Photoacoustic and ultrasound imaging with a gas-coupled laser acoustic line detector

Jami L Johnson[a], Kasper van Wijk[a], James N Caron[b], and Miriam Timmerman[c]

[a]The Dodd-Walls Centre for Photonic and Quantum Technologies and Department of Physics, University of Auckland, 38 Princes Street, Auckland, New Zealand
[b]Research Support Instruments, 4325-B Forbes Boulevard, Lanham, MD 20706, USA
[c]Department of Biomechanical Engineering, University of Twente, Postbus 217, 7500 AE Enschede, The Netherlands

ABSTRACT

Conventional contacting transducers are highly sensitive and readily available for ultrasonic and photoacoustic imaging. On the other hand, optical detection can be advantageous when a small sensor footprint, large bandwidth and no contact are essential. However, most optical methods utilizing interferometry or Doppler vibrometry rely on the reflection of light from the object. We present a non-contact detection method for photoacoustic and ultrasound imaging– termed Gas-Coupled Laser Acoustic Detection (GCLAD) – that does not involve surface reflectivity. GCLAD measures the displacement along a line in the air parallel to the object. Information about point displacements along the line is lost with this method, but resolution is increased over techniques that utilize finite point-detectors when used as an integrating line detector. In this proceeding, we present a formula for quantifying surface displacement remotely with GCLAD. We will validate this result by comparison with a commercial vibrometer. Finally, we will present two-dimensional imaging results using GCLAD as a line detector for photoacoustic and laser-ultrasound imaging.

Keywords: photoacoustic imaging, laser-ultrasound, beam deflection, gas-coupled laser acoustic detection, ultrasonics

1. INTRODUCTION

Non-contact acoustic detection methods for photoacoustic and laser-ultrasound imaging are of interest for applications where contact with the sample could cause harm, or access to the sample is desired. Gas-Coupled Laser Acoustic Detection (GCLAD) is a non-contact detection method that was first demonstrated for materials evaluation.[1] Here, we explore the use of this method as a line detector for medical photoacoustic and laser-ultrasound imaging.

Interferometers,[2] Fabry-Pérot detectors,[3] Fiber Bragg sensors,[4] and micro-ring resonators[5] are examples of optical detectors that measure acoustic waves at a focused point on the sample surface. Most optical methods rely on a reflection of light from the object, therefore a reflective medium is required. GCLAD detects acoustic waves through air, and there is no need for a reflective layer – allowing for a completely non-contacting imaging system. GCLAD is a line detector, thus, information at each point along the line is lost. Nonetheless, when the line is much larger than the target, reconstruction algorithms for line detectors are more accurate than those for point detectors that poorly approximate a point, such as transducers.[6]

In this proceeding, we present a formula for quantifying measurements of surface displacements from waves detected with GCLAD in the surrounding air. Calibration scans for obtaining the attenuation coefficient and speed of sound of the air are outlined, as well as methods for determining a calibration constant for the detector electronics. Next, we compare GCLAD to a commercial optical point detector in order to validate the amplitude quantification. Finally, two-dimensional photoacoustic imaging of an absorber on the order of the size of major arteries is presented.

Further author information: (Send correspondence to J.L.J.)
J.L.J.: E-mail: jami.johnson@auckland.ac.nz, Telephone: +64 (0)9 373 7599 ext 88747

