Speckles in laser Doppler perfusion imaging

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We report on the quantitative influence of speckles in laser Doppler perfusion imaging. The influence of speckles on the signal amplitude and on the Doppler spectrum is demonstrated experimentally for particle suspensions with different scattering levels and various beam widths. It is shown that the type of tissue affects the instrumental response through the effect of lateral light diffusion on the number of speckles involved in the detection process. These effects are largest for narrow beams. © 2006 Optical Society of America

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Laser Doppler perfusion imaging (LDPI) is a noninvasive technique for measuring blood flow maps on an area of tissue. The fundamental output quantity of a laser Doppler flowmetry (LDF) instrument is the first moment M_1 of power spectrum $P(\omega)$ of the photocurrent fluctuations, where the ith moment is defined as[1]

$$M_i = \int_0^\infty P(\omega)\omega^i \mathrm{d}\omega, \tag{1}$$

often normalized with the square of direct current, DC^2, to cancel out laser power fluctuations. In LDF, multiple waves illuminating the detector generate a dynamic speckle pattern, which causes fluctuations in the detector signal. Other techniques for measuring perfusion from dynamic speckles are based on calculation of differences between subsequent speckle fields[2] and speckle contrast analysis.[3] Briers[4] compared speckle contrast and laser Doppler techniques and showed the common physical basis of laser Doppler and laser speckle techniques. In this Letter we show that speckles have an essential effect on LDPI signals. Quantifying the speckle phenomenon and its spatial properties is essential to account for the system's response to different tissue optical properties and beam sizes. The influence of the number of speckles on the LDPI signal as influenced by the beam size was mentioned by Wårdell *et al.*[5] However, the effect of scattering on the number of speckles and on the LDPI signal was not reported. Piederrière *et al.*[6] reported the effect of particle size and concentration on the static speckle size as a function of polarization. Other studies were concerned mainly with the effect of scattering on temporal field fluctuations in situations of constant speckle size. Bonner and Nossal[1] included the coherence in an instrumental response factor β, which they determined by calibration. Boas and Yodh,[7] following Bonner and Nossal, used a constant β, which depends on the experimental setup. Also, Binzoni *et al.*[8] identified β as an instrumental factor that depends on the optical coherence of the signal at the detector's surface. Those studies[1,7,8] excluded the effects of scattering on the spatial field correlation function and on β. They used fiber optic systems in which the spatial field correlation function is an instrumental constant, which is not the case in a laser Doppler perfusion imager for which a free beam geometry is used. In this Letter we demonstrate the effect of multiple scattering on spatial field correlation at the tissue surface when a free laser beam is used for illumination and a detector collects the backscattered photons propagated through free air.

It can be derived[9] that the AC signal generated by a fluctuating speckle pattern, normalized by DC^2, is equal to

$$\frac{\langle i_{\mathrm{AC}}{}^2\rangle}{\langle i_{\mathrm{DC}}\rangle^2} = \frac{1}{N} f_D(2 - f_D), \tag{2}$$

where $\langle i_{\mathrm{AC}}{}^2\rangle$ is the mean square of the photocurrent fluctuations, $\langle i_{\mathrm{DC}}\rangle$ is the mean photocurrent, N is the number of speckles on the detector, and f_D is the Doppler shifted fraction of all detected photons. It is assumed that the photodetector surface is, on average, homogeneously illuminated.

In LDPI instrumentation, two modes of detection are adopted: The first illuminates a photodetector without focusing the light by a lens. This is the mode of detection used in the device developed by Wårdell *et al.*[5] The other focuses the light by using a lens, as in the LDPI instrument of Essex and Byrne.[10] We show below that, for both modes of detection, the relation between the number of speckles within the detector and the geometric properties of the tissue spot that emits diffusely reflected light is identical.

