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Glossary
Cerebrovascular autoregulation Maintenance of cerebral blood flow during blood pressure changes to provide optimal brain perfusion.
Chromophore Light-absorbing molecule.
Encephalopathy Disease of various types affecting the structure or function of the brain.
Endarterectomy Surgical removal of blockage in the lining of an artery.
Ischemia Insufficient blood supply to tissue leading to oxygen and glucose shortage.
Hypoxia Deficiency of oxygen in tissue.
**Introduction**

Continuous monitoring of physiological functions in the intensive care unit is crucial for improving patient outcome. Standard techniques include measuring temperature, respiration, brain activity, nutrition, and cardiovascular dynamics. It is essential to maintain a sufficient supply of oxygen to tissue; both oxygenation and perfusion must be adequate for metabolic needs to be sustained. The failure of such supply can lead to tissue hypoxia, shock or organ failure, and possibly patient death. Some techniques used in the intensive care unit focus on monitoring central cardiovascular dynamics, such as electrocardiography (ECG) or echocardiography. ECG records the electric activity of the heart and echocardiography images cardiac anatomy and assesses flow dynamics. Other methods focus on the supply of blood to peripheral structures and their function. One of these is electroencephalography (EEG), which can noninvasively track the electric activity of the brain. Pulse oximetry and near-infrared spectroscopy (NIRS) are optical techniques that can inform on the oxygen status in the blood. Pulse oximetry is a well-established tool for monitoring arterial blood oxygenation, and NIRS is a promising tissue oximetry tool spreading among clinics. Both methods benefit from being noninvasive and inexpensive and providing real-time data. Another useful clinical tool for hemodynamic monitoring based on light is laser Doppler, which measures blood flow (Fig. 1).

**Light in the Clinic**

The use of visible and near-infrared light in medical care is desirable as it has several advantages: it is nonionizing, it can yield molecular information on the illuminated structures, and the broad range of available optical devices make light controllable. Light can be transported using optical fibers, light sources of various wavelengths are available, lasers produce coherent light, and the

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**Fig. 1** The use of light-based techniques (NIRS and pulse oximetry) among other monitoring tools in the neonatal intensive care unit in infants with hypoxic–ischemic encephalopathy. The transcutaneous monitor measures the partial pressure of oxygen and carbon dioxide. The cooling blanket and thermometer are used to cool the baby to treat encephalopathy.
intensity of light can be adjusted. Most optical methods are also noninvasive. Nevertheless, a major drawback of using light in imaging and therapy is the limited transparency of biological tissue in the visible range of wavelengths. This is caused by the strong attenuation of light by tissue. Two interactions contribute to this process: absorption and scattering. The level of attenuation depends on the properties of light (wavelength) and properties of the medium (scattering particle size, type of present chromophores, and structure of the medium) (Table 1).

Tissue chromophores
Absorption occurs when the energy of a photon matches some of the energy levels of a molecule of the medium, and if so, the photon is absorbed, and the energy is used for excitation of the molecule. Such absorbing molecules are called chromophores. Each chromophore has a different absorption spectrum; how much light is absorbed by a chromophore is wavelength-specific. The transmission of broadband light through a material with several distinct chromophores (such as tissue) results in an irregular decrease of light intensity across the wavelength range. Strong absorbers of visible light in tissue are melanin, lipids, water, and hemoglobin, and therefore, visible light cannot penetrate in tissue further than about 1 cm. Scattering of light occurs mainly on cells and their membranes. It is the main contribution to attenuation in most biological tissue; for example, in brain tissue, scattering is approximately 50 times more likely than absorption, and thus, tissue is known as a diffuse medium.

Fig. 2 shows the absorption spectra of hemoglobin and water. The specific extinction coefficient describes the absorption properties of the medium. Absorption of light by melanin and hemoglobin dominates in shorter wavelengths than water, which is the main absorber above 950 nm. Tissue is most transparent in the near-infrared (NIR) range as absorption by hemoglobin, melanin, and water is minimal. The range of 650–1350 nm is referred to as the optical window, and it has been shown that light of these wavelengths can travel as far as 8 cm deep in tissue. NIRS and pulse oximetry use light wavelengths in this range to achieve maximal depth of light penetration.

Hemoglobin absorption
Pulse oximetry and NIRS inform on the adequacy of respiratory function and tissue perfusion by tracking the oxygenation state of hemoglobin. The monitoring principle relies on the absorption properties of hemoglobin that depend on the oxygenation of the chromophore. The difference between the oxygenated (HbO₂) and deoxygenated (HHb) hemoglobin spectra is apparent in the optical window (Fig. 2); note the isosbestic point at 800 nm where the two spectra intersect. Attenuation of light due to absorption is directly proportional to chromophore concentration (see "Light Transport in Tissue" section, Eq. 9). Pulse oximetry and NIRS use wavelengths where hemoglobin is the main absorber and the two forms are optically distinct; measuring the changing attenuation of light emerging from tissue allows the quantification of concentration changes. The influence of other chromophores on the measurement can be considered negligible; lipids, melanin, and water concentrations do not change rapidly and can be assumed to be constant during the measurement period.

Chapter Objectives
The purpose of this article is to provide an overview of clinical monitoring instruments that use methods based on the absorption of NIR light in tissue. The focus is on pulse oximetry and NIRS; laser Doppler is also mentioned. The theoretical background of the principles of the methods is provided, including a brief description of typical instrumentation. The use of the techniques in clinical care is discussed, including the advantages and disadvantages of the methods. Promising state-of-the-art techniques, such as the measurement of metabolism, functional activation, and novel methods of monitoring blood flow, will also be introduced.