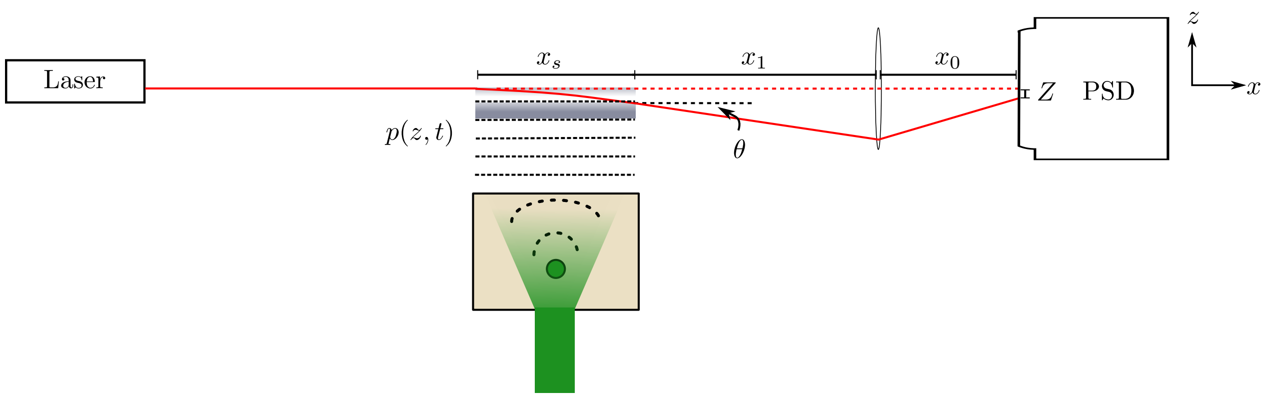

Figure 1. Setup for bias-free gas-coupled laser acoustic detection (GCLAD). A photoacoustic wave is generated inside of tissue and continues to propagate into the air as a plane wave. The acoustic wave travels perpendicular to the GCLAD probe beam, causing it to deflect upon interaction. The deflected wave is detected with a position-sensitive detector (PSD). A lens is placed one focal length from the PSD to remove the dependence of Z on x_1.

2. GAS-COUPLED LASER ACOUSTIC DETECTION

When a pulse of light is rapidly absorbed by a tissue chromophore, an acoustic wave is generated. This generation can occur at the surface (laser-ultrasound, LU) or deep in the tissue (photoacoustic, PA). These waves continue to travel through the tissue and displace the surface upon incidence, compressing the volume of the surrounding air. The air pressure is proportionally increased, and the waves are propelled into the air.

GCLAD probes the pressure variations of a propagating acoustic wave in the air with an optical beam perpendicular to the direction of acoustic propagation (Fig. 1). The pressure gradient across the optical beam causes the beam to deflect, and we record this deflection in real-time with a position-sensitive detector (PSD). In order to remove the dependence of the measured signal on the location of interaction, a convex lens is placed one focal length from the PSD. With the lens included, a formula relating the displacement of the probe beam at the position-sensitive detector (Z) and the displacement of the sample surface (δ) has been derived:[7]

$$Z = \frac{x_0 x_s}{2} \frac{(n_0 - 1)}{n_0} \delta e^{-\alpha z} (k^2 \cos(kz - \omega t) - \alpha k \sin(kz - \omega t)), \tag{1}$$

where α is the attenuation coefficient in air, z is the distance between the probe beam and sample surface, $\omega = 2\pi\nu$ (where ν is the frequency of the waveform), $k = \frac{\omega}{c}$, and x_s, x_1 and x_0 are the distances labeled in Fig. 1.

3. EXPERIMENTAL VALIDATION OF GCLAD QUANTIFICATION

In this section, we explore the validity of Eq. 1 for quantifying surface displacement by comparing a GCLAD measurement to a line of point-measurements recorded by a commercial laser-Doppler vibrometer (OFV-505, Polytec, Irvine, CA, USA). First, we obtain the attenuation coefficient and speed of sound of the air with a laser-ultrasound (LU) experiment in transmission mode (Fig. 2). A tissue phantom of 1% Intralipid (Fresenius-Kabi, Uppsala, Sweden) and 1% agar gel (A0930-05, U.S. Biological, Swampscott, MA, USA) is created with a tape with an unknown absorption coefficient placed over the surface of the sample. An unfocused 1064 nm Nd:YAG laser (Quanta-Ray, Spectra Physics, Newport, Irvine, CA, USA) with a fluence of $150\,\mathrm{mJ/cm^2}$ is incident on the tape, generating a sound wave that travels through the phantom. The LU waves are detected with GCLAD every 1 mm up to 10 cm from the detection surface. The speed of sound of the air is found by applying a linear regression relating the time-of-arrival of the LU waves to the distance between the sample and GCLAD beam. The slope of this line is the acoustic speed in the air, $c = 344\,\mathrm{m\,s^{-1}}$. The amplitude of each successive wave is found to decrease exponentially with an attenuation coefficient of $0.6\,\mathrm{dB\,cm^{-1}}$ (Fig. 2); however, the attenuation coefficient has been shown to be dependent on the frequency of the acoustic waves.[7]

