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Photoacoustics with coherent light

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Abstract

Since its introduction in the mid-nineties, photoacoustic imaging of biological tissue has been one of the fastest growing biomedical imaging modality, and its basic principles are now considered as well established. In particular, light propagation in photoacoustic imaging is generally considered from the perspective of transport theory. However, recent breakthroughs in optics have shown that coherent light propagating through optically scattering medium could be manipulated towards novel imaging approaches. In this article, we review the recent works showing that it is also possible to exploit the coherence of light in conjunctions with photoacoustics. We illustrate how the photoacoustic effect can be used as a powerful feedback mechanism for optical wavefront shaping in complex media, and conversely show how the coherence of light can be exploited to enhance photoacoustic imaging. Finally, we discuss the current challenges and perspectives down the road towards practical applications in the field of photoacoustic imaging.

Keywords: Photoacoustic imaging, Coherent light, Multiple scattering, Speckle Illumination, Optical wavefront shaping

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1. Introduction

Photoacoustic imaging of biological tissue is a fast developing multi-wave imaging modality that couples light excitation to acoustic detection, via the photoacoustic effect, to yield images of optical absorption [1, 2, 3, 4]. The photoacoustic effect consists in light absorption followed by acoustic emission, via thermo-elastic stress-generation. It was first used in the field of optical absorption spectroscopy, and has been introduced for biomedical applications in the mid-90s [5, 6, 7]. The general principle of photoacoustic imaging is the following: the sample to be imaged is illuminated by pulsed light (for most implementations), and acoustic waves generated from illuminated absorbing regions are detected by acoustic sensors. Depending on the situation, the resolution can be limited either by the acoustic or by the optical wavelength. Photoacoustic imaging was first developed for deep tissue optical imaging in the so-called acoustic-resolution regime, to overcome the loss of optical resolution caused by optical scattering. Due to multiple scattering of light in biological tissue, optical-resolution imaging based on ballistic

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light is limited to depths typically less than one millimeter, while the resolution of technique based solely on multiply scattered light (such as Diffuse Optical Tomography) is on the order of the imaging depth. On the other hand, ultrasound is very weakly scattered in biological tissue, and therefore photoacoustic waves can be used to reconstruct images of optical absorption with the resolution of ultrasound, which inversely scales with tissue thickness. The resolution and penetration depth of deep tissue photoacoustic imaging is ultimately limited by the attenuation of light and sound. In the spectral region 600-900 nm, the so-called “optical window” where absorption is minimal in tissues, the amount of multiply scattered light decreases exponentially with an effective attenuation length of about 1 cm. The acoustic attenuation in tissue increases linearly with frequency, with a typical value of 0.5 dB cm$^{-1}$ MHz$^{-1}$. As a consequence, the penetration depth of photoacoustic imaging scales linearly with the acoustic-resolution, with a maximum depth-to-resolution ratio of about 200 [1, 4]. Another regime of photoacoustic imaging is optical-resolution photoacoustic microscopy, for which light is focused and raster-scanned over the sample to make a point-by-point photoacoustic image with a resolution given by the optical spot size [9]. This regime is only possible at shallow depth, where ballistic light is still present and can be focused to the optical diffraction limit. Over both the optical- and acoustic-resolution regimes, the depth-to-resolution ratio of photoacoustic imaging is typically in the 100-200 range, a combined consequence of both optical and ultrasound attenuation. Because multiple scattering of light is an inescapable process during the propagation of light in complex media such as biological tissue, it has long been considered a nuisance to get rid of. In the last decade, it has however been demonstrated that it could actually be exploited for optical imaging at unprecedented depth. This blooming field of research leveraged on the coherence properties of multiple scattered light (the optical speckle [10], sec. 2.3) and the possibility to control such properties thanks to the manipulation of light impinging on the medium: optical waveform shaping has allowed focusing and imaging at optical resolution through strongly scattering materials [11] (sec. 2.4). In the field of photoacoustics, imaging, up until recently, light has usually been considered from the sole point of view of the absorption of optical energy. Lasers have therefore been widely used as powerful and flexibles sources of light energy. In optical-resolution microscopy, their spatial coherence was the necessary condition to focus them to a diffraction spot. However, coherence properties of lasers also provide specific properties for multiple scattering, at the core of phenomena such as the formation of optical speckle patterns, and offer the possibility of manipulating scattered light with optical waveform shaping. This paper reviews the recent research efforts led over the past few years that exploit and take advantage of the photoacoustic effect in conjunction with coherent illumination in the multiple scattering regime. We first introduce general concepts regarding both photoacoustics and light propagation in scattering media (Sec. 2), which will be extensively used in the rest of the paper. The following sections then review the use of the photoacoustic effect as a feedback mechanism for optical wavefront shaping (Sec. 3) and how coherent light may enhance photoacoustic imaging with speckle illumination or optical wavefront shaping (Sec. 4). We finally discuss the current limitations and envision some perspectives in the field.

2. Background

2.1. Photoacoustics: from light absorption to sound generation

In the context of photoacoustic imaging of soft biological tissue, one of the simplest and widely used theoretical description of the photoacoustic effect can be summarized by the following equation [12, 3]

$$\rho(r,t) = I(r,t) \frac{\partial H}{\partial t}(r,t)$$

where $\rho(r,t)$ is the photoacoustic pressure field, and $H(r,t)$ is a heating function that corresponds to the thermal energy converted from optical absorption, per unit volume and time per unit time. Eq. (1) assumes that the medium is acoustically and thermally homogeneous (with $c_0$ the speed of sound and $I$ the Gruneisen coefficient [3]), while the optical properties of the medium (hence $H$) may vary spatially. It also assumes that thermal diffusion may be neglected over the spatial and temporal scales of interest (i.e. heat-confinement assumption [3]), which is usually true for most situations encountered in photoacoustic imaging and will be considered fulfilled in this paper. This equation simply states that the heating following (optical) absorption appears as a source term in the acoustic wave equation, and therefore leads to the generation and propagation of acoustic waves. $H(r,t)$ is proportional to the optical intensity $I(r,t)$, with some coefficient representative of the optical absorption. Importantly, time $t$ in Eq. 1 refers to the time evolution of the optical intensity, which by definition is proportional to the square of the electric field averaged over a few optical periods. Using the complex notation for electric fields with slowly time-varying envelopes, the optical intensity may be written as $I(r,t) \propto |E(r,t)|^2$, where the proportionality constant reflects local dielectric properties. In strongly scattering media such as biological tissue, there is no simple description for $I(r,t)$ and $E(r,t)$. The propagation of the electric field may be described by Maxwell’s equations in which material properties strongly vary in space, with scattering caused by local variations of the index of refraction. While it is impossible in practice to obtain a full description of $E(r,t)$ at the microscopic level, due to the very complex propagation process, light propagation in multiply scattering media may however be
described with statistical approaches, further discussed in the following sections.