Light Transport in Tissue

In order to use light in the clinic, we must understand the physical principles behind the interactions of light and biological tissue. This article will introduce the basic theory of light transport in diffuse media.
Light traveling through a medium is attenuated through scattering and absorption. NIRS and pulse oximetry use models of light transport to quantify these changes of light intensity due to attenuation and link them to chromophore concentration changes.

**Beer–Lambert Law**

Assuming a light beam of intensity, $I_0$, traveling a short distance, $d$, through an attenuating medium, it will emerge with intensity $I$; attenuation is defined as Eq. (1):

$$ A = \log_{10} \left( \frac{I_0}{I} \right) $$

Scattering is the main attenuation process occurring in biological tissue. It is caused by inhomogeneities in the medium; light is scattered when refractive indexes mismatch, for example, tissue and a small particle within it. The scattering properties of a medium are described by the scattering coefficient, $\mu_s$. If a light beam travels through a purely scattering medium, the intensity loss is described by Eq. (2):

$$ I = I_0 e^{-\mu_s d} $$

where $I_0$ is the emitted intensity and $I$ is the intensity after traveling through a medium of thickness $d$. The scattering coefficient, $\mu_s$, does not account for the anisotropy of scattering. In the case of multiple scattering, the anisotropy factor, $g$, is used to describe the mean scattering angle; a high $g$ means mostly forward scattering. It is then possible to assume isotropic scattering and describe it using the transport scattering coefficient, $\mu'_s$ (Eq. 3):

$$ \mu'_s = \mu_s (1 - g) $$

In the case of absorption, a similar relationship to Eq. (2) is used to describe the attenuation due to the absorption of light with intensity $I_0$ traveling through a purely absorbing medium (Eq. 4). It is called the Lambert–Bouguer law:

$$ I = I_0 e^{-\mu_a d} $$

It is also possible to quantify absorption properties in base-10 logs, using the extinction coefficient, $k$ (Eq. 5):

$$ \mu_a = \ln(10) k $$

The Lambert–Bouguer law is expressed in terms of $k$ in Eq. (6):

$$ I = I_0 10^{-kd} $$

Absorption properties can also be described in terms of chromophore concentration with the specific extinction coefficient, $e$; $e$ is molar concentration:
\[ e = \frac{k}{c} = \frac{\mu_a}{\ln(10)c} \]  

(7)

Assuming the medium is nonscattering, combining the definition of attenuation Eq. (1) with Eq. (6) gives

\[ A = \log_{10} \left( \frac{I_0}{I} \right) = kd = \varepsilon cd \] 

(8)

This is called the Beer–Lambert law. It states that the change of intensity is directly proportional to the concentration of different chromophores in the medium. It can be applied to a medium containing several different chromophores; their contributions can be summed using Eq. (9):

\[ A = \log_{10} \left( \frac{I_0}{I} \right) = \sum_n \varepsilon_n c_n d \] 

(9)

where \( A \) is the absorbance of the medium, \( \varepsilon_n \) is the specific extinction coefficient of the \( n \)th chromophore that describes its absorption properties, and \( c_n \) is the \( n \)th chromophore concentration. Performing measurements of intensity changes and applying the Beer–Lambert law could be used to quantify chromophore concentrations \( c_n \). Nevertheless, it is applicable only to a purely absorbing medium; this does not hold for biological tissue.

Radiative Transfer Equation

A more general approach to studying the behavior of light in any medium is using the radiative transfer equation (Eq. 10). It describes the transfer of energy through a volume and is an expression of the conservation of energy:

\[
\frac{1}{c} \frac{\partial L(\vec{r}, \hat{s}, t)}{\partial t} = -\hat{s} \nabla L(\vec{r}, \hat{s}, t) - \mu_t L(\vec{r}, \hat{s}, t) + S(\vec{r}, \hat{s}, t) + \int_{4\pi} L(\vec{r}, \hat{s}', t)P(\hat{s}' \cdot \hat{s}) d\Omega'
\]

(10)

where \( L \) is radiance in position \( r \), direction \( \hat{s} \), and time \( t \), which describes the flow of radiation energy through a small area; \( P \) is the phase function describing the probability of light traveling with direction \( \hat{s}' \) being scattered into the solid angle \( d\Omega \) around \( \hat{s} \); \( \mu_t \) is the total attenuation coefficient (\( \mu_t = \mu_a + \mu_s \)); and \( S \) is the light source. The RTE consists of five main terms; the term on the left stands for the change in radiance that is caused by these four events:

- \(-\hat{s} \nabla L(\vec{r}, \hat{s}, t)\) stands for the net inflow of light entering or leaving the volume.
- \(-\mu_t L(\vec{r}, \hat{s}, t)\) accounts for intensity losses due to attenuation.
- \( S(\vec{r}, \hat{s}, t) \) is the presence of a source injecting light into the volume.
- \( \mu_s \int_{4\pi} L(\vec{r}, \hat{s}', t)P(\hat{s}' \cdot \hat{s}) d\Omega' \) is light scattered into direction \( \hat{s} \) from another direction \( \hat{s}' \).

