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Correction for Frequency-Dependent Hydrophone Response to Nonlinear Pressure Waves Using Complex Deconvolution and Rarefactional Filtering: Application with Fiber Optic Hydrophones

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Abstract

Nonlinear acoustic signals contain significant energy at many harmonic frequencies. For many applications, the sensitivity (frequency response) of a hydrophone will not be uniform over such a broad spectrum. In a continuation of a previous investigation involving deconvolution methodology, deconvolution (implemented in the frequency domain as an inverse filter computed from frequency-dependent hydrophone sensitivity) was investigated for improvement of accuracy and precision of nonlinear acoustic output measurements. Time delay spectrometry (TDS) was used to measure complex sensitivities for six fiber-optic hydrophones. The hydrophones were then used to measure a pressure wave with rich harmonic content. Spectral asymmetry between compressional and rarefactional segments was exploited in order to design filters used in conjunction with deconvolution. Complex deconvolution reduced mean bias (for 6 fiber-optic hydrophones) from 163% to 24% (peak compressional pressure, p_+), 113% to 15% (peak rarefactional pressure, p_-) and 126% to 29% (pulse intensity integral, PII). Complex deconvolution reduced mean coefficient of variation (COV) (for 6 fiber optic hydrophones) from 18% to 11% (p_+), 53% to 11% (p_-), and 20% to 16% (PII). Deconvolution based on sensitivity magnitude or the minimum phase model also resulted in significant reductions in mean bias and COV of acoustic output parameters but was less effective than direct complex deconvolution for p_+ and p_- . Therefore, deconvolution with appropriate filtering facilitates reliable nonlinear acoustic output measurements using hydrophones with frequency-dependent sensitivity.

Keywords

hydrophone; sensitivity; deconvolution; fiber-optic

I. INTRODUCTION

Accurate and precise measurements of acoustic output from medical ultrasound transducers is required to ensure that ultrasound exposure to tissues is restricted to acceptable levels. These measurements are often performed with hydrophones [1]. When the sensitivity (frequency response) of the hydrophone is uniform over the usable bandwidth of the acoustic signal, acoustic pressure amplitude may be accurately estimated using the common approach of taking the ratio of the hydrophone output voltage to the value of the hydrophone sensitivity at the “acoustic working frequency” [2, 3]. However, this traditional approach may become problematic, for example, for acoustic waveforms that have significant energy in harmonic frequencies (which arise from nonlinear propagation) and are therefore affected by the hydrophone sensitivity over an extended band of frequencies.

When hydrophone sensitivity is not sufficiently uniform over the usable bandwidth of the acoustic signal, an improved measurement of the acoustic pressure waveform may be obtained by deconvolving the measured hydrophone waveform with the hydrophone impulse response (the inverse Fourier transform of the sensitivity). The need for deconvolution is increasing in medical ultrasound as source transducers move toward higher frequencies (e.g., 15 MHz and above) with concomitant increases in bandwidths [4, 5]. An International Electrotechnical Commission (IEC) standard states that in the presence of short pulses or significant nonlinear distortion, the ± 3 dB bandwidth of the hydrophone should be at least 8 times the acoustic working frequency in order to keep measurement errors below about 5% [2]. When such a bandwidth is not achievable, the standard recommends that deconvolution be performed if “the uncertainty in the measurement becomes unacceptably large due to limited bandwidth of the hydrophone” [2, 3].

One potentially important application of deconvolution is for improving acoustic output measurements performed with Fabry-Perot interferometric fiber-optic (FPIFO) hydrophones. One FPIFO hydrophone design is relatively new and has many attractive features, including high sensitivity, small sensitive element size (10 μm), the ability to withstand moderately-high intensity therapeutic fields, the ability to be implanted into tissues and phantoms, and the ability to measure temperature as well as pressure [6]. This hydrophone has been demonstrated to have potential as a cavitation detector, particularly in applications where space is limited or during magnetic resonance-guided studies [7]. However, as the originators of this FPIFO device acknowledge, these hydrophones have nonuniform sensitivity, varying by a factor of 5-10 over the range from 1 to 50 MHz (see Ref. [6], Figure 7). An earlier fiber optic hydrophone with Fabry-Perot structure has also been reported to have nonuniform sensitivity, varying by a factor of 7 over the band from 5 to 45 MHz (Ref. [8], Figure 3) [8, 9]. A fiber-optic displacement sensor, which is interrogated by a heterodyne interferometer and is capable of withstanding high intensity therapeutic fields, also exhibits non-uniform frequency response, varying by a factor of 17 over the band from 1 to 100 MHz (see Ref. [10], Figure 2). The originators of this design therefore recommend deconvolution in conjunction with this device [10]. Another fiber-optic design, based on commercially available standard single mode 10/125 μm fiber, however, shows relatively uniform response [5].

The benefits of deconvolution have been demonstrated for membrane and needle hydrophones [4, 11, 12] and for fiber-optic hydrophones not based on the Fabry-Perot interferometer design [13, 14]. A previous study from this laboratory investigated deconvolution for three membrane hydrophones, two needle hydrophones, one capsule hydrophone, and one FPIFO hydrophone [12], and it concluded that the need for deconvolution was far greater for the FPIFO hydrophone than for the other designs. That previous study motivated the present investigation, which focuses on FPIFO hydrophones because of their challenging nature. The FPIFO design is particularly challenging because its sensitivity can vary by an order of magnitude or more over a short span of just a few MHz. In addition, FPIFO hydrophones (especially non-tapered ones) can exhibit significant structure in the phase response, with phase shift magnitudes approaching 180 degrees [12]. Since FPIFO hydrophone sensitivities vary considerably from fiber to fiber (from limited ability to control the delicate fabrication process), the present study considers 6 FPIFO probes in order to achieve a statistical analysis of the quantitative effects of deconvolution in the measurement of a pressure signal designed to mimic a typical clinical pulsed Doppler waveform. Deconvolutions were implemented in the frequency domain as inverse filters and were performed using both complex sensitivity and the magnitude of sensitivity (since many investigators only have access to the latter). The acoustic output parameters considered were the peak compressional pressure (p_+), peak rarefactional pressure (p_-), and the pulse intensity integral (PII). (The symbol for the latter is PII in [15] and pii in [2]). 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 [15]. 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 [16].

II. METHODS

A. Data Acquisition

Six FPIFO hydrophones (Precision Acoustics, Dorset, U.K.) were tested. Two of the six had tapered tips. One single-layer polyvinylidene fluoride (PVDF) membrane hydrophone (NTR Systems Inc., Seattle, WA, USA) was used for a reference. The complex sensitivity of the reference membrane hydrophone was measured from 0.95 to 50 MHz by an independent laboratory (Physikalisch-Technische Bundesanstalt, Braunschweig, Germany). Complex sensitivities were measured at the FDA for all hydrophones using a time-delay spectrometry system (TDS) previously described [12, 17]. 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 frequency 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 seven hydrophones were used to receive a pressure waveform that was designed to mimic a pulsed Doppler waveform in a water tank. A Tektronix (Beaverton, OR, USA) AFG 3021B function generator produced a signal of approximately 4 cycles at 3.5 MHz to drive a ROHE 5519 transducer (Rohe Scientific Corp., Santa Ana, CA, USA, center frequency: 3.5

MHz, diameter: 19 mm, with focus at approximately 9 cm). The hydrophones were placed 9 cm from the source transducer, which roughly corresponded to the location of maximum pulse intensity integral. At this depth, the -6 dB beam diameter was approximately $1.4\lambda z/d \approx 3$ mm, where λ is the wavelength and d is the transducer diameter. These measurements were digitized and stored (sampling rate = 2.5 GHz, vertical resolution = 11 bits) using a Tektronix DPO 3054 digital oscilloscope.

B. Data Analysis

Data analysis was performed using Matlab (Natick, MA, USA). Deconvolutions were implemented in the frequency domain as inverse filters computed from frequency-dependent hydrophone sensitivities. Complex sensitivities of the six fiber-optic hydrophones were measured using a substitution method with the membrane hydrophone serving as the reference. Broadband (1-40 MHz) sensitivity functions were constructed by concatenating data acquired with the four broadband source transducers. (See Appendix of [12].) The concatenation process includes adding linear functions of frequency ($\phi = b_i f$ where ϕ is an additive phase change, b_i is a constant, i is an index corresponding to the measurement frequency band, and f is frequency) to the phases measured over the four frequency bands in order to impose continuity of phase values at the transition frequencies between adjacent frequency bands (*i.e.*, 3, 6, and 25 MHz) [12]. Note that the addition or subtraction of this linear phase function in the frequency domain corresponds only to a shift in the time domain but not a change in shape of the time domain signal (according to the Fourier shift theorem). Therefore, this operation compensates for discrepancies in axial positions of two hydrophones in the substitution experiment but does not affect pressure pulse parameters [11]. In this substitution experiment, even a slight difference in position, on the order of four microns (approximately $\lambda/10$ at 40 MHz), can produce a measurable effect. Koch performed a similar linear adjustment to TDS data by analyzing a low-frequency portion of the spectrum over which the hydrophone phase response was assumed on theoretical grounds to be constant [16]. Wilkens and Koch performed a similar linear phase shift correction procedure in the time domain using a cross-correlation maximization method for the pulses measured for hydrophone calibration [11].”

As previously reported [Ref. 12, Figure 3], FPIFO hydrophone sensitivity could have a steep phase transition near 20 MHz. This transition could complicate comparison between direct phase measurements and minimum-phase-model-based estimates of phase. (See Ref. [12] for method of estimation of minimum phase model.) In order to facilitate comparison between measured and model-based estimates of phase, the following processing was performed. First, a linear phase function ($\phi = c_i f$, producing only a shift in time domain) was added to the concatenated direct sensitivity phase measurement in order to minimize the average difference between the phases obtained by measurement and by the minimum phase model over the range from 10 to 18 MHz. Second, a multiple of 360 degrees was added to each phase value in order to achieve a result between -180 and 180 degrees. (This is a mathematical contrivance with no physical significance). Third, the direct measurements of phase were convolved with a rectangular smoothing filter with width equal to 2 MHz. Finally, another linear phase function (again, $\phi = d f$) was added to the direct sensitivity

phase measurement in order to minimize the average difference between the phases obtained by measurement and by the minimum phase model over the range from 10 to 38 MHz.

Sensitivities at negative frequencies (which were required to compute inverse filters) were obtained by assuming that the sensitivity magnitude was an even function of frequency and the sensitivity phase was an odd function of frequency. This Hermitian form for the frequency response follows from assuming that the hydrophone impulse response is a real function [12].

The measurement band of frequencies (f_{min} to f_{max}) of hydrophone complex sensitivity was limited by the finite bandwidths of the source transducers, the sensitivities of the hydrophones, and the signal-to-noise ratios (SNRs) of the digitized output voltage signals. The values of f_{min} and f_{max} varied among the hydrophones but were typically approximately 1 MHz and 40 MHz respectively. The frequency range for the minimum phase model was the same as the frequency range for the direct measurement of sensitivity magnitude.

In order to compute inverse filters, it was necessary to extrapolate complex sensitivities for frequencies $|f| < f_{min}$. For the FPIFO hydrophones, the log magnitude of sensitivity was fit to a linear function of frequency with the form $\log |M_L(f)| = A|f| + B$ over the range of frequencies from f_{min} to 5 MHz. Then the fitted formula was used to extrapolate $M_L(f)$ for frequencies below f_{min} . For the membrane hydrophone, the value of the magnitude of the sensitivity at f_{min} was used for all frequencies for which $|f| < f_{min}$. For all hydrophones, the phase at zero frequency was assumed to be zero (as is required for an odd function). The phase at frequencies below f_{min} was found by linearly interpolating between zero at zero frequency and the phase measured at f_{min} . Finally, the magnitude and phase of sensitivity were each convolved with a rectangular smoothing filter with a width of 2 MHz. As will be seen in the next section (Figure 2), the test signal for this investigation had very little energy for frequencies $|f| > f_{max}$, and therefore extrapolation of complex sensitivity to frequencies higher than the measurement band was not critical.

The most elementary approach for deconvolution of the time-domain pressure waveform, $p(t)$, from the voltage waveform is to use an inverse filter in the frequency domain followed by an inverse Fourier transform,

$$p(t) = FFT^{-1} \left\{ \frac{U_L(f)}{M_L(f)} \right\} \quad (1)$$

where FFT denotes Fast Fourier Transform, FFT^{-1} denotes inverse FFT, $U_L(f)$ is the FFT of the time-domain hydrophone voltage signal, $u_L(t)$, and $M_L(f)$ is the hydrophone end-of-cable loaded sensitivity [2]. Practical limitations arise at frequencies at which the hydrophone sensitivity is not known with high certainty. Appropriate filtering of the voltage signal can mitigate this problem and therefore improve performance of pressure measurements [2]. As a first step, a Gaussian global low-pass filter, $G(f) = \exp(-f^2/2\sigma^2)$, was applied to measured voltage signals in order to suppress potential amplification of high-frequency noise due to division by $M_L(f)$. The value for σ was chosen to be 35 MHz, which is approximately ten

times the fundamental frequency in this experiment, so that the global low-pass filter would be expected to have minimal effect on the signal of interest.

Application of the Gaussian global low pass filter is equivalent to performing a convolution in the time domain with a Gaussian smoothing function that produces a weighted average of the signal with neighboring values at times occurring in the immediate past and future. Therefore the Gaussian global low-pass filter is not causal. Causality is not essential for this application, however, because the purpose of the filter is not to replicate an operation that could be realizable with a physical device. Rather, the purpose is to obtain an improved estimate of the signal by averaging over a range of times that is both compact and relevant in order to suppress additive noise.

A more general approach to deconvolution in the frequency domain is Wiener filtering

$$p(t) = FFT^{-1} \left\{ \frac{U_L(f)\Phi(f)}{M_L(f)} \right\} \quad (2)$$

where

$$\Phi(f) = \frac{|U_L'(f)|^2}{|U_L'(f)|^2 + |N(f)|^2}, \quad (3)$$

$U_L'(f)$ is the voltage spectrum that would be measured in the absence of noise and $N(f)$ is the noise spectrum so that the measured voltage spectrum $U_L(f) = U_L'(f) + N(f)$ [19]. In order to implement a Wiener filter for FPIFO hydrophone measurements, $N(f)$ was estimated by taking the FFT of a measurement during a gated time interval prior to the onset of the signal waveform and therefore assumed to contain pure noise. A fit to the noise power spectrum, $NPS_{fit}(f)$, was obtained from a power-law fit of $|N(f)|^2$ vs. f . $U_L'(f)$, which is required in Equation (3), cannot be known with certainty and therefore must be approximated. In the numerator of Equation (3), $|U_L'(f)|^2$ was approximated using $|U_L'(f)|^2 = \max\{|U_L(f)|^2 - NPS_{fit}(f), 0\}$. The denominator of Equation (3) was approximated by $|U_L(f)|^2$ as suggested in [19]. The latter approximation is supported by taking the squared magnitude of the defining equation, $U_L(f) = U_L'(f) + N(f)$, applying an expectation operator, and then assuming zero-mean noise that causes the expected values of cross terms, $U_L'(f)N(f)$, to be zero.

Although the global low-pass filter and the Wiener filter can improve deconvolution performance, neither one exploits a salient feature of nonlinear acoustic signals, which is that the spectral content of compressional segments tends to be spread over a considerably broader frequency range than that for rarefactional segments. In other words, in the time domain, compressional segments typically exhibit sharper peaks while rarefactional segments tend to be more rounded [20,21]. This spectral asymmetry suggests that deconvolution performance may be further improved by applying additional filtering (beyond the global low-pass or Wiener filters described above) only to rarefactional segments, without degrading the signal of interest. This may be achieved, after estimation of

$p(t)$ using Equation (1) or (2), by convolving rarefactional segments in $p(t)$ with a time-domain smoothing function, $r(t)$. In the present application, a rectangular smoothing function was chosen: $r(t) = \text{rect}(t/T)$ where $\text{rect}(x) = 1$ for $|x| < 1/2$ and 0 otherwise. The Fourier transform is $R(f) = (1/F) \text{sinc}(f/F)$ where $F = 1/T$ [22]. The central lobe of $R(f)$ achieves a maximum at $f = 0$ and goes to zero at $f = \pm F$. For the present application, F was chosen to be 20 MHz, corresponding to $T = 0.05 \mu\text{s}$.

Following the procedure introduced in [12], pressure waveforms were estimated using 4 methods:

1. Traditional method (ratio of the hydrophone output voltage to the magnitude of the hydrophone sensitivity at the acoustic working frequency) (“Scale”)
2. Deconvolution based on the direct measurement of the complex sensitivity (“Direct”)
3. Deconvolution based only on the magnitude of the sensitivity (“Mag”)
4. Deconvolution based on the estimate of the complex sensitivity assuming that the hydrophone measurement system (including source electronics, source transducer, diffraction, hydrophone, and receiving electronics) is minimum phase (“MP”) (see [12], [17], and [23]).

The third method, deconvolution based on the magnitude of the sensitivity, was previously proposed by Hurrell [4] as a potential improvement over the traditional method when phase information is unavailable. The acoustic output parameters (p_+ , p_- , PII) were computed using each of the 4 methods for each of the 6 FPIFO hydrophones.

III. RESULTS

For the reference membrane hydrophone, the acoustic parameter values obtained using complex deconvolution were $(2.4 \pm 0.07) \text{ MPa}$ (p_+), $(1.1 \pm 0.03) \text{ MPa}$ (p_-), and $(0.054 \pm 0.003) \text{ mJ/cm}^2$ (PII). The standard deviations correspond to precision levels of 3% for p_+ and p_- and 6% for PII. Precision estimates are based on three measurements from three different TDS experiments with repositioning.

Figure 1 shows the magnitude (left column) and phase (right column) of measurements of sensitivities for the membrane reference hydrophone and the 6 FPIFO hydrophones. The quasi-linear appearance of the FPIFO hydrophone sensitivity magnitudes at low frequencies on the log-linear plot supports the low-frequency extrapolation formula of the form $\log |M_L(f)| = A \log f + B$. From comparison of repeated sensitivity measurements (with repositioning), it was found that precision (standard deviation) of sensitivity magnitude averaged 10% at 3 MHz and 21% at 20 MHz and that precision of sensitivity phase averaged 3 degrees at 3 MHz and 11 degrees at 20 MHz. Averaging two sensitivity measurements prior to deconvolution was expected to improve measurement precision by a factor of $\sqrt{2}$. Hydrophones PA04 and PA06, which had tapered tips, were associated with smoother sensitivity magnitudes and less complicated phase responses than the other FPIFO hydrophones. The root-mean-square-error (RMSE) between direct measurements of phase

and minimum-phase-model-based estimates of phase between 0 and 30 MHz was 25 ± 5 degrees.

Figure 2 shows a power spectrum of a measured pulsed Doppler waveform (“Signal + Noise”), a power spectrum of noise (“Noise”), and a power law fit to the noise power spectrum (“NPS fit”). These measurements were used to construct the Wiener Filter.

Figure 3 shows the spectral asymmetry of compressional and rarefactional segments of the nonlinear test waveform. The left column shows the RF waveform measured with the reference membrane hydrophone (top left) decomposed into compressional (middle left) and rarefactional (bottom left) components. The magnitudes of the spectra of the three waveforms are shown in the right column. As stated previously, the spectral content of compressional segments is spread over a broader frequency range than that for rarefactional segments. Figure 3 supports the parameter choices for the global low-pass filter (G) ($\sigma = 35$ MHz) since there is little signal energy beyond 35 MHz and for time-domain smoothing filter (R) applied to rarefactional segments only ($F = 20$ MHz) since there is little energy in the rarefactional spectrum beyond 10 MHz.

Figure 4 shows the effect of deconvolution and filtering on a waveform measured with a FPIFO hydrophone. The top panel shows a raw FPIFO hydrophone voltage output divided by the magnitude of the sensitivity at the acoustic working frequency. The middle panel shows the FPIFO hydrophone signal filtered with the inverse of the sensitivity (Equation 1) and filtered with the Gaussian global low-pass filter (G). The bottom panel shows the result of applying the time-domain rarefactional smoothing filter (R) on the waveform in the middle panel. The time-domain rarefactional smoothing filter suppresses oscillations in rarefactional segments that can affect estimates of p_- . Such rapid oscillations during rarefactional segments seem inconsistent with the vast majority of reported measurements of nonlinear ultrasound signals. Therefore, the oscillations are likely due to uncertainties in the hydrophone sensitivity or imperfections in the initial deconvolution. Comparison of the middle and bottom panels of Figure 4 suggests that the rarefactional smoothing filter (R) introduces slight discontinuities in the slopes of compressional-to-rarefactional transitions near the zero line, as might have been expected.

Table I gives bias, coefficient of variation (COV), and RMSE for the acoustic pulse parameters (expressed as percentages) using “Scale,” “Mag,” “Direct,” and “MP” methods. Bias = $[(X - X_{REF}) / X_{REF}] \times 100\%$ (where $X = p_+$, p_- , or PII) was assessed by taking the membrane hydrophone measurements for reference values, using the same filters used for the FPIFO hydrophones in each particular case. The filters affected acoustic output parameters for the membrane hydrophone by a few percent. Because of uncertainty in the reference sensitivity, which led to uncertainties in estimates of acoustic pulse parameters for the reference hydrophone, estimates of bias were accurate only to within approximately ten percent. Comparison of the first two rows indicates that addition of the global low-pass filter (G) to the traditional scaling method reduced bias, COV, and RMSE for p_+ , p_- , and PII. The third row (Direct, G) shows substantial reduction of bias, COV, and RMSE for the complex deconvolution (direct method) compared with the traditional scaling method. The fourth row (Direct, G, W) shows that addition of the Wiener filter offered negligible change. This

implies that uncertainties in estimates of acoustic output parameters were determined more by deterministic errors (*e.g.*, positioning errors, calibration errors, temporal or temperature-dependent drift of hydrophone properties) than by inadequate SNR below 35 MHz. The fifth row (Direct, G, R) shows substantial reduction of bias, COV, and RMSE for p_- after application of the rarefactional low-pass filter (R). The last two rows (Mag, G, R, and MP, G, R) show that alternative methods for handling the phase of the sensitivity usually resulted in lower bias, COV, and RMSE than the scaling method but usually not as low as those obtained using direct complex deconvolution. The estimate of 3 percent mean bias in MP-deconvolution measurement of p_- is likely a statistical fluke, given the large COV. (In addition, as stated above, estimates of bias were only accurate to within approximately 10 percent anyway.)

Figure 5 shows estimates of peak compressional pressure (p_+), peak rarefactional pressure (p_-), and pulse intensity integral (PII) obtained using the traditional scaling method (“Scale”) and using deconvolution based on the magnitude of the sensitivity (“Mag”), the direct measurement of the complex sensitivity (“Direct”), and the minimum phase estimate of the complex sensitivity (“MP”). The dashed lines correspond to values measured using the reference membrane hydrophone (after complex deconvolution). All three methods of deconvolution resulted in substantial reductions in bias and variance of measurements of p_+ , p_- , and PII. As noted previously [6], PII (unlike p_+ and p_-) does not depend on phase. (This is a consequence of Parseval’s theorem.) Therefore, the three methods of deconvolution yielded identical results for PII for each hydrophone. Both G and R filters were used for Figures 5-7.

Figure 6 shows reconstructed pressure waveforms obtained by dividing the hydrophone output voltage by the magnitude of the sensitivity at the acoustic working frequency (left column) and by deconvolving the hydrophone output voltage using an inverse filter based on the complex hydrophone sensitivity (right column). As can be seen in Figure 1, all of the FPIFO hydrophones had high-pass filter behavior up to about 10 MHz. This high-pass filtering, which boosted harmonics relative to the fundamental frequency resulted in higher, sharper compressional peaks as shown in Figure 6 (left column). By comparing the left and right columns in Figure 6, it can be seen that deconvolution improved the consistency of the pressure waveforms. Deconvolution was far from perfect in restoring waveforms, however, as is evidenced by the strange reconstructed waveform shape, including narrow rarefactional excursions, for PA15.

Figure 7 shows reconstructed pressure waveforms obtained by deconvolving the hydrophone output voltage using an inverse filter based on the magnitude of the hydrophone sensitivity (left column) and the complex sensitivity based on the minimum phase assumption (right column). Comparison of Figure 6 (right column) and Figure 7 (both columns) indicates that the shapes of the reconstructed pressure waveforms depend on the form chosen for the sensitivity phase. Discrepancies are particularly evident for hydrophone PA15 (second from bottom). This dependence is much greater than that previously reported for membrane, needle, and capsule hydrophones [12]. One reason for this is that phase responses for FPIFO hydrophones have greater variations in phase (with magnitudes approaching 180 degrees

[12]) than phase responses for typical membranes, needles and capsules used in medical ultrasound exposimetry.

IV. DISCUSSION

Deconvolution based on frequency-domain inverse filtering with FPIFO hydrophone sensitivity improves accuracy and precision of measurements of peak compressional pressure (p_+), peak rarefactional pressure (p_-), and pulse intensity integral (PII) from a waveform that mimics a diagnostic pulsed Doppler pulse. Deconvolution improves the consistency of reconstructed RF traces. Explanations for remaining discrepancies among deconvolved signals include imperfect hydrophone positioning, hydrophone stability, and calibration uncertainties. (Regarding imperfect hydrophone positioning, note that while addition of a linear-with-frequency phase function corrects for differences in axial alignment, it does not correct for differences in lateral or elevational alignment).

In its best implementation (Direct, G, R), deconvolution reduced mean (over the 6 fiber-optic hydrophones) bias from 163% to 24% (p_+), 113% to 15% (p_-) and 126% to 29% (PII), and reduced mean COV from 18% to 11% (p_+), 53% to 11% (p_-), and 20% to 16% (PII). Note that the COV values for p_+ and p_- (11%) are comparable to the precision estimate for sensitivity magnitude measurements near the fundamental frequency (10% at 3 MHz).

The present study showed that proper filtering can improve deconvolution performance. This requires setting parameters for the global low-pass filter (σ) and the rarefactional low-pass filter (T). A choice for σ in the G filter may be facilitated when the fundamental frequency of the signal is known with a high level of certainty, which is commonly the case when the source transducer is driven to operate near its resonant frequency. Hydrophone voltage spectral plots viewed alongside frequency-dependent hydrophone sensitivity should indicate how many spectral harmonics make significant contributions to the signal. The value of σ may be chosen to include all significant harmonic content while excluding frequencies beyond the significant harmonic content. A choice for T in the R filter may be facilitated in some applications by empirical observation of the frequency of oscillations in rarefactional segments. For example, if rarefactional oscillations are judged to be nonphysical, they could be suppressed by choosing T to be greater than the oscillation period. The additional filtering applied to rarefactional segments (R filter) after the filtering applied to both compressional and rarefactional segments (G filter) may partially explain why the bias was found to be lower for p_- than for p_+ . Similar methodology may be useful for other sensors with nonuniform response that are used to measure nonlinear signals, including high intensity therapeutic ultrasound signals that have very high bandwidth and place high demands on hydrophone bandwidth [25]. Added filtering of the rarefactional segment in conjunction with deconvolution might be useful to enhance overall bandwidth without corrupting the rarefactional parts of the waveforms.

Deconvolution affected measurements of p_+ , p_- , and PII differently. This is related to the waveform shape. The test waveform, which mimicked a pulsed Doppler waveform, exhibited effects of nonlinear propagation as compressional segments tended to have sharper peaks while rarefactional segments tended to be more rounded (Figure 6, top row). This kind

of asymmetry between compressional and rarefactional segments is common for many diagnostic ultrasound signals [20, 21]. It means that higher harmonics of the spectrum manifested themselves more on compressional segments than rarefactional segments. Therefore, compressional features of the nonlinear waveform would be expected to be dominated by higher frequencies while rarefactional features would be expected to be dominated by lower frequencies. Differences in the effectiveness of the deconvolution process for p_+ and p_- arise as they are dependent on different spectral characteristics of the acoustic field. The effects of imperfect hydrophone positioning, which increase with ultrasonic frequency, presumably had a greater impact on estimates of p_+ than estimates of p_- since p_+ exhibits greater influence from high frequencies than p_- .

Sensitivity phase, which is required to perform the complex deconvolution, is often not available because hydrophone manufacturers often provide only sensitivity magnitude. When sensitivity phase measurements are unavailable, then deconvolution can be performed with the magnitude of the sensitivity. Alternatively, a complex deconvolution can be performed by assuming that the hydrophone measurement system obeys the minimum phase principle, which has previously been shown to be a good approximation for some membrane, needle, and capsule hydrophones [17, 23]. The minimum-phase model has previously been applied to a Michelson-interferometric fiber-optic system used in extracorporeal shock-wave lithotripsy [24]. Table I and Figure 5 show that all three deconvolution methods tested in the present investigation resulted in substantial improvements in accuracy and precision of p_+ , p_- , and PII. Since the Wiener filter offered little (< 1 percent) improvement, remaining uncertainties in estimates of acoustic output parameters are likely determined more by deterministic errors (*e.g.*, positioning errors, calibration errors, temporal or temperature-dependent drift of hydrophone properties) than by insufficient SNR below 35 MHz.

Figure 1 might suggest that the agreement between direct measurements of phase and minimum-phase-model-based estimates of phase was considerably better than results published for another FPIFO hydrophone [Ref 12, Figure 3]. Figure 1 provided motivation to revisit the previously published phase results. The previously-published direct phase measurements were re-analyzed by the methodology explained in the Methods section of the present paper (see Section II. B). In particular, this re-analysis included addition of a multiple of 360 degrees to each phase value in order to achieve a result between -180 and 180 degrees followed by convolution with a rectangular smoothing filter with width equal to 2 MHz (two steps that were not used previously [12]). (Note that this new methodology would not have much effect on the non-fiber-optic hydrophones considered in Ref. [12] because their phase responses tended to be fairly smooth and they produced phase shifts much less than 180 degrees in magnitude.) The result is shown in Figure 8. The two phase functions exhibit similar oscillations between 16 and 32 MHz. The RMS difference between the two phase functions over the range from 2 to 37 MHz is 30 degrees. The sharp upward sloping portion of the direct measurement of phase near 18 MHz in Figure 8 arises from adding 360 degrees to the phase shown in Ref 12, Figure 3 when it drops below -180 degrees.

The validity of the minimum phase model was tested for another fiber optic hydrophone, the FOPH 2000 (RP Acoustics, Germany) with an active diameter of 100 μm . The manufacturer provides the impulse response for this system. The sensitivity was obtained from the FFT of the impulse response. The phase of the sensitivity was obtained 1) by computing the phase of the FFT of the impulse response, and 2) by computing the Hilbert transform of the logarithm of the magnitude of the sensitivity [17]. Figure 9 shows good agreement between both phase functions, supporting the minimum phase model for this fiber optic hydrophone design.