2.2. Light transport in multiple scattering media

The most widely used approach to model light propagation for the photoacoustic imaging of biological tissue is based on transport theory. In this theory, the physical quantity of interest is the ensemble averaged optical intensity, or fluence rate, that describes the flux of the optical energy. Depending on the desired accuracy and scales of interest, several approaches may be used to describe the flux of optical energy with a transport approach. Numerical approaches include Monte-Carlo simulations of random walks used to describe the paths followed by the optical energy [13], and analytical models include the radiative transfert equation or the diffusion equation [14]. These approaches all have in common to describe the propagation of the optical energy based on scattering and absorption, defined as macroscopic values such as the absorption coefficient $\mu_a$ and the scattering coefficient $\mu_s$. The simplest form of the transport theory is given by the following diffusion equation [14]

$$\frac{1}{c} \frac{\partial}{\partial t} + \frac{1}{3(\mu_s + \mu_a)} \nabla^2 \phi(r, t) = -\mu_a(r) \phi(r, t) \tag{2}$$

where $\phi(r, t)$ is the optical fluence rate, defined as the energy flux per unit area per unit time regardless of the flux direction. Eq. 2 states that the fluence rate obeys a classical diffusion equation, with a loss term that reflects optical absorption, and a diffusion coefficient $D = \frac{1}{3(\mu_s + \mu_a)}$ that only depends on scattering and absorption. In $D$, $\mu_s$ is the reduced scattering coefficient, defined as $\mu_s' = \mu_s(1 - g)$ where $g$ reflects the scattering anisotropy [14]. The transport mean free path $\lambda = 1/\mu_s'$ and the absorption length $\lambda_a = 1/\mu_a$ are also often used as the spatial scales relevant respectively for multiple scattering and absorption.

In biological tissue in the near infrared (the "optical window"), $\lambda$ and $\lambda_a$ are of the order of 1 mm and 10 cm respectively [15]. Eq. (2) can be derived from the radiative transfert equation (RTE), which is a more elaborate (and large scale) description of the energy transport based on the radiance $L(r, s, t)$, i.e. a quantity that takes into account the direction $s$ of the energy flux. It is out of the scope here to discuss the RTE (further details may be found in [14] for instance), but suffice it to mention that the fluence rate (that obeys Eq (2) under the diffusion approximation) is defined from the radiance by $\phi(r, t) = \int L(r, s, t) ds$.

Under the assumption that light propagation may be described by the transport theory, the fluence rate is the important physical quantity for photoacoustic imaging as the heating function $H(r, t)$ may be readily expressed as

$$H(r, t) = \mu_a(r) \phi(r, t) \tag{3}$$

When the fluence rate $\phi(r, t)$ may be decomposed as $\phi(r, t) = \Phi(r) f(t)$, the following widely used form of the photoacoustic wave equation is obtained:

$$\left[ \frac{\partial^2}{\partial t^2} - c_a^2 \nabla^2 \right] p(r, t) = \Gamma \mu_a(r) \Phi(r) \frac{\partial f(t)}{\partial t} \tag{4}$$

In most practical implementations of photoacoustic imaging, $f(t)$ is a pulsed function (normalized such that $\int f(t) dt = 1$), and $\mu_a(r) \Phi(r)$ is the amount of absorbed energy per unit volume. For very short pulse (such as to verify the so-called stress-confinement condition [3, 16]), it can be shown that the forward photoacoustic problem described by Eq. (2.1) may be re-formulated as a source-free initial value problem, with an initial condition given by

$$p(r, t = 0) = p_0(r) = \Gamma \mu_a(r) \Phi(r) \tag{5}$$

The stress-confinement condition is fulfilled when the pulse duration is much longer that the characteristic acoustic propagation time within the medium, which for nanosecond pulses is verified with absorbers with typical dimensions larger than a few micrometers. The solution $p(r, t)$ corresponding to pulses $f(t)$ with finite duration may be obtained straightforwardly from the temporal convolution of the solution to Eq. (5) with $f(t)$. This formulation shows that the appropriate resolution of the inverse problem based on the measurements of pressure waveforms provides a reconstruction of $\mu_a(r) \Phi(r)$. In other words, under the stress-confinement assumption, the initial pressure build-up is proportional both to the local absorption and to the local fluence.

Although the formulation of the photoacoustic effect based on Eqs. (3) and (4) is one of the most widely used in photoacoustic imaging, it is inherently limited to situations where the propagation of light may be described appropriately by use of the light fluence $\phi(r)$. While such situations are indeed the most commonly encountered in photoacoustic imaging, there however exist situations where the light fluence is not appropriate to describe phenomena of interests. As may be demonstrated from rigorous derivations of the diffusion equation from first principles in disordered media [17, 18], $\Phi(r)$ corresponds to a theoretically averaged value of the optical intensity, averaged over an ensemble of realizations of the disorder. In practice where experiments are performed with one given medium (one realization), a good approximation to $\Phi(r)$ is a spatial average of the optical intensity $I(r, t)$ over a volume with typical linear dimensions of the order of a few wavelengths. As a consequence, $\Phi(r)$ is a physical quantity that does not take into account higher-order spatial correlations of the optical field. In particular $\Phi(r)$ does not take into account phenomena such as speckle patterns that exist when interference takes place between various propagation paths followed by sufficiently coherent light, as introduced in the following section.

2.3. Optical speckle

Definition. The phenomenon commonly called "speckle" refers to the granular structure of the intensity field $I(r, t)$.
that results from the seemingly random interference of a multitude of field amplitudes from different propagation paths [19, 20, 10]. Speckle patterns are observed in various configurations, including scattering by rough surfaces, propagation through scattering media and propagation inside multiple scattering media such as biological tissue. A typical speckle pattern is shown in Fig. 5. As further discussed below, speckle patterns are only observed when the light source has sufficient temporal and spatial coherence.

Mathematically, the intensity at a given point \( I(r, t) \) may be written as a sum of a large number of complex amplitudes contributions as

\[
I(r, t) \propto \left| \sum_{\text{path } i} A_i(r, t) e^{i\phi_i(r, t)} \right|^2
\]  

\( (6) \)

**Properties of an ideal speckle**. We first consider the ideal case of perfectly coherent (monochromatic) light with angular frequency \( \omega_0 \) that has undergone multiple propagation paths. Under this assumption, the intensity at a given point is stationary with \( I(r) \propto \left| \sum_{\text{path } i} A_i(r) e^{i\phi_i(r)} \right|^2 \).

