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A Semiconductor Laser Used for Direct Measurement of the Blood Perfusion of Tissue

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Abstract—An instrument consisting merely of a semiconductor laser in its own housing was used to measure the blood perfusion in tissue. Use is made of the feedback of Doppler-scattered light to the photodiode in the laser housing. A recording perfusion of a finger under occlusion of blood flow in the arm is shown.

A recent method for velocimetry is the use of direct-mixing laser Doppler velocimetry. With this method, light emitted by a laser is scattered or reflected by a moving target. Due to the Doppler effect the scattered or reflected light is slightly shifted in frequency. This light is partly fed back to the laser housing (which contains a photodiode), where it is mixed coherently with the original (non-Doppler shifted) light of the laser. As a result, a Doppler frequency signal can be observed, not only in the voltage across the laser (at constant current) but also in the signal emerging from the photodiode, positioned in the laser housing at the back side of the laser crystal [1]–[4]. Normally, this photodiode signal is used to control the temperature-current-wavelength-intensity performance of the laser (with the help of a small Peltier element), but in the present setup this signal is recorded directly. It turns out that the amount of light fed back into the cavity of the laser is of crucial importance. Too much light fed back can lead to coherence collapse and mode hopping of the laser [5]–[7].

Laser Doppler velocimetry using conventional methods has been used before for blood perfusion measurements, either in the form of a fiber-coupled laser with separate detector(s) [8]–[10] or in the form of a semiconductor laser with detectors mounted together in a small

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Fig. 1. Sketch of the semiconductor laser, directly placed on the tissue; L: laser crystal (dimensions $\approx 300\mu\text{m}$); D: photodiode; C: cooling block; G: cover glass; T: tissue (i.e., finger tip).

probe [11], [12]. Light entering the tissue is either absorbed or it is scattered by tissue particles at rest or by moving particles such as blood cells. The latter scattering gives rise to a Doppler frequency shift. The light emerging from the tissue contains Doppler-shifted and non-Doppler shifted contributions, which will mix coherently at the detector. In this way the Doppler frequency can be measured.

We used this mixing technique for the direct measurement (i.e., without the use of a fiber or other optical equipment) of the perfusion of blood in tissue. The instrument consists of the semiconductor laser in its housing, together with a photodiode at the back facet of the laser crystal (Fig. 1). A very cheap and light instrument results, which can easily be handled in clinical situations.

The Doppler frequency spectrum to be measured at the detector is, in case of a single velocity value, peaked at the frequency corresponding with the velocity. In the case of scattering from tissue, it will be smeared out to a functional form resembling an Gaussian in time, due to the random character of the velocity vector of the blood perfusion in the capillaries and due to multiple scattering in the tissue [8]. This will also be the case with self-mixing velocimetry. Also, the finite divergence of the laser beam of semiconductor lasers (typically $10\text{--}20^\circ$) will contribute to this randomization.

The relation between the Doppler frequency Δf to be observed in the photodiode output and the velocity v is given by

$$\Delta f = 2vn \cos \vartheta / \lambda \quad (1)$$

where n is the refractive index, ϑ is the angle between the direction of the laser beam and the velocity vector, and λ is the wavelength of the laser light. In case of random velocities or of multiple scattering the value of v and ϑ will be randomized. It can be shown [8] that in this case the first moment $\langle \omega \rangle$ of the power spectrum $S(\omega)$ of the frequency distribution of all possible Doppler frequencies, given by

$$\langle \omega \rangle = \int \omega \cdot S(\omega) \cdot d\omega \quad (2)$$

is proportional to the averaged velocity $\langle v \rangle$ times the number of Doppler scattering particles, i.e., to the flow of particles. The second moment

$$\langle \omega^2 \rangle = \int \omega^2 \cdot S(\omega) \cdot d\omega \quad (3)$$

(proportional to the averaged squared flow) can be used also. The zeroth moment $\langle \omega^0 \rangle$, which is merely the integrated power spectrum, can be shown to be proportional to the number of Doppler scattering

Fig. 2. Recording of the blood perfusion of the finger tip of a healthy volunteer. (a) Second moment; (b) zeroth moment; (c) square-root of the normalized second moment of the frequency power spectrum. Three subsequent occlusions of the upper arm have been performed (indicated by "Oc" in the figure). This is accompanied by a rapid drop in the signal level to almost zero. The heart beat is clearly distinguishable in the un-occluded regions. The slower fluctuations (period ≈ 20 s) are mainly due to vasomotion (i.e., involuntary movements of the blood vessels). The zero signal levels, obtained with the probe attached to a stationary scatterer, are indicated with (o). The noise level in the zero signal does not exceed the line thickness.