Light Transport Summary

Pulse oximetry and NIRS are based on the dependence of light absorption and chromophore concentration described by the Beer–Lambert law. However, as its assumption of no scattering does not hold in biological tissue, both methods have used different approaches on how to include scattering. These will be described in further sections of this article. The RTE describes intensity losses of a light beam through the conservation of energy.

Pulse Oximetry

Pulse oximetry is the first established, noninvasive continuous method of blood oxygenation monitoring based on the measurement of light absorption in tissue. It was introduced to clinics at the end of 1980s and quickly spread worldwide. It measures the heart rate and oxygen saturation of arterial blood, which is used as an indicator of oxygen supply to tissue.

Oxygenation Measurement

Pulse oximetry is based on the difference of light absorption by oxygenated and deoxygenated hemoglobin and on the change of blood volume within vessels during the cardiac cycle. The display of a pulse oximeter gives an estimate of arterial oxygen saturation and heart rate.
**Arterial oxygen saturation**

Arterial oxygen saturation (SaO₂) is a measure of hemoglobin oxygenation in the arterial compartment of the circulatory system. It is not a measure of the total oxygen content in the arterial blood because a small fraction of oxygen (about 2%) is dissolved in the plasma. SaO₂ is defined as the ratio of the concentration of oxygenated hemoglobin \([HbO₂]\) and the concentration of total hemoglobin, \([HbT] = [HbO₂] + [HHb]\):

$$\text{SaO}_2 = \frac{[HbO₂]}{[HbT]} \times 100\%$$

(11)

The value of SaO₂ in health is the same throughout the whole arterial system. It is directly related to the oxygen supply to organs, and normal values lie between 95% and 100%.

The concept of measuring SaO₂ by quantifying the differences in the absorption properties of oxygenated and deoxygenated hemoglobin is hindered by the unknown influence of light scattering, the absorption by other chromophores, and the absorption by hemoglobin in the venous compartment. Pulse oximetry overcomes this by combining the measurement of intensity changes due to absorption (using the Beer–Lambert law, Eq. 9) with the change of the arterial blood volume during the cardiac cycle (arterial pulsation). As the Beer–Lambert law assumes no scattering, its influence is taken into account by instrument calibration.

The chromophores of interest in pulse oximetry are oxygenated and deoxygenated hemoglobin. The Beer–Lambert law enables linking the intensity loss to the oxygen content of the medium (biological tissue). As the concentrations of the two chromophores are unknown, measurements at two wavelengths give a pair of two equations. A typical choice is 660 or 690 nm and 940 nm as there is a big difference between the absorption properties of HHb and HbO₂ at these wavelengths.

**Arterial pulsation and pulse oximeter calibration**

The arterial blood volume in the tissue of interest changes during the cardiac cycle. It is maximal during systole, when blood is ejected from the left ventricle, and decreases during diastole, when blood fills the ventricles. As the absorption by other chromophores and the venous compartment can be assumed to stay constant, temporal absorbance changes can be attributed to the fluctuations in arterial blood volume (Fig. 3).

Measurements of these fluctuations at two wavelengths can isolate the contribution of arterial blood to the total attenuation. Pulse oximeters achieve this by separating the AC component of the signal, the pulsatile expansion, from the DC component that...

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**Fig. 3** (A) Change of blood volume in blood vessels during the cardiac cycle. AC is the pulsatile component of the signal, and DC is the constant component. (B) Fluctuations of the absorption of light by blood vessels during the cardiac cycle. The high peak occurs at systole and absorption reaches minimum during diastole. The small dip between systole and diastole is the dicrotic notch that occurs when the aortic valve closes and causes a short rise of aortic pressure.
stays constant (Fig. 3). Ratios of these components are obtained at two wavelengths; \( \text{SaO}_2 \) is derived from the ratio of the ratios, \( R \) (Eq. 12):

\[
R = \left( \frac{AC_1}{DC_1} \right) \div \left( \frac{AC_2}{DC_2} \right)
\]

The influence of scattering on the measurement makes a direct determination of \( \text{SaO}_2 \) from \( R \) impossible. The Beer–Lambert law cannot be used as human tissue is a scattering medium. Thus, every pulse oximeter needs to be calibrated. In this process, \( \text{SaO}_2 \) is measured simultaneously with \( R \) in healthy volunteers, and an empirical relationship between these values is obtained. As this calibration introduces errors by assuming little intersubject variability, the reading of a pulse oximeter is no longer denoted as \( \text{SaO}_2 \) but is referred to as \( \text{SpO}_2 \). The standard deviation of the differences between \( \text{SaO}_2 \) and \( \text{SpO}_2 \) is claimed to be 2% by most manufacturers. Another drawback of the calibration process is the limit of the physiologically achievable values of \( \text{SaO}_2 \) induced in the volunteers; lowering \( \text{SaO}_2 \) to values below \( \sim 75\% \) can be dangerous, and thus, most pulse oximeters stop being reliable at values below 80%.

**Instrumentation**

Pulse oximeters are small, often battery-operated devices, and are required to be light and compact. The two main hardware components of a probe are a light source and a detector. The light source must emit pulsed, high-intensity light of the selected wavelengths; light-emitting diodes (LEDs) are usually the light source of choice because they are small and cheap and can emit light at the required wavelengths. Standard solid-state photodetectors such as photodiodes are sufficient to detect the light that has traveled through the tissue as they are small, their sensitivity is sufficient, and their response is sufficiently linear. Pulse oximeters can be built to operate either on the transmission or on the reflection of light. In transmission pulse oximetry, the detector is placed opposite the light source with the investigated tissue in between (see Fig. 4A). They are the most commonly used probes as they can be very simple, clip-on devices for use on the finger or earlobe. In reflection pulse oximetry, the detector is placed next to the light source, and the measurement relies on the backscatter of the light (see Fig. 4B). They are usually placed on the forehead. Pulse oximetry can be also performed on the nose, cheeks, tongue, or neonatal foot.

