows distortions in simulated peak rarefactional pressure before and after application of the model sensitivity function. As with peak compressional pressure, d/λ_0 is an important factor in determining distortion of peak rarefactional pressure. In contrast with peak compressional pressure, fractional bandwidth, BW_0 , is a more important determinant of distortion of peak rarefactional pressure than SI . Therefore, Fig. 7 is drawn with four panels corresponding to different values for BW_0 instead of SI (as in Fig. 6). Experimental data are not shown in Fig. 7 because they would obscure simulation results but are shown on a zoomed scale in Fig. 8 instead.

Figure 8 shows distortions in simulated peak rarefactional pressure before and after application of the model sensitivity function for $BW_0 = 17\%$, along with experimental data. The biggest discrepancies occur for $0.2 < d/\lambda_0 < 0.6$.

Fig. 9 shows percent changes in simulated pulse intensity integral before and after application of the model sensitivity function. As with peak compressional pressure, the distortion in pulse intensity integral caused by the non-uniform hydrophone sensitivity is primarily due to d/λ_0 and SI . Comparison of Figs. 6 and 9 show similar trends of distortions of p_+ and p_{ii} as functions of SI . This may be explained by the fact that as SI increases, the time-domain waveform becomes increasingly asymmetric with respect to the time axis (see Fig. 2) and therefore p_+ has an increasing relative influence (compared with p_-) on p_{ii} .

V. Discussion

Needle and fiber optic hydrophones are desirable for many applications in medical ultrasound but can distort acoustic pressure pulses. The rigid cylindrical rod model presented here is accurate for predicting responses of the needle and fiber-optic hydrophones investigated for frequencies up to at least $1.5 f_{\max}$. This high-pass filter behavior has been previously shown for needle [5, 26, 27, 29, 40, 41, 60] and fiber optic [8, 14] hydrophones. This work identifies conditions for which nonuniform sensitivity of needle and fiber optic hydrophones results in substantial distortions in acoustic pressure pulse measurements. Knowledge of these conditions can help inform the appropriate choice for a hydrophone for a particular measurement.

Comparison of Figs. 1a and 4 shows that the cylinder model was consistent with experimental sensitivities for all four hydrophones investigated at frequencies up to and beyond the maximum sensitivity magnitude frequency, $f_{\max} = 2.4c/(x d)$. However, two needle hydrophones showed sensitivity magnitude reductions compared to the model for frequencies above 10 MHz. As can be seen from Fig. 3, the test signals used in this

investigation contained relatively little energy above 10 MHz so the differences between theoretical and experimental sensitivities were not consequential for this investigation. However, for signals with substantial fractions of energy above 10 MHz (say, over 25%), a modified model or the empirical sensitivity might improve accuracy of predictions of output pressure pulses. Still, the cylinder model faithfully modelled sensitivities for frequencies up to at least $1.5 f_{max}$ for the needle hydrophones investigated. Moreover, for HITU signals, fiber optic hydrophones are generally preferred due to their superior robustness to high pressures (*e.g.* > 10 MPa) [17–24], and the cylinder model performs very well for frequencies up to 100 MHz for the fiber optic hydrophone investigated, as may be seen in Fig. 10 (maximum relative magnitude error = 23% and maximum phase error = 13 degrees).

Distortions in measurements of peak positive pressure, p_+ , due to nonuniform sensitivity are more sensitive to d/λ_0 and SI than BW_0 . They are highest when $d/\lambda_0 < 0.5$ and increase with increasing SI . Distortions in measurements of peak negative pressure, p_- , due to nonuniform sensitivity tend to be smaller than p_+ and are highest when $d/\lambda_0 < 1$. Figure 11 illustrates why p_+ is much more sensitive to nonlinearity than p_- . When a nonlinear medical ultrasound signal, which typically has sharp compressional peaks and rounded rarefactional segments, is decomposed into compressional and rarefactional components, spectral analysis shows that the nonlinear content of the signal is carried disproportionately by the compressional component.

