SFH 7050 - Photoplethysmography Sensor

Application Note   draft version - subject to change without notice

1   Introduction

This application note describes the use of the SFH 7050 (see Fig. 1) as the sensor element for a photoplethysmography system.
The sensor is designed for reflective photoplethysmography (PPG) as advances in signal processing and high efficient LEDs enable small and powerful reflective photoplethysmography sensors (like the SFH 7050, see Fig. 1). This is especially important as reflected PPG signals can be measured on body areas where transmissive PPG can't be applied allowing wearable PPG sensors.

By analyzing the PPG signal various parameters can be derived. Among them is the heart rate. In addition the oxygenation saturation level of arterial blood can be determined by measuring the absorption at two different wavelengths. Oxygen saturation (SpO_2) is a vital parameter as it is the level of oxyhemoglobin (HbO_2) in arterial blood (usually SpO_2 expresses the percentage of saturation). In the human body, SpO_2 is defined as the ratio of HbO_2 concentration to the total haemoglobin concentration present in the arterial blood.

2   SFH 7050

The SFH 7050 is a fully integrated optoelectronic sensor, specially designed and optimized for reflective pulse oximetry. It features three LEDs – green (535 nm), red (660 nm) and IR (940 nm) - and a large area photodiode (PD) to maximize signal level (see Fig. 2 for LED spectra / PD responsivity). The device design includes a light barrier to minimize internal crosstalk thus enhancing the signal-to-noise ratio.

Fig. 1: *SFH 7050: Sensor with integrated LEDs and photodiode for heart rate and pulse oximetry applications.*

The sensor allows measurement of heart rate only (by powering only one LED), pulse oximetry by using the red and infrared LEDs or both.

2.1   Emitter Wavelength

To understand the impact of the LED specification on the overall device performance in a PPG application a few definitions are important (see also Fig. 3):

Peak wavelength (λ_{peak}) is the wavelength of the peak of the spectral density curve (in most applications it is of little significance).
Full-width at half-maximum (FWHM, $\Delta\lambda$): sometimes also called spectral bandwidth. It is the wavelength distance between the spectral points where the spectral density $S(\lambda)$ is 50 % of the peak value.
Center wavelength ($\lambda_{0.5m}$) is the wavelength halfway between the two spectral points with spectral density of 50 % of the peak value.
Centroid wavelength ($\lambda_{centroid}$) is the mean wavelength (see Eq. (1)). It divides the spectrum in two equal parts. It is the **most important definition** for non-visual systems
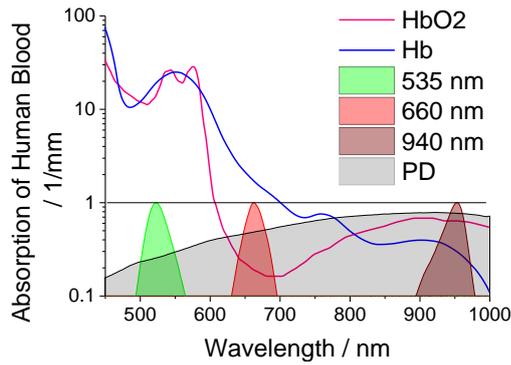
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Fig. 2: Absorption of human blood (oxyhemoglobin, HbO_2 and hemoglobin, Hb) vs. wavelength of light. Also included are the spectral responsivity of the photodiode (PD) and the normalized emission spectra of the LEDs (SFH 7050). For pulse oximetry usually 660 nm and 940 nm are used as they have opposite absorption levels (Hb vs. HbO_2). For pure heart rate monitoring 535 nm is a good choice (depending on the body location).
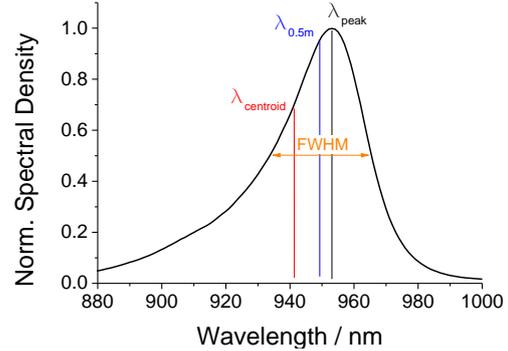
(like sensors) and relevant for this kind of PPG application.

$$\lambda_{centroid} = \frac{\int\limits_{\lambda_1}^{\lambda_2} \lambda \cdot S(\lambda) \, d\lambda}{\int\limits_{\lambda_1}^{\lambda_2} S(\lambda) \, d\lambda} \qquad \text{Eq. (1)}$$

Dominant wavelength (λ_{dom}) is a colorimetric quantity. It is an important description for visual illumination systems as it describes the human perception of the color of an LED[1].

