Digital Implementation of a Laser Doppler Perfusion Monitor

Examensarbete utfört inom Elektroniksystem, Electronics Systems av Ola Larsson

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Abstract

For 20 years Perimed AB have been developing and manufacturing LDPM and LDPI instruments based on an analog filter construction. The analog components in the construction are complex and suffer from non-linear frequency dependency and temperature drifts. The functionality of the design is also heavily depending on analog components which need to be trimmed in the production.

In this thesis, an alternative design employing a digital signal processor is presented. The signal processing method used is based on well established laser Doppler perfusion theories. The implemented signal processing algorithm calculates the perfusion from a sampled photodetector current, pre-filtered into AC and DC components by an analog detector card. The algorithm produces a raw perfusion signal by calculating a frequency weighted sum of the power spectral density, PSD, of the photocurrent. Detector noise compensation and light intensity normalization of the signal has also been implemented.

The presented digital implementation has been verified using the PF 5010 LDPM unit as a reference. In vitro measurements have shown similar behaviour as the reference in a highly perfused reference fluid as well as for a static scatterer. Furthermore, the DSP implementation has been verified on in vivo measurements of skin, showing nearly identical signal levels and response to heat provocation as the reference.

The demonstrated invention improves the manufacturability of the instruments since it reduces the number of electronic components significantly and thus, the amount of manufacturing tests. The DSP also reduces the temperature sensitivity of the instrument since it replaces several analog components sensitive to temperature changes.
Acknowledgements

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## Terminology

<table>
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<tr>
<th>Acronym</th>
<th>Full Form</th>
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<tbody>
<tr>
<td>ACF</td>
<td>Autocorrelation Function</td>
</tr>
<tr>
<td>ADC</td>
<td>Analog to Digital Converter</td>
</tr>
<tr>
<td>CMBC</td>
<td>Concentration of Moving Blood Cells</td>
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<tr>
<td>CPU</td>
<td>Central Processing Unit</td>
</tr>
<tr>
<td>DAQ</td>
<td>Data Acquisition</td>
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<td>DSP</td>
<td>Digital Signal Processor</td>
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<tr>
<td>EVM</td>
<td>Evaluation Module</td>
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<tr>
<td>FFT</td>
<td>Fast Fourier Transform</td>
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<tr>
<td>JTAG</td>
<td>Joint Test Action Group</td>
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<tr>
<td>LDF</td>
<td>Laser Doppler Flowmetry</td>
</tr>
<tr>
<td>LDPI</td>
<td>Laser Doppler Perfusion Imager</td>
</tr>
<tr>
<td>LDPM</td>
<td>Laser Doppler Perfusion Monitor</td>
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<tr>
<td>McBSP</td>
<td>Multichanneled Buffered Serial Port</td>
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<tr>
<td>MSPS</td>
<td>Mega Samples Per Second</td>
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<tr>
<td>PSD</td>
<td>Power Spectral Density</td>
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<tr>
<td>Pu</td>
<td>Perfusion units</td>
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<tr>
<td>RAM</td>
<td>Random Access Memory</td>
</tr>
<tr>
<td>RBC</td>
<td>Red Blood Cells</td>
</tr>
<tr>
<td>ROM</td>
<td>Read Only Memory</td>
</tr>
<tr>
<td>SAR</td>
<td>Successive Approximation Register</td>
</tr>
<tr>
<td>SNR</td>
<td>Signal to Noise Ratio</td>
</tr>
<tr>
<td>SPI</td>
<td>Serial Peripheral Interface</td>
</tr>
<tr>
<td>tcpO₂</td>
<td>Transcutaneous oxygen pressure</td>
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Chapter 1
Introduction

This chapter presents the background of Laser Doppler Flowmetry, LDF, and the aim of this thesis. A short description of Perimed AB is also included. The background chapter is a compilation of laser Doppler perfusion history found in Shepherd and Öberg's book, Laser Doppler Blood Flowmetry [5] and Biophotonics handbook chapter 15: Laser Doppler Perfusion Monitoring and Imaging, G. Nilsson et al. [7]. Previous patented methods are also presented.

1.1 Background

Laser Doppler flowmetry (LDF) is a widely used method for measuring microvascular blood perfusion. The first measurements using LDF were conducted in 1972 by Riva et al. [12], who studied red blood cell velocities in the retinal vessels of rabbits. In the following years the method was refined so the laser power could be decreased to eye safe levels and measurements on retinal vessels on humans, showing signals varying with blood flow, were made. Experiments included setups with free laser beams as well as laser beams guided by optical fibres.

The first measurements were all made on single vessels. M. D. Stern [13] was the first to suggest that LDF could be used to measure tissue perfusion. In 1973 he began his work to develop a device for measuring tissue perfusion by studying reflected laser light from a fingertip. When he couldn’t find a way to prove that the microvascular blood flow in the fingertip gave rise to speckle pattern due to Doppler shifts, he nearly gave up, but after consulting Navy physicist Ron Atkinson, he realized that it simply was a matter of insufficient measuring equipment and understanding of the underlying physical process. With help from Ralph Nossal an experimental setup was designed, consisting of a, low noise, He-Ne laser and a red-sensitive photomultiplier mounted behind a small pinhole collimator.

After travelling and demonstrating the experimental flow meter for researchers the commercial phase was initiated when Stern met A. Holloway and R. Rushmer during a visit in Seattle. Holloway was very enthusiastic about the new technique and helped convincing a local company, Medpacific to manufacture the flow meter. Shortly after Sterns visit, P. Å. Öberg visited Seattle and learned about the new method from Holloway. He brought his new knowledge to Sweden and Linköping University where he developed and patented a new flow meter together with G. Nilsson and T. Tenland. This flow meter was subsequently manufactured by Perimed, a company formed by S. Malmström and three inventors.

Early implementations of the flow meter suffered from a major limitation in the form of laser mode interference noise. This problem was overcome when a differential detector system was introduced. This system uses two detectors connected to a differential amplifier so that the
common mode modal interference of the laser is suppressed and the uncorrelated perfusion-related signal is amplified. [7]

The LDF project initiated by Stern was later inherited by R. Bonner. Together with R. Nossal he developed a theoretical model of light scattering in perfused tissues [14]. This model serves as the fundamental theoretical model on which present LDF systems are based. The model suggests that the microvascular perfusion is proportional to the first moment of the photocurrent and is together with the differential amplifier design mentioned above the most widely used method to produce a signal proportional to the product of the average speed and the concentration of blood cells in the microvascular beds of tissue.

As LDF instruments became commercially available, the properties of microvascular blood perfusion were studied by several researchers and clinicians. They discovered that microvascular blood perfusion possesses substantial spatial and temporal variability. This leads to great variations between measurement values taken from different measurement sites. This limitation of single point LDF instruments led to the development of LDF imagers at the end of the 1980s. LDF imagers or scanners use mirrors to scan the laser beam across the measurement site and a colour scaled perfusion map over the measurement site is produced.

At this time, most implementations of LDF instruments where based around an entirely analog design using the differential detector system described earlier in this chapter. This design led to big improvements compared to earlier designs but it still suffered from temperature dependency and non-linear dependency with respect to the frequencies produced by the photodetector system. In 1986, R. J. Adrian et al. [9] patented an alternative LDF monitor prototype with a digital signal processor which calculates the perfusion using the autocorrelation function, thus eliminating the need to calculate the Fourier transform of the photodetector signal. This reduced the need for a DSP sufficiently fast for real-time Fast Fourier Transform (FFT) calculation, which at this time was considered to be too expensive.

LDF instruments also suffer from movement artefacts which often can be observed as sharp spikes in the perfusion signal. In 2001, D. Bogget et al. [10] patented a method to compensate for movement artefacts using an FFT-based DSP algorithm to calculate the perfusion. FFT-based DSP algorithms also make it possible to divide the perfusion signal into velocity components as suggested by M. Larsson and T. Strömberg in their recently published article [11]. The method is based on the power spectral density, PSD, of the signal, which requires real-time FFT calculation.