**Clinical Use**

Pulse oximetry is a standard inpatient observation as the measurement of arterial saturation is a valuable indicator of a possible respiratory dysfunction through hypoxemia detection.

Pulse oximetry has a very broad application within a clinic; it is standard use in critical care units, emergency departments, and perioperative care. Correct interpretation of a pulse oximeter reading is a valuable tool in clinical decision-making; it can be used for the diagnosis of various conditions, such as acute lung injury, acute respiratory distress syndrome, or sleep-related breathing disorders.

Pulse oximetry is also a common tool in pediatric care. \( \text{SpO}_2 \) is often monitored immediately after birth as respiration and circulation establishment problems can arise in the change from intra- to extrauterine life, especially in preterm neonates. Pulse oximetry could soon also be used to detect congenital heart defects in neonates with better results than ultrasonography and clinical examination alone.

Even though pulse oximetry is popular in clinical care, there is limited systematically derived evidence supporting its clinical significance, and little data are available on whether \( \text{SpO}_2 \) readings influence the rate of complications such as mortality, intensive care admission, or length of stay. Even with optimal patient pulse oximeter interface and settings, a normal value of \( \text{SpO}_2 \) does not rule out any respiratory problems as it monitors only the arterial compartment of the blood system. Therefore, it is necessary to take observations acquired by other monitoring techniques into account before making a clinical decision (Fig. 5).

![Fig. 4 Pulse oximetry in transmission mode (A) and reflection mode (B).](image)
Future Development and Use of Pulse Oximetry

Even though pulse oximetry has been used for over 30 years now and has shown its practicality, its major disadvantages lie in the need for empirical calibration, which introduces an error of up to 4%, and the unclear clinical significance of the SpO₂ signal. Current research focuses on modifying the process of relating the signal to SaO₂ as it is desirable to find an alternative to the empirical calibration. Nevertheless, a decline of the use of pulse oximetry is not expected as there is currently no other well-established, simple, real-time, noninvasive, and cheap blood oxygenation monitoring technique to replace it. Additionally, as pulse oximeters are small and cheap devices, they can be found in households, and recently, even pulse oximetry smartphone applications have been introduced (which should not be used as a substitute for clinical devices).

Near-Infrared Spectroscopy

Near-infrared spectroscopy is a noninvasive regional oxygenation monitoring method based, similarly to pulse oximetry, on the different absorption properties of hemoglobin. The main advantage of NIRS over pulse oximetry is its ability to quantify temporal concentration changes of each monitored chromophore separately. Many NIRS instruments can also measure tissue oxygen saturation, often indicated as the tissue oxygenation index (TOI) or StO₂, which is an absolute indicator of oxygenation. NIRS was introduced in 1977 and was first used for cerebral monitoring. Even though this application is the most widely used, NIRS has also been used to monitor also the oxygenation of other organs, such as the muscle or liver.

NIRS Techniques in Oximetry

Three main techniques are used in research and clinical instruments to link the intensity change between the emitted and detected light to concentration changes. These are continuous-wave (CW), time-resolved (TR), and frequency-domain (FD) measurements. CW instruments emit light at a constant intensity and measure only the changes in intensity of the light passed through tissue (Fig. 6A). The method does not measure the scattering and absorption properties of the medium independently.
FD methods (Fig. 6B) are based on emitting modulated light and measure the intensity change and phase shift, which correspond to the time of flight. Combining this with light transport in tissue models allows measuring the attenuation properties of the medium separately.

TR measurements (Fig. 6C) apply ultrashort pulses and measure the time of flight directly. Computational models are then used to give attenuation properties of the tissue. TR is not only the most informative method but also the most complex one.

As the CW is the least complex and clinically most used technique, only CW algorithms will be further described.

**Continuous-Wave Technique**

As the CW measurement cannot directly measure the light transport properties of the tissue, it uses a modified version of the Beer-Lambert law to quantify chromophore concentration changes. It differs from Beer–Lambert law defined in Eq. (9) by no longer assuming a purely absorbing medium. The modified Beer–Lambert law (MBLL) (Eq. 13) includes two extra terms: G, an unknown wavelength-dependent term that represents light losses due to scattering, and B, the dimensionless differential pathlength factor that accounts for pathlength increases due to multiple scattering in diffuse media:

$$A = \log \left( \frac{l_0}{T} \right) = \sum_n a_n e_n dB + G$$  \hspace{1cm} (13)

The MBLL assumes homogenous tissue and no time variations in the scattering properties of the medium so that detected light intensity fluctuations can be attributed mainly to hemoglobin concentration changes.