For a symmetrical spectrum λ_{peak}, $\lambda_{0.5m}$ and $\lambda_{centroid}$ are identical. However, the high efficient LEDs inside the SFH 7050 feature slightly asymmetrical spectra. For the SFH 7050 λ_{peak} as well as $\lambda_{centroid}$ are defined (with $\lambda_{centroid}$ as being the important wavelength concerning the application, i.e. the blood absorption coefficients).

All LEDs inside the SFH 7050 have very tight wavelengths specifications and no

Fig. 3: Wavelength definition of an LED (example: 940 nm IR-LED). $\lambda_{peak} \neq \lambda_{centroid}$ (due to the asymmetric spectrum).

secondary peak, e.g. $\lambda_{centroid}$ is within ± 3 nm for the red wavelength (660 nm). This ensures reproducible signal readings as the slope of the blood absorption coefficient ($d\alpha/d\lambda$) is highest and subsequently wavelength stability is most critical at 660 nm (see also Fig. 2 and 4). Additionally, the LEDs feature low temperature dependent drift (0.13 nm/K) as well as narrow spectral bandwidth (typ. 18 nm for the 660 nm emitter). Fig. 5 and 6 presents the typical wavelength behaviour vs. ambient temperature / drive current.

Using short pulses also minimize any temperature dependent wavelength shift as well as spectral broadening due to internal heating of the LED (e.g. pulse width < 300 µs and repetition rate > 2 ms). This is especially important for SpO_2 calculation.

Fig. 4 presents the influence of the emitter wavelength shift vs. **SpO_2 measurement** accuracy. The wavelength stability of the red LED is most critical. In general, the emitting wavelength depends on the ambient temperature and the driving conditions (pulse peak current, pulse width and duty cycle), see Fig. 5 and 6.

For highly accurate measurements it is recommended to compensate the wavelength shift of the (red) LED. This can be done by e.g. monitoring the LEDs junction temperature ($\Delta\lambda = f(T_j)$).

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1) λ_{dom} definition makes only sense for LEDs within the visible spectrum.
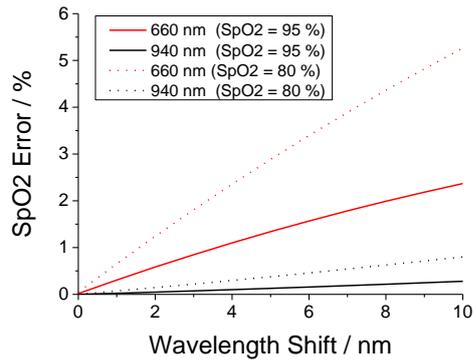
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Fig. 4: *SpO$_2$ error due to wavelength shift of emitter from its nominal value. The lower the SpO$_2$ level, the higher the overall error due to wavelength shift. In general the wavelength stability of the 940 nm IR-LED is uncritical compared to the 660 nm LED.*

Implementation can be realized via ambient temperature measurement. Either with a temperature sensor or e.g. via the junction voltage of an external Si-diode located close to the LED. This voltage is correlated to the ambient temperature, indicative for the wavelength shift of the LED during the particular operating conditions (calibration needs to be e.g. during final device testing to obtain room temperature reference). Another, more complex method, is via direct junction voltage measurement immediately after the LED pulse (e.g. biasing the LED with 1 µA and measure the forward voltage drop, similar to Si-diodes).