We further consider the case of a fully-developed speckle, i.e. the phases \( \{\phi_i\} \) are uniformly distributed over \([0; 2\pi]\), which has extremely well defined statistical properties. The first-order statistics of a fully-developed speckle field is described by the distribution of its intensity, which obeys the following negative exponential statistics [20, 10]

\[
p_l(I) = \begin{cases} \frac{1}{I} \exp \left( -\frac{1}{I} \right) & I \geq 0 \\ 0 & \text{otherwise} \end{cases}
\]  

\( (7) \)

An important properties of the above probability distribution is that its standard deviation \( \sigma_I \) is equal to its mean \( I \). As a consequence, fully developed speckle patterns have a contrast \( \sigma_I/I \) = 1. While this probability function refers to an ensemble statistics over realizations of different order, it is often realistic in practice to assume ergodicity and to consider that this ensemble statistics also describes the statistics over spatial position in the speckle field. This statement is an example of a simple though fundamental fact of multiply scattered coherent light which is discarded by the transport theory: a homogeneous speckle field \( p_l \) independent of \( r \) translates into a constant fluence rate \( \Phi = \langle I \rangle \) in the transport theory (N.B. The fluence rate is also often called accordingly the optical intensity, although it represents only an averaged intensity strictly speaking). A useful property of speckle is that the addition of \( N \) uncorrelated speckle intensity patterns will result in a speckle with a reduced contrast of \( 1/\sqrt{N} \) [19].

As a consequence, with spatially or temporally incoherent illumination, the intensity distribution is smoothed toward the mean intensity value from the transport theory. Furthermore, the analysis of the spatial autocorrelation of a stationary speckle pattern provides the typical dimensions of a speckle “grain”, another major property of a speckle, which depend on the considered geometry. Two most configurations are of particular interest in the context of this review. The first one is a free-space propagation geometry, which corresponds for instance to the observation at some distance of the scattering by a rough surface or propagation through a scattering layer. In this case, the typical transverse linear dimension of a speckle grain is given by [10]:

\[
\phi_s \sim \frac{\lambda}{D}
\]  

\( (8) \)

where \( \lambda \) is the optical wavelength, \( z \) is the distance from the scatterer to the transverse imaging plane, \( D \) is the typical linear dimension of the illuminated surface of the scattering object. The exact value of the numerical prefactor (close to one) in the expression above Eq. (8) depends on the illumination distribution on the scattering object. Along the main direction of propagation, the typical longitudinal dimension is given by [10]:

\[
l_s \sim \frac{7\lambda}{D}
\]  

\( (9) \)

The exact value of the numerical prefactor in Eq. (9) depends on the illumination distribution on the scattering object. This prefactor is close to 7 for a circular aperture of diameter \( D \), close to 5 for a square of side \( D \). The dimensions given by the formulas (8) and (9) (valid only for small values of \( \frac{\lambda}{D} \)) are identical to those of the diffraction-limited focal spot of a lens with aperture \( D \) and focal distance \( z \). The second important situation for the speckle grain size is inside a multiply scattering medium. There, due to the fact that the speckle is formed from contributions from all directions, the speckle grain is isotropic, with a typical linear dimension dictated solely by the wavelength and given by

\[
\phi_s^{\text{inside}} \sim \frac{\lambda}{2}
\]  

\( (10) \)

which is also the dimension of a diffraction-limited focal spot obtained with a full 4\( \pi \) aperture.

The contrast value of 1 discussed above is in fact only true in a scalar model, i.e. for linearly polarized light. For fully polarized light undergoing multiple scattering [10], the polarization is also mixed [21]. In paraxial free-space configuration geometry with \( \frac{\lambda}{D} \ll 1 \), one can consider that there are two uncorrelated and fully developed speckle intensity patterns associated to two orthogonal polarizations in the imaging plane, which add up incoherently, and the resulting contrast is reduced to \( 1/\sqrt{2} \). Deep inside a multiple scattering medium, the 3-D speckle intensity results from the incoherent summation of the three possible polarizations and the contrast is further reduced to \( 1/\sqrt{3} \). Moreover, it has been assumed so far that the propagation medium is stationary, and that the speckle pattern is therefore stationary in time. For media whose properties may vary in time, such soft matter or biological tissue, the previous description of speckle patterns with monochromatic light remains valid provided that the intensity field are measured over integration time much smaller than any characteristic time of motion in the scattering medium.
Speckle with partially coherent light. A general condition to observe speckle patterns is that the coherence length of the light is larger than the largest path differences involved in the interference patterns. The coherence length $l_c$ may be defined as the maximum length difference between two different paths in order to still observe interference, and is a direct consequence of the coherence time $\tau_c$ of a light source ($l_c = c \times \tau_c$) [20]. The temporal coherence of a light source is related to the spectral linewidth $\Delta \nu$ of its spectral power density, with $\tau_c$ being proportional to $\frac{1}{\Delta \nu}$ (with a proportionality constant that depends on the shape of the linewidth). For a Lorentzian line, $\tau_c = \frac{1}{\pi \Delta \nu}$, and the coherence length is therefore

$$l_c = c \times \tau_c = \frac{c}{\pi \Delta \nu}$$  \hspace{1cm} (11)

If the coherence length is too short compared to the typical range of propagation paths, some of the partial waves corresponding to the terms in the summation in Eq. (6) cannot interfere coherently at position $r$, giving rise to incoherent sums of speckles, thus leading to a loss of contrast. If one considers light propagation through a slab of thickness $L$, the range of propagation paths in the multiple scattering regime (i.e. $L \gg l_c^*$) scales as $\frac{L^2}{l_c^*}$. As a consequence, a condition to obtain a well contrasted speckle pattern after the propagation of light with coherence length $l_c^*$ through a thickness $L$ of a multiply scattering media with transport mean free path $l^*$ is

$$l_c^* \gg \frac{L^2}{l^*}$$  \hspace{1cm} (12)

This condition may also be written in the time or frequency domain as $\tau_c \sim \frac{1}{\Delta \nu} \gg \frac{L^2}{l^*}$, where $\frac{L^2}{l^*}$ is the Thouless time [23], corresponding to the light storage time in the medium and to the temporal spreading of a light pulse after a diffusive propagation through a distance $L$. As an order of magnitude, the coherence length required to obtain a well contrasted speckle pattern inside or through $3\text{ cm}$ of biological tissue is typically $l_c \sim \frac{(3\text{ cm})^2}{1\text{ mm}} \sim 1\text{ m}$. For pulsed light, a coherence length $l_c \sim 1\text{ m}$ corresponds to a minimal pulse duration $\tau_p \sim \frac{l_c}{c} \sim 3\text{ ns}$. Therefore, whereas light coherence is generally neglected in photoacoustics, sufficiently coherent pulsed light does lead to coherent effects such as speckle patterns through or inside strongly scattering media. Before reviewing the recent investigations aimed at coupling photoacoustics and coherence effects, we briefly introduce the main principles of optical wavefront shaping in complex media, a field that has developed very rapidly over the past few years [11].