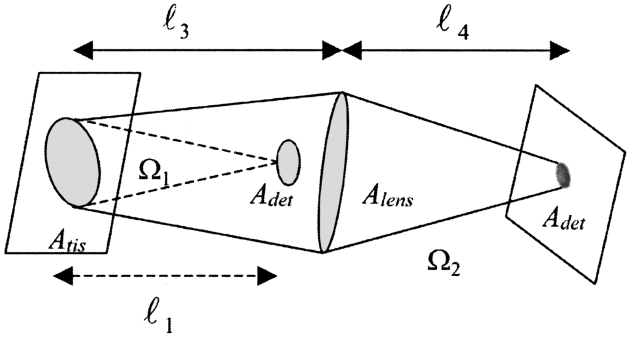

Fig. 1. Schematic of diffusely reflected focused (solid double-headed arrows) and unfocused (dashed double-headed arrows) light illuminating a photodetector.

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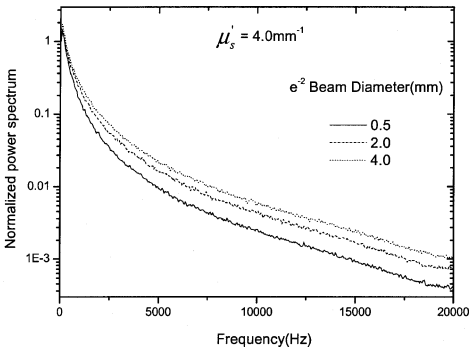

Fig. 2. Normalized power spectra for $\mu_s' = 4.0\ \mathrm{mm}^{-1}$.

For simplicity, we assume that the diffuse reflection by the tissue happens within a confined circular region with area A_{tis}, which has uniform irradiation. A photodetector with area A_{det} (not necessarily circular) is placed at distance l_1 from the tissue (Fig. 1, dashed double-headed arrow). The speckle area on the photodetector, A_{sp}, can be written as[9,11] $A_{\mathrm{sp}} = \lambda^2/\Omega_1$, with Ω_1 the solid angle subtended by the irradiating tissue spot on the photodetector. For l_1 much larger than the size of the light-emitting spot, this quantity is taken as $\Omega_1 = A_{\mathrm{tis}}/{l_1}^2$; substituting this equation into Eq. (2) will give

$$\frac{\langle {i_{\mathrm{AC}}}^2 \rangle}{\langle i_{\mathrm{DC}} \rangle^2} = \frac{\lambda^2 {l_1}^2}{A_{\mathrm{det}} A_{\mathrm{tis}}} f_D(2 - f_D). \tag{3}$$

Now we place a lens with area A_{lens} a distance l_3 from the tissue. This lens produces an image of the light-emitting tissue spot on the photodetector plane, which is placed a distance l_4 from the lens (Fig. 1, solid double-headed arrows). We assume that this image is completely within the physical borders of the photodetector. Hence the effective photodetector area A_{det} that must be taken into account is the area of the image on the detector, which depends on magnification $M = l_4/l_3$ obtained with the lens, giving $A_{\mathrm{det}} = A_{\mathrm{tis}} M^2$. Furthermore, the speckle area now is governed by solid angle $\Omega_1 = A_{\mathrm{lens}}/{l_4}^2$ subtended by the lens instead of by a solid angle subtended by the irradiating spot on the tissue. Thus Eq. (2) can be written as

$$\frac{\langle {i_{\mathrm{AC}}}^2 \rangle}{\langle i_{\mathrm{DC}} \rangle^2} = \frac{\lambda^2 {l_3}^2}{A_{\mathrm{lens}} A_{\mathrm{tis}}} f_D(2 - f_D). \tag{4}$$