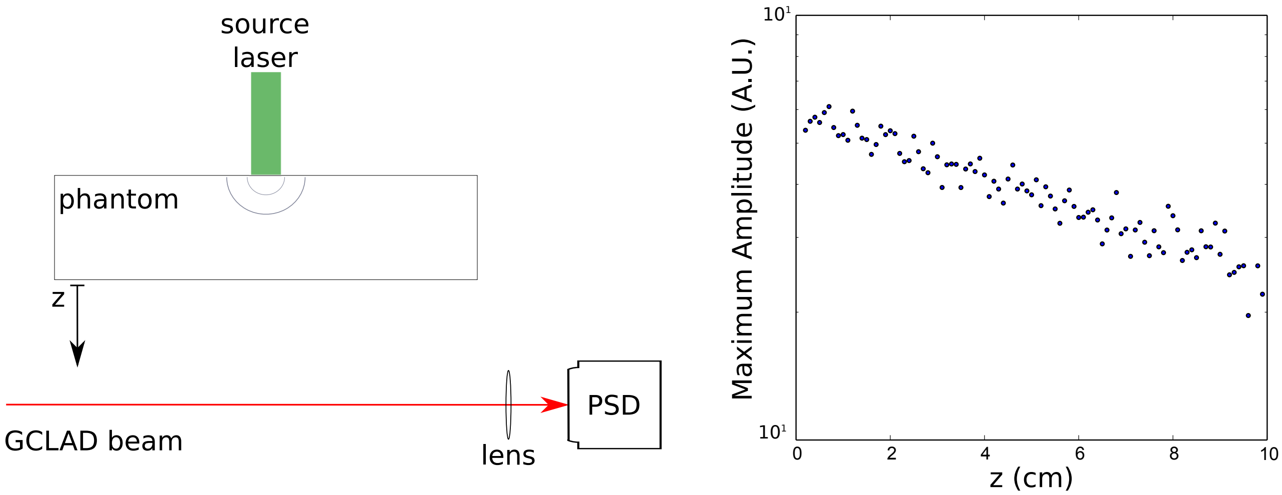

Figure 2. Experiment to measure the attenuation coefficient and speed of sound in air using GCLAD. Left: laser-ultrasound waves are generated on the surface of a tissue phantom and recorded every 1 mm up to 10 cm from the detection surface. Right: The amplitude of the direct wave decreases exponentially with distance between the sample surface and detector.

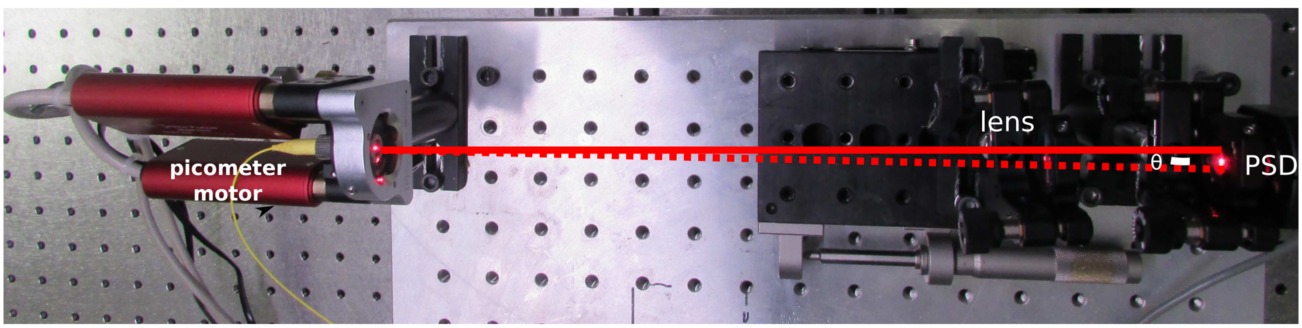

Figure 3. Setup for calibration scan of position-sensitive detector. The probe beam is rotated through an angle θ with high-precision actuators and the corresponding voltage from the detector is recorded.