**Differential Spectroscopy**

Measurements at two wavelengths in the NIR range are necessary for resolving temporal HHb and HbO2 concentration changes. Two wavelengths are required to solve Eq. (14) for two unknown concentrations:

$$A_{ij} = \log \left( \frac{l_{0,j}}{l_{T,j}} \right) = e_{HHb,j} c_{HHb,j} dB + e_{HbO2,j} c_{HbO2,j} dB + G_{ij}$$  \hspace{1cm} (14)

The measurement is conducted at two times, $t_1$ and $t_2$, which yields Eq. (15). The difference in absorbance is denoted as $\Delta A_{ij} = A_{ij}(t_1) - A_{ij}(t_2)$ and the change in concentrations as $\Delta[HHb]$ and $\Delta[HbO2]$:

$$\Delta A_{ij} = e_{HHb,j} c_{HHb,j} dB + e_{HbO2,j} c_{HbO2,j} dB + G_{ij}(t_1) - G_{ij}(t_2)$$  \hspace{1cm} (15)

Applying the MBLL for each wavelength and assuming no change in scattering losses over time ($G(t_1) = G(t_2)$) yields

$$\frac{\Delta A_{i1}}{\Delta A_{i2}} = \frac{\Delta[HHb]}{\Delta[HbO2]}$$  \hspace{1cm} (16)

Solving Eq. (16) for $\Delta[HHb]$ and $\Delta[HbO2]$ gives the change of each chromophore concentration in the time interval $t_2-t_1$.

**Spatially resolved spectroscopy**

The concentration changes calculated by differential spectroscopy give no information on the amount of oxygen in tissue as the changes are monitored from an arbitrary zero at the beginning of the measurement period. Spatially resolved spectroscopy (SRS) enables measuring the absolute oxygenation of tissue. It calculates the TOI, which is the ratio of the scaled oxygenated hemoglobin concentration and scaled total hemoglobin concentration:

$$\text{TOI} = \frac{k[HbO2]}{k[HB]} \times 100\%$$  \hspace{1cm} (17)

It utilizes at least two different source–detector distances to estimate the spatial gradient of absorbance. This is achieved by using several detectors/light sources to give multiple optical pathlengths (Fig. 7).

The theory of calculating TOI is not based on Beer–Lambert law but on the RTE (Eq. 10).

Solving the RTE (Eq. 10) is difficult, so SRS applies a simplification, the diffusion equation (Eq. 18). It assumes that scattering is much more probable than absorption and that the attenuation in the medium is space-invariant:

$$\frac{1}{c} \frac{\partial \Phi(r, t)}{\partial t} + \mu_s \Phi(r, t) - D \nabla^2 \Phi(r, t) = S(r, t)$$  \hspace{1cm} (18)

where $\Phi$ is fluence rate, $S$ is the light source, and $D$ is the diffusion coefficient that describes the attenuation properties of the medium. The solution to the diffusion equation leads to defining absorbance as
where $\rho$ is the distance light has traveled and $\mu_s^\prime$ is the reduced scattering coefficient (see "Light Transport in Tissue" section). This derivative is calculated experimentally by SRS with multiple attenuations measured at multiple distances (Fig. 7). $\mu_s^\prime$ is considered a constant $k$ that changes slightly for NIR wavelengths:

$$\mu_s^\prime(\lambda) = k(1 - h\lambda)$$  \hspace{1cm} (20)

where $h$ is the slope, $h = 6.3 \times 10^{-4}$ mm$^{-1}$/nm. Combining Eq. (19) and Eq. (20) gives

$$k \mu_s = \frac{1}{3(1 - h\lambda)} \left[ \ln(10) \left( \sqrt{3\mu_s \mu_s^\prime} + \frac{2}{\rho} \right) \right]^2$$  \hspace{1cm} (21)

It is thus possible to obtain the relative absorption coefficient $k \mu_s$ by measuring attenuation changes. The relative concentrations $k[HbO_2]$ and $k[HHb]$ can be calculated using Eq. (22) using at least two wavelengths that can then be used to express TOI (Eq. 17):

$$\begin{bmatrix} k[HbO_2] \\ k[HHb] \end{bmatrix} = \begin{bmatrix} e_{HbO_2} & e_{HbO_2} \\ e_{HHb} & e_{HHb} \end{bmatrix}^{-1} \begin{bmatrix} k \mu_s^{i_1} \\ k \mu_s^{i_2} \end{bmatrix}$$  \hspace{1cm} (22)

**Instrumentation**

NIRS brain oximetry instruments operate in reflectance mode as the adult human skull is too thick to be imaged in transmission mode (as seen for pulse oximetry in Fig. 4A). The main components of a basic NIRS system are a light source and detector. An NIRS system can have several source–detector pairs to cover multiple brain regions at once.

The light source and detector either can be small and be placed directly on the skin or can be connected to the skin using optical fibers. Not only are NIRS probes usually placed on the forehead for frontal lobe measurements (as this avoids optical attenuation due to hair), but also bigger probes for imaging of the whole head are available. The distance between the source and detector is related to the depth of light penetration. The light that reaches the detector travels an elliptical path (Fig. 7); increasing source–detector distances enables deeper light penetration but is also a source of bigger intensity losses; thus, it is key to find a compromise between sufficient detected intensity, preventing tissue damage from optical power, and sufficient depth penetration. For adults, commonly used source–detector distances are around 3 cm, which enable imaging up to a depth of 2 cm and thus provide information about the oxygenation of gray matter.