1.2 Task

Perimed AB manufactures and develops Laser Doppler Perfusion Monitor (LDPM) instruments and Laser Doppler Perfusion Imaging (LDPI) instruments. The present implementation producing the first moment of the photocurrent or the raw perfusion is similar in both instrument types and consists entirely of analog amplifiers and filters. The aim of this project is to design a laser Doppler perfusion system which produces the raw perfusion signal by using a perfusion algorithm implemented in a digital signal processor mounted on an evaluation module.
The analog implementation in present instruments suffers from temperature drifts and non linear frequency dependency within the bandwidth of interest. Another drawback is that the analog filters need to be trimmed at production for better functionality. The digital implementation has the advantage of less temperature dependency and more linear frequency dependency. Another advantage is the possibility to calculate the power spectral density, PSD, of the photocurrent in real-time. A feature which increases the possibilities for signal processing related to the PSD such as movement artefact compensation suggested by Bogget et al. [10] or velocity resolved flowmetry as suggested by M. Larsson and T. Strömberg [11]. A DSP implementation provides the means to implement such methods. Software implementations also have the advantage of being easily changed without significant hardware changes.

The first goal and basic requirement of this project was to calculate the raw perfusion in a DSP algorithm. After the first goal was achieved the development continued with the implementation of detector noise compensation. The work included a pre-study phase, hardware and software design phase, an evaluation phase and a report writing phase.

1.3 About Perimed AB

Perimed AB develops and manufactures laser Doppler instruments for micro vascular blood perfusion measurement and tcpO₂ instruments for tissue oxygenation measurement. The company was founded in 1981 and the headquarters are located in Järfalla, Sweden. Perimed has subsidiaries in Europe, North America and Asia.
Chapter 2

Theory

This chapter describes the necessary theoretical background of the most essential parts of a complete laser Doppler flowmetry system including a brief description of microcirculation in tissue.

2.1 Microcirculation

The cardiovascular system includes the blood, the heart and the blood vessels. The main functions of the cardiovascular system are:

- To carry oxygen to cells and to carry carbon dioxide from the cells to the lungs
- To regulate body temperature and to regulate the pH level of the cells
- To protect against blood loss after an injury

The blood vessels which are carrying blood from the heart to the cells are called arteries. The arteries branches into the smaller arterioles, which branches into the microscopic capillaries, when entering tissue. In the capillary beds of the tissue, oxygen and metabolites are diffused into the interstitial fluid of the cells before the blood is continuing its way back to the heart via the venules and later the larger veins. [4]

Figure 1 shows a cross-section of the skin. The top layer of the skin, the epidermis, consists of keratinized cells, which protects underlying tissue from heat, microbes and chemicals. Oxygen and metabolites are supplied to the epidermis via the dermal papillae’s in the dermis layer. The dermal papillaries contains loops of capillaries and touch sensitive nerve endings. The flow of blood from small arteries (arterioles) to small veins (venules) trough capillaries in the dermal papillae’s is called the microcirculation. [4, 7]
The blood flow in the capillary beds cannot be given a well defined direction as in the case for larger vessels. The movement of the blood cells is therefore not referred to in terms of flow but rather in terms of **blood perfusion**. The blood perfusion is defined as the concentration of moving blood cells, CMBC, multiplied by their average velocity.

Microvascular blood perfusion in the skin is influenced by external factors such as heat or applied vasocative substances. Consumption of beverages and drugs also affects the blood perfusion, as well as smoking and mental stimuli. A number of common diseases, such as inflammatory conditions, allergic reactions and tumours, also affect the blood flow in the skin [7].

### 2.2 Laser Doppler Perfusion theory

#### 2.2.1 The Doppler effect

The Doppler effect, named after the Austrian scientist Johann Christian Doppler (1803-1853) [5], can be applied to both sound waves and electromagnetic waves. In both cases the Doppler effect causes a frequency shift of the sound or light wave. This effect can be observed for example when an ambulance passes. The tone of the ambulance will be higher when it’s moving towards you compared to when it’s moving away from you.

![Figure 2 The Doppler effect](image)

In the case of a static sound source, an observer receives \( \frac{v_s T}{\lambda} \) wavelengths in time T. The frequency which is observed is \( f_s = \frac{v_s}{\lambda} \). If the observer is moving with the velocity \( v \), an additional \( \frac{v T}{\lambda} \) wavelengths will be received. The observer will now hear the frequency

\[
f_0 = \left( \frac{v_s T}{\lambda} + \frac{v T}{\lambda} \right)/T = \frac{(v_s + v)/\lambda}{(v_s + v)/v_s} = f_s \left( \frac{v_s + v}{v_s} \right)
\]

**Equation 1**

The Doppler effect formula for sound is depending on weather it is the sound source or the observer that is moving. If the source was moving instead another formula would be derived.

The formula above is derived for the case when the observer is moving directly towards the sound source. If, for example the observer is moving instead the formula would be different. The shift also depends on the angle between the observer and the direction, in which the sound source is moving. If the sound source is moving with velocity \( v_{sa} \), and the vector
between the observer and the source is forming the angle $\theta$ with the direction of $\mathbf{v}_{sa}$, the expression for $v_s$ is:

$$v_s = v_{sa} \cdot \cos(\theta)$$

The fundamental difference between sound waves and light waves is that the speed of light is independent of the medium, in which the wave is travelling. Therefore, it’s only the relative speed between the observer and the source which determines the Doppler shift. The Doppler formula for light can be expressed as

$$f_0 = f_s \frac{1 - v_{os}/c}{\sqrt{1 - (v_{os}/c)^2}}, \quad \text{Equation 2}$$

where $v_{os}$ is the relative velocity between the observer and the source.

By studying Doppler shift in light omitted from stars the speed and direction of the star can be determined. If the star is moving towards an observer the light will appear as blue shifted and if it’s moving away it will appear as red shifted. The magnitude of the shift can be used to determine the velocity.

Electromagnetic waves scattered on the surface of an object will also be Doppler shifted if the object is moving. A positive shift will occur when the scattering object is moving towards the light source and a negative shift will occur if it’s moving away.

### 2.2.2 Single photon scattering

A light beam can be viewed as a stream of photons, each undergoing a frequency shift when being scattered by a moving object. The relation between an incoming wave, the speed of the scattering object and the scattered wave, $E_s$, can be described as follows:

$$E_s = E_{so} e^{i\alpha_{so}} e^{-iqvt}, \quad \text{where } q \text{ is defined as magnitude of the scattering vector } q = k_i - k_s$$

and $E_{so}$ is the magnitude of the scattered wave

![Figure 3 Single scattering event](image)

While $k_i \approx k_s$, the magnitude of the scattering vector can be described as:

$$q = 2k \sin\left(\frac{\alpha}{2}\right) = \frac{4\pi}{\lambda_s} \sin\left(\frac{\alpha}{2}\right) \quad \text{Equation 3}$$

where $\lambda_s$ is the wavelength in tissue.
The Doppler shift can then be determined by the time-variant phase factor \( qv = v \cos(\theta) \), which can be expressed as:

\[
\sigma_D = \frac{4\pi}{\lambda} \sin\left(\frac{\alpha}{2}\right) v \cos(\theta) \quad [7] \tag{Equation 4}
\]

In the case of light with a wavelength of 780 nm and red blood cells, the scattering angle, \( \alpha \) is very small. This is due to the relatively large size of the red blood cells. The angle between \( q \) and \( v \) is arbitrary due to the random nature of the blood flow in capillaries and light scattering in tissue (see next chapter). Therefore the maximum shift that can occur can be described as

\[
\sigma_{\text{max}} = \frac{4\pi}{\lambda} \sin\left(\frac{\alpha}{2}\right) v \quad \tag{Equation 5}
\]

The speed of blood cell travelling in capillaries is less than 10 mm/s and when 780 nm laser light is used the maximum frequency shift is ca 25 kHz. However, the average shift is much lower since \( \alpha \) is small.