Second, we calibrate the position-sensitive detector. High-precision actuators (8310, NewFocus, Newport, Irvine, CA, USA) are used to rotate the probe beam through an angle of 720 µrad at 3.6 µrad increments (Fig. 3). The voltage from the position-sensitive detector is recorded at each increment. In the paraxial range, a calibration coefficient of 110 mV μm^{-1} is found, which is used to convert the signal recorded at the detector to the displacement of the probe beam, Z.

Finally, an LU experiment is performed (Fig. 4) to compare GCLAD measurements to a commercial vibrometer. Laser-ultrasound waves are generated on a tissue phantom composed of 1% Intralipid, 1% agar, and 0.35% India ink with a 1064 nm laser (8 mm beam diameter, 100 mJ/cm^2 fluence). A single GCLAD waveform is recorded (64 averages). The average of 64 vibrometer waveforms is recorded at 161 positions across 4 cm of the surface (250 µm increments). The average of the 161 vibrometer waveforms and the GCLAD waveform are shown in Fig. 4. A correlation coefficient of 0.94 is found between the two measurements for the time window shown.

4. TWO-DIMENSIONAL IMAGING

To demonstrate the feasibility of imaging in two-dimensions with GCLAD, a photoacoustic imaging experiment is performed (Fig. 5). A 1% Intralipid and 1% agar tissue phantom is created with a tube embedded 1.2 cm deep (5.6 mm diameter, 25.4 µm wall). The tube is filled with infrared absorbing dye with an absorption coefficient of $\sim$10 cm^{-1} at 1064 nm (Epolight 2057, Epolin, Newark, NJ, USA). A collimated, unfocused Nd:YAG laser beam

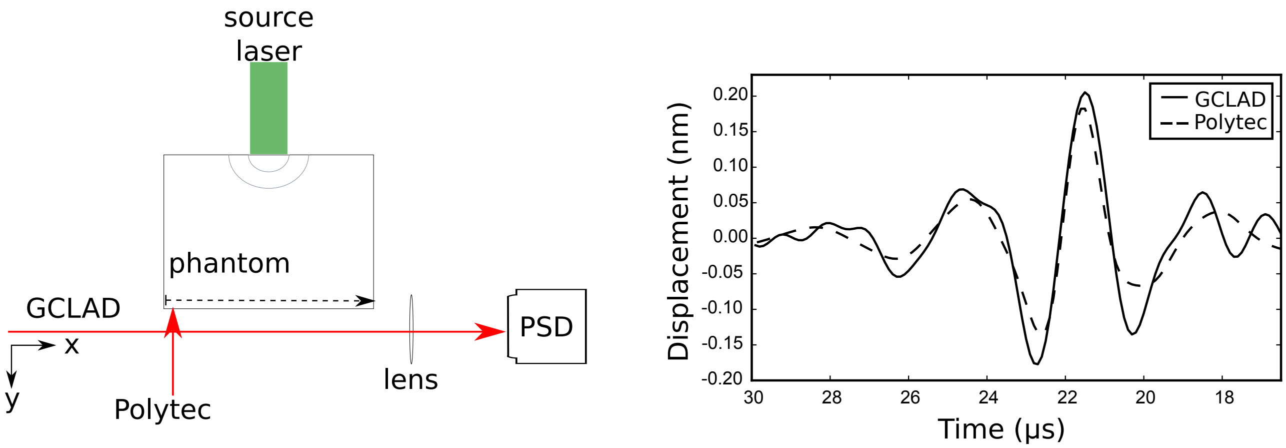

Figure 4. Left: Experimental setup for laser-ultrasound experiment to compare measurements made with a commercial vibrometer (Polytec) with a GCLAD measurement. A single GCLAD waveform is recorded and 161 Polytec waveforms are recorded at 250 μm increments along the surface, annotated with the dashed line. Right: The average of the Polytec measurements is compared to a GCLAD measurement.

with an 8 mm diameter and 100 mJ/cm^2 fluence is incident on the phantom surface. The longitudinal axis of the GCLAD beam is aligned with the source beam on the opposite side of the phantom with a 5.8 cm air gap between the detector and phantom. A two-dimensional scan is performed by translating the phantom in the x-direction with a high-resolution linear stage (M-IMS300LM, Newport, Irvine, CA, USA). The average of 64 waveforms is recorded at 250 μm increments, covering a total of 8.5 cm.