2.2 Detector

The photodiode features a low dark current, suitable for low noise applications. Additionally the photodiode is highly linear to enable accurate SpO$_2$ measurements. The photodiode current is typically amplified and converted into a voltage with an external transimpedance amplifier. The low capacitance and the fast response of the photodiode make it suitable for short pulse operation to minimize power consumption.
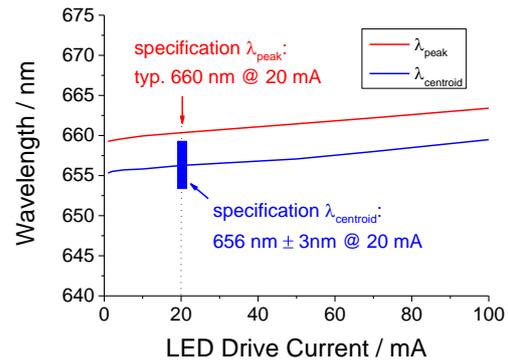

Fig. 5: *Drive current dependency of the 660 nm emitter wavelength and their datasheet definition (100 ms pulsed operation at T_a = 25 °C).*
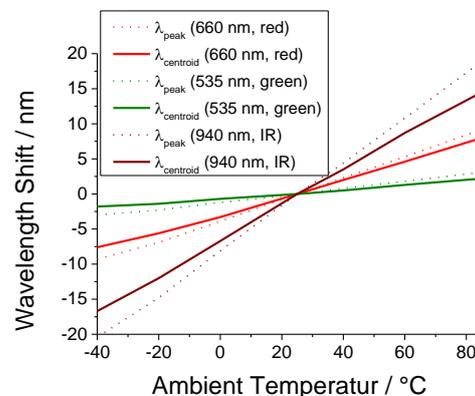

Fig. 6: *Temperature dependence of the emitter wavelength (relative to T_a = 25 °C, 20 ms pulse). The application relevant wavelength is $\lambda_{centroid}$. The temperature coefficients for the centroid wavelength are: $TK_{660\,nm}$ = 0.13 nm/K, $TK_{535\,nm}$ = 0.03 nm/K and $TK_{940\,nm}$ = 0.25 nm/K.*

2.3 Application Environment

The SFH 7050 is designed to operate close to human skin, any additional air-gap between human skin and the sensors surface can reduce the signal strength.

Additionally, the infrared LED can be used as a proximity sensor to indicate the presence of skin. This allows to start measurements when the sensor is close to the skin or to display an out of reach message. Operating the SFH 7050 with a cover glass might cause optical crosstalk.

Crosstalk needs to be reduced or avoided as it reduces the signal-to-noise ratio. For larger air-gaps a proper optical aperture design or light baffles between the LEDs and the sensor might be required. Crosstalk reduction is discussed in application notes for proximity sensors like OSRAMs SFH 7741, SFH 7776 or SFH 7771.

3   Operating the SFH 7050

There are (slightly) different measurement requirements concerning the application scenario:

- heart rate only
- heart rate plus pulse oximetry

In case of heart rate only designs the DC component of the photocurrent can be neglected; only the periodicity of the AC component (I_{max} – I_{min}, frequency) is of interest. For pulse oximetry the DC as well as the AC components (I_{min}, I_{max}) are needed. Thus in general, a pure heart rate device is easier to implement as it requires only one LED. For most body locations the green (535 nm) LED might be the preferred choice. However, there is the option to drive the red (660 nm) as well as the IR (940 nm) LED. The IR-LED might be of advantage as its light is invisible to the human eye. This can be a key criteria as in dark environments the green or red glow - if not shielded properly – might distract the user. In addition, the 940 nm IR-LED features the lowest forward voltage (slightly lower than the 660 nm LED).

3.1   General Considerations

The signal level (signal quality) is affected by the measurement system as well as by biological characteristics.
The complete system includes the SFH 7050 (the optical engine, sensor) with LEDs for illumination and a photodiode for signal detection. The photocurrent can be split into a DC component (no information if only the heart rate is of interest) and the AC component. In addition, ambient light might be present (considered as AC+DC noise). Especially IR light can penetrate deep into / through the skin, i.e. IR light from pulsed light sources (fluorescence or incandescent) and / or light from DC sources (like the sun) modulated by body movements can contribute to the optical signal as noise. The DC component of optical noise is usually subtracted due to an ambient light measurement immediately prior or after the LED light on measurement, resulting in an effective signal of:

$$I = I_{DC(signal)} + I_{AC(signal)} + I_{AC(noise)} \qquad \text{Eq. (2)}$$

Important operating parameters of the SFH 7050 influencing the signal quality are:

- LED current
- LED on-time
- LED repetition rate

Increasing the LED current results in the following:
- $AC_{(pk-pk)}$ (= I_{max} – I_{min}) increases
- DC (= I_{min}) increases
- AC/DC - ratio stays the same[2]
- $AC_{(pk-pk)}$ to ambient light ratio increases
- SNR increases ($AC_{(pk-pk)}$ to electronic noise ratio or dark current increases)
- energy consumption increases

Reducing the on time of the LED delivers:
- comparable AC/DC - ratio
- comparable AC to ambient light ratio
- comparable AC to dark current ratio
- SNR might decrease due to (potentially) more electronic noise from a shorter integration time
- higher bandwidth required for the transimpedance amplifier (TIA)
- larger system bandwidth makes it difficult to filter out (ambient) noise (in case ambient subtraction is performed in

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2) Using an even larger photodiode would not result in an improved AC/DC – ratio.