### 2.2.3 Light scattering in tissue

When light enters the skin it will be scattered by moving particles in the bloodstream and by static tissue. The photons migrate through the skin in a random pattern and are frequency shifted according to the Doppler principle when scattered by moving particles.

![Figure 4 Photon scattering in tissue](image)

The reflected light can be divided into unshifted and shifted light. The interference of the scattered light results in a fluctuating interference pattern which can be observed on the detector surface. The resulting photocurrent can be divided into three parts, which arises from:

- Stationary, unshifted light with no interference from shifted light
- Heterodyne mixing, shifted and unshifted light mixed
- Homodyne mixing, shifted light mixed
Most of the reflected light is unshifted light, which only have been reflected by static tissue, only a small portion of the light is Doppler shifted according to equation 4. [7]

The optical properties of tissue are complex to describe. Therefore, the behaviour of the photons reflected in the capillary bed cannot be described analytically. Instead, Monte Carlo simulations of photons in a tissue model known as “turbid media” has been used to simulate the behaviour of the photons.

The random Doppler shifts of the photons results in a continuous broadening of the optical spectrum of the reflected light. The frequency shifts follows a random distribution which only can be simulated using the method mentioned above.

![Figure 5 Continuous broadening of the optical spectrum](image)

Figure 5 symbolizes how the optical spectrum ($S(f)$) of the reflected light is broadened around the frequency ($F_L$) of the incoming light [2].

### 2.2.4 Signal processing

On the detector an interference pattern is formed due to the interaction of the reflected light. The resulting photocurrent consists of a static part, $i_{DC}$, and a fluctuating part, $i_{AC}$. The light is interfering coherently on a coherence area, which is defined as:

$$A_c = \frac{\lambda^2}{\Omega} = \left(\frac{Ar}{d}\right)^2$$  \hspace{1cm} \text{Equation 6}$$

where $\lambda$ is the wavelength in vacuum, $\Omega$ is the solid angle under which the laser spot on the object is seen from the detector surface and for fibre based systems $r$ is the radius of the fibre and $d$ is the distance from the detector to the fibre. In LDPM applications the coherence area can be seen as constant.

The total contribution to the photocurrent from all coherence areas can be written as

$$i(t) = \sum_{j=1}^{N} \left(\langle i_{DC}\rangle + \Delta i_{AC}(t)\right)$$  \hspace{1cm} \text{where $N$ is the number of coherence areas} \hspace{1cm} \text{Equation 7}$$
The total signal is stochastic and can be described with the autocorrelation function, which is directly related to the PSD of the signal. The total autocorrelation can be simplified to:

$$\langle i(0) i^*(t) \rangle = N^2 \langle i_{DC}^2 \rangle + N \langle \Delta i_{AC}(0) \Delta i_{AC}^*(\tau) \rangle,$$  \hspace{1cm} \text{Equation 8}

where \( \Delta i_{AC}(\tau) \) represents the average fluctuating part of a single coherence area. Furthermore, the autocorrelation can be divided into terms of the origin of the current,

$$\langle i(0) i^*(t) \rangle = \frac{2 i_{Re} i_{Sc}}{\text{Stationary}} + i_{Re}^2 \left( \frac{e^{i\text{opt}} + e^{-i\text{opt}}}{2} \right) + i_{Sc}^2 \left( \frac{e^{i\text{opt}} + e^{-i\text{opt}}}{2} \right) + i_{Sc} \sum_{k=1, k \neq l}^{S} \left( e^{i(\delta_k \nu_l - \delta_l \nu_k)} \right) \tau.$$  \hspace{1cm} \text{Equation 9}

\textbf{Equation 9} Autocorrelation function of the current produced by a single coherence area

where \( i_{Re} \) and \( i_{Sc} \) represents the average of the currents produced by the unshifted light and the shifted light within a coherence area, respectively. For low and moderate CMBC, \( i_{Sc} \) is proportional to the total current produced by a single coherence area and \( i_{Sc} = CMBC \cdot i_{Re} \). The autocorrelation function of the heterodyne part can therefore be expressed as:

$$r(t) = CMBC \cdot i_{Re}^2 \left( e^{i\text{opt}} + e^{-i\text{opt}} \right)$$  \hspace{1cm} \text{Equation 10}

In applications including low and moderate blood volumes the heterodyne part dominates over the homodyne part. This is the case for most applications which is why the homodyne part mostly is disregarded.

According to the Wiener-Khintchine theorem, the Fourier transform of the autocorrelation function, \( r(t) \), is equal to the power spectral density, PSD, of the input. That is

$$P(f) = \int_{-\infty}^{\infty} r(\tau) e^{-i2\pi f \tau} d\tau$$  \hspace{1cm} \text{Equation 11}

which for the heterodyne part equals

$$P(f) = CMBC \cdot i_{Re}^2 \left( e^{i\text{opt}} + e^{-i\text{opt}} \right) e^{-\pi f^2/2} d\tau.$$  \hspace{1cm} \text{Equation 12}

This is the basis of the proof of the perfusion formula. By further derivation of this expression it can be shown that the quantity \( \int f P(f) df \) scales with the product of the CMBC and the average speed \( \langle v \rangle \). This product is defined as the \textit{perfusion} in the scattering volume.

$$\text{Perfusion} = CMBC \cdot \langle v \rangle \propto \int f P(f) df \hspace{1cm} [7]$$ \hspace{1cm} \text{Equation 13}

where \( P(f) \) is the PSD of the heterodyne term of the photocurrent and \( \langle v \rangle \) is the mean
speed of the blood cells. For the full proof of the perfusion formula, please refer to reference no. 7.

The integral limits are determined by the bandwidth of the active signal, which usually is less than the maximum Doppler shift. The lower limit is usually around 20 Hz to avoid spectral leakage from the DC, which is of no interest.

The autocorrelation function and the PSD scales with \( i_{pe}^2 \), so to make the formula independent of the total light intensity at the detector surface, the PSD is normalized with the factor \( 1/i_{pe}^2 \). (In the rest of this document the term \( I \) will be used for the DC instead of \( i_{pe} \)). The normalized perfusion formula can then be expressed as

\[
\text{Perfusion} = \text{CMBC} \* \{v\} = \frac{k}{I^2} \int_{f_L}^{f_H} fP(f)df
\]

**Equation 14**

The proof of the perfusion formula is based on these assumptions:

- The total number of shifted photons is much lower than the number of non shifted photons.
- Most of the shifted photons are only shifted one time before returning to the photodetector.
- The speed and direction of the blood cells follows a random velocity distribution.
- The angle between the scattering vector and the velocity is arbitrary.
- The scattering angle is small and statistically known.

### 2.2.5 Photo detector and amplifiers

In LDPM instruments the laser light is led to a measuring probe via an optical transmitter fibre and the reflected light is led back via a receiver fibre. At the other end of the receiver fibre, a two-element Si-detector is placed so that the remitted light is illuminating the two elements evenly.

![Figure 6 The measuring probe](image-url)
The elements of the photo detector produce two currents, which are converted to voltages and filtered into AC and DC. The AC signals are then amplified using a differential amplifier, which rejects the common mode currents and amplifies the fluctuating currents. The filters and amplifiers are further described in chapter 3.1.2.