On comparison of Eqs. (3) and (4) we can conclude that, when we use a photodetector without focusing, we obtain the same modulation depth as when the photodetector is replaced by a lens of equal size that focuses the light within the physical borders of another detector. Furthermore, we can see that the influence of the size of the irradiating tissue spot is the same for both configurations. When no lens is used, a change in A_{tis} will change the modulation depth through the effect on the speckle size, whereas, when a lens is used, the modulation depth will change because the effective size of the photodetector (the image on the photodetector surface of the light-emitting spot on the skin) will change. Equations (3) and (4) indicate that in LDPI those factors that affect the irradiating tissue area will determine the response of the system. Serov *et al.*[9] give a more rigorous treatment of the problem, including a definition of the effective speckle size and its relation to the angular distribution of the detected light. From this it follows that the number of speckles will depend on the size and shape of the laser beam used for scanning the tissue and on the amount of lateral broadening that the light will undergo before escaping the tissue. The latter will depend on the optical properties of the tissue. Hence the beam's shape and size, and the optical properties, might affect perfusion maps through the effect of these quantities on the spatial properties of the dynamic speckle pattern that is created on the photodetector. In this Letter, by studying the amplitudes and Doppler spectra of photocurrents generated by well-defined particle suspensions that have a range of scattering levels and by using a range of laser beam diameters, we show experimental evidence that this effect is significant for practical LDPI instrumentation.

The experimental setup consists of a simple backscattered configuration in which we illuminate the medium with a perpendicular laser beam and collect the backscattered photons by using a lens and a photoreceiver. A linearly polarized Uniphase 1125P He–Ne laser, of 632.8 nm with output power of 5mW, is used as the source. A beam expander made from two positive lenses of focal lengths 20 and 30 mm is used to vary the beam diameter. The beam diameter is measured with a commercial beam profiler. An 8 ml glass cuvette is used as a sample holder. A microscopic glass slide (500 μm) is used to cover the cuvette. A lens ($f = 50$ mm) is placed a distance of 25 cm from the sample to collect the backscattered light, with a photoreceiver on focus. Detection is performed with a New Focus Model 2001 photoreceiver, with an effective detector area of 0.81 mm^2. The AC signal is amplified by 40 dB and then applied to an antialiasing low-pass filter (Butterworth) (fifth-order sampled capacitor, $f_c = 20$ kHz). The filtered signal is then applied to a 12-bit analog-to-digital converter, where it is sampled at 40 kHz. A water suspension of polystyrene microspheres (Polysciences, Inc.) of $\varnothing 0.771$ μm ($g = 0.9$) of known concentration is used to make the phantoms, which mimic tissue optical

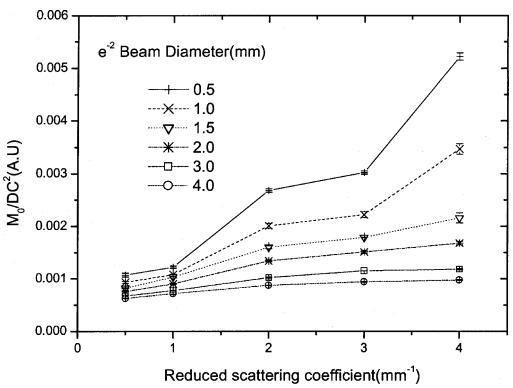

Fig. 3. M_0/DC^2 versus reduced scattering coefficient for several beam widths.

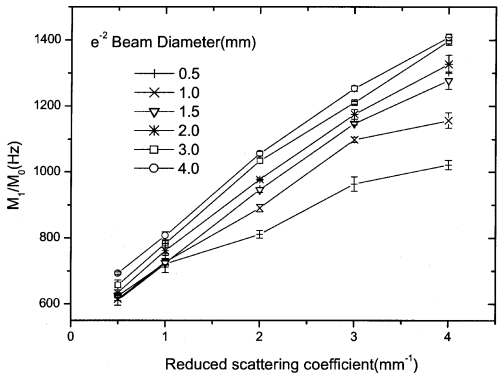

Fig. 4. M_1/M_0 versus reduced scattering coefficient for several beam widths.

properties. The polystyrene suspension is further diluted to make samples with reduced scattering coefficients (μ_s') of 0.5, 1, 2, 3, and 4 mm^{-1}, following Mie theory calculations. Ecoline Black dye (Talens) is added to get an absorption that is realistic for tissue ($\mu_a = 0.02$ mm^{-1}).