The time-travel of the acoustic waves through air is removed and time reversal is used to reconstruct the photoacoustic image.[8,9] The top and bottom edges of the tube are highlighted, due to the relatively large photoacoustic source, non-uniform light distribution, and strong acoustic contrast between the tube wall and phantom material.

5. CONCLUSIONS

In this proceeding, we have presented a formula for quantifying surface displacement with GCLAD measurements. This result was validated experimentally using calibration scans of a GCLAD system to obtain the attenuation coefficient and speed of sound of the air, and the voltage-to-displacement calibration constant applied to Eq. 1. The GCLAD measurement compared well to the average of a line of point-measurements recorded by a commercial laser-Doppler vibrometer that relies on a highly-reflective surface (R = 0.94).

Subsequently, we demonstrated the feasibility of two-dimensional imaging of a 5.6 mm diameter absorber embedded 1.2 cm deep in a tissue phantom using GCLAD with a large air-gap (5.8 cm). These results show promise for a photoacoustic imaging system with GCLAD for applications where a non-contacting system is desirable.

REFERENCES

[1] Caron, J. N., Yang, Y., Mehl, J. B., and Steiner, K. V., "Gas-coupled laser acoustic detection at ultrasonic and audio frequencies," *Review of Scientific Instruments* **69**(8), 2912–2917 (1998).

[2] Speirs, R. W. and Bishop, A. I., "Photoacoustic tomography using a Michelson interferometer with quadrature phase detection," *Applied Physics Letters* **103**, 053501 (2013).

[3] Zhang, E., Laufer, J., and Beard, P., "Backward-mode multiwavelength photoacoustic scanner using a planar Fabry-Perot polymer film ultrasound sensor for high-resolution three-dimensional imaging of biological tissues," *Applied Optics* **47**(4), 561–577 (2008).

[4] Rosenthal, A., Razansky, D., and Ntziachristos, V., "High-sensitivity compact ultrasonic detector based on a pi-phase-shifted fiber Bragg grating," *Optics Letters* **36**(10), 1833–1835 (2011).