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the digital domain)
- energy consumption decreases

Increasing the repetition rate of the LED might result in a higher accuracy of the signal on the expense of higher energy consumption. In essence the sampling rate needs to be high enough not to miss the peaks / valleys of the pulsatile signal.

In order to increase the battery life of wearable / mobile devices the LED on time and LED current (depending on the noise level and the signal amplification) should be reduced as much as possible.

3.2 Biological Influence

The obtained signal level in general is strongly affected by the physical implementation (cover glass, air-gap) and by biological characteristics as such:

- measuring location
- skin tone
- applied pressure to the skin

In order to point out the design challenge some measurements were done with the SFH 7050 and operating the green (535 nm) LED with 8 mA. Additionally a 0.25 mm cover glass was right on top of the sensor. The photodiode signal was amplified and low-pass filtered:

Caucasian skin type (location: wrist):
- detected AC signal: 7 $nA_{(pk-pk)}$
- detected DC offset: 2000 nA
- AC/DC – ratio: 0.35%

African american skin type (location: wrist):

- detected AC: 2.2 $nA_{(pk-pk)}$
- detected DC: 1240 nA

- AC/DC – ratio: 0.18%

Caucasian skin type (location: finger tip):
- low skin pressure
- detected AC: 80 $nA_{(pk-pk)}$
- detected DC offset: 1680 nA
- AC/DC – ratio: 4.8 %

Caucasian skin type (location finger tip):
- high skin pressure[3]
- detected AC signal: 450 $nA_{(pk-pk)}$
- detected DC offset: 2360 nA
- AC/DC – ratio: 19.1 %

In essence, especially measurements at the wrist will result in low AC signal levels and low AC/DC ratios (e.g. 2.2 $nA_{(pk-pk)}$ at 8 mA LED current with AC/DC - ratios as low as 0.1 %). Using identical LED pulse currents *the **AC signal** level from the **red and infrared** are **comparable to the green** but the **DC level** for red and infrared can be a factor of 10 higher and subsequently the AC/DC – ratio a factor of 10 smaller.*

3.3 System Design

Resolving signals with such a low AC/DC – ratio is challenging. A simple consideration should highlight the general issue:

It is assumed that a signal with an AC/DC – ratio of 0.1 % is converted with a digital resolution of at least 1 % for the AC component. This leads to a required overall resolution (AC + DC) of 0.001 % or at least 16 bits. Effectively only 6 bits are used for the AC signal (assuming the full range of the ADC is used, what is the most optimistic scenario). In the following some other strategies for achieving that resolution are presented.

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3) If skin pressure is too high it can lead to so called venous congestion. This leads to artifacts due venous pulsation. As the venous blood has lower HbO_2 concentration (typ. 75 % saturation) compared to arterial HBO_2 concentration (SpO_2) the derived SpO_2 from the measurement might be impaired (i.e. lower than in reality).
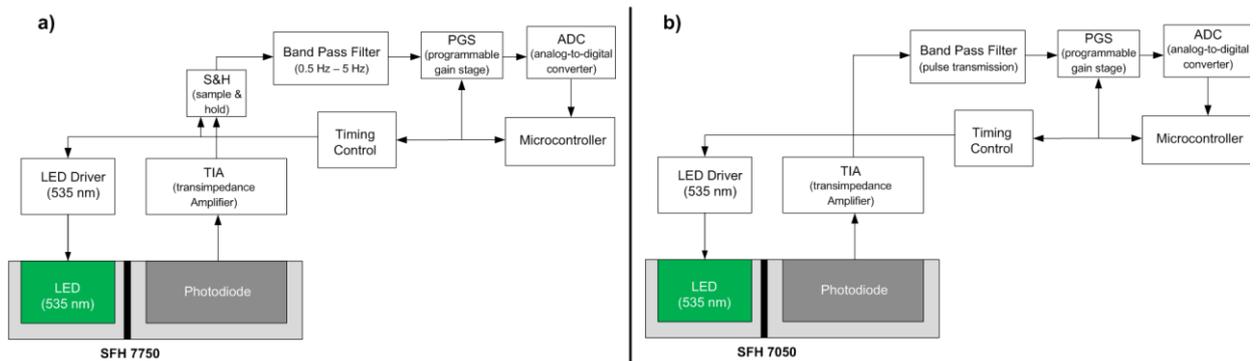
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Fig. 7: *Typical block diagrams of a heart rate only detection system. a) Synchronous demodulation principle: The ambient subtraction is done in the analog domain via the band-pass filter. b) Ambient subtraction is done in the digital domain (requires larger system bandwidth but allows an individual ambient measurement).*