The light source of choice depends on the applied wavelengths. Some instruments use only a few (at least two are needed to resolve concentration changes in the two hemoglobin chromophores); others apply broadband sources with tens of wavelengths. The sources can be laser diodes, LEDs, or white light sources, such as halogen lamps. Another essential requirement for the light source is that the light cannot cause any thermal damage to the skin. It is especially important to keep this in mind for neonatal NIRS—measurements are often conducted on only a-few-hour-old prematurely born infants with very fragile skin. The emitted light is delivered to the tissue, travels through the skull, and is emitted by detector fibers. It is then detected by a CCD camera or photodiode (Fig. 8).
Commercial NIRS system

NIRS systems for brain oximetry in clinical care are available from several manufacturers. The following list provides a selection of those. All the mentioned are suited both for adult and pediatric/neonatal monitoring and are all, except one (OxiplexTS), based on CW measurements (Table 2).

Clinical Use of NIRS

Brain oximetry is being increasingly used in clinics as it is a low-cost, noninvasive, real-time, and relatively easy-to-use monitoring tool popular among hospital staff. The following section provides an overview of what the NIRS signal means and how it is used in clinical care (Fig. 9).

Physiology of the NIRS Signal

The fluctuations of oxygenated and deoxygenated hemoglobin concentrations carry information on the presence of oxygen in the monitored tissue. NIRS monitors the concentration in both the arterial and venous compartment of cerebral circulation, including signal originating in extracranial tissue, such as skin. The concentration changes of hemoglobin can be triggered by various physiological processes. These include changes in arterial pressure of oxygen, in blood pressure, or in the cerebral blood flow. As the brain circulation is a system with strong autoregulation, the effect of these changes can differ outside of the autoregulation limits.

TOI is clinically important as it contains information on the balance of cerebral oxygen delivery and oxygen metabolism. It can respond differently than hemoglobin concentration changes; for example, both TOI and [HbO₂] rise during arterial vasodilation and [HHb] decreases, while only TOI rises during venous passive contraction and both [HHb] and [HbO₂] decrease.

The main advantage of measuring TOI over hemoglobin concentration is that it is an absolute value. Monitoring [HHb] and [HbO₂] cannot inform on a pathology that has occurred before the beginning of the measurement unless it is followed by abnormal cerebral hemodynamic processes. Average values of TOI in adults are in a range from 65% to 85%, which is lower than

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**Table 2** Example of commercial NIRS systems

<table>
<thead>
<tr>
<th>Product</th>
<th>Manufacturer</th>
<th>Measures</th>
<th>Wavelengths (nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NIRO 200NX</td>
<td>Hamamatsu (Japan)</td>
<td>TOI, ΔHb²</td>
<td>735, 810, 850</td>
</tr>
<tr>
<td>FORE-SIGHT ELITE</td>
<td>Casmed (the United States)</td>
<td>TOI</td>
<td>690, 730, 770, 810</td>
</tr>
<tr>
<td>INVOS 5100C</td>
<td>Medtronic (the United States)</td>
<td>TOI</td>
<td>730, 810</td>
</tr>
<tr>
<td>OxiplexTS (frequency-domain system)</td>
<td>ISS (the United States)</td>
<td>TOI, ΔHb, absorption and reduced scattering coefficient, intensity, phase</td>
<td>690, 830</td>
</tr>
<tr>
<td>SenSmart X-100</td>
<td>Nonin (the United States)</td>
<td>TOI, SpO₂</td>
<td>730, 760, 805, 830</td>
</tr>
</tbody>
</table>

²ΔHb stands for hemoglobin concentration changes, including oxygenated, deoxygenated, total, and hemoglobin difference.
average SpO2 values (around 95%). This is due to the differences between TOI and SpO2: while SpO2 is based only on the concentration fluctuations in arterial blood, TOI includes absorption by the venous, arterial, and capillary components of cerebral circulation. It is important to note this as their volume and average oxygenation differ; arterial blood is 98% oxygen saturated, while venous blood is 70%. It is the cause of SpO2 having generally a higher value than TOI. TOI is sensitive to changes in the arterial:venous volume ratio, relative oxygenation of blood, oxygen consumption, and cerebral blood flow.

Clinical significance of the NIRS signal

As the NIRS signal carries information on cerebral oxygenation and metabolism, it has huge potential to be used in the monitoring of brain injury—traumatic brain injury (TBI), brain injury following birth, stroke, or hemorrhage. Monitoring with NIRS is also applied in orthopedic surgeries, such as shoulder repair or hip replacement, or general surgery—abdominal and bariatric surgery or sinus endoscopy. Intensive care units can apply NIRS during vasospasm or hyperemia.

Nevertheless, the complicated physiological background of the signal hinders NIRS from becoming a self-contained tool. Comparing NIRS measurements with results obtained by different, established hemodynamic monitoring techniques helps validate the physiological meaning of the signal. These standard methods include jugular bulb oxygen saturation, computed tomography, intracranial pressure measurement, or intracranial Doppler. Another difficulty of NIRS lies in a lack of standardization among devices from different suppliers, resulting in nonidentical algorithms to quantify the same observed variable leading to unequal values of the same measure. This makes setting intervention thresholds difficult.

A drawback of NIRS is the poor precision of the TOI signal; repeating a measurement about five times can achieve a precision comparable with pulse oximetry. Causes of this inaccuracy can be the placement of the probe (as the arterial, venous ratio of the illuminated tissue varies depending on the underlying anatomy) or that the assumptions of light transport models applied in the signal processing algorithms are not fulfilled in the tissue. These include assumptions regarding cerebrospinal fluid geometry, metabolic rate, or attenuation properties.