2.4. Optical wavefront shaping with multiply scattered light

Principles. Although multiple scattering may appear stochastic, as illustrated by the random appearance of speckle patterns, it is deterministic in nature. However, the deterministic propagation of coherent light through strongly scattering media is driven by a huge number of parameters that reflect the complex nature of the multiple scattering process. For instance, the speckle pattern that arises from an illumination area $A$ after propagation through a thick scattering medium is typically described by a number of parameters $N$ (often referred to as the number of modes) that scales as $\frac{2A \lambda^2}{\Delta \nu}$, which for visible light corresponds typically to 10 million modes per square millimetre [11]. As a consequence, it has long been thought that the techniques of adaptive optics (which involves measuring and controlling the phase and/or amplitude of the wavefronts of light with a given number of degrees of freedom (DOF)) were limited to situations where the distortions of optical wavefront could be described or compensated for with a relatively small number of modes, comparable to the number of DOF provided by the optical devices. The pioneering demonstration of spatial focusing through a strongly scattering layer by Vellekoop and Mosk [22] has however shown that adaptive optics could in fact be extended to situations where one controls only a limited numbers of DOF compared to the total number of mode involved in the propagation: it was demonstrated in this work that optical wavefront shaping with $N_{\text{DOF}}$ degrees of freedom allowed enhancing the intensity of a single speckle grain by a factor $\eta \propto N_{\text{DOF}}$ relatively to the intensity of

Figure 1: General illustration of optical wavefront shaping through a strongly scattering sample. (a) A coherent plane is multiply scattered through a strongly scattering sample, yielding a speckle pattern propagating in free-space to the observation plane. (b) Optical wavefront shaping of the incident wave allows focusing light through the scattering sample. (c) Experimental setup used to perform optical wavefront shaping in the pioneer experiment by Vellekoop and Mosk [22]. The values of the phase on each pixel of the spatial light modulator (SLM) were found one by one with an optimization algorithm based on a feedback signal measured on the camera (CCD). Figure reproduced with permission from [22], OSA.
each speckle grain in the diffuse background, while the ratio $N_{\text{DOF}}/N \ll 1$ only dictates the ratio of the intensity within the enhanced spot to the total transmitted intensity.

Schematics of the experiment performed by Vellekoop and Mosk [22] are shown in Fig. 1. The key principle at the core of this experiment is that the transmitted electric field $E_m$ in the CCD camera plane is a linear combination of the electric fields $E_m = A_ne^{i\phi_n}$ coming from the $N_{\text{DOF}}$ pixels of the spatial light modulator (SLM):

$$E_m = \sum_{n=1}^{N_{\text{DOF}}} t_{mn} A_ne^{i\phi_n}$$ (13)

where $A_n$ and $\phi_n$ are the amplitude and phase of the light reflected from the $n^{th}$ input pixel, and $t_{mn}$ is the complex transmission matrix between the transmitted (output) field and the SLM (input) field [22]. Optical wavefront shaping essentially consists in first measuring transmitted output values, followed by appropriately setting the phase and/or amplitude (depending on the type of control provided by the spatial light modulator) of the input field in order to obtain a targeted pattern in the output field. Several approaches have been investigated to implement optical wavefront shaping with strongly scattering media, based either on optimization or measurement of a transmission matrix, as discussed in the two following sections.

Optimization-based optical wavefront shaping. In their pioneering experiment, Vellekoop and Mosk [22] demonstrated focusing towards a single speckle grain by use of an optimization approach: with the typical dimension of speckle grains matched to that of the measurement pixel size, the phases $\phi_n$ of each input electric field $E_n$ corresponding to the $n^{th}$ mode were cycled sequentially from 0 to $2\pi$, and the phase values that maximized the intensity on a given pixel of the camera are recorded for each input mode. After this procedure, the phases of all the input modes are set simultaneously to their recorded optimal values, resulting in a strong constructive interference at the chosen speckle grain as all the terms $t_{mn}A_ne^{i\phi_n}$ are in phase [22], effectively forming a very strong focus. The authors were able to enhance the light intensity of a targeted speckle grain by a factor 1000 through a 10-μm thick layer of rutile (TiO$_2$) with a transport mean free path of 0.55 μm.

A key parameter when optimizing the light intensity is the dimension of the targeted detection area relatively to that of the speckle grain. When a number $N_s$ of speckle grains are contained within the targeted area, the intensity enhancement factor is typically divided by $N_s$ as the focusing is spread over the $N_s$ speckle grains, and therefore scales as $\eta \propto \frac{N_s}{N_{\text{DOF}}}$ [24]. Moreover, when the targeted area contains several speckle grains, the global effect of a phase modulation of a single input mode is decreased compared to that obtained for a single speckle grain, as the phases on each speckle grain are uncorrelated. As a consequence, the possibility to detect intensity modulation in the target region depends on the signal-to-noise ratio and decreases with the number $N_s$ of independent speckle grains in the detection area. Note that stemming from the initial work of Vellekoop and Mosk [22], several algorithms have been proposed, in order to improve the focusing efficiency in low SNR scenarios [25], to improve focusing speed [26, 27] or to adapt to different modulation schemes [28]. The main limitation of optimization approaches is that the whole optimization procedure has to be repeated for each desired target pattern, leading to very long measurement time in practice if several target patterns are required.

Wavefront shaping with the transmission matrix. Following the initial demonstration of optical wavefront shaping by use of an optimization approach, Popoff and coworkers demonstrated the first measurement with a strongly scattering layer of an optical transmission matrix $t_{mn}$ with over 60,000 elements [29]. To do so, the transmitted speckle patterns were measured over the camera plane for a set of orthogonal input modes that forms a full basis for all the possible SLM modes. As the camera records only the optical intensity, an interferometric approach was implemented to retrieve the phase and amplitude information from intensity measurements: an unshaped part of the beam reflected off an unmodulated region on the SLM is used as a reference beam, and the phase of each controlled SLM input mode is varied from 0 to $2\pi$ in order to retrieve the amplitude and phase of the matrix element. For each input mode $n$, the phases and amplitudes of the intensity modulations measured on all the output pixels of the camera provide a measurement of the column $t_{mn} = |t_{mn}|e^{i\phi_{mn}}$ of the transmission matrix. Repeating these measurements for all possible input mode provides the transmission matrix between the pixels of the camera and the pixels of the SLM. From the transmission matrix, one can predict the amplitude and phase required on each input mode in order to obtained any desired output pattern in the camera plane, via inversion or phase-conjugation of the transmission matrix [30]. As the simplest example, Eq. 13 shows that focusing light onto the $m^{th}$ output pixel may simply be obtained with a phase-only SLM by setting the phase of each $n^{th}$ input mode to $\phi_n = -\phi_{mn}$; doing so, all the terms in Eq. 13 end up in phase, resulting in focusing towards the $m^{th}$ output pixel. The result is therefore similar to that obtained with the optimization approach; however the key advantage of the transmission matrix approach is that once the transmission matrix is measured, input patterns may be computed for any desired transmitted pattern, while the optimization approach requires to measure the optimized input patterns for each desired output pattern.