The following discussion splits between systems which detect SpO$_2$ or heart rate only. Additionally the ambient light subtraction can be done in the analog or digital domain requiring different systems.

For systems without SpO$_2$ detection (**heart rate only**) only the AC-component is of interest. A block diagram of such a data acquisition system based on synchronous demodulation is presented in Fig. 7a). The gain stage following the TIA and a S&H (sample and hold) can be comprised of a band-pass filter to reduce the large DC content and amplify the signal to match the analog-to-digital converters (ADC) input range (eventually using a programmable / variable gain stage to use the full dynamic range of the ADC). It might be useful to employ an offset voltage for this gain stage to match the signal level with the input range of the ADC or to set the ADC reference voltages accordingly. For a typical system a pass-band between 0.5 Hz to 5 Hz allows effective removal of electrical noise and optical noise from ambient light in the analog domain. The advantage of using synchronous demodulation lies in the fact that slow and high resolution ADCs can be used. On the other hand it costs effort to implement a separate ambient light measurement (requiring one more S&H plus filter channel for the ambient light). Fig 7b)

shows a common setup where the ambient subtraction is performed in the digital domain. Here the pulses are amplified and converted on the expense of higher system bandwidth (e.g. high speed ADCs). Additionally there are advantages in detecting motion artifacts.

For a **pulse oximetry system** the DC level needs to be detected as well. There are different implementation schemes possible.

One strategy is to drive the two LEDs with different currents in order to achieve equal baseline DC – levels. A typical block diagram featuring synchronous demodulation for such a setup is illustrated in Fig. 8a). The DC-tracker (usually realized in the digital domain) measures the average DC value of sequential pulses and adjusts the (660 nm and 940 nm) LED currents to equalize the DC signals. This information can also be used to offset the second gain stage to reduce or eliminate the DC component in the signal. The influence of ambient light is suppressed by filtering in the analog domain. Ambient light can be subtracted as well by adding another S&H channel plus filter.