2.2.6 Detector noise compensation

The AC signal produced by the photo detector and filters also contains noise produced by the detector. This noise adds an offset to the perfusion, which must be accounted for.

\[
\text{Perf}_{\text{raw}} = \int_{f_c}^{f_H} f \cdot (P_{\text{signal}}(f) + P_{\text{noise}}(f)) df = \int_{f_c}^{f_H} f \cdot P_{\text{signal}} df + \int_{f_c}^{f_H} f \cdot P_{\text{noise}}(f) df = \text{Perf} + \text{Perf}_{\text{noise}}
\]

Equation 15

The detector noise consists of a static part, independent of the light intensity at the detector surface, and one part which increases linearly with the intensity. These parts are referred to as dark noise and shot noise, respectively. Therefore the perfusion offset can be fitted to a first order polynomial, i.e.

\[
\text{Perf}_{\text{noise}}(I) = C_1 I + C_0.
\]

Equation 16

The perfusion offset can therefore be derived from the DC level. This is convenient while only two calibration constants are needed to perform the noise compensation. The resulting calibrated perfusion formula equals:

\[
\text{Perf}_{\text{cal}} = \frac{k}{f^2} \left( \int_{f_c}^{f_H} f \cdot P(f) df - (C_1 I + C_0) \right) = \frac{k}{f^2} (\text{Perf}_{\text{raw}} - \text{Perf}_{\text{noise}})
\]

Equation 17

\[
k = \text{instrumental constant or gain factor}
\]

\[
I = \text{DC produced by the photodetector}
\]

\[
f_c \text{ and } f_H \text{ are low and high cut off frequencies respectively}
\]

\[
P \text{ is the power spectral density of the photocurrent (AC)}
\]

The instrumental constant, \(k\), is used to adjust the gain of the instrument.

2.3 Perfusion units

Perfusion cannot be measured in an absolute unit, for example in ml/min/100 gram tissue. This is mainly because simulated models of photon scattering in tissue is not sufficiently accurate to exactly define the measuring depth and volume.

LDPM instruments manufactured by Perimed are though calibrated using a special motility standard. The motility standard is a suspension of latex spheres with a well-defined Brownian motion. The latex spheres are smaller than red blood cells, RBC, which leads to bigger
frequency shifts than for a corresponding velocity distribution of RBC. However, the corresponding PSD is rather stable, which makes the motility standard a suitable reference. By calculating the perfusion when the tip of the probe is in the motility fluid, the instrument gain can be adjusted to show a desired perfusion value. The perfusion in the motility standard is defined as 250 Pu.

2.4 Digital signal processing

2.4.1 The Discrete Fourier Transform

The discrete Fourier transform, DFT, is a fundamental block in the digital signal processing area. It defines the discrete counterpart of the Fourier transform theory. While the continuous Fourier transform always is applied to a periodic signal, or a non-periodic signal extended to infinity, the discrete Fourier transform is defined for non-periodic signals. The DFT is defined as:

\[
X[k] = \sum_{n=0}^{N-1} x[n] e^{-j2\pi kn/N} = \sum_{n=0}^{N-1} x[n] W_N^{kn} \quad \text{Equation 18}
\]

As the use of the DFT and the data flow increased, the need for more efficient algorithms grew bigger. The DFT implemented “as it is”, is not a very efficient algorithm, while it has redundancy due to the periodicity of the factor \( W_N^{kn} \). The number of calculations of the DFT grows with the square of the signal length, which is notated \( \mathcal{O}(N^2) \) for order.

2.4.2 The Fast Fourier transform

In the 1960s, scientists utilized the redundancy of the DFT to write a more efficient algorithm, and the first FFT algorithm was developed. Today, there are a number of different standard FFT algorithms with different features, but with one common factor: the order is \( \mathcal{O}(N \log N) \).

One widely used FFT algorithm is the radix-2, decimation in time, algorithm. This method is using the redundancy of the DFT to divide the input signal into subsequences, thus dividing the DFT sum into smaller sums. This method is known as decomposition.
2.4.3 Power spectral density

Power spectral density, PSD, or power spectrum measures how the total power of a signal is distributed over the frequency domain. The most direct method to calculate the PSD of a discrete signal is to calculate the so-called periodograms of the input sequence. The method provides a direct and easy method to derive an estimate of the PSD from the DFT. The periodogram of an input sequence, \( x[n] \), is defined as:

\[
P_{xx}(\Omega) = \frac{1}{N} |X(\Omega)|^2 = \frac{1}{N} \left| \sum_{n=0}^{N-1} x[n] e^{-j\Omega n} \right|^2
\]  

\[\text{Equation 19}\]

PSD estimation using periodogram

Basically, this means that the PSD estimate can be calculated by squaring the absolute values of the DFT and divide by the number of input samples, \( N \). Another, indirect method is to calculate the autocorrelation function of the sequence and then transform it to the frequency domain. In this thesis the periodogram method is used.

2.4.4 Windowing

The finite input sequence to the DFT is often referred to as the observation window. While the sequence is finite, the infinite impulse response of the sequence will be truncated. The finite sequence, \( x_t[n] \), can be seen as an infinite sequence \( x[n] \), multiplied by a finite rectangular window function, \( w[n] \): 

\[
x_t[n] = x[n] * w[n]
\]

The consequence of windowing is so-called spectral leakage due to the sinc form of the finite step response.

![Spectral leakage of the rectangular window](image)

Figure 7 Spectral leakage of the rectangular window [3]

2.4.4.1 Window functions

To reduce the spectral leakage, several window functions have been developed and characterized by scientists. Two of those functions have been implemented and tested in the software of this implementation: The Hamming, and Bartlett (triangular) windows.
The equation of an N-point Hamming window function is shown below

\[ w[n] = 0.54 + 0.46 \cos\left(\frac{(2n - N + 1)\pi}{N}\right) \]  

[Equation 20]

Figure 8 shows the form of the Hamming window function and the corresponding frequency response.

![Figure 8 Frequency response of the Hamming window [3]](image)

The equation of an N-point Bartlett window function is shown below

\[ w[n] = 1 - \frac{|2n - N + 1|}{N} \]  

[Equation 21]

Figure 9 shows the form of the Bartlett window function and the corresponding frequency response.

![Figure 9 Frequency response of the Bartlett window [3]](image)

The theoretical difference between these two window function is that the frequency response of the Hamming window shows a slightly narrower peak down to ca -40 dB compared to that of the Bartlett window, which on the other side decreases faster at that point and beyond.
2.4.5 Welch's method

The Welch method is a special syntax used in digital signal processing to estimate an averaged PSD of an input sequence. It is a method which provides an efficient way to reduce the noise of the periodogram. The PSD estimation is calculated using the following syntax:

- Divide the input vector to segments of equal length. Each segment is overlapping the prior by 50% percent of the segment length.
- Apply a window to each segment
- Calculate the DFT of each segment
- Calculate the periodogram of each segment
- Calculate the average of all periodograms
Chapter 3
Experimental setup

This chapter describes the hardware used in the experimental setup. Each module will be described in terms of their function and how they are used. For more information about the devices, refer to the list of components in Appendix A.

Figure 10 shows the test setup, which has been used throughout the project. The modules included are:
- PF 5000 system with a PF 5010 LDPM unit and PF 5020 temp unit, *from Perimed*
- NI-PCI 6023E, 200 kS/s, 16-bit Data acquisition card, DAQ, *from National Instruments*
- Circuit board with amplifiers, anti-aliasing filter and ADC, *Self-made*
- TMS320VC5509 DSP evaluation module (EVM) and XDS5510PP Plus JTAG emulator, *from Spectrum Digital*

![Figure 10 The experimental instrument setup](image)

3.1.1 PF 5010 LDPM unit

To be able to focus on the task, rather than constructing new filters to extract DC and AC from the photocurrent, the PF 5010 LDPM unit was used to perform this task. The detector card which produces the AC and DC signals is described in the next section.
3.1.2 Detector card

The detector card in the LDPM unit is calculating the raw perfusion using a chain of analog filters illustrated in Figure 11. The calibrated perfusion is then calculated in the DSP software.

The amount of reflected light collected is usually very small compared to the amount of emitted light. This property and the low laser effect allowed for safety regulations results in a very small photo detector current, which needs to be amplified with a very high gain. These amplifiers are called the I/V stage of the signal processing chain while they are also converting the current to a voltage.

The two signals are then filtered into DC and AC components using low and high pass filters respectively. The AC signals are then amplified using a differential amplifier, which rejects common mode currents of the two signals. This means that the actual Doppler shift information in the signal is amplified more than signal that are common for the two photodetector elements, such as variations in the surrounding light intensity or variations in the intensity of the laser light.

