Basic Theory and Operating Principles of Laser Doppler Blood Flow Monitoring and Imaging (LDF & LDI), Issue 1.

Introduction

Laser Doppler is a standard technique for the non-invasive blood flow monitoring and measurement of blood flow in the microcirculation. The strength of the technique is in looking at changes in flow - either over time or differences in flow over an area of skin or other exposed tissue.

For example you might compare flow in normal tissue with flow in a burnt area of tissue.

You might also use a provocation (stimulus) to change flow. This could be skin heating, drug delivery by iontophoresis, pressure cuffs etc.

Operating Principles

The laser Doppler technique measures blood flow in the very small blood vessels of the microvasculature, such as the low-speed flows associated with nutritional blood flow in capillaries close to the skin surface and flow in the underlying arterioles and venules involved in regulation of skin temperature. The tissue thickness sampled is typically 1mm, the capillary diameters 10 microns and the velocity spectrum measurement typically 0.01 to 10mm/s. The technique depends on the Doppler principle whereby low power light from a monochromatic stable laser (**a**), e.g. a Helium Neon gas laser or a single mode laser diode, incident on tissue is scattered by moving red blood cells and as a consequence is frequency broadened (**b**). The frequency broadened light, together with laser light scattered from static tissue, is photodetected and the resulting photocurrent processed to provide a blood flow measurement. Please note, where laser light is scattered for tissue with a low red blood cell concentration the average Doppler frequency shift is proportional to the average speed of red blood cells.

(**a**)   (**b**)
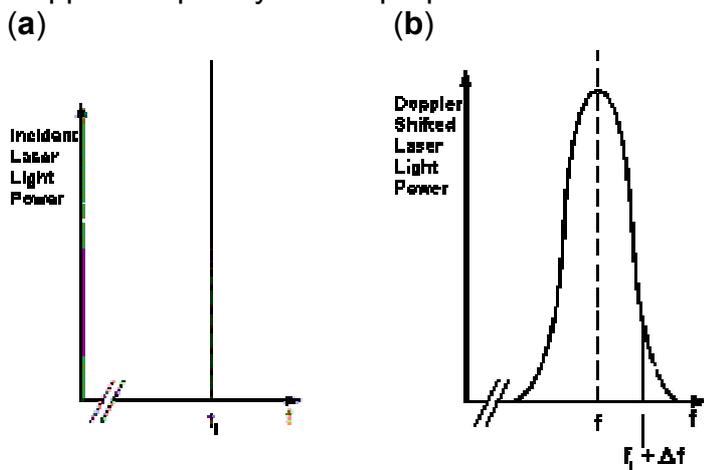

A laser operating in a single mode.

Δf: Doppler shift frequency is typically in the range of 20Hz to 20KHz for a well perfused region of the microcirculation. e.g. finger tips

Laser light can be directed to the tissue surface either via an optic fibre (c) or as a light beam (d). For 'fibre optic' monitors (**LDF** instruments) the optic fibre terminates in an optic probe which can be attached to the tissue surface. One or more light collecting fibres also terminate in the probe head and these fibres transmit a proportion of the scattered light to a photodetector and the signal processing electronics. Normal fibre separations in the probe head are a few tenths of a mm, consequently blood flow is measured in a tissue volume of typically $1mm^3$ or smaller. When a larger volume of tissue is stimulated to vasodilate or vasoconstrict, or where for example a healing process results in increased blood flow, the measured blood flow changes in the small tissue volume are generally taken to be representative of the larger volume.
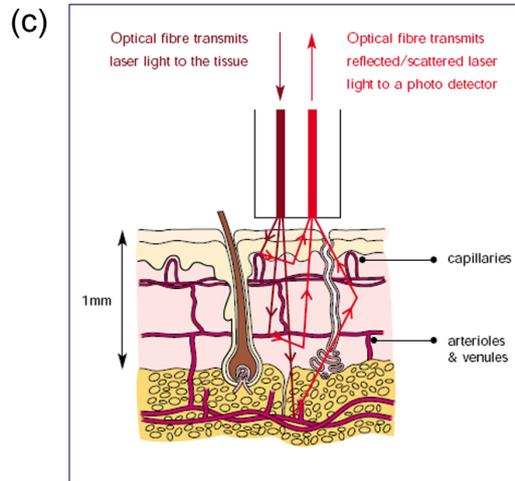
In a Laser Doppler blood flow Imager (LDI) the low intensity laser beam is scanned across a tissue surface in a raster fashion using a moving mirror. There is no direct contact with the tissue being assessed. The basic elements of the moorLDI are shown schematically in the following figure.
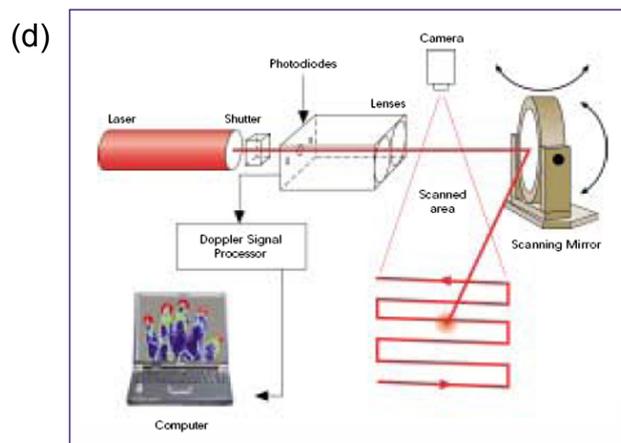
Both large areas (a full torso) and small areas (part of a finger) can be scanned enabling the blood flow to be mapped and colour coded images of the blood flow displayed. Regions of interest can be defined and statistical data calculated and recorded.