Fig. 8b) presents a block diagram where ambient subtraction is performed in the digital domain on the expense of higher system bandwidth to allow undistorted
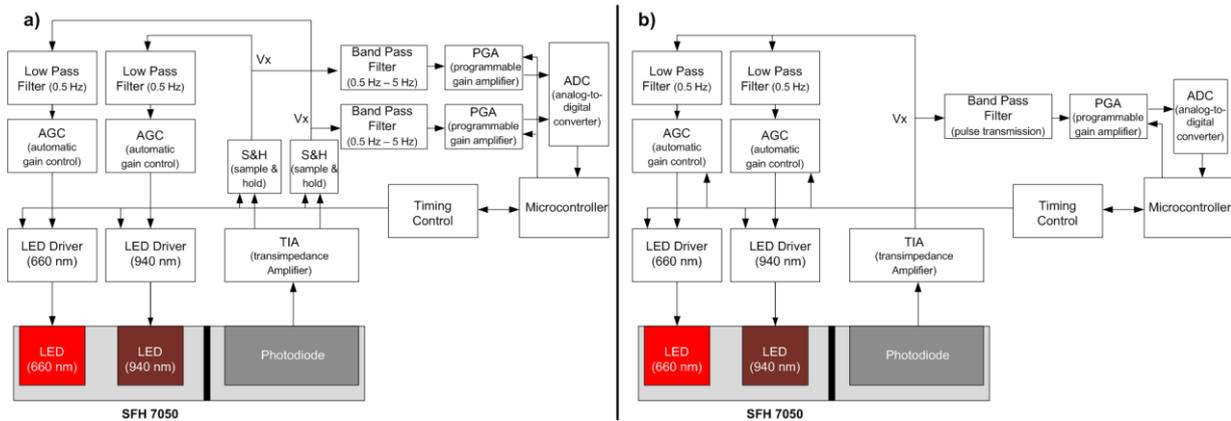
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Fig. 8: *Typical block diagrams of a pulse oximeter with equalization of the DC levels (660 nm / 940 nm after TIA). a) Synchronous demodulation: The ambient subtraction is done in the analog domain via the band-pass filter. b) Ambient subtraction is done in the digital domain (it features only one ADC, but requires larger system bandwidth and allows an individual ambient measurement).*
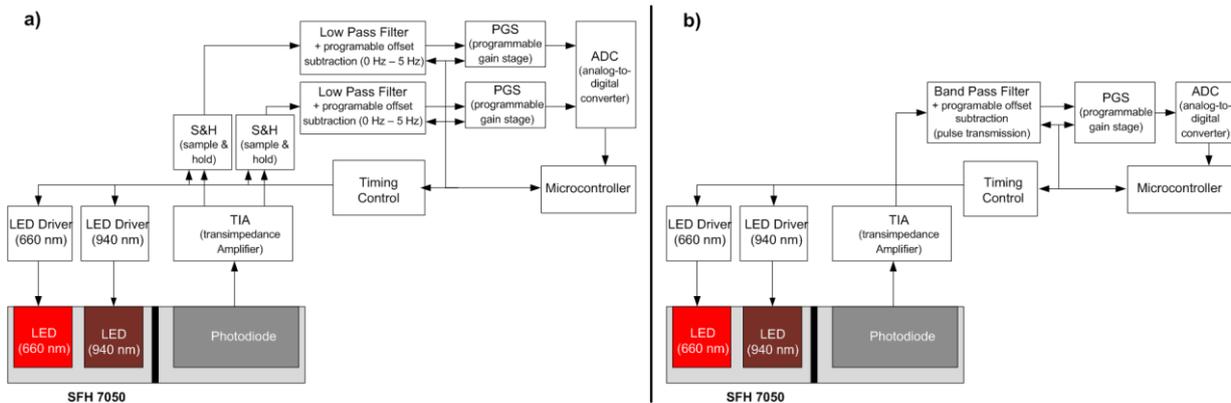

Fig. 9: *Typical block diagrams of a pulse oximeter measuring AC + DC signal level (660 nm / 940 nm). A programmable offset is subtracted and later on added again in the digital domain to get high resolution of the AC component (e.g. signal offset via a programmable DC tracker). a) Synchronous demodulation: The ambient subtraction is done in the analog domain via the band-pass filter. b) Ambient subtraction is done in the digital domain (it features only one ADC, but requires larger system bandwidth and allows an individual ambient measurement).*

transmission of the µs-long pulses. One advantage is the use of only one channel for ambient, red and infrared.

A second strategy (*non-pulsatile component elimination and addition*) simply subtracts a large DC component from the signal after the TIA. Fig 9a) presents a principle block diagram of such an approach (synchronous demodulation incl. an S&H stage). A programmable subtractor (offset amplifier controlled by e.g. the microcontroller)

removes a substantial part of the non-pulsatile signal. This allows the use of a low-pass filter for the tiny AC signal (I_{max}-I_{min}). The signal can then be amplified by a programmable gain amplifier to best match the input signal range of the ADC. Later on, in the digital domain, the non-pulsatile component is added back again to reconstruct the original signal containing the correct DC + AC signal levels. Again, the ambient light subtraction can be done in the
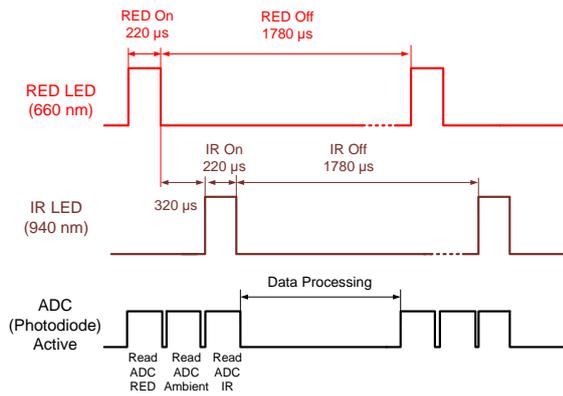
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Fig. 10: *Timing diagram with a sampling rate of 500 Hz and pulse width (integration time, measurement time) of 220 µs. Note the ambient measurement is used to subtract the ambient value from the red channel as well as from the IR channel.*

analog domain (Fig. 9a) or in the digital domain (Fig. 9b)).