The rest of the analog signal processing chain is calculating the raw perfusion signal using specialized filters. The AC signal produced by the differential amplifier and one of the intensity (DC) signals is connected to the signal adjustment board and sampled by the DSP.

![Figure 11 Detector card filter chain](image-url)
3.1.3 Calibration

The PF 5010 can perform a full calibration using a light emitting diode to illuminate the photo detector surface with a number of selected intensity levels for a short time at each level. The intensity is regulated by a regulatory system where the DC level of the photodetector is used as feedback to the system to produce a stepwise, linearly increasing intensity function as shown in Figure 12 (right image). The raw perfusion and DC level is measured and averaged at each intensity level. The resulting data points are then fitted to a first order polynomial to produce the two constants $C_0$ and $C_1$ mentioned in chapter 2.2.6.

![Figure 12 Raw perfusion (left) and DC level (right) during calibration](image)

By running the DSP during this calibration process, a similar calibration can be made for the DSP. By exporting the perfusion and DC vectors to Matlab the mean perfusion at each DC level can be calculated. These mean values and DC levels are then used as input to a curve fitting algorithm in the DSP. The reason why the calibration algorithm isn’t entirely made in the DSP is the difficulties to sync the calibration process of the PF 5010 with the DSP algorithm or to control the calibration LED from the DSP.

3.2 Data acquisition card

The DAQ card is used to sample AC, DC, perfusion and raw perfusion, signals from the PF 5010 unit to Matlab. In the temperature experiments it was also used to sample the temperature signal from the PF 5020 unit.
3.3 Signal adjustment and A/D conversion

The AC and DC signals from the PF 5010 unit are sampled into the DSP, but first they are adjusted to fit the reference voltage of the analog to digital converter, ADC. The relatively high bandwidth of the AC signal requires an ADC with a fast conversion rate and an ability to clock the data at a fast rate. The ADC is a two-channel, 1 MSPS, Successive Approximation Register ADC with an SPI-interface towards the DSP. The DSP samples the AC at 50 KHz and the DC at 12.5 kHz.

Before the AC signal is sampled it is processed with an 8\textsuperscript{th} order anti-aliasing filter, with a cut-off frequency at 20 kHz. The sampling frequency and cut-off frequency was chosen to fit the bandwidth of the photodetector current, which is expected to be below 20 kHz according to the discussion in chapter 2.2.2.

The schematic layout of the signal adjustment board can be viewed in Appendix B.

3.4 TMS320VC5509 EVM

The EVM provides an easy way to get started with the evaluation of the DSP. The board contains external RAM, Flash ROM and JTAG interface towards the debugger among other things. The TMS320C5509 is a fixed point DSP with a 16-bit core.
Chapter 4
Software design

This chapter describes the software written for the DSP. The most essential parts of the code are presented without getting into details. More details can be found in the code flow charts in Appendix D.

4.1 Basic design considerations

The first software that was implemented in the project was made in Matlab to quickly get an idea of how a good DSP algorithm could be designed. In Matlab, the `pwelch` function was used to calculate the PSD. The `pwelch` function is an implementation of Welch’s method described in chapter 2.4.5. The `pwelch` function reduces noise in the PSD very efficiently, since it is averaging spectras over the sample time period. Of course, in a real-time DSP application it can be hard to average spectras over a long time period, while averaging of long data sequences are both time- and space consuming. However, the DSP software design was based on a modification of Welch’s method.

4.1.1 Overlap

To be able to calculate a new perfusion value, a new periodogram must be calculated and to increase the rate in which perfusion (and periodograms) values are calculated, overlap can be used as illustrated in Figure 13. The perfusion rate can be calculated using this equation:

Perfusion rate = \( \frac{F_s}{NFFT \cdot (1 - \text{overlap})} \), where overlap is (overlap in percent) / 100  \[ \text{Equation 22} \]

For example, a 2048 point spectra with 50% overlap, and 50 kHz sampling frequency yields:

\[
\text{Perfusion rate} = \frac{50 \times 10^3}{2048 \cdot (1 - 0.5)} = 48.828125 \text{ Hz}
\]

Figure 13 Overlapping FFT windows

Of course, an alternative way to increase the periodogram rate would be to increase the sampling rate or reduce the FFT size to the cost of decreased frequency resolution.
4.1.2 Frequency resolution limitation

The frequency resolution of the FFT equals

\[ R_f = \frac{F_s}{N_{FFT}} \]

\text{Equation 23}

A high resolution is convenient because the lower cut-off frequency of the perfusion integral can then easily be set by excluding elements in the periodogram which corresponds to frequencies lower than the low cut-off frequency. To be able to set the lower cut-off frequency to 25 Hz, the resolution must be 25 Hz or lower. A sampling frequency of 50 kHz and a 2048 point FFT yields:

\[ R_f = \frac{50e3}{2048} = 24,4140625 \text{ Hz} \]

While the FFT algorithm discussed in the next chapter has a maximum input length of 2048 samples the lower limit of the spectral resolution is 24,4140625 Hz. Preferably, this limit should be lower. However, the first points of the FFT can contain leakage from the DC component so when setting the lowest possible limit the signal might contain that leaking signal.

Another approach to set the lower cut-off frequency would be to filter the AC signal using a high pass filter with a cut-off frequency at the desired lower limit of the integral. This reduces the need for a higher frequency resolution.

4.1.3 FFT algorithm

Texas Instruments provides a free software library called \texttt{dsplib}, which includes a number of optimized FFT algorithms. The FFT algorithms evaluated in this project are the \texttt{rfft} and \texttt{rfft32} algorithms. These are Radix-2 two algorithms with 16-bit and 32-bit integer input, respectively. The later is used in the present implementation because it reduces the need for scaling in the FFT algorithm, thus keeping the resolution at its maximum. Of course the 32-bit version is about half as fast as the 16-bit version.

The FFT algorithms are written in assembly and can be called from a C program. This is an \texttt{rfft32} call:

\texttt{rfft32(fftstore, NFFT, NOSCALE);}  

The first argument is a pointer to the input sequence and to where the result is to be stored. The second argument is the input length and the third is a flag which indicates that no scaling should be done to prevent overflow. \texttt{NFFT} must be a power of 2.
The FFT for frequencies from zero to the Nyqvist frequency is stored in the `fftstore` buffer using the following format.

<table>
<thead>
<tr>
<th>Memory location*</th>
<th>Data before call, (x[n])</th>
<th>Data after call, (X[k])</th>
</tr>
</thead>
<tbody>
<tr>
<td>(N)</td>
<td>(x[0])</td>
<td>(\text{Re}(X[0]))</td>
</tr>
<tr>
<td>(N+2)</td>
<td>(x[1])</td>
<td>(\text{Im}(X[0]))</td>
</tr>
<tr>
<td>(N+4)</td>
<td>(x[2])</td>
<td>(\text{Re}(X[1]))</td>
</tr>
<tr>
<td>(N+6)</td>
<td>(x[3])</td>
<td>(\text{Im}(X[1]))</td>
</tr>
<tr>
<td>...</td>
<td>...</td>
<td>...</td>
</tr>
<tr>
<td>(N+4*(\text{NFFT}-1)-1)</td>
<td>(x[\text{NFFT-2}])</td>
<td>(\text{Re}(X[\text{NFFT/2-1}]))</td>
</tr>
<tr>
<td>(N+4*(\text{NFFT}-1))</td>
<td>(x[\text{NFFT-1}])</td>
<td>(\text{Im}(X[\text{NFFT/2-1}]))</td>
</tr>
</tbody>
</table>

Table 1 Format of FFT storage buffer

*(While numbers are stored in 32-bit format, each number occupies two memory locations)*

The FFT algorithms are using in-place calculation which means that the result is stored in the input buffer. While the FFT windows are overlapping it’s not possible to use a part of the input buffer as input to the FFT algorithm. Therefore, the FFT window is always copied to the `fftstore` buffer before the FFT is calculated.