Discussion. Wavefront shaping in biological tissues is currently a very active field of research. While this review focuses on photoacoustics-related works, other imaging modalities are explored in parallel, in particular mul-
3. Photoacoustic-guided optical wavefront shaping

All implementations of optical wavefront shaping require some feedback signal from the targeted region. A feedback mechanism for optical wavefront shaping should provide some sensing of the optical intensity. Appropriate mechanisms include direct intensity measurement with a camera or optical detector, or the use of some “guide star” following the approach in adaptive optics for astronomy [52]. While the use of a camera or detector limits wavefront shaping towards region outside the scattering media [22, 29, 31, 53], the “guide star” approach may be implemented inside a scattering sample. Fluorescent or second-harmonic “guide stars” have been successfully investigated as feedback mechanisms [34, 24], but these approaches, in addition to being invasive, only allows focusing in the vicinity of a single static target. Ultrasound tagging via the acousto-optic effect is a promising approach that offers dynamic and flexible control, which has been the subject of several recent investigations [55, 56, 57, 58, 59, 60, 37]. This approach has the advantage that it allows single shot digital phase conjugation, i.e., finding the optimal wavefront to refocus on the guide-star without a long learning process (like optimization or transmission matrix), and therefore refocusing in a single refresh frame of the spatial light modulator. This was for instance demonstrated by Liu and coworkers who demonstrated focusing in tissues with 5.6 ms decorrelation time [37]. Although this approach is in principle compatible with in vivo imaging, the activation of a local guide star by acoustic tagging is limited to a single ultrasound focal zone, and scanning is required to focused light at various direction, requiring in turns long acquisition times.

As introduced in Sec. 2.1, the photoacoustic effect is sensitive to the absorption of optical energy, and therefore provides a mechanism to sense both the optical absorption and the optical intensity inside multiple scattering media. Based on its sensitivity to optical absorption, photoacoustic-guided wavefront shaping has first been investigated for ultrasound wavefront shaping, to focus acoustic waves towards optical absorbers with time-reversal approaches [61, 62]. In 2011, Kong and coworkers first demonstrated the use of the photoacoustic effect as a feedback mechanism for optical wavefront shaping [51], triggering significant research efforts towards photoacoustic-guided optical wavefront shaping. Analogous to wavefront shaping with the other feedback mechanisms introduced above, two main approaches have been used to implement photoacoustic-guided optical wavefront shaping (PA-WFS), either based on optimization or transmission matrix, as reviewed in the two following sections.

3.1. Photoacoustic-guided optical wavefront shaping with optimization

In their pioneering work [51], Kong and coworkers followed the optimization approach initially proposed by Vellekoop and Mosk [22], as illustrated on Fig.2. The target plane consisted of a glass layer covered with graphite particles, placed behind the scattering layer. Different concentrations and types of absorbers were used to demonstrate photoacoustic-guided wavefront shaping: the au-
Interestingly, it was shown that with a speckle size of 1
for the photoacoustic signal ranged from 5 to 10, with the
for the optical enhancement factor. Typical enhancement

target, in qualitative agreement with what is predicted

\( \mu (10\) in clear water. Experiments with single microparticles

(10\( \mu \)m or 50\( \mu \)m in diameter) isolated within the 90\( \mu \)m-
diameter focal zone of a 40 MHz ultrasound transducer
confirmed that the enhancement of the optimized photoa-
coustic signal decreased with the number of optical speckle\(^{534}\)
grains (with grain size about 2\( \mu \)m) within the absorbing\(^{535}\)
target, in qualitative agreement with what is predicted\(^{536}\)
for the optical enhancement factor. Typical enhancement\(^{537}\)
for the photoacoustic signal ranged from 5 to 10, with the\(^{538}\)
larger enhancements observed for the smallest particles\(^{539}\).

Interestingly, it was shown that with a speckle size of 1\( \mu \)m\(^{540}\)
and a 10\( \mu \)m-diameter graphite particle, it was not possible\(^{541}\)
to observe any enhancement with the available 140 degree
of freedom provided by the deformable mirror used for the
experiment [51].

This pioneering work was rapidly followed by several\(^{542}\)
investigations of photoacoustic-guided optical wavefront\(^{543}\)
shaping. In all the works reviewed in this section, the ex-
perimental setups are similar to that introduced by Kong\(^{544}\)
and coworkers: in particular, photoacoustic feedback sig-
nals are measured from speckle patterns produced in a\(^{545}\)
free-space geometry after propagation through a scatter-
ing sample. Importantly, the size of the speckle grains is\(^{546}\)
ystematically adjusted to match the typical dimension of
the ultrasound focus by setting the distance between the
scattering sample and the measurement plane (see Sec. 2.3) on
the properties of optical speckle patterns). The spatially
light modulators or deformable mirrors used to perform
wavefront shaping were used in a reflection configuration\(^{548}\)
as in Figs. 1 and 2. Following the approach proposed\(^{549}\)
by Kong et al. [51], Caravaca-Aguirre and coworkers used\(^{550}\)
a genetic algorithm to perform PA-WFS and enhance the
light intensity behind a scattering layer by one order of
magnitude [63], as illustrated in Fig. 3. This study, aimed\(^{551}\)
at improving photoacoustic imaging, is further discussed\(^{552}\)
in Sec. 4. Chaigne et al. [64] further demonstrated that\(^{553}\)
the large bandwidth of photoacoustic signals could be ex-
plotted in the frequency domain to adjust the dimension\(^{554}\)
of the photoacoustic focal zone. By iterative optimization
of the highest frequency components (55-70 MHz band)
of the broadband photoacoustic signals measured with a
transducer with central frequency 27 MHz, the authors ob-
tained a photoacoustic enhancement factor of about \(\times 12\),
higher than the enhancement obtained with optimization
in lower frequency bands (ranging from \(\times 4\) to \(\times 8\)) or from
peak-to-peak amplitude measurements (\(\times 8\)), as illustrated
in Fig. 4. To maximize the sensitivity of photoacoustic
measurement to phase modulation of the light beam, the
optimization algorithm used a Hadamard basis vectors as
the basis for the input modes (instead of the canonical
pixel basis) of 140-element deformable mirror [64]. More-
over, by simultaneously monitoring the evolution of the
speckle pattern during the optimization process, it was
confirmed experimentally that the optimization with the
highest photoacoustic frequencies lead to a tighter optical
focus than what was obtained by optimization with the
lower frequency components.

A key advantage of the photoacoustic effect as a feed-
back mechanism is that the sensing may be performed si-
multaneously over the whole measurement volume, by use
of imaging ultrasound arrays. With a spherical matrix
array of 256 piezoelectric transducers, De´ an-Ben and co-
workers demonstrated photoacoustic-guided optical wave-
front shaping by optimizing photoacoustic signals from se-
lected targets of a 3D photoacoustic image, by means of a

genetic algorithm [65]. PA-WFS is usually limited in speed
by either the laser pulse repetition frequency or the refresh
rate of the adaptive optics device. In the context of pho-

toacoustic flowmetry, Tay and coworkers investigated the
potential of digital micromirror devices (DMD), which are
binary amplitude modulators, towards rapid PA-WFS [66]
: a combination of Hadamard multiplexing with multiple

\[ 0.25 \times \frac{0.25}{0.25} \]

(a) Enhancement (b) Flattop PA signal (c) Optimized PA signal

Figure 3: Illustration of the photoacoustic signal enhancement obtained with optimization-based photoacoustic-guided optical wavefront shaping. (a) Evolution of the photoacoustic enhancement with the optimization process, based on a genetic algorithm. (b) Photoacoustic signal prior to wavefront shaping. (c) Enhanced photoacoustic signal obtained for the optimal input wavefront. Figure reproduced with permission from [63], OSA.