For pulse oximeter applications the exact removal of the ambient light level (DC + AC) is even more important, making an ambient light subtraction mandatory.

3.4 Interfacing the LEDs

In a **pure heart rate monitoring system** it is sufficient to drive only one LED. In order to compensate for ambient light and to save power the LED can be operated in pulsed mode. During the LED off time the ambient light can be measured and subtracted from the signal obtained during LED on time. The LED pulse repetition rate and pulse width are a trade off between signal quality (AC) and overall power consumption. Usual systems sample with rates between 25 and 500 Hz per channel with a pulse width ranging from 500 µs down to 5 µs.

In a **pulse oximetry application** the red and infrared LEDs are driven alternately. Between the red and infrared light pulse an ambient light measurement can be performed as well. In terms of sampling characteristics the same consideration apply as for heart rate monitoring. However, a low

noise level is more critical here due to the different use of the signal (e.g. absolute signal levels). Fig. 10 shows a typical timing diagram.

As a **key characteristic** for pulse oximetry the **LED driver** circuit must generate minimal noise as any noise inside the signal bandwidth will degrade the overall performance (SNR, especially of the tiny AC signal, which may vary for wrist applications between 0.1 % and 5 % of the total signal level) in terms of accuracy and resolution. Therefore special LED driver solutions are recommended with low current ripple (in contrast to standard LED drivers used in solid state lighting or backlighting applications). Strong smoothing of the LED current during the pulse is advised.

Using the approach of driving the LEDs to achieve equal DC - signal level at the receiving side (e.g. 2 V) for the red and IR signal LED drivers with a wider dynamic range and high accuracy are required (usually driven via a high resolution digital signal from the DC-tracker circuit).

Other driving options use bursts of pulses instead of one single pulse. This allows better DC and AC ambient light suppression with a high-pass filter as the burst rate can be in the hundreds of kHz region with only µs long individual pulses.

Some systems drive the two LEDs in an antiparallel configuration.

3.5 Interfacing the Photodiode

There are numerous interface options to connect the photodiode of the SFH 7050 to an analog-to-digital converter (e.g. microcontroller) for further signal processing.

The most prominent is the use of a transimpedance amplifier (TIA) followed by a gain stage with filtering before analog-to-digital conversion.

Fig. 11 illustrates a typical TIA setup. The gain (i.e. feedback resistor value) of the TIA stage should be set as large as possible to
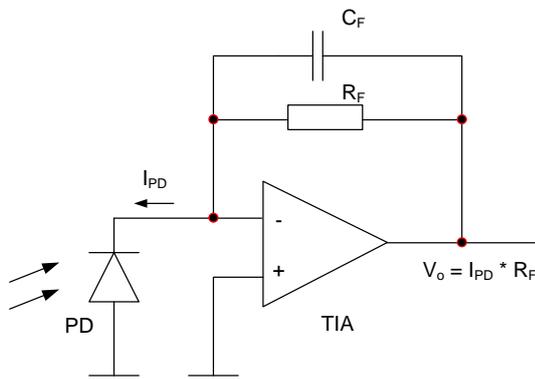
Fig. 11: *Transimpedance amplifier (TIA) setup for interfacing the photodiode of the SFH 7050.*

Fig. 12: *Capacitance of the photodiode vs. reverse bias.*
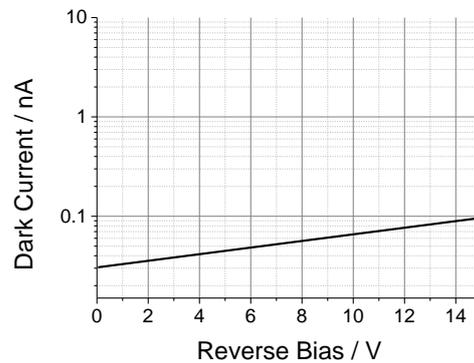
Fig. 13: *Typ. dark current vs. reverse bias (datasheet max. 10 nA at 10 V).*

optimize the SNR. On the other hand a high feedback resistor reduces the available bandwidth. The typ. bandwidth of a TIA is determined by the following equation:

$$f_{3dB} = \sqrt{\frac{GBP}{2\pi R_F (C_F + C_D + C_{TIA})}}$$

Eq. (3)

with *GBP* as the Gain-Bandwidth-Product of the TIA, R_F the feedback resistance, C_F the feedback capacitance, C_D the photodiode capacitance (15 pF at 0 V bias, see Fig. 12) and C_{TIA} the input capacitance of the TIA. The above Eq. is valid for µs / kHz application as the intrinsic speed of the photodiode is much faster.