Even though validating the NIRS signal against other cerebral monitoring techniques can help with proving its reliability and significance, there is still the issue of the physiological background of the signal. As already mentioned, hemoglobin concentrations and TOI can respond differently to hemodynamic changes. Moreover, TOI is a quantity influenced by cerebral oxygen delivery and consumption; it is dependent on several processes occurring at once, some of which can cancel out contributions to the TOI signal or can be balanced out by another mechanism. For example, in the case of cerebral blood flow interruption during brain cell death, metabolism decreases, so the venous compartment could be filled with oxygenated blood, thus rendering the pathophysiology invisible to the TOI signal. It is crucial for doctors to bear this in mind when justifying any clinical decision based on the NIRS signal.

It is also unknown how stable the SRS algorithm (Eq. 21) assumptions are during pathological processes, such as hypothermia, hypovolemia, or sympathetically mediated vasoconstriction from pain, and how much they influence the extracranial component of the reading. The "normal values" and reference ranges for TOI have yet to be validated. Based on common cerebral blood flow assumptions, TOI for an adult should be approximately 66%. Values measured in healthy adults in several studies are usually higher in a range from 65% up to 85%. Currently, NIRS system providers have intervention thresholds recommended specifically for use with their products; for example, the INVOS system by Covidien has a range of 58%–82% for healthy TOI and recommends...
intervention at a 20% drop from baseline. The problem with setting injury thresholds in terms of NIRS signal is the variance caused by differences in age, cerebral metabolism, anesthetic and sedative regimen, body temperature, and other factors, which influence the response of the body to the injury and its severity.

**Brain Oximetry in Intensive Care**

The potential application of NIRS is in the critical care for monitoring brain injury. Cerebral hypoxia–ischemia is commonly the final pathophysiologic process of brain injury caused by various diseases, for example, trauma or cardiac surgery. Hypoxia–ischemia originates in a mismatch between cerebral metabolic demand and blood flow. These events can be quite common in the postoperative period; the knowledge of their frequency and severity is limited. They can occur in the operating room or in closely monitored postanesthesia care but are more frequent in the postoperative period; it is not clear whether the events of the postoperative period have a bigger influence on outcomes than events during operation. Maintaining real-time monitoring of the balance of cerebral hemodynamic processes is critical for managing secondary brain injury, yet it remains problematic. Standard techniques currently used to assess cerebral blood flow include jugular venous oximetry, transcranial Doppler, or EEG. However, jugular venous oximetry is invasive and global and thus can miss regional ischemia. Both transcranial Doppler and EEG require skilled and experienced users. NIRS is currently the only noninvasive method of cerebral oxygenation monitoring and has the potential to be the answer to the demand for a real-time bedside cerebral hemodynamic monitoring tool providing data from several brain regions simultaneously. Another advantage is its portability and relatively low cost. It is becoming a valuable tool in the intensive care community, yet more clinical evidence is needed for it to become a routine technique in critical care.

**NIRS as a predictor of outcome**

Another valuable use of the NIRS signal is its possible association with global outcomes that has been shown in many studies and is one of the main aims of current research. Arterial desaturations occurring in one region (also interventions to correct the desaturation) can relate to global outcomes; the frontal cortex and muscle tissue have shown their response to processes arising in remote regions. The ability to autoregulate determines the response of brain tissue and the muscle to these mechanisms. Autoregulation of blood differs in both types of tissue; it is greater in the brain than in the muscle during hypotension. Thus, muscle hypoxia/desaturation could be a warning sign of impending impaired oxygen delivery. On the contrary, as the cerebral cortex has a high autoregulation capacity, decreased global brain oxygenation could be a sign of a systemic problem with low cardiac output or low vascular resistance. This response is an indicator of changes already occurring in remote tissue.

**Traumatic brain injury**

Research in monitoring TBI with NIRS has shown promising results. TBI includes several pathophysiological changes occurring following trauma. These can be intracranial bleeding, inadequate cerebral oxygenation, cerebral blood flow changes, and problems with cerebral metabolism and cerebrovascular autoregulation. As these processes can lead to irreversible changes leading to permanent disability or death, prompt diagnosis and monitoring are vital for successful TBI management. NIRS could replace the use of measuring brain-tissue oxygen tension and jugular venous oxygen saturation, which are invasive but standard in TBI management. Even though the potential for NIRS use in neurology is great, the lack of clinical evidence hinders its acceptance. Another delaying factor is the problem of defining a hypoxia/ischemia threshold and the existence of a NIRS gold standard. The influence of the presence of posttraumatic changes in brain tissue, such as intracranial pooling of extravascular blood, subdural air, or extracranial contusion on the NIRS signal, must be evaluated to improve the reliability of the NIRS signal. Currently, more clinical proof is needed for standardization of NIRS use for managing brain injury.

**Cardiac surgery**

The application of NIRS in cardiac surgery is of great interest as poor neurological outcome is a major concern. 1 to 3% of patients undergoing cardiopulmonary bypass suffer from stroke, and more than 50% of patients develop long-standing postoperative cognitive dysfunction. NIRS is used as a guiding tool during cardiac surgery as the likely cause of these issues includes emboli and cerebral hypoperfusion. NIRS is also used for monitoring during valve replacement or transplants. Several studies have been conducted supporting the positive impact of NIRS during cardiac surgery; they have shown correlation between cerebral desaturation and adverse outcome after cardiopulmonary bypass and outcome improvement, including a reduction in stroke occurrence. TOI can also be an alternative marker for poor outcome and it is likely that the duration and degree of cerebral desaturation determine the severity of ischemia.