B
Figure 4: Evolution of the photoacoustic enhancement factor by optical wavefront shaping during the optimization process, with different feedback values. In addition to the usual peak-to-peak photoacoustic amplitude as feedback signal, RMS values over several frequency bands computed from a Fourier analysis were used as alternative photoacoustic feedback values. For each feedback quantity, the photoacoustic enhancement factor was computed by normalizing the optimized quantity by its value under homogeneous illumination. Figure reproduced with permission from [64], OSA.
binary-amplitude illumination patterns was implemented to perform wavefront shaping based on the photoacoustic signal measured with a 10 MHz spherically focused transducer, and an intensity enhancement of a factor 14 was obtained. Although the DMD refresh rate was as high as 22 kHz, the optimization approach remained very long (typically two hours) because of a SNR issue. This study however demonstrated the potential of using DMD for PA-WFS.

One specific feature of photoacoustic sensing for optical wavefront shaping arises from the possibility to create an optical focus smaller than the ultrasound resolution spot, the feedback signal mixes the information coming from individual speckles. However, based on the non-uniform spatial sensitivity across the ultrasound focal region, it has been shown that the spatially non-uniform photoacoustic feedback tends to localize the optimized optical intensity to a single speckle smaller than the acoustic focus, by preferentially weighting the single optical speckle closest to the center of the ultrasound focus during the optimization [67]. As a consequence, an optical enhancement factor of 24 was reported for the optimized optical grain, about three times higher than the photoacoustic enhancement factor which averages the optical enhancement over all the optical speckles present in the focal spot. Possible applications of this sub-acoustic resolution optical focusing are further discussed in Sec. 4.2. While this effect was first reported in the context of linear photoacoustics, where the photoacoustic amplitude is proportional to the absorbed optical intensity as described by Eq. 1, Lai and coworkers introduced a nonlinear PA-WFS with a dual-pulse illumination scheme [68]. In short, this approach exploits the change in photoacoustic conversion efficiency between two consecutive intense illuminations to produce a feedback signal that is nonlinearly related the optical intensity; the first illumination pulse creates a photoacoustic signal that is linearly related to the optical intensity, but also changes the value of the Grüneisen coefficient \( \Gamma \) involved in the second illumination pulse. The change in the Grüneisen coefficient is caused by the temperature increase that follows the first illumination pulse [69, 68]. As a consequence, the feedback signal defined as the difference of the photoacoustic amplitudes of the two consecutive pulses varies nonlinearly with the optical intensity. As a result, optimization based on such a nonlinear feedback signal strongly favors focusing towards a single optical speckle grain rather than distributing the optical intensity evenly over all the speckle grains inside the acoustic focus spot. This effect had first been demonstrated with optical wavefront shaping based on nonlinear feedback from two-photon fluorescence [33, 32]. With nonlinear PA-WFS, Lai and coworkers achieved focusing to a single optical speckle grain 10 times smaller than the acoustic focus, with an optical intensity enhancement factor of \( \sim 6000 \) and a photoacoustic enhancement factor of \( \sim 60 \).

3.2. The photoacoustic transmission matrix

Following the transmission matrix approach proposed in optics [29], introduced in Sec. 3.2, the measurement of a photoacoustic transmission matrix was demonstrated for PA-WFS with both 1D and 2D photoacoustic images [71, 70]. The concept is strictly similar to that in optics, except that the pixels of the optical camera are replaced by the pixels of the photoacoustic image, which values are linearly related to the local optical intensity. The method was first implemented with the time-resolved photoacoustic signal from a single-element transducer processed as a 1D photoacoustic image [71], and
was rapidly extended to 2D photoacoustic images reconstructed from signals acquired with a conventional linear ultrasound array [70]. The typical experimental setup used to acquire the photoacoustic transmission matrix from 2D photoacoustic images is shown in Fig. 6, along with typical results. Fig. 6(b) shows the photoacoustic image of an absorbing leaf skeleton (photograph shown in Fig. 6(a), obtained by averaging the various photoacoustic images obtained during the measurement of the transmission matrix. The absence of horizontally oriented features in Fig. 6(b) is a consequence of the limited view configuration, where the ultrasound probe mostly detect waves propagating upwards, and is further discussed in Sec. 4. As opposed to the optimization approach, the photoacoustic transmission matrix approach can be used to focus light at any desired location after the matrix has been measured: Fig. 6(c) is a zoom on the blue region of Fig. 6(b), showing a target region outlined in red. Fig. 6(d) illustrates the light intensity enhancement (typically 6 times) after the SLM input pattern has been set to focus light towards the target region based on the knowledge of the photoacoustic transmission matrix. As an additional illustration of the power of the matrix approach, it was also shown that a singular value decomposition (SVD) of the photoacoustic transmission matrix provides a mean for automatically identifying signals from the most absorbing targets [71]. In contrast with optimization approaches, the transmission matrix approach rely on the assumption that the measured signal is proportional to the intensity. It therefore cannot fit from non-linearities or non-uniformities of the acoustic response for sub-acoustic resolution focusing.

4. Enhancing photoacoustic imaging with coherent light

In the previous section, we reported results for which the photoacoustic effect was used as feedback mechanism for optical wavefront shaping of coherent light. In this section, we now illustrate how photoacousting imaging may directly benefit from effects based on the coherence of light, such as speckle illumination or optical wavefront shaping. Generally speaking, the ultimate objective of photoacoustic imaging is to quantitatively reconstruct the distribution of optical absorption, described via the absorption coefficient \( \mu \) by considering that the optical absorption may still be described by some function \( \mu(r,t) \). As opposed to the fluence rate \( \Phi(r,t) \) is the relevant quantity, where the fluence rate \( \Phi(r,t) \) is a spatially smooth function, in particular usually smoother than \( \mu(r) \). However, if the coherence of light is to be taken into account, the local optical intensity \( I(r,t) \) is the appropriate physical quantity, as discussed in Sec. 2.3. From a theoretical point of view, the relevant photoacoustic equation for coherent light should then reads

\[
\left[ \frac{\partial^2}{\partial t^2} - c_s^2 \nabla^2 \right] p(r,t) = I \mu_a(r) \frac{\partial I}{\partial t}(r,t) \tag{14}
\]

where \( I(r,t) \) is generally a speckle pattern (and assuming that the optical absorption may still be described by some function \( \mu_a(r) \)). As opposed to the fluence rate \( \Phi(r,t) \), \( I(r,t) \) is strongly spatially varying over the typical dimensions of the optical speckle grain (i.e. half the optical wavelength inside scattering media), and can vary from pulse to pulse. In particular, in many cases \( I(r,t) \) will
Figure 7: Photoacoustic imaging with multiple speckle illumination. The experimental setup is similar to that of Fig. 6. Three types of absorbing samples are illuminated with multiple speckle illumination, by either using a SLM or a moving diffuser. (a1), (b1) and (c1) are photographs of the absorbing samples. (a2), (b2) and (c2) are conventional photoacoustic images equivalent to those obtained under homogeneous illumination. (a3), (b3) and (c3) are fluctuation images computed from the photoacoustic images obtained under all the multiple speckle illuminations. The fluctuations images reveal features otherwise invisible on the conventional images, because of directivity or frequency bandwidth issues. Figures (a) and (c) adapted with permission from [72], 2013 OSA. Figure (b) adapted with permission from [70], 2014 OSA.