It is commonly thought that it is desirable to use a hydrophone with sensitive element as small as possible (provided sufficient sensitivity is achieved) in order to minimize spatial averaging artifacts and to allow finely-spaced contiguous measurements in a pressure field scan. However, Figs 6 – 9 show that reductions in sensitive element size (d) can actually increase signal distortion for needle and fiber optic hydrophones.

Common robust fiber-optic hydrophones have sensitive element sizes around $d = 100 \mu\text{m}$ [10, 17, 21, 23, 62]. Highly nonlinear waveforms in combination with values of d/λ_0 near 0.1–0.4 can result in considerable potential distortion in peak compressional pressure (see Fig. 6, lower right panel). Such values are achievable in well-validated, extremely promising approaches including histotripsy (*e.g.*, $d/\lambda_0 \approx 0.1$ at 1.5 MHz [23] and $d/\lambda_0 \approx 0.07$ at 1 MHz [62]) and other HITU applications (*e.g.*, $d/\lambda_0 \approx 0.14$ at 2 MHz [17] and $d/\lambda_0 \approx 0.08$ at 1.2 MHz [21]).

When distortions reach unacceptable levels, they can be suppressed by applying deconvolution of the frequency-dependent hydrophone sensitivity. The value of deconvolution in the context of hydrophone measurements is appreciated in the ultrasound metrology community [17, 19, 21, 25–30, 63, 64] and to some extent outside that community [65, 66]. Generally speaking, in order to perform a deconvolution, the magnitude and the phase of the hydrophone sensitivity must be known over all frequencies present in the signal spectrum. In the event that the hydrophone manufacturer provides only limited sensitivity information (*e.g.* magnitude from 1 – 20 MHz but not phase information), the model presented in this paper provides a method to estimate sensitivity phase and to extrapolate magnitude to unspecified frequencies (*e.g.* 0 – 1 MHz, and above 20 MHz).

Limitations of this work include the following. First, this analysis only considers uncertainties (or pressure discrepancies) due to nonuniform sensitivity. It does not consider discrepancies due to uncertainties in the magnitude of sensitivity, which are on the order of 10 percent [59]. Second, this analysis does not consider spatial averaging effects, which can be a problem for highly focused nonlinear beams.

Previous investigations of pressure pulse distortion by membrane hydrophones due to nonuniform response [67, 68] played an influential role in acoustic output measurement standards [69–71]. The present paper extends these ideas to needle and fiber-optic hydrophones. While low-frequency (*e.g.*, $f < 1$ MHz) response of membrane hydrophones is dominated by capacitance of the hydrophone and associated electronics [70, 72, 73], low-frequency ($f < f_{\max}$) responses of needle and fiber optic hydrophones are dominated by diffraction effects related to the dimensions of the needle or fiber.

VI. Conclusion

Sensitivities for many needle and fiber-optic hydrophones commonly used in medical ultrasound can be accurately predicted using a rigid rod model for frequencies up to $1.5 f_{\max}$ and beyond. Distortions in measurements of peak positive pressure due to nonuniform sensitivity can exceed 20% when $d/x_0 < 0.5$ and $SI > 7\%$ and can exceed 40% when $d/\lambda_0 < 0.5$ and $SI > 14\%$, where d is the sensitive element size, λ_0 is the wavelength of the fundamental component, and SI (spectral index) is the fraction of power spectral density contained in harmonics. Distortions in peak negative pressure due to nonuniform sensitivity can exceed 15% when $d/\lambda_0 < 1$ and are particularly high for very broadband signals. When these experimental circumstances cannot be avoided, deconvolution of hydrophone sensitivity from measurements can be used to suppress these distortions. The rigid rod model provides a method to extrapolate sensitivity magnitude data and to generate sensitivity phase data to facilitate complex deconvolution.