4.1.4 Timing

It is critical that the time needed to calculate the perfusion is less than the time it takes to fill the FFT window, the buffering time. Using 2048 FFT points, 50 percent overlap and 50 kHz sampling rate results in a buffering time of \(2048*0.5/50e3 \approx 20\) ms. The implemented code results in an execution time of ca 2 ms for this case using 16-bit resolution and ca 4 ms using 32-bit resolution. This means that it is possible to use a larger overlap to increase the perfusion rate.

4.2 Sampling

The maximum frequency shift that can be expected is ca 25 kHz but the average shift is much lower according to the theory discussed in chapter 2.2.2. A sampling frequency of 50 kHz and an anti-aliasing filter with a cut-off frequency at 20 kHz should be well adapted to the measurement situation. Signal information above 20 kHz is of no interest. The sampling routine is further illustrated in Appendix D.

4.2.1 Timer

Sampling is synchronized using timer interrupts. The timer interrupt rate is set to 100 kHz to be able to sample both AC and DC at 50 kHz. However, it’s not required to use every DC sample, which is why only every fourth sample is stored in the DC input buffer. The timer interrupt routine initiates every conversion by writing a serial word, containing channel information, to the ADC.
4.2.2 Multichanneled Buffered Serial Port

The serial interface of the C5509 DSP, McBSP, can be configured for a number of standard serial interfaces including the SPI interface. This capability is provided when the McBSP is put in clock stop mode. This means that the SPI clock (SCLK) will only run for 16 clock periods when a CS signal is issued. Figure 14 shows the SPI-compatible ADC used in this project.

![Figure 14 ADC interface towards DSP](image)

After each AD conversion, a receive interrupt is generated where AC or DC data is stored in buffers. In the interrupt routine a flag is set when the input buffer contains NFFT/2 new input samples. This flag is read in the main loop and when it is set a perfusion calculation sequence is executed.

4.3 Windowing

For better spectral resolution, windowing is required. Each FFT window is windowed using an integer window stored as a vector. This is to avoid calculating window data for each FFT window. The window is a weighting of the FFT windowing and is containing only numbers from 0 to 1, so to be able to use integers the window is scaled by the window length divided by two, NFFT/2. The windowed data is then scaled again by the factor 2/NFFT. Figure 15 shows the Hamming and Triangular windows which have been implemented and evaluated.

![Figure 15 Hamming (left) and triangular (right) window functions with integer values](image)
The window functions are calculated and stored when the program starts. This is done to avoid calculating the window function for each new FFT. The input sequence is simply multiplied by the window stored in the memory.

The expected effect of applying these window functions to the input sequence is described in chapter 2.4.4.1.

4.4 Periodogram

The periodogram of the FFT is calculated simply by squaring the real and imaginary part and adding them according to the theory presented in chapter 2.4.3. No scaling is done.

4.5 Averaging

Averaging of the periodograms is implemented for noise reduction. The averaging is very space and time consuming and is for now only implemented as a moving average of the two latest periodograms. The number of averaging periodograms can easily be changed by just changing a constant.

4.6 Perfusion

The discrete counterpart of the perfusion formula (equation 15) can be expressed as

$$\text{Perf}_{\text{cal}} = \frac{K}{I^2} \left( \sum_{k_L}^{k_H} f_k P(f_k) - \text{Perf}_{\text{noise}} \right), \quad \text{Equation 24}$$

where \( f_k \) can be written as

$$f_k = k \cdot f_R, \text{ where } f_R \text{ is the frequency resolution } f_s / \text{NFFT}$$

Replacing \( f(k) \) yields

$$\text{Perf}_{\text{cal}} = \frac{Kf_R}{I^2} \left( \sum_{k_L}^{k_H} kP_k - \text{Perf}_{\text{noise}} \right) = \frac{M}{I^2} \left( \sum_{k_L}^{k_H} kP_k - \text{Perf}_{\text{noise}} \right), \text{ where } M \text{ is the gain} \quad \text{Equation 25}$$

This solution provides a mean to use integer indexes instead of a floating point vector \( f(k) \), thus providing a faster solution. The only floating point number needed is the gain factor M.
The perfusion is calculated according to the calibrated perfusion formula following these steps:

- Calculate the raw perfusion: \( \text{Perf}_{\text{raw}} = \sum_{k} kP_{k} \) (here scaling is necessary to prevent overflow)

- Calculate the mean DC level: \( V_{DC} \)

- Subtract noise: \( \text{Perf}_{\text{raw}} - (C_{0} + C_{1}V_{DC}) \), where the constants are calculated in the calibration process described in chapter 5.7.

- Scale and normalize: \( \text{Perf}_{\text{cal}} = \frac{M}{V_{DC}^{2}} (\text{Perf}_{\text{raw}} - (C_{0} + C_{1}V_{DC})) \)

The scaling has been adjusted to fit the range of the input signal during measurement. It is not guaranteed that overflow is prevented for any given input. The final implementation has a low cut-off frequency at 25 Hz and a high cut-off frequency at 16 kHz.

### 4.7 Detector noise calibration

Noise calibration is performed partly in Matlab. The DSP is calculating and storing raw perfusion values during the execution of the calibration sequence run in the LDPM unit. The DC and perfusion buffers are then exported to Matlab where the perfusion is averaged at each DC level. The DC and perfusion points are used in the DSP as input to a curve-fitting function which uses the least-square method. Figure 16 shows the result of the curve-fitting function.

![Figure 16 Noise curve-fitting](image-url)
Chapter 5
Experiments

This chapter describes the experiments and measurements done to evaluate the functionality of different parts of the laser Doppler algorithm. The tests were made using Matlab calculations or the PF 5010 LDPM unit as a reference.

5.1 PSD test

In this test Matlab was used as reference to verify the PSD calculated in the DSP. The test was to calculate and store PSD:s for ca 5 seconds, using the DSP. The same test was then done using Matlab. Afterwards, the stored PSD:s were averaged and plotted. The PSD was evaluated for two reference objects: motility and delrin. Figure 17 shows the results:

Figure 17 PSD for motility (left) and delrin (right)

Note: The periodograms have been scaled differently to be able to separate them from each other.

The test showed that the PSD calculated in the DSP contains low frequent noise, roughly around 250-600 Hz (1.), not present in the Matlab implementation. Some attempts were made to reduce this noise without success. The low frequency noise could probably be reduced by better circuit board layout. The PSD of the DSP also shows a sudden increase (2.) when the frequency is close to the Nyqvist frequency. No big efforts were put on solving this error while it has no, or a very small, effect on the calculated perfusion. The overall result of the test is that the DSP implementation produces a PSD which is well correlated to the Matlab reference.
5.2 Frequency dependency test

The analog filter chain shown in figure 11 produces a perfusion signal which suffers from non-linearity with respect to frequency. The effect can be observed mainly for frequencies above ca 14 kHz. This test was performed to verify that the perfusion algorithm produces a signal which is linearly depending on the input frequency. To verify this property a signal generator was used to generate a sine signal which was used as input directly to the ADC. The sine signal was manually increased in steps between the low and high cutoff frequencies and the corresponding perfusion values was calculated and stored. To be able to show the frequency dependency up to 20 kHz the upper cutoff frequency was set to 20 kHz in this test. Figure 18 shows the results of the measurements compared to a typical frequency response curve of the PF 5010.

![Perfusion vs. frequency](image1)

**Figure 18 Perfusion vs. input frequency**

The optimal line is chosen as a linear curve fitted to the perfusion at 1 kHz and 12 kHz. The relative error to the optimal line is presented in Figure 19.

![Relative error vs. frequency](image2)

**Figure 19 Relative error vs. frequency**
The non-linearity of the DSP solution arises from the anti-aliasing filter and possibly other filters in the chain. An anti-aliasing filter with a higher cut-off frequency would probably lead to improved linearity at higher frequencies.