**Carotid endarterectomy**

Carotid endarterectomy, the unblocking of a carotid artery, is another surgical procedure possibly leading to insufficient cerebral oxygen supply. It can occur during a period of carotid occlusion. The risk of stroke during carotid endarterectomy is about 2% and is caused by emboli or cross-clamp-related ischemia. Cerebral perfusion can be maintained by inducing hypertension or intracarotid shunt placement, yet these methods can also cause complications in oxygen supply. Therefore, a shunt is placed only when ischemia is imminent during the cross clamping period. The simplest tool of monitoring this is the alteration of mentation, however, that is possible only under regional anesthesia. In the case of general anesthesia, cerebral blood supply can be measured using EEG or transcranial Doppler. Nevertheless, NIRS has the advantage of simplicity and has been used as the main monitoring tool during carotid endarterectomy for more than 15 years.
**Pediatric and neonatal intensive care**

Babies are the most suitable group for NIRS monitoring as their skulls are thinner and there is less of problem with hair; thus, NIR light can penetrate deep into gray and white matter. Another advantage is the compactness of the probes, which can be placed securely on infant’s heads without causing any damage to the delicate skin. Bedside NIRS monitoring can provide vital real-time information on cerebral oxygenation of infants without disturbing routine clinical care. Several causes can lead to issues with cerebral oxygen supply in the neonatal brain; examples are birth injuries and congenital malformations causing cardiopulmonary compromise. The applications of NIRS in the pediatric and neonatal intensive care units include monitoring during congenital heart surgery, bowel ischemia, inhaled nitric oxide therapy, therapeutic hypothermia, hypoxic–ischemic encephalopathy, sepsis, and transfusion. It has been demonstrated that monitoring cerebral blood flow and blood pressure can identify neonates with impaired cerebral autoregulation and perfusion. That can help preventing the development of intracranial hemorrhage and poor outcomes. NIRS can also aid in the detection of intracranial hematomas or changes in intracranial pressure.

NIRS has proved to be useful in monitoring prematurely born infants. They are in great risk of cerebral hypoxia and hyperoxia during the hemodynamic instability caused by the transitions of their respiratory and circulation systems. Ongoing NIRS monitoring from the first hours of life can detect of potentially harmful oxygenation changes that can be treated and reduce the burden of hypo- or hyperoxia.

A disadvantage of NIRS in neonatal and pediatric care is the lack of standardized TOI values; establishing them is difficult as the range of TOI values among children is big and changes with age, for neonates even in the range of hours (Fig. 10).

**Noncerebral NIRS Oximetry**

Sometimes, it is useful to monitor the oxygenation of other tissue than the brain. Liver oxygenation can give useful information on gastrointestinal circulation, as 25% of the liver blood supply comes from the hepatic artery, which supplies next to the liver also the duodenum, stomach, and pancreas. It is useful for neonatal monitoring as the liver is directly beneath the skin. NIRS can also be used for muscle oxygenation measurements. It either can be a predictor of global outcome, as previously described, or can inform on some occurring pathology in the muscle tissue, which can be caused by, for example, sepsis or cardiovascular diseases. Another use is in monitoring the muscle oxygen metabolism during exercise.

**NIRS Conclusion**

It is evident that NIRS is a source of clinically significant information about the hemodynamic state of gray matter of both adults and children. Its ability of real-time noninvasive assessment of microcirculatory oxygenation changes applicable in various clinical scenarios is in favor of it spreading among clinics and its growing significance as a valuable indicator of patient’s well-being. It is certain that the future will bring more possible applications of NIRS monitoring in the clinical care, such as anesthesia monitoring and weaning patients from a ventilator in intensive care or in peripheral arterial disease patients.

Nevertheless, one should be still aware of the scarcity of clinical evidence supporting the positive impact of NIRS monitoring before, during, and after surgery. Currently, the NIRS signal can be taken as a guiding tool, a trend monitor, but not for setting injury thresholds. There are still many obstacles preventing it from becoming a sufficient measure of oxygen supply, and NIRS is still an area of ongoing research. It focuses on several issues that range from the algorithms used for processing data to hardware issues, achieving a better precision and explaining the physiological background of the signal. It is desirable to evaluate the stability of the TOI signal during pathological cerebral events and standardize the commercially manufactured NIRS systems to achieve an agreement between the readings obtained by different machines. Light transports in tissue models are constantly being improved in order to realistically predict and evaluate light behavior. Additionally, the problem of the uncertainty of the physiological cause of TOI change must be solved. One of the possible solutions could be finding another marker of cerebral hemodynamics independent of TOI and drawing conclusions based on both. Another ongoing problem of NIRS is the attenuation of light from human hair.

![Fig. 10 Monitoring a neonate with hypoxic–ischemic encephalopathy in the neonatal intensive care unit. (A) NIRS probe placed on the forehead. The instrument is a broadband NIRS UCL in-house developed research instrument with four detector optodes and one light optodes (all optical fibers). (B) The NIRS system among other monitoring devices.](image)
This is not an issue with neonates, so the whole head can be covered with optodes; however, it is an issue in adults as it is difficult to achieve optimal optode–skin contact. Hardware development also focuses on creating portable headgear; some manufacturers produce wireless instