Title
Acoustic-based measurements of material absorption coefficients: Relationship between laser pulse duration and stress confinement time

Permalink
https://escholarship.org/uc/item/0s97x0q0

Journal
JOURNAL OF APPLIED PHYSICS, 94(12)

ISSN
0021-8979

Authors
Choi, B
Jansen, ED
Welch, AJ

Publication Date
2003-12-15

DOI
10.1063/1.1627464

License
https://creativecommons.org/licenses/by/4.0/ 4.0

Peer reviewed
Acoustic-based measurements of material absorption coefficients: Relationship between laser pulse duration and stress confinement time

Bernard Choa)
Department of Biomedical Engineering, The University of Texas at Austin, Austin, Texas 78712
and Beckman Laser Institute, University of California, Irvine, California 92612

E. Duco Jansen
Department of Biomedical Engineering, Vanderbilt University, Nashville, Texas 37235

Ashley J. Welch
Department of Biomedical Engineering, The University of Texas at Austin, Austin, Texas 78712

(Received 21 July 2003; accepted 24 September 2003)

Ideally, to use photoacoustics to determine the absorption coefficient \( \mu_a \) of a medium, the laser pulse duration \( \tau_p \) is much shorter than the stress confinement time \( \tau_{sr} \). The laser-induced stress wave propagates a distance equal to the light penetration depth \( \delta \). However, without prior knowledge of \( \delta \) (equal to \( 1/\mu_a \)), it is not clear whether a given photoacoustic measurement is indeed performed under stress-confined conditions. The purpose of this study was to explore the effects of \( \tau_p \) and \( \tau_{sr} \) upon efforts to obtain estimates of \( \mu_a \) using photoacoustics. A numerical model was developed to simulate stress signals and investigate how measurements of \( \mu_a \) are related to the ratio \( \tau_r = \tau_{sr}/\tau_p \). Experimental photoacoustic measurements at several values of \( \tau_r \) were performed to estimate \( \mu_a \) of water, and a deconvolution model was applied to correct the measured \( \mu_a \) without prior knowledge of \( \tau_r \). Under the conditions simulated in this study, \( \tau_p \) must be less than \( \sim 0.1 \tau_{sr} \) for optimal photoacoustic measurements of \( \mu_a \). Since it is difficult to achieve such conditions at midinfrared wavelengths for accurate soft tissue characterization due to strong water absorption in blood vessels, a numerical deconvolution technique was implemented to overcome this limitation of conventional photoacoustics, resulting in up to a 30% improvement in photoacoustic-based estimates of the sample \( \mu_a \). © 2003 American Institute of Physics. [DOI: 10.1063/1.1627464]

I. INTRODUCTION

Accurate knowledge of tissue optical properties is critical for development of light-based diagnostic and therapeutic technologies. Both direct and indirect techniques are employed to determine these parameters. With indirect techniques such as spectrophotometry and pulsed photothermal radiometry, complex algorithms are required to determine absorption and scattering properties from the measured signals. One direct means to measure tissue properties is to use photoacoustics to detect stress waves induced by short laser pulses. Ideally, the laser pulse duration \( \tau_p \) is much shorter than the stress confinement time \( \tau_{sr} \), required for a laser-induced stress wave to propagate a distance equal to the light penetration depth \( \delta \).  

\[
\tau_{sr} = \frac{\delta}{c_s},
\]

where \( c_s \) is the speed of sound in the medium (1.5 \( \times 10^3 \) cm/s in water, which is similar to the value for biological soft tissue). Under stress confinement conditions (e.g., \( \tau_p \ll \tau_{sr} \)), the measured stress signal shape represents exactly the light distribution in the medium. Absorption and scattering coefficients (\( \mu_a \) and \( \mu_s \), respectively) are determined from these profiles. For homogeneous media, these parameters are identified from examination of the peak value and exponential decay of the profiles. Viator et al. derived a relatively simple technique for using photoacoustics to determine \( \mu_a \) values of layered tissue phantoms.

At visible and near-infrared wavelengths (\( \lambda = 0.4–1.1 \mu m \)), \( Q \)-switched lasers emitting pulses of \( \tau_p = 1–10 \) ns are readily available. At midinfrared wavelengths (\( \lambda = 2–10 \mu m \)), conventional \( Q \)-switched solid-state lasers provide longer pulses of \( \tau_p \geq 100 \) ns. Without prior knowledge of \( \delta \) (equal to \( 1/\mu_a \)), it is not clear whether a given photoacoustic measurement is indeed performed under stress-confined conditions. For experiments at midinfrared laser wavelengths on tissue with high water content, \( \mu_a \) values range over several orders of magnitude, from \( \sim 10 \) to over \( 10000 \) cm\(^{-1} \), corresponding to \( \tau_{sr} \) between 0.1 and 1000 ns. Under these conditions, the relationship between \( \tau_p \) and \( \tau_{sr} \) is of paramount importance. The purpose of this study is to explore the effects of \( \tau_p \) and \( \tau_{sr} \) on \( \mu_a \) values ascertained with photoacoustics. A numerical model was developed to simulate stress signals and investigate how measurements of \( \mu_a \) are related to the ratio \( \tau_r = \tau_p/\tau_{sr} \). Experimental photoacoustic measurements at several values of \( \tau_r \) were performed to estimate \( \mu_a \) of an absorbing medium (water), and a deconvolution model was applied.
to correct the measurements of \( \mu_a \) without prior knowledge of \( \tau \).

II. MATERIALS AND METHODS

A. Thermoeelastic stress wave generation

An ideal acoustic signal induced by a stress-confined laser pulse in a homogeneous medium is shown in Fig. 1. In this example, four conditions are required: (1) light absorption dominates over light scattering; (2) the acoustic transducer is at the backside of the medium and the laser pulse is incident on the front side; (3) acoustic impedance of the medium is greater than that of the surroundings; and (4) \( \delta \) is at least an order of magnitude less than the spot size, providing a basis for a one-dimensional geometry assumption. Upon pressure generation, half of the pressure signal propagates towards the transducer and the other half towards the medium surface. The first half of the signal is a compression wave. The shape of the compression wave resembles the absorbed energy density distribution inside the medium; the peak at 450 ns (Fig. 1) corresponds to the medium surface and the initial portion of the rise represents deeper regions of the medium. A subsequent tensile wave occurs due to reflection of the acoustic signal from the front surface. Since the compression and tensile waves are exact replicas of the absorbed energy density distribution, \( \mu_a \) can be calculated either from the peak pressure amplitude or from a Beer’s law exponential fit to the data because

\[
P(z) = \frac{1}{k'} \frac{\mu_a H_0 \exp(-\mu_a z)}{k},
\]

where \( P(z) \) is the pressure (atm) at depth \( z \), \( \Gamma \) is the Gruneisen coefficient (\( \Gamma \)), \( H_0 \) is the radiant exposure (J/cm\(^2\)), and \( k' = 101 \text{ 325 Pa/atm} \) is a unit conversion factor. \( \Gamma \) represents a measure of heat-to-acoustic energy conversion and is defined as

\[
\Gamma = \frac{\beta c_p^2}{c_p},
\]

where \( \beta \) is the thermal coefficient of volume expansion \( (\text{K}^{-1}) \) and \( c_p \) is the specific heat at constant pressure \( (\text{J/g/K}) \).

Each of these parameters is temperature dependent; for water, \( \Gamma \) can be estimated from the following empirically-derived equation:

\[
\Gamma = 0.0043 + 0.0053T,
\]

where \( T \) is temperature \( (\text{C}) \).

If \( \tau_p > \tau_{sr} \), sufficient time exists for the generated stress wave to propagate beyond \( \delta \). Under this condition, \( \mu_a \) calculations from measured stress signals may underestimate the actual values. As \( \tau_p \) increases, the degree of underestimation increases and the stress signal shape resembles more closely the laser pulse temporal profile. For \( \tau_p \approx \tau_{sr} \), the degree of error associated with photoacoustic-based absorption coefficient measurements is largely unknown.

B. Relationship between \( \tau_p \) and \( \tau_{sr} \)—Numerical model

Our model convolves numerically a laser pulse of arbitrary shape with a pressure distribution of arbitrary shape, resulting in a theoretical estimate of the expected stress signal. In this study, the laser pulse temporal profile was assumed to be gaussian shaped, which is typical for many laser systems. The profile was divided arbitrarily into “impulses” with duration 0.01 \( \tau_{sr} \). In this study, the pressure distribution was assumed to follow the shape

\[
P(z) = \sum_i \Gamma H_{i,j} \mu_a \exp(-\mu_a z)
\]

where \( P(z) \) is depth-resolved pressure (bars) induced by the laser pulse and \( H_{i,j} \) is input radiant exposure \( (\text{J/cm}^2) \) of each impulse.

In this study, we assumed the following: (1) medium absorption coefficients \( \mu_{a,med} \) with values ranging between 10 and 1000 \text{ cm}^{-1}; and (2) \( \tau_p \) between 1 and 5000 ns. Stress signals were computed and the theoretical estimate of the photoacoustic-based absorption coefficient measurement \( \mu_{a,pre} \) calculated and compared to \( \mu_{a,med} \).

C. Relationship between \( \tau_p \) and \( \tau_{sr} \)—Experiments

In this phase of the study, we acquired photoacoustic-based measurements of \( \mu_a \) for a range of \( \tau \) and \( \mu_a \). De-ionized water was used as a homogeneous absorbing medium since values of water \( \mu_a \) are abundant in the literature and it is the primary constituent of soft tissue. Since water \( \mu_a \) exhibits a strong wavelength dependence in the midinfrared, different \( \mu_{a,med} \) values were obtained by using different laser wavelengths. Two laser systems were used. A commercial Ho:Yttrium–aluminum–garnet (YAG) laser emitted light at 2.1 \( \mu \text{m} \), and was used in \( Q \)-switched mode with a measured pulse duration of 819 \( \pm \) 145 ns. The Vanderbilt free electron laser (FEL) was used as a tunable light source at wavelengths of 2.4 and 3.7 \( \mu \text{m} \).

For the Ho:YAG laser experiment, the optical setup (Fig. 2) consisted of an intracavity acousto-optic modulating \( Q \) switch, mirrors to steer the beam, and beam expansion optics to achieve a laser spot diameter of \( \sim 3 \text{ mm} \). Portions of the incident beam were sampled with an InAs photodiode (J12-18C-R250U, Judson Technologies, Montgomeryville, PA) and an energy detector (JP25, Moleclectron Detector, Inc., Port-
land, OR) to obtain measurements of pulse temporal profile and pulse energy, respectively. The ratio between incident energy and sampled energy was calculated by replacing the acoustic transducer with a second energy detector and simultaneously measuring pulse energies over a wide energy range.

De-ionized water was placed in a 1.4 cm diameter plastic ring located on top of an acoustic transducer (described in the next paragraph). A known volume of water was poured into the ring. Water layer thickness was estimated by dividing the volume by the area of the ring.

The acoustic transducer (WAT-19, LaserSonix, Inc., Houston, TX) consisted of a piezoelectric ceramic element with an aluminum acoustic conductor. It was sensitive to frequencies ranging between 0.5 and 40 MHz. Stress waves reaching the transducer front surface were converted to electrical signals that were subsequently amplified with a low-noise preamplifier (SR445, Stanford Research Systems, Sunnyvale, CA) and acquired with a digital oscilloscope (TDS640A, Tektronix, Beaverton, OR). Laser pulse temporal profiles and acoustic wave forms were stored on the oscilloscope and transferred to PC for postprocessing. To reduce pulse-to-pulse noise, each acquired stress signal was an average of 25 stress waves induced at a pulse repetition rate of 2 Hz, for both lasers.

The FEL is a tunable infrared laser capable of emitting light at wavelengths between 2 and 10 μm. The FEL macro-pulse consists of 1 ps long micropulses that are spaced 350 ps apart (e.g., 2.85 GHz micropulse repetition rate). A sequence of micropulses results in a macro-pulse with a duration of approximately 4 μs.

In one set of experiments, the entire FEL macro-pulse was delivered to the water sample (Fig. 3, excluding region enclosed by the dashed line). A portion of the beam was sampled with pyroelectric detectors (J8LP and P3-01, Molecron Detector Inc.) to measure pulse energies and temporal profiles, respectively. The beam was focused to the water sample surface using a 200 mm focal length planoconvex CaF₂ lens. The FEL was tuned to a wavelength of 2.4 μm, corresponding to a water μₐ of 50 cm⁻¹.⁷

To investigate the effects of τₚ on photoacoustic-based μₐ measurements, an alternate experimental setup was used (Fig. 3, including region enclosed by a dashed line). The FEL laser wavelength was set at either 2.4 or 3.7 μm (μₐ of water = 50 and 122 cm⁻¹, respectively). A germanium filter was used to remove harmonics of the incident radiation. The beam was then reduced in diameter with a 1:10 telescopic CaF₂ lens system. At 3.7 μm, a CdTe Brewster-cut Pockels cell was used to obtain τₚ ranging between 100 ns and 2 μs. The Pockels cell served as a fast electro-optic shutter; application of a high-voltage (~3 kV) pulse to the Pockels cell resulted in a transient rotation of the plane of incident linearly polarized light by 90°. A polarizer-attenuator (PAX-20-AC-4, II-VI, Inc., Saxonburg, PA) placed behind the Pockels cell served as a fast electro-optic shutter; application of a high-voltage (~3 kV) pulse to the Pockels cell resulted in a transient rotation of the plane of incident linearly polarized light by 90°. A polarizer-attenuator (PAX-20-AC-4, II-VI, Inc., Saxonburg, PA) placed behind the Pockels cell served as a fast electro-optic shutter; application of a high-voltage (~3 kV) pulse to the Pockels cell resulted in a transient rotation of the plane of incident linearly polarized light by 90°. A polarizer-attenuator (PAX-20-AC-4, II-VI, Inc., Saxonburg, PA) placed behind the Pockels cell served as a fast electro-optic shutter; application of a high-voltage (~3 kV) pulse to the Pockels cell resulted in a transient rotation of the plane of incident linearly polarized light by 90°. A polarizer-attenuator (PAX-20-AC-4, II-VI, Inc., Saxonburg, PA) placed behind the Pockels cell served as a fast electro-optic shutter; application of a high-voltage (~3 kV) pulse to the Pockels cell resulted in a transient rotation of the plane of incident linearly polarized light by 90°.

For both setups, the spot size at the target plane was measured to be ~7 mm diameter, using the knife-edge technique.⁹ The transducer and acquisition electronics used in the Ho:YAG experiments described earlier were also used in these experiments.

Software written in LabVIEW (Version 6i, National Instruments, Austin, TX) and MATLAB (Version 6.1, The MathWorks, Natick, MA) was used to process the photoacoustic wave forms. High-frequency noise was removed with a digital second-order Chebyshev low-pass filter [cutoff frequency of 0.01fₛ, where fₛ was the oscilloscope sampling frequency used during signal acquisition (50–500 MHz)]. This filter type was selected to maximize attenuation of frequencies in the stop band and was observed to preserve extremely well the fidelity of the acoustic wave forms. Each time-resolved photoacoustic signal was converted to P(z)
profiles by converting time values to depth values using the relationship \( z = c \cdot t \).

First-order exponential decay profiles were fit to the acquired acoustic wave forms using a Levenberg–Marquardt-based nonlinear curve-fit routine.\(^\text{10}\) The exponential decay constant was the measured absorption coefficient \( \mu_{a, \text{meas}} \). Confidence intervals of 95% of \( \mu_{a, \text{meas}} \) were calculated. Box and Whisker plots were used to identify outliers; these data points were not considered in subsequent calculations.

D. Deconvolution model for improved accuracy of absorption coefficient measurements

During the experiments, each laser temporal profile was acquired and saved. A model was developed to predict the stress signal that would result for a given pulse profile and for a range of assumed medium absorption coefficients \( (\mu_{a, \text{guess}}) \). For a given stress signal measurement, the associated laser pulse temporal profile was reduced to a series of impulses. The duration \( \Delta t \) of each impulse was equal to the shorter of the following values: \( 1/f_s \), where \( f_s \) is the oscilloscope sampling frequency, or 0.01 \( \tau_s \). The factor 0.01 was used because it is reasonable to assume that a pulse with a duration of 0.01 \( \tau_s \) would be considered as stress confined. That is, the pulse would be short enough to create an impulse response. Radiant exposures of each impulse were computed as the energy contained in each impulse divided by the laser spot size.

For each impulse, the resulting bipolar stress wave was modeled as follows. An exponential decay curve calculated with Eq. (2) was generated as a function of time with a decay coefficient equal to \( \mu_{a, \text{guess}} \). \( \Gamma \) was assumed to equal 0.121, the value calculated with Eq. (4) at \( T = 22 \degree \mathrm{C} \). The decay curve was flipped in time to become an exponential growth curve corresponding to the compression stress wave induced by the impulse. Since a free-surface boundary condition and ideal stress generation conditions were assumed, the shape and absolute amplitude of the tensile wave were modeled to be identical to those of the compression wave. Under these assumptions, a bipolar stress wave was created. The stress signal amplitude was scaled according to the fractional energy contained in each impulse.

With each impulse, the corresponding bipolar stress wave was assumed to be generated instantaneously. The onset of subsequent stress signals was separated in time by an integer multiple of \( \Delta t \). The resulting sequence of impulse responses was summed to form one composite bipolar stress wave.

For direct comparison between measured and modeled stress signals, the modeled signal was “sampled” at frequency \( f_s \) and filtered with the same digital low-pass filter used on the measured data. The x axis was converted from units of time to space and an exponential fit applied to the initial rise of the modeled stress signal. The absorption coefficient calculated from the exponential fit represented the prediction of the measured absorption coefficient \( (\mu_{a, \text{pred}}) \) for a given guess of the sample absorption coefficient \( \mu_{a, \text{guess}} \).

For example, consider a scenario in which a 4 \( \mu \text{s} \) long pulse with a gaussian temporal profile irradiates a homoge-
the model. The range of $\mu_{a,\text{guess}}$ was 2–220 cm$^{-1}$, and corresponding $\mu_{a,\text{pred}}$ ranged between 2 and 80 cm$^{-1}$. Note that for larger values of $\mu_{a,\text{guess}}$, the curve appears to approach an asymptotic value for $\mu_{a,\text{pred}}$.

A summary of experimental and deconvolved data is provided in Table I. $\tau_{sr}$ at each wavelength was calculated with Eq. (1) using the published values of $\mu_{a,\text{med}}$ for water and $c_s$ of 150 000 cm/s. $\tau$ ranged between 1.8 and 37, indicating that all data were taken under conditions of $\tau_{p} > \tau_{sr}$. For each laser wavelength, $\mu_{a}$ values determined directly from measured data ($\mu_{a,\text{med}}$) and deconvolved model results ($\mu_{a,\text{pred}}$) are tabulated alongside $\mu_{a,\text{med}}$. The improvement resulting from use of the deconvolution model was determined using the following equation:

$$\% \text{ improvement} = (|\mu_{a,\text{med,exp}} - \mu_{a,\text{med}}|/\mu_{a,\text{med}}) \times 100\%.$$  

(6)

IV. DISCUSSION

In this study, we investigated the relationship between $\tau_p$, $\tau_{sr}$, and $\mu_{a,\text{med}}$ for photoacoustic-based measurements of $\mu_a$. A numerical model was employed to determine theoretically the effects of $\tau_{sr}$ and $\tau_p$ on $\mu_{a,\text{pred}}$. Stress signal measurements were obtained at various $\tau$ and $\mu_{a,\text{med}}$ was extracted from the data. A numerical deconvolution routine was applied to the experimental data to improve the accuracy of the measurements of $\mu_a$. The advantage of using such an approach to determine $\mu_a$ is that no prior information about $\tau$ is required, which is the case when materials of unknown optical properties at a given laser wavelength are probed, such as kidney stones and thermally denatured skin.

Assuming a Gaussian-shaped laser pulse temporal profile and ideal stress signal generation and detection, photoacoustic-based measurements of $\mu_a$ are optimal for $\tau_p$ that are at least one order of magnitude shorter than $\tau_{sr}$ (Fig. 5). Such conditions are readily achieved at visible and near-infrared wavelengths due to the availability of commercial Q-switched lasers emitting pulses of $\tau_p \approx 1–10$ ns. However, in the midinfrared, typical solid-state lasers emit laser pulses that are too long to satisfy the criterion $\tau \ll 0.1 \tau_{sr}$. The Q-switched Ho:YAG and Er:YAG ($\lambda = 2.94$ μm) lasers used in this and other photoacoustics studies emitted pulses of $\tau_p \approx 100$ ns. At these wavelengths, water $\mu_a$ is 28 and $\sim 10000$ cm$^{-1}$, respectively, corresponding to $\tau_{sr}$ of 239 and $\sim 0.7$ ns, respectively. Thus, the lasers available in our laboratories were not able to emit sufficiently short laser pulses for optimal stress confinement. An alternative laser system to use is a tunable optical parametric oscillator (OPO). Unfortunately, at the time of these measurements, the OPO sys-

| Wavelength [μm] | $\tau_p$ [ns] | $\tau_{sr}$ [ns] | $\tau$ (s) | $\mu_{a,\text{med}}$ [cm$^{-1}$] | $\mu_{a,\text{pred}}$ [cm$^{-1}$] | $\mu_{a,\text{med}}$ [cm$^{-1}$] | % improvement |
|-----------------|-------------|----------------|----------|----------------------------|----------------|----------------------------|--------------|
| 2.1             | 815         | 239           | 3.4      | 21.87 ± 4.29               | 33.64 ± 4.33 | 27.92                      | 1            |
| 2.4             | 4000        | 133           | 30.0     | 21.29 ± 1.96               | 64.32 ± 14.70 | 50.06                      | 29           |
| 3.7             | 100         | 55            | 1.8      | 77.20 ± 3.15               | 100.18 ± 10.10 | 122.27                      | 19           |
| 3.7             | 200         | 55            | 3.7      | 72.58 ± 2.17               | 147.60 ± 22.89 | 122.27                      | 20           |
| 3.7             | 500         | 55            | 9.2      | 69.94 ± 1.85               | 159.70 ± 22.05 | 122.27                      | 12           |
| 3.7             | 1000        | 55            | 18.3     | 70.81 ± 3.78               | 176.56 ± 30.73 | 122.27                      | 2            |
| 3.7             | 2000        | 55            | 36.7     | 70.91 ± 2.47               | 159.82 ± 23.42 | 122.27                      | 11           |
tems at our institutions did not emit pulses of sufficient energy to generate measurable stress signals at midinfrared laser wavelengths. A recent study by Kostli et al.\textsuperscript{14} employed an OPO for photoacoustic characterization of cartilage and chicken breast at wavelengths between 1.86 and 1.94 µm, close to the wavelengths investigated in this study. Further development of OPO technology will provide hopefully the means to characterize soft tissue $\mu_a$ at wavelengths in the 2–10 µm range without the need for deconvolution.

Two regions can be identified from the representative data set shown in Fig. 6. The first region involves a monotonic linear relationship between $\mu_{a,\text{pred}}$ and $\mu_{a,\text{guess}}$. The second region consists of a relatively large scatter in the data points. The values of $\mu_{a,\text{pred}}$ seem to fall around a line with a positive slope, but this slope is considerably smaller than that of the first region. For smaller values of $\tau$, the model predictions $\mu_{a,\text{pred}}$ and measured values match in the first region. As $\tau$ increases, $\mu_{a,\text{pred}}$ and $\mu_{a,\text{meas}}$ tend to match in the second region, resulting in larger error in the estimate of $\mu_{a,\text{med}}$.

Oraevsky et al.\textsuperscript{5} noted that for experimental conditions in which $\tau_p \gg \tau_{sr}$, the stress signal resembles the laser pulse temporal profile. When this limit is reached, all input values of $\mu_{a,\text{guess}}$ into the deconvolution model results in similar values of $\mu_{a,\text{pred}}$. The presence of the second region (Fig. 6) indicates that this limit is being approached for larger values of $\mu_{a,\text{guess}}$. Thus, the model is limited in its ability to estimate $\mu_{a,\text{meas}}$ for extremely large values of $\tau$. Further studies need to be conducted to determine the value of $\tau$ over which the utility of the model is limited due to excessive stress wave propagation during the laser pulse.

Using photoacoustics, measurements of water $\mu_a$ were obtained for different $\tau$ (Table I). In general, as $\tau$ increased, the discrepancy between $\mu_{a,\text{meas}}$ and $\mu_{a,\text{med}}$ increased due to stress wave propagation during the laser pulse to regions outside of $\delta$. Thus, standard photoacoustics provide limited information under conditions of $\tau_p \gg \tau_{sr}$. Application of a numerical deconvolution routine (Fig. 4) to the data resulted in equivalent or markedly improved knowledge of $\mu_{a,\text{med}}$. With the model, the relative error was reduced by up to 30%.

In this study, we allowed $\mu_{a,\text{guess}}$ values with corresponding $\mu_{a,\text{pred}}$ values falling within the 95% confidence interval of $\mu_{a,\text{meas}}$ to be used as possible guesses for $\mu_a$. Other approaches may improve further the accuracy of this model-based approach for correction of $\mu_a$ measurements with photoacoustics. By incorporating the acoustic transducer impulse response in the deconvolution computation,\textsuperscript{15} further improvement is expected. Nevertheless, the improved accuracy of our approach is apparent from the results in Table I.

V. CONCLUSIONS

The purpose of this study was to investigate the effects of $\tau_{sr}$ and $\tau_p$ on $\mu_a$ values determined from photoacoustic-based measurements. For one representative set of experimental conditions, theoretical computation suggested that $\tau_p \approx 0.1 \tau_{sr}$ for optimal determination of $\mu_a$. Since it is difficult to achieve such conditions at midinfrared wavelengths due to the relatively long $\tau_p$ of commercial lasers and high $\mu_a$ of water, we developed a numerical deconvolution scheme to overcome this limitation of conventional photoacoustics. Use of the deconvolution-enhanced technique resulted in up to a 30% improvement in photoacoustic-based estimates of $\mu_{a,\text{med}}$. Future studies are planned to apply this technique to measurements of stress signals from biological hard and soft tissue.

ACKNOWLEDGMENTS

The authors thank Michelle Baltz, Mike Papantonakis, Stephen Uhlinhorn, Dr. Richard Haglund, and Dr. Bill Gabella for their assistance with the FEL setup; and Dr. Alexander Oraevsky and Dr. Alexander Karabutov for their instruction on TRSD measurements. Funding for this research was provided in part by grants from the Air Force Office of Scientific Research through MURI from DDR&E (F49620-98-1-0480), the Texas Higher Education Coordinating Board (BER-ATP-253), the Albert and Clemmie Caster Foundation, and the “Medical Free Electron Laser Beam Delivery for Human Care” part of the “Medical & Materials Research with Free Electron Laser,” ONR Contract No. N00014-94-1-1023.

1 J. W. Pickering, S. A. Prahl, N. van Wieringen, J. F. Beek, H. J. C. M. Sterenborg, and M. J. C. van Gemert, Appl. Opt. 32, 399 (1993).
2 S. A. Prahl, M. J. C. van Gemert, and A. J. Welch, Appl. Opt. 32, 559 (1993).
3 E. Milner, D. M. Goodman, B. S. Tanenbaum, and J. S. Nelson, J. Opt. Soc. Am. A 12, 1479 (1995).
4 S. A. Prahl, I. A. Vitkin, U. Bruggemann, B. C. Wilson, and R. R. Anderson, Phys. Med. Biol. 37, 1203 (1992).
5 A. A. Oraevsky, S. L. Jacques, and F. K. Tittel, Appl. Opt. 36, 402 (1997).
6 J. A. Viator, S. L. Jacques, and S. A. Prahl, IEEE J. Sel. Top. Quantum Electron. 5, 989 (1999).
7 G. M. Hale and M. R. Querry, Appl. Opt. 12, 555 (1973).
8 K. Becker, J. B. Johnson, and G. Edwards, Rev. Sci. Instrum. 65, 1496 (1994).
9 J. M. Khosrofian and B. A. Garetz, Appl. Opt. 22, 3406 (1983).
10 W. H. Press, B. P. Flannery, S. A. Teukolsky, and W. T. Vetterling, Numerical Recipes in C (Cambridge University Press, Cambridge, 1988).
11 K. F. Chan, B. Choi, J. M. H. Teichman, and A. J. Welch, Lasers Surg. Med. Suppl. 12, 49 (2000).
12 B. Choi, K. F. Chan, and A. J. Welch, Lasers Surg. Med. Suppl. 12, 4 (2000).
13 B. Choi, Ph.D. dissertation, The University of Texas at Austin, 2001.
14 K. P. Kostli, M. Frenz, H. P. Weber, G. Paltauf, and H. Schmidt-Kloiber, J. Appl. Phys. 88, 1632 (2000).
15 G. Ku and L. V. Wang, Med. Phys. 27, 1195 (2000).