V. CONCLUSION

For hydrophones with highly frequency-dependent sensitivity and acoustic signals with high harmonic content arising from nonlinear propagation, complex deconvolution can substantially reduce bias and variability of estimates of acoustic output measurements. Spectral asymmetry between compressional and rarefactional segments may be exploited in order to design optimal filters used in the deconvolution algorithm. Reliable acoustic output measurements may be performed with FPIFO hydrophones in conjunction with complex deconvolution and appropriate filtering. The methodology presented here is likely to be useful for other sensors with nonuniform responses.

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Figure 1. Magnitude (left column) and phase (right column) of sensitivity for the membrane reference hydrophone and the six FPIFO hydrophones. The hydrophone identifications are given in the upper left corner of the magnitude plots. Sensitivities have been smoothed with a rectangular smoothing filter with width = 1 MHz.

Figure 2.

Power spectra of measured pulsed Doppler waveform (“Signal + Noise”) and of a measurement during a gated time interval prior to the onset of the signal waveform and therefore assumed to contain pure noise (“Noise”). Also shown is a power law fit to the noise power spectrum (“NPS fit”).

Figure 3.

RF waveform measured with reference membrane hydrophone (top left) decomposed into compressional (middle left) and rarefactional (bottom left) components. The magnitudes of the spectra of the three waveforms are shown in the right column. G: global low-pass filter. |R|: magnitude of rarefactional low-pass filter.

Figure 4.

Raw FPIFO hydrophone (PA03) output divided by the magnitude of the sensitivity at the acoustic working frequency (top), deconvolved using an inverse filter based on the complex hydrophone sensitivity without rarefactional low-pass filter (middle) and with rarefactional low-pass filter (bottom). G: global low-pass filter. R: rarefactional low-pass filter. The G filter is applied to the entire waveform (in the frequency domain), and the R filter is applied to rarefactional segments only (in the time domain) (see Section II.B).

Figure 5.

Estimates of peak compressional pressure (p_+), peak rarefactional pressure (p_-), and pulse intensity integral (PII) obtained using the traditional scaling method (“Scale”) and using deconvolution based on the magnitude of the sensitivity (“Mag”), the direct measurement of the complex sensitivity (“Direct”), and the minimum phase estimate of the complex sensitivity (“MP”). The dashed lines correspond to values measured using the reference membrane hydrophone, which are based on using direct measurements and the G and R filters.

Figure 6.

Raw hydrophone output divided by the magnitude of the sensitivity at the acoustic working frequency (left column) and hydrophone output deconvolved using an inverse filter based on the complex hydrophone sensitivity (right column). Deconvolved waveforms utilized the global low-pass filter and the rarefactional low-pass filter.

Figure 7.

Raw hydrophone output deconvolved using an inverse filter based on the magnitude of the hydrophone sensitivity (left column) and the complex sensitivity based on the minimum phase assumption. Deconvolved waveforms utilized the global low-pass filter and the rarefactional low-pass filter.

Figure 8.

Direct measurements and minimum-phase-model-based estimates of sensitivity phase for a FPIFO hydrophone used in a previous publication [10].

Figure 9.

Impulse response (top), sensitivity magnitude (middle), and sensitivity phase computed directly and with the minimum phase model (bottom) for the FOPH 2000 fiber-optic hydrophone system.

Table I.

Relative bias, coefficient of variation (COV), and root mean squared error (RMSE) of acoustic pulse parameters for FPIFO hydrophones expressed as percentages, obtained using the traditional scaling method (“Scale”) and deconvolution based on three forms of sensitivity: direct complex measurement (“Direct”), magnitude (“Mag”), and phase estimated from minimum phase principle (“MP”). G: Gaussian global low-pass filter $\exp(-f^2/2\sigma^2)$ with $\sigma = 35$ MHz. W: Wiener Filter. R: Rectangular rarefactional low-pass filter with $F = 20$ MHz, corresponding to $T = 0.05 \mu\text{s}$.

Method	Filters	p_+			p_-			PII		
		Bias	COV	RMSE	Bias	COV	RMSE	Bias	COV	RMSE
Scale		163	18	169	113	53	161	126	20	134
Scale	G	108	13	111	81	51	123	97	19	104
Direct	G	24	11	27	48	21	58	33	17	40
Direct	G, W	24	11	27	48	22	58	33	17	40
Direct	G, R	24	11	27	15	11	20	29	16	36
Mag	G, R	38	14	41	11	29	35	29	16	36
MP	G, R	42	12	46	3	24	25	29	16	36