A quick analyze of the results showed a relative error of ca 4.5 % for the DSP and 21.1 % for the PF 5010 at 16 kHz. The test showed that the perfusion is close to linear with respect to input frequency up to the upper cut-off frequency. For the final implementation the maximum relative error is ca 4.5 percent, but it can probably be improved by better filter design in the whole signal chain and especially by using an anti-aliasing filter with a higher cut-off frequency.

5.3 Amplitude dependency test

The periodogram is a square sum of the FFT and therefore the perfusion should be a function of the square of the input amplitude. This test was performed to verify this property. The input signal was a sine with constant frequency generated by a signal generator. The amplitude was increased manually.

![Figure 20 Perfusion vs. input amplitude](image)

The data points where fitted to a second order polynomial using *polyfit* in Matlab.
To prove that the perfusion is a function of the square of the input amplitude, the raw perfusion was plotted versus the square of the input amplitude and fitted to a first order polynomial. The result is shown in Figure 21.

![Figure 21 Raw perfusion vs. squared input amplitude](image)

To estimate how well data points are fitted to the line the following error estimation can be used:

$$R = 1 - \frac{\sum_{i=1}^{n} (y_i - f(x_i))^2}{\sum_{i=1}^{n} (y_i - \text{Mean}(y_i))^2} = 0.9999$$

where R=1 corresponds to a perfect fit.

The test showed that amplitude is clearly a function of the square of the input amplitude.
5.4 Evaluation of window functions

This test was performed to compare two window functions of triangular and Hamming type. The basic approach was to compare the periodograms calculated on a test signal for: rectangular (no window), triangular and Hamming window applied. The test signal was a 5 kHz sine signal generated by a signal generator. This makes it easier to evaluate how the frequency resolution is affected by the window function.

The results of the test plotted in Figure 22 shows that the windowing has a clear effect on the spectral leakage. By looking closer around the tone it was concluded that the two window functions have practically the same effect. But the Hamming window decreases faster further away from the tone. This does not correspond to the theory presented in chapter 2.4.4.1. Further investigations need to be done to be able to tell why.

5.5 Signal quality test

To evaluate the signal quality of the perfusion signal the signal to noise ratio, SNR, was calculated using the DSP, the PF 5010 unit and Matlab. The three results were then compared. The most interesting comparison is the one between Matlab and the DSP. It’s harder to compare the SNR of the PF 5010 unit to the other methods because the methods used to calculate the perfusion is significantly different. The analog calculation components of the perfusion unit also add unknown offsets to the signal.
The basic approach to measure the signal quality was to calculate the ratio between the unscaled, unnormalized perfusion and the perfusion added by the noise, i.e.

\[ NF = \frac{\text{Perf}}{\text{Perf}_{\text{Noise}}} \], \text{ NF = Noise Factor } \]

where \( \text{Perf} \) and \( \text{Perf}_{\text{Noise}} \) are noise compensated perfusion and perfusion caused by noise, respectively (refer to chapter 2.2.6).

This method provides a simple way to see how much the noise is contributing to the whole raw perfusion signal. However, it doesn’t tell us so much about the actual SNR of the AC signal. This is because the perfusion is a sum of a frequency weighted PSD, which means that the signal quality will increase if the spectrum contains more high frequency components. This implies that the noise factor would increase if the high cut off frequency should be lowered. This hypothesis was also tested and verified. The upper cutoff frequency was changed from 20 to 16 kHz, which resulted in an increase in the noise factor from ca 1 to ca 1.8. This change was predicted by calculating the weighted frequency sum from 16 to 20 kHz for a recorded PSD. However the Matlab values did not correspond well to the DSP values so it was hard to draw any conclusions from the experiment.

5.6 Perfusion: DSP vs. PF 5010

This test was performed to see how well the perfusion calculated in the DSP corresponded to the perfusion calculated by the PF 5010 unit. At first, some basic measurements were made to see how well the perfusion curves corresponded to each other. Then, several heat provocation measurements were conducted and the percentile change of the perfusion before and after the heat provocation was compared for the two instruments. The PSD before and after each provocation was also calculated and stored using both Matlab and the DSP.

The PF 5010 can filter the perfusion signal with three different low pass filters which can be altered on the front panel of the instrument. Which filter to be used is determined by the level of detail required. During this experiment the “medium detail” filter was used. This filter is designed to show the frequency of the heart beats and suppress faster physical changes. The DSP signal is filtered digitally in Matlab. The Matlab filter was designed to reassemble the filter of the PF 5010 as much as possible. The Matlab filter used is a 2\(^{nd}\) order Butterworth filter with a cut off frequency at 2 Hz.
5.6.1 In vitro measurements

These basic measurements were made to compare the perfusion signals of the two methods quantitatively. Figure 23 and Figure 24 shows the perfusion of both instruments on the reference objects.

The gain was adjusted to show approximately 250 Pu for the two instruments. Both instruments calculates approximately zero in the zeroing disc which indicates that the noise compensation implemented in the DSP works accurately.

5.6.2 In vivo measurements

This chapter describes measurements made on actual persons to verify that the DSP implementation calculates the same perfusion as the PF 5010. At first, some basic measurements were made to roughly see how the results of the DSP implementation corresponded to the PF 5010 unit. Then, a heat provocation experiment was made to verify that the DSP implementation scales equally as the PF 5010.
5.6.2.1 Basic perfusion measurement

Figure 25 shows the results of a test measurement made on my palm.

The measurements showed that the software implementation is well correlated to the analog implementation. No attempts to statistically compare the two signals were made, while the two signals have been filtered through different filters. Such a comparison would make no sense. A more qualitative investigation is made in the next chapter.

5.6.2.2 Heat provocation measurements

A common method when making LDP measurements is to apply a heat provocation to the measuring area to get a fast increase in perfusion. The percentile change in perfusion before and after the provocation is used as an indication of how well the cardiovascular system responds to the provocation. In this case, it was used to compare the linearization of the two different measurement systems.

It is also interesting to study how the PSD is changed when a provocation is applied. By studying the PSD, the necessary bandwidth of the measurement can be observed, for example. For each measurement, the PSD was calculated for ca 5 seconds before and after the provocation using both Matlab and the DSP.

The procedure of these measurements was to measure the perfusion continuously for 15 minutes. After the first five minutes the measurement site is heated to ca 44 °C using a standard heat probe designed for use with PF 5010 and PF 5020. The heat is then kept at this level for the rest of the measurement. The mean perfusion was averaged over an interval where the perfusion had reached a steady level. The average times shown in Appendix C are denoted TB and TA in for averaging time before and after the provocation, respectively.
The plots of the tests are placed in Appendix C and the tables below lists the results.

<table>
<thead>
<tr>
<th></th>
<th>After/Before</th>
<th>Olia</th>
<th>After/Before</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>PF 5010</td>
<td>DSP</td>
<td>PF 5010</td>
</tr>
<tr>
<td>Arm</td>
<td>32.5035</td>
<td>31.3715</td>
<td>9.0342</td>
</tr>
<tr>
<td>Thumb</td>
<td>4.8843</td>
<td>4.667</td>
<td>1.5641</td>
</tr>
<tr>
<td>Foot</td>
<td>8.9454</td>
<td>8.5204</td>
<td>5.6924</td>
</tr>
<tr>
<td>Forehead</td>
<td>4.3506</td>
<td>3.8476</td>
<td></td>
</tr>
</tbody>
</table>

Table 2 Results from heat provocation measurements

<table>
<thead>
<tr>
<th></th>
<th>PF5010</th>
<th>DSP</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Before</td>
<td>After</td>
</tr>
<tr>
<td>Arm</td>
<td>5.4179</td>
<td>173.86</td>
</tr>
<tr>
<td>Thumb</td>
<td>42.09</td>
<td>183.09</td>
</tr>
<tr>
<td>Foot</td>
<td>32.215</td>
<td>177.41</td>
</tr>
<tr>
<td>Forehead</td>
<td>114.43</td>
<td>497.82</td>
</tr>
</tbody>
</table>

Table 3 Perfusion values from heat provocation measurements

In all measurements the absolute perfusion values were higher for the DSP compared to the PF 5010, except for the case of Fredrik’s forehead where the perfusion calculated by the DSP is higher before the provocation and lower after the provocation. This exception can possibly be explained by the different frequency properties of the two instruments. The DSP ignores all signal information above 16 kHz unlike the PF 5010 which also uses signal information above 16 kHz as seen in Figure 18. This difference might be a cause of error for perfusion values as high as in this particular case.
The percentile change of the DSP is plotted vs. the percentile change of the LDPM unit in Figure 26. This result was considered to be good. For better verification of the correlation of the two methods more measurements should have been done. But the tests are rather time consuming so seven measurements were considered to be sufficient at this time.