Figure 2 shows the normalized power spectra for the suspension of $\mu_s' = 4.0$ mm^{-1} for three beam sizes. In Fig. 3 the DC-normalized values[2] of zero-order moment M_0 are given (which by definition are the same as the modulation depth values $\langle i_{AC}^2\rangle/\langle i_{DC}\rangle^2$); Fig. 4 shows the ratio M_1/M_0, which actually is the mean Doppler shift.

It was observed (Fig. 3) that the variation in scattering level results in a considerable variation of M_0. Moreover, the variation in M_0 that is due to variation in scattering level decreases with increasing beam diameter. For a beam of 0.5 mm diameter, M_0 shows a variation by a factor of 5, whereas for a 4 mm beam the variation is only approximately a factor of 1.5. For a 4 mm beam the limiting value of M_0 seems to be reached for the highest level of scattering, while for the narrowest beam M_0 is likely still to increase on a further increase of μ_s'. The observed variations clearly show the speckle-related effects of the scattering level and the beam diameter on the LDPI signal. As the scattering level increases, the number of speckles on the effective detector decreases, because the higher scattering level results in a narrow backscattered intensity distribution. This variation is larger for narrow beams than for wide beams because the relative change of the width of the intensity variations by variations in the scattering level is larger for narrow beams than for wide beams. For wide beams, which result in wide intensity distributions and more speckles on the detector, variation of speckle number caused by scattering level variations is suppressed.

In Fig. 4 the spectral width (M_1/M_0), which represents the average Doppler frequency of the spectrum, is plotted. The spectral width increases with the beam diameter. This is a remarkable observation, because the Doppler shifts of individual photons will depend only on the cumulative effect of a series of scattering events of these photons by particles, processes that are completely independent of beam diameter. We can explain the effect by stating that photons with large Doppler shifts have a wider lateral distribution than photons with small Doppler shifts. If this is true, then for a given scattering level the sensitivity of M_1/M_0 to beam size should be largest for the smallest beams, where the lateral distributions are governed by the multiple scattering process rather than by the beam size itself. This can be observed in Fig. 4, in particular for the highest scattering levels where the curves for small beams diverge.

In summary, we have shown that the scattering level of the tissue strongly affects a laser Doppler imager signal based on the number of speckles involved in the detection. This speckle-related cross talk between the scattering level and the LDPI signal can be suppressed by use of a sufficiently large beam size but at the expense of decreased signal-to-noise ratio and spatial resolution. Both the signal modulation depth and the frequency content are affected by this speckle-related influence. This result demonstrates that the effect of speckles should be modeled to predict the system's response to various tissue optical properties and particle velocities.

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References

1. R. Bonner and R. Nossal, Appl. Opt. **20**, 2097 (1981).
2. H. Fujii, K. Nohira, Y. Yamamoto, H. Ikawa, and T. Ohura, Appl. Opt. **26**, 5321 (1987).
3. J. D. Briers and S. Webster, Opt. Commun. **116**, 36 (1995).
4. J. D Briers, J. Opt. Soc. Am. A **13**, 345 (1996).
5. K. Wårdell, A. Jakobsson, and G. E. Nilsson, IEEE Trans. Biomed. Eng. **40**, 309 (1993).
6. Y. Piederrière, J. Cariou, Y. Guern, B. Le Jeune, G. Le Brun, and J. Lotrian, Opt. Express **13**, 5030 (2005).
7. D. A Boas and A. G. Yodh, J. Opt. Soc. Am. A **14**, 192 (1997).
8. T. Binzoni, T. S Leung, M. L. Seghier, and D. T. Delpy, Phys. Med. Biol. **49**, 5445 (2004).
9. A. Serov, W. Steenbergen, and F. F. M. de Mul, J. Opt. Soc. Am. A **18**, 622 (2001).
10. T. J. H. Essex and P. O. Byrne, J. Biomed. Eng. **13**, 189 (1991).
11. A. T. Forrester, R. A. Gudmudsen, and P. O. Johnson, Phys. Rev. **99**, 1691 (1955).