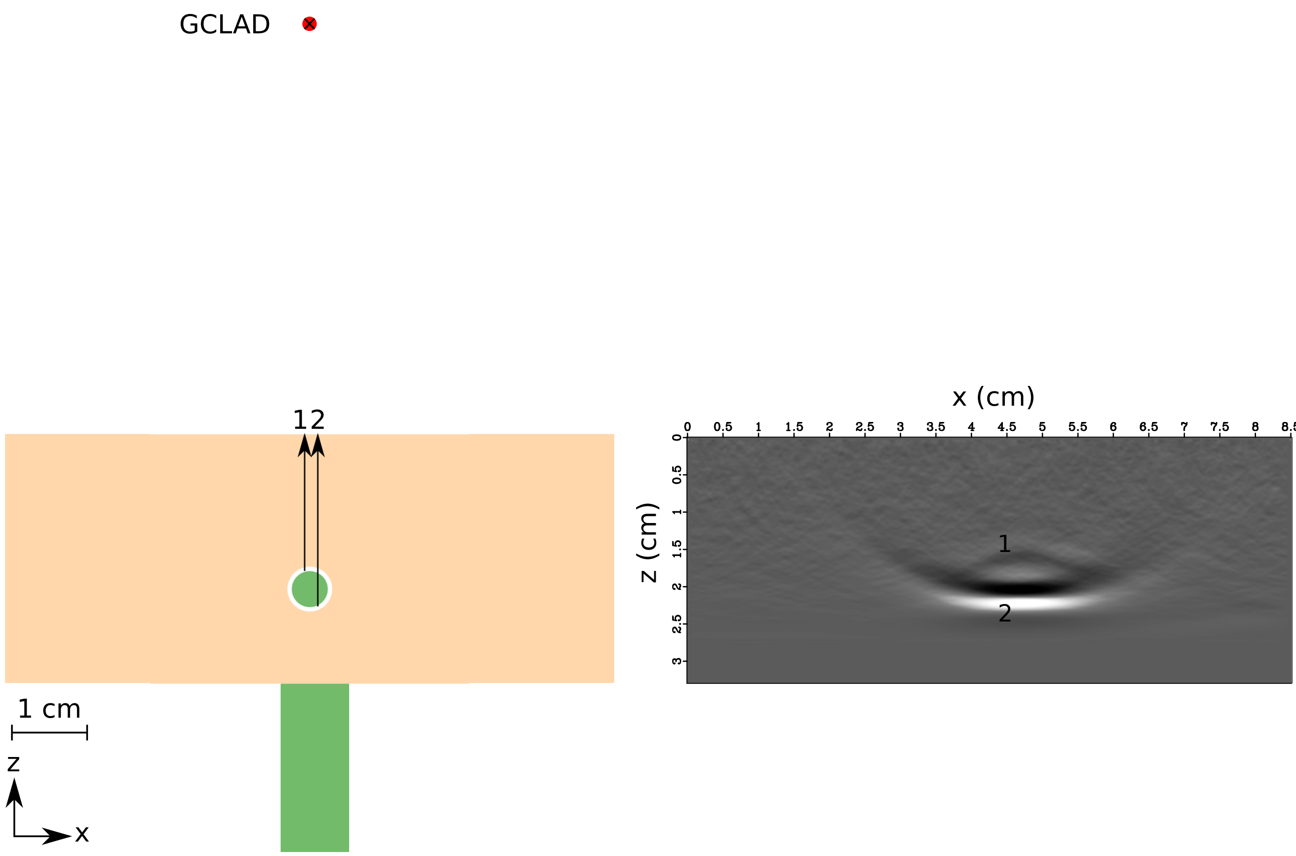

Figure 5. Left: setup for two-dimensional photoacoustic imaging experiment. A 5.6 mm diameter tube filled with absorbing dye is embedded 1.2 cm deep. The source laser and GCLAD are arranged in transmission mode. The phantom is translated in the x-direction and the resulting photoacoustic waves are detected at each 250 µm increment along the surface. The ray paths for photoacoustic waves generated at the top and bottom edges of the tube are denoted with arrows. Right: photoacoustic image reconstructed with time-reversal. The top and bottom edges of the tube are highlighted due to non-uniform light distribution and the strong acoustic contrast of the tube walls.

[5] Ling, T., Chen, S.-L., and Guo, L. J., "Fabrication and characterization of high Q polymer micro-ring resonator and its application as a sensitive ultrasonic detector," *Optics Express* **19**(2), 861–869 (2011).

[6] Paltauf, G., Nuster, R., Haltmeier, M., and Burgholzer, P., "Experimental evaluation of reconstruction algorithms for limited view photoacoustic tomography with line detectors," *Inverse Problems* **23**(6), S81–S94 (2007).

[7] Johnson, J. L., van Wijk, K., Caron, J. N., and Timmerman, M., "Gas-coupled laser acoustic detection as a non-contact line detector for photoacoustic and ultrasound imaging," *Journal of Optics* **18**(2), 024005 (2016).

[8] Shragge, J., Blum, T. E., van Wijk, K., and Adam, L., "Full-wavefield modeling and reverse time migration of laser ultrasound data: A feasibility study," *Geophysics* **80**(6), D553–D563 (2015).

[9] Johnson, J. L., Shragge, J., and van Wijk, K., "Image reconstruction of multi-channel photoacoustic and laser-ultrasound data using reverse time migration," in [*SPIE BiOS*], 932314, International Society for Optics and Photonics (2015).