4.1. Exploiting multiple speckle illumination

As discussed above, photoacoustic waves generated from a sample illuminated with an optical speckle pattern bear some information on \( \mu_a(\mathbf{r}) \). As a consequence, the general features of photoacoustic sources such as their frequency content or directivity may be strongly affected by a speckle illumination. By using multiple speckle illumination, Gateau and coworkers have shown that both limited-view and wavefront shaping may be exploited to improve photoacoustic imaging.

While multiple speckle illumination was initially used in photoacoustic to palliate visibility issues, it also has a tremendous potential for super-resolution imaging. Indeed, when a sample is illuminated with multiple uncorrelated speckle patterns, optical absorbers distant from more than one speckle size behave as uncorrelated sources of fluctuating photoacoustic signals. 

The super-resolution would be obtained with homogeneous illumination with incoherent light (in practice, they were obtained by averaging the photoacoustic signals obtained under various illumination patterns with coherent light). These images illustrate the limited-view artefacts associated with directive photoacoustic source: the ultrasound array located above the samples can only measure the photoacoustic waves that propagates upwards, i.e. the waves emitted by horizontally-oriented elements or boundaries. Moreover, image (c2) illustrates how the low frequency content associated to the low spatial frequency content of the large and homogeneous absorbing disk is filtered out by the high-frequency transducer array (central frequency about 20 MHz [72]). 

However, when multiple speckle illumination is used, the heterogeneous spatial distribution of the light intensity breaks the amplitude correlation among the ultrasound waves generated by each point-like absorber throughout the structure: the fluctuation of the photoacoustic signals from one illumination to the other may be interpreted as fluctuation signals emitted from fluctuating point-like sources (with size that of the speckle grain) that generate high-frequency and omnidirectional photoacoustic waves. Images (a3), (b3) and (c3) are fluctuation images computed from the photoacoustic images obtained under all the multiple speckle illuminations, illustrating how both high-pass filtering and limited view artefact can be overcome by taking advantage of multiple speckle illumination enabled by the use of coherent light.
Although the photoacoustic effect has first been proposed in the context of optical wavefront shaping as a way to provide a feedback mechanism to improve photoacoustic imaging. Because coherent light can be scanned across an absorbing sample, it has not been reported yet.

The first demonstrations of the potential of optical wavefront shaping (OFS) techniques developed to improve the spatial resolution of photoacoustic imaging have been reported in two publications from the same group [63, 67]. The resolution is thermally increased to break the acoustic diffraction limit [73]. As illustrated in Fig. 5, the resolution enhancement is expected to increase the penetration depth and resolution, whereas Fig. 9(c) is the photoacoustic image obtained by scanning the sample relatively to a fixed scattering layer. The resolution is thermally increased to break the acoustic diffraction limit [73]. As illustrated in Fig. 5, the resolution enhancement is expected to increase the penetration depth and resolution.

The first demonstrations of the potential of optical wavefront shaping to improve photoacoustic imaging were reported in two publications from the same group [63, 67]. The resolution is thermally increased to break the acoustic diffraction limit [73]. As illustrated in Fig. 5, the resolution enhancement is expected to increase the penetration depth and resolution.

The first demonstrations of the potential of optical wavefront shaping to improve photoacoustic imaging were reported in two publications from the same group [63, 67]. The resolution is thermally increased to break the acoustic diffraction limit [73]. As illustrated in Fig. 5, the resolution enhancement is expected to increase the penetration depth and resolution.
Although promising, these preliminary results were never obtained in a rather unrealistic configuration for imaging where the purely absorbing object to image was scanned relatively to a static scattering object. Additional promising preliminary results were also reported by Conkey et al. [67], obtained by scanning the transducer in stead of scanning the scattering sample: it was shown that by optimizing the photoacoustic amplitude at each point of a 1D scan over a simple 1D absorbing pattern, the photoacoustic image obtained from optimized signals exhibits an improved resolution as compared to the conventional image under homogeneous or single random illumination. This improved resolution was attributed to the narrower spatial point spread function, similarly what was observed on a homogeneously absorbing sample (see Fig. 5(a)). Achieving enhanced photoacoustic imaging by performing wavefront shaping inside the object image however remains to be demonstrated. In addition to the approach investigated by Conkey et al. [67], alternative approaches towards this goal include the use of the acousto-optic effect to first enhance the optical intensity and then scan the optimized spot to form a photoacoustic image, or the development of iterative approaches where an initial conventional photoacoustic image could then be used to perform PA-WFS and consequently improve the signal-to-noise ratio of further photoacoustic images to improve their resolution.

4.3. Photoacoustic microendoscopy with multi-mode optical waveguides

In the field of photoacoustic imaging, optical-resolution photoacoustic endoscopy was first introduced by use of bundles of single-mode fibers by Hajireza et al. [76, 77], and was further investigated with various approaches including multiple optical and acoustic components [78, 79] or all-optical components [80] assembled in a cathether housing. With these approaches, the diameter of the probes typically ranges from 1 mm to 4 mm and the resolution ranges from 5 µm to 20 µm. As introduced in Sec. 2.4, optical wavefront shaping has been investigated to manipulate or deal with light propagation in multi-mode optical fibers [47, 48, 49, 50], an important step towards the miniaturization of optical endoscopes. The principle of a miniaturized photoacoustic endomicroscope endowed with optical wavefront shaping was first demonstrated by focusing and scanning pulsed coherent light through a 220µm-diameter multimode fiber [75], based on a phase conjugation approach [50]. As illustrated in Fig. 10, an absorbing wire was imaged with diffraction limited optical resolution (around 1.5 µm) at the distal tip of a multimode fiber. However, the photoacoustic signals were detected through water only with a 20 MHz ultrasound transducer, a situation not relevant for imaging inside biological tissue which strongly attenuates high-frequency ultrasound. Consequently, Simandoux et al. [81] demonstrated the use of a water-filled silica capillary as a multi-mode optical waveguide for optical excitation and a quasi-monomode acoustic waveguide to collect the photoacoustic wave with a reduced attenuation, through a 3-cm thick fat layer. The use of such a capillary to simultaneously perform optical wavefront shaping with optical digital phase conjugation and photoacoustic detection was further demonstrated in a recent study highlighting the potential of such capillaries for multi-modal optical imaging [82].