Figure 26 Percentile change: DSP vs. PF 5010
Chapter 6
Conclusion

This thesis work presents a DSP solution replacing an analog signal processing chain. The DSP algorithm has been verified in experiments using Matlab and the PF 5010 LDPM unit as a reference. The experiments made shows very promising results but also some drawbacks in the construction. This chapter presents the results of the work and discusses possible changes and the future outlook of the implementation.

6.1 Review

The result of this thesis work is a real-time implementation of a perfusion algorithm based on the 2048-point power spectral density, PSD, of the photocurrent, giving a frequency resolution of roughly 25 Hz. The demonstrated implementation shows improvements compared to the previous analog construction present in the PF 5010 LDPM unit in terms of linearity with respect to frequency. The implementation should also be less depending on temperature and component trimming made at production.

Measurements made have shown that the perfusion produced by the implementation is well correlated to the perfusion produced by the PF 5010 unit. In vitro measurements on the motility standard and a static scatterer have verified that the implementation shows the same absolute perfusion values as the PF 5010 unit used as reference. In vivo measurements have also verified that the implementation shows the same response to heat applied on the measurement site.

6.2 Recommendations

During the work some alternative solutions were thought about but never realized. Here are some of them presented.

The 2048-point FFT solution provides a good frequency resolution and makes it easy to set the low cut-off frequency of the algorithm, but it requires quite a lot of CPU speed. To be less depending on CPU speed, the low cut-off frequency can be set using a high pass filter implemented in the DSP or outside. That way, the design would not be so much depending on the number of FFT points. However it is doubtful if such as filter could be implemented given that the sampling time is limited and the required cut-off frequency is very low. An alternative is an external, high order, analog filter.

The present frequency resolution is limited to 25kHz/2048 since 2048 is the highest numbered FFT algorithm available in the software library provided by the DSP manufacturer. A higher frequency resolution is only required to be able to set a lower low cut-off frequency of the perfusion algorithm than in the present implementation. This is the only reason why a higher resolution is needed, which is why an alternative way to set the low cut-off frequency as discussed above is motivated.
The present implementation produces a perfusion value every 20.48 ms, but the calculation takes less than a fourth of that time when the CPU is running at 144 MHz. To speed up the algorithm even more, several things can be done. To start with, the code is written to produce a full spectrum, which really isn’t necessary but it is convenient to have the whole spectrum during evaluation. The code is written entirely in C. Several functions can be optimized further in C or, instead, be written in assembly. Assembly offers the possibility to use hardware implemented circular pointers – a function that can be useful while most buffers need to be circular.

6.3 Future leads

To make the digital laser Doppler implementation fully functional a completely automatic calibration routine needs to be implemented. Most of the work needed to realize such a function has already been made but the hardware needed is not constructed.

The hardware and software can still be further optimized in terms of signal gain and the resolution of the integer operations made in the DSP.

The fact that the PSD is calculated in real-time generates new possibilities to investigate the PSD during measurement. Current work by M. Larsson and T. Strömberg [11] shows how the PSD could be used to divide the perfusion signal into velocity components. This is an interesting new possibility. Movement artefact compensation suggested by Bogget et al. [10] is also possible with this DSP implementation.

The DSP solution makes the construction more flexible and makes it possible to present more data from the measurements. One possibility is to alter the cut-off frequencies while running the program. This can be done since the noise is white and the level of the noise is linearly depending on the DC level of the photocurrent. Of course, more experiments must be made to verify this property.
## Appendix A

### List of components

<table>
<thead>
<tr>
<th>Component</th>
<th>Name &amp; description</th>
<th>Manufacturer</th>
</tr>
</thead>
<tbody>
<tr>
<td>DSP</td>
<td>TMS320VC5509A, Fixed-Point DSP, 16-bit core</td>
<td>Texas Instruments</td>
</tr>
<tr>
<td>DSP Evaluation module</td>
<td>TMS320VC5509A Evaluation module</td>
<td>Spectrum Digital</td>
</tr>
<tr>
<td>Perfusion unit</td>
<td>PF 5010 LDPM unit</td>
<td>Perimed</td>
</tr>
<tr>
<td>Temperature unit</td>
<td>PF 5020 Temperature unit</td>
<td>Perimed</td>
</tr>
<tr>
<td>A\D converter</td>
<td>AD7922, 1 MSPS, SAR, Two channel, 4-wire SPI-interface</td>
<td>Analog Devices</td>
</tr>
<tr>
<td>Anti-aliasing filter</td>
<td>Max 291, 8\textsuperscript{th} order Butterworth switched capacitor filter</td>
<td>Maxim IC</td>
</tr>
<tr>
<td>Crystal oscillator</td>
<td>EXO-3, tuneable oscillator, 16 MHz base frequency</td>
<td>C-MAC</td>
</tr>
<tr>
<td>Operational amplifier x 2</td>
<td>TL072</td>
<td>Texas Instruments</td>
</tr>
<tr>
<td>Voltage regulator + 5V</td>
<td>L7805</td>
<td>ST Microelectronics</td>
</tr>
<tr>
<td>Voltage regulator -5V</td>
<td>L7905</td>
<td>ST Microelectronics</td>
</tr>
<tr>
<td>Voltage reference 3.3 V</td>
<td>REF196 Precision micropower, low dropout</td>
<td>Analog Devices</td>
</tr>
</tbody>
</table>
Appendix B

Signal adjustment board schematics

[Diagram showing signal adjustment board schematics with component labels and values.

R1 = 100 kΩ
R2 = 6.34 kΩ
R3 = 11 kΩ
R5 = 39.2 kΩ
R6 = 11 kΩ]
Appendix C

Heat provocation experiment

Perfusion
Fredrik: Arm

Foot

Forehead

Ola: Arm

Thumb(inside)
Finger

Note: TB and TA are averaging times before and after provocation, respectively.

Power spectral density
Fredrik: Arm

Foot

Forehead

Thumb (inside)
Ola: Arm

Periodogram before and after heat provocation: Arm, Ola

Finger

Periodogram before and after heat provocation: Finger, Ola

Note: All periodograms have been normalized and the periodogram of the dark noise has been subtracted from each periodogram. This way all periodograms can be compared in the same scale. The drawback of subtracting the dark noise spectrum is that several values become zero or lower in the regions where no change occurs. This is why the plots looks distorted in those regions.
Appendix D
Program flowcharts

Main program

Initiations of buffers, vectors, variables and CPU registers

Is input buffer ready?

Main: Perfusion calculation (next page)

McBSP1 receive interrupt

Current AD channel?

CH0: Put sample in AC input buffer

CH1: Put every fourth sample in DC input buffer

NFFT/2 new AC samples?

NO

YES

Set buffer ready flag

Return

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Perfusion calculation

- Copy AC and DC data from buffer
- Subtract DC offset from AC data
- Apply triangular window to AC data
- Calculate in-place fft
- Calculate and store periodogram
- Average two latest periodograms
- Calculate raw perfusion
- Calculate average of DC data
- Calculate perfusion offset from DC average
- Calculate noise compensated perfusion
- Scale perfusion
- Display averaged perfusion and DC level
- Clear data ready flag
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10. D. Bogget et al., Apparatus for measuring microvascular blood flow, US patent no. 6,173,197
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In English

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