The C_F capacitance is critical as it minimizes gain peaking and improves the overall circuit stability. In general it is recommended to refer to the TIA datasheet for recommendations

Note that the bandwidth of the TIA must ensure that the short pulses (typ. 5 µs to 50 µs) will get amplified without amplitude distortion.

Further it is advised to operate the photodiode of the SFH 7050 without reverse voltage bias to minimize any dark current related noise. Fig. 13 presents a typ. dark current graph (datasheet limit: max. 10 nA at 10 V at room temperature).
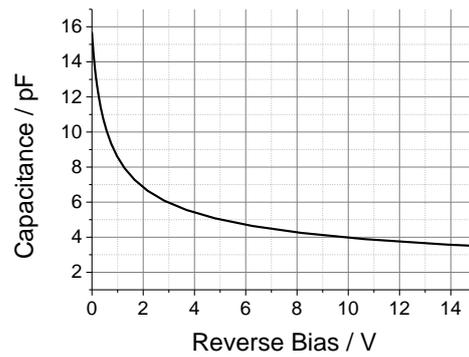
The **key specifications** for the **TIA** are extremely low input current, input current noise, and input voltage noise, as well as high-voltage operation. These characteristics are necessary to maximize the SNR so that the small currents of interest can be measured amid the large ambient currents from the reflected LED light. High-voltage operation means that a larger feedback resistor can be used to easily amplify the ambient and received LED originated current before removing the reflective DC and ambient light portion with a high-pass filter (after the sample & hold circuit). The remaining small signal of interest is then amplified to maximize the use of ADC's dynamic range. This gain stage should be programmable to compensate for changing environmental factors and the aging of optical components.

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The **key specifications** for the **ADC** are high resolution, SNR and short acquisition time. The acquisition time should be short enough to capture the modulated signal with the required resolution.

Other hardware realizations include e.g. differential current sensing TIA configuration and so called zeroing circuits to remove most of the ambient light contribution (DC component).

As a general design rule the SFH 7050 should be placed as close as possible to the TIA and the pcb tracks should be kept away from the LED supply lines to minimize any electrical crosstalk (noise). Additionally good electromagnetic shielding is recommended.

Finally **chipsets are available** which include the complete analog signal processing (incl. LED driver) as well as analog-to-digital conversion (e.g. from Texas Instruments or Analog Devices). The SFH 7050 can be directly connected to these chipsets to enable fast and easy evaluation and design.

4 Summary

The SFH 7050 is a component specially designed to allow heart rate resp. pulse oximetry measurements. The user can choose to employ one of the three available wavelengths for pure heart rate monitoring, depending on the body location where the sensor is applied. For pulse oximetry the 660 nm / 940 nm LED combination is used to extract the SpO_2 value out of the measurements. The tight wavelength specification as well as compact size (compared to traditional discrete realizations) make the SFH 7050 ideally suited for the next generation of integrated sensor systems.

Integration into a measurement system can be done by individually optimizing the drive and detector circuits. As an alternative the SFH 7050 can be directly interfaced to the off-the-shelf available heart rate / pulse oximeter solutions which already contain the TIA, gain stage, drive circuit as well as analog-to-digital conversion.

Further on, the application of the SFH 7050 is completely safe for humans as well as pets. The radiated light doesn't present any harm to the human skin / body (no ultraviolet light content) and the radiation is well below any critical level concerning eye safety regulations (at typical pulse currents below 1 A).

5 Literature

For further information concerning PPG and pulse oximetry the following reading is recommended:

[1] J. G. Webster, *"Design of Pulse Oximeters"*, Series in Medical Physics and Biomedical Engineering, Taylor & Francis, New York, USA, 1997.

Authors: Hubert Halbritter, Rolf Weber, Stefan Strüwing

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Appendix

LED Light for you
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