5. Discussion and conclusion

The several recent investigations reviewed above illustrate how coupling photoacoustics and light coherence enables new horizons in several directions. On the one hand, the photoacoustic effect provides a valuable feedback mechanism for optical wavefront shaping, that allows in principle sensing inside scattering media via remote ultrasound detection. On the other hand, photoacoustic imaging may take advantage from the properties of coherent, via the possibility to use multiple speckle illumination or to manipulate with optical wavefront shaping.

Although recent publications demonstrated promising proof-of-concepts experiments, several challenges lay
The principles reviewed above inside scattering samples [91] the photoacoustic effect is used to sense speckle patterns, the typical size of the optical speckle grain was much larger than $\lambda/2$ and comparable to the ultrasound resolution. Doing so, the number of independent optical speckle grains within the ultrasound resolution cell was kept relatively small, either to allow sensing fluctuations from multiple speckle illumination with a sufficient signal-to-noise ratio or to demonstrate significant light intensity enhancement by wavefront shaping with a relatively low number of degrees of freedom. However, controlling the size of the optical speckle grains is only possible with free-space propagation, usually by adjusting the distance between the scattering object to the sample plane. Inside biological tissue, the typical speckle size cannot be controlled anymore as it is dictated solely by the optical wavelength $\lambda_{\text{optics}}$. If one considers a 3D ultrasound focal spot with typical linear ear dimension $\lambda_{\text{ultrasound}}$, the number $N_s$ of independent 3D optical speckle grains within this focal spot is expected to scale as $N_s \sim \left(\frac{\lambda_{\text{ultrasound}}}{\lambda_{\text{optics}}}ight)^2$. For photoacoustics sensing with several tens of MHz ultrasound, which has been demonstrated up to several mm in tissue [83], $\lambda_{\text{ultrasound}}$ is of the order of a few tens of microns, and the number of independent speckle grains within the ultrasound focal spot may be as high as several thousands to ten thousand. The photoacoustic detection of speckle fluctuations with grain size as small as 2$\mu$m was demonstrated through a diffusing scatterer with 20 MHz ultrasound propagating through water with sufficient SNR to provide fluctuation images [72], but exploiting multiple speckle illumination inside scattering media has yet to be demonstrated. Similarly, photoacoustic-guided optical wavefront shaping has only been demonstrated with significant optical enhancement factor through scattering samples with speckle grain enlarged by free-space propagation [71, 67, 63, 70, 64, 66, 65, 68], as for acousto-optic-guided wavefront shaping experiments [55, 56, 57, 58, 59, 60]. By studying the influence of the absorber size with a fixed speckle grain size, it was recently confirmed experimentally that the efficiency of photoacoustic-guided wavefront shaping decreases rapidly when the typical absorber dimension is large compared to the speckle size: with a speckle size of about 30$\mu$m (generated via free-space propagation), the photoacoustic enhancement was reduced to less than 1.5 for spherical absorbers 400$\mu$m in diameter [84], in agreement with earlier qualitative observations by Kong et al. [51].

There are several possible directions towards enabling the principles reviewed above inside scattering samples [91] The photoacoustic effect, as opposed to acousto-optic modulation, only takes place in the presence of optical absorption. While this is certainly one drawback of photoacoustic sensing of light intensity, as no information can be retrieved from absorption-free regions, it may however be turned into an advantage for PA-WFS: for PA-WFS, the relevant number of independent speckle grains to consider within the ultrasound focal spot is that overlapping the distribution of optical absorbers. Therefore, if optical absorbers are sparse enough at the scale of the ultrasound focal spot, it is expected that the number of relevant speckle grains to sense or control with wavefront shaping may remain relatively low. Sparse distributions of absorbers may occur in tissue for instance either for blood microvessels or exogenous contrast agents at relatively low concentrations. For a given distribution of photoacoustic sources, reducing the size of the ultrasound sensing region via increasing the detection frequency is the most straightforward option, but this remains limited by the ultrasound attenuation. As the signal-to-noise ratio is the fundamental limitation, either because a small fluctuation has to be detected over a large signal (large number $N_s$ of independent relevant speckle grains) or because a small signal is involved (very high ultrasound frequency to reduce $N_s$), there is a clear need for highly sensitive ultrasound detectors optimized for photoacoustic sensing. The transducers that have been used so far are commercially available ones, with standard technologies usually developed for pulse-echo measurement and not necessarily optimized for photoacoustic detection. The tremendous development of biomedical photoacoustic imaging will hopefully trigger the development of dedicated transducers, which could bring photoacoustics with coherent light closer to practical applications.

Regarding optical wavefront shaping, fast light manipulation is needed for $\textit{in vivo}$ tissue application, in which various types of motion leads to speckle decorrelation with time scales as short as a few millisecond [85]. The most recent research efforts towards fast wave front shaping involve the use of digital micromirrors (DMD) [27, 86, 87, 88, 36]. It is expected that the significant research efforts and very rapid progresses made in the field will continue to stimulate the development of new devices with both fast refresh rates and millions of pixel with flexible amplitude and/or phase control, that will in return benefit the field of photoacoustics with coherent light. Exploiting light coherence in photoacoustics also requires appropriate laser sources. For pulsed light, a minimal coherence length $l_c \sim 1$ m corresponds to a minimal pulse duration $\tau \sim \frac{c}{l_c} \sim 3$ ns. In the context of photoacoustic imaging with multiply scattered coherent light, this shows that pulses of at least a few nanoseconds must be used if the effect of coherence is to be exploited at centimeters depth in biological tissue. However, not all nanosecond-pulse laser have a nanosecond temporal coherence. For instance, the Q-switched nanosecond-pulse lasers widely used for deep photoacoustic imaging usually have a coherence length no longer that a few millimeters. While lasers such as Nd:YAG pulsed lasers may be injected with a single-longitudinal-mode seed laser with a large coherence length to obtain pulses with a coherence time of a few nanoseconds, this approach may not be extended to tunable laser sources based on optical parametric oscillators (OPO). So far, 532 nm is the only wavelength
that has been used to perform the proof-of-concepts experiments reviewed in this paper. There is thus a clear need of new tunable and coherent laser sources in the so-called therapeutic window (600-900 nm) where light can penetrate deep into biological tissue.

In summary, the coupling between the photoacoustic effect and propagation of multiply scattered coherent light opens up new horizons for both optical wavefront shaping and photoacoustic imaging. On the one hand, the photoacoustic effect offers a unique feedback mechanism optical wavefront shaping or optical imaging with speckle illumination. On the other hand, the possibility to exploit the enormous number of degrees of freedom of multiply scattered coherent light with optical wavefront shaping and/or multiple speckle illumination for photoacoustic imaging offers exciting opportunities to break the current depth-to-resolution ratio of non-invasive photoacoustic imaging and/or to make photoacoustic endomicroscopy minimally invasive.

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7. Conflicts of interest

The authors declare that there are no conflicts of interest.

8. References


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