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Keith A. Wear received his B.A. in Applied Physics from the University of California at San Diego in 1980. He received his M.S. and Ph.D. in Applied Physics with a Ph.D. minor in Electrical Engineering from Stanford University in 1982 and 1987. He was a post-doctoral research fellow with the Physics department at Washington University, St. Louis from 1987–1989. He has been a research physicist specializing in biomedical ultrasound at

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Yunbo Liu received B.S. in Applied Mechanics from the Peking University, Beijing, China in 2001 and Ph.D. in Mechanical Engineering from Duke University, Durham, North Carolina in 2006. His research activity was focused on therapeutic ultrasound for cancer treatment. Since 2006 he has been employed as a research engineer by the Center for Devices and Radiological Health of Food and Drug Administration, Silver Spring, MD. His main research area is experimental evaluation of biomedical ultrasound dosimetry and safety.

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Fig. 1.

a) spectrum of a nonlinear signal (blue) and the hydrophone frequency-dependent sensitivity (red). b) unfiltered waveform (blue) and a waveform distorted by the hydrophone (red dash). The hydrophone boosts harmonics relative to the fundamental, resulting in a waveform with reduced rarefactional pressure and exaggerated high-frequency content, which is manifested as sharper, taller, compressional peaks.

Fig. 2.

Signals tested in the simulation. SI = spectral index. BW_0 = fractional bandwidth of the fundamental lobe. The center frequencies of the fundamental lobes are given in the right column of figures.

Fig. 3.

Spectra of signals tested in the simulation. SI = spectral index. BW_0 = fractional bandwidth (-6dB) of the fundamental lobe. The center frequencies of the fundamental lobes are given in the right column of figures.

Fig. 4.

Theoretical and experimental forms for hydrophone sensitivities for the fiber optic hydrophone (top row) and 3 needle hydrophones. The two models are “Model 1” [38] and “Model 2” [31]. The arrows show the maximum response at $ka = 2.4$ or, equivalently, $f = 2.4$ $c / (\pi d)$.

Fig. 5.

Metrics to quantify nonlinearity for the signals studied in this paper, obtained from simulation and experiment. Power law fits to simulated data ($y = A SI^n + B$; solid lines) are shown where A and n were fitting parameters and the intercept value B was forced to the theoretical value for a purely linear signal (0, 1, 0, and $\sqrt{2}$ respectively). Cubic polynomial fits ($y = a_3 SI^3 + a_2 SI^2 + a_1 SI + B$; dashed lines) are shown where a_3 , a_2 , and a_1 were fitting parameters.

Fig. 6.

Percent changes in simulated peak compressional pressure before and after application of the model sensitivity function. Experimental data points correspond to measurements acquired with the Onda HNC-200 (circles), Onda HNA-400 (asterisks), and Danish Institute of Biomedical Engineering (squares) needle hydrophones respectively.

Fig. 7.
Percent changes in simulated peak rarefactional pressure before and after application of the model sensitivity function.

Fig. 8.

Percent changes in simulated peak rarefactional pressure before and after application of the model sensitivity function for $BW_0 = 17\%$. Experimental data points correspond to measurements acquired with the Onda HNC-200 (circles), Onda HNA-400 (asterisks), and Danish Institute of Biomedical Engineering (squares) needle hydrophones respectively.

Figure 9.

Percent changes in simulated pressure pulse intensity integral before and after application of the model sensitivity function.

Fig. 10.

Experimental and theoretical sensitivity for the RP Acoustics FOPH 2000 fiber-optic hydrophone from 0 to 100 MHz. The experimental sensitivity of the fiber-optic hydrophone was obtained from the Fourier transform of the impulse response provided by the manufacturer.

Fig. 11.

A nonlinear tone burst decomposed into compressional and rarefactional components (left). Spectra (right) show that the nonlinearity is carried disproportionately by the compressional component.

Table I

Hydrophones Used For Experimental Measurements

Manufacturer	Model	Type	Sensitive Element Size $d(\mu\text{m})$	
			Nominal	Fit
RP Acoustics (Leutenbach, DE)	FOPH 2000	Fiber Optic	100	128
Onda (Sunnyvale, CA)	HNC-200	Ceramic Needle	200	176
Onda	HNA-400	Ceramic Needle	400	308
Force Technology Inst. (Brøndby, DK).	Custom	PVDF Needle	600	584