particles, when integrated for $\omega \neq 0$. Therefore, the ratio of the first or second moment and the zeroth moment (the normalized or weighted moments) can be considered as being related to the averaged velocity or the averaged squared velocity of the particles respectively [13]:

$$\langle r \rangle \sim \langle \omega \rangle / \langle \omega^0 \rangle, \quad \text{and} \quad \langle r^2 \rangle \sim \langle \omega^2 \rangle / \langle \omega^0 \rangle. \quad (4)$$

We developed a digital signal processor unit to derive these moments and the mentioned ratio from the measured signal on an on-line basis [13], [14]. After preamplification the photodiode signal is fed via a programmable amplifier, a bandpass filter (set at 30 Hz–30 kHz to select the moving particles of interest), and a fast A-to-D converter into a TMS-32025 processor unit. Then the moments are calculated from repeatedly taken samples of the digitized photodiode output signal (duty cycle 60 kHz), using zero-time autocorrelation of the signal (for the zeroth moment) and of its time derivative (for the second moment) with a moving time-average technique. The result has an overall time resolution of 1/40 s. This is sufficient in order to be able to measure the heart beat frequency and its specific details. The unit is also capable of correcting for noise contributions (detected in a representative band of 30–40 kHz), but use of this option was not needed in the experiment described here.

In the specific experiments described here we choose a Sharp semiconductor laser, type LT080MD0, with a wavelength of 780 nm, operated at 44 mA, producing 3 mW laser light. This particular laser was chosen because of its high wavelength stability, caused by the presence of an extra semi-transparent mirror behind the rear laser crystal facet. The extra mirror has a strong line-narrowing effect. The laser was attached directly to the tissue under investigation, the finger tips of several healthy volunteers. The divergence of the beam is about 20° , thus a considerable velocity randomizing effect is obtained. Focusing the beam or imaging was not necessary. Another favorable effect of using an unfocused beam is the reduction of the change of damage or injury by local heating of the tissue. This effect may also disturb measurements, since part of the light power of 3 mW is converted into heat, depending upon the absorption characteristics.

In Fig. 2 we show some typical recordings of the blood flow in the finger tip of one of the volunteers. The laser housing was attached to the finger with minimum pressure using adhesive tape. The laser housing and the finger tip did not move with respect to each other. The square-root of the second moment (curve a; proportional to the rms flow), the zeroth moment (curve b; proportional to the number

of scattering moving particles) and the square root of the ratio of the second and the zeroth moments (curve c ; proportional to the rms flow velocity) are shown. Also the effect of occlusions of the upper arm is seen in the figure. The frequency of the heart beat can be detected easily. It is absent during the occlusions. In those periods the flow drops almost immediately, but the number of blood cells shows a slower decrease, almost to zero. Therefore with an occlusion the normalized moment (curve c) becomes unreliable and shows no heart frequency. The noise in the normalized moment is relatively large [see (4)]. The slower variations in the signal are most probably due to vasomotion (i.e., involuntary elastic motion of the vessel walls with a period of about 20 s).

Since the mixing effect can result from reflection as well as from scattering, the tiny movements of the tissue surface at the frequency of the heart beat may also contribute to the Doppler signal. Since the velocity causing this signal is not randomized, the signal will lead to a more peaked frequency spectrum than the contribution from internal scattering. However, it can be shown [14] that these movements, which in first approximation will have an oscillatory character, will give rise to mixing signals with double heart frequency. This is caused by the fact that the power spectrum cannot distinguish between motions to and from the laser. In the recordings, no indications for this contribution from reflection was observed.

We also tested this point in another manner. Besides the recording of the blood perfusion of the bare finger tip, shown in Fig. 2, the finger tip was covered with a thin microscope slide cover glass (thickness 170 μm). A small amount of refractive index matching oil was used between the cover glass and the finger tip, and the laser housing was positioned on the cover glass, carefully avoiding extra pressure. In doing so we observed similar spectra as in Fig. 2, which observation excludes possible contributions from moving skin reflections.

We feel we have developed an instrument that, when combined with a signal processing unit on-a-chip, will be very cheap and easy to apply, and will have the potential to deliver a significant contribution in the field of medical and clinical blood perfusion measurements.

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