Single point measurements give a high temporal resolution (40Hz data rates are typical) enabling rapid blood flow changes to be recorded, whereas the laser Doppler imager can provide spatial information and has the ability to average blood flow measurements over large areas. Fibre optic systems can measure at tissue sites not easily accessible to a laser beam. For example measurements in brain tissue, mouth, gut, colon, muscle and bone.

Perfusion measurements using single and multiple channel fibre optic laser Doppler monitors have been made on practically all tissues and applied in most branches of medicine and physiology. The technique and its application have been described in numerous publications. A representative selection of these are included in 'Laser-Doppler Blood Flowmetry', ed. A.P. Shepherd and P.Å. Oberg, Kluwer Academic Publishers 1990 ISBN 0-7923-0508-6 and also 'Laser Doppler', ed. G.V. Belcaro, U. Hoffmann, A. Bollinger and A.N. Nicolaides, Med-Orion Publishing Co. 1994.

Definitions

The term commonly used to describe blood flow measured by the laser Doppler technique is 'flux': a quantity proportional to the product of the average speed of the blood cells and their number concentration (often referred to as blood volume). This is expressed in arbitrary 'perfusion units' and is calculated using the first moment of the power spectral density. Standardisation of LDF or LDI instrument measurements in perfusion units can be achieved by measuring a flux due to the Brownian motion of particles in a motility standard comprising polystyrene microspheres in water:

$$\text{flux} = k_1 \int_{\omega_1}^{\omega_2} \omega . \frac{P(\omega)}{I^n} d\omega - \text{noise}$$

$$= k_2 \times \textbf{Average Speed of blood cells} \times \textbf{Number Concentration of Blood Cells}$$

k_1 and k_2 are constants used to scale the raw output to a pre-determined calibration point.

ω is the Doppler shift in angular frequency units.

$\omega = 2\pi\Delta f$ where Δf is the Doppler shift frequency measured in frequency units Hertz
$\pi(\omega)$ is the optical power density at frequency ω.

I is the light intensity, used to normalise the signal to help eliminate the effects of gross variations in scattered light intensity (due to laser power differences, skin colour and beam angle of incidence).

n is an index which normally takes the value 2 if normalisation is being used to compensate for laser power differences. To compensate for tissue reflection changes in imaging a value of **n = 1** is a better normaliser.

ω_1 is the lower cut off angular frequency.

$\omega_1 = 2\pi f_1$ Typically f_1 is 20Hz in LDF instruments and 20Hz to 250Hz in LDI instruments depending on scan speed.

ω_2 is the upper cut off angular frequency: it is limited to reduce overall noise.

$\omega_2 = 2\pi f_2$ Typically f_2 is 15KHz for measurement of blood cell speeds up to approximately 6mm/s. For low blood cell speeds (~ 1mm/s), and for an optimum signal to noise ratio, 3KHz can be used. For speeds up to 10mm/s a 22KHz bandwidth is used. (These estimated speeds assume the laser wavelength is 785nm). For a given bandwidth the speed scales with laser wavelengths. The shorter the wavelength the shorter the speed range measured.

noise is the dark and shot noise components. Dark noise is a constant for an instrument and the shot noise level is proportional to **I**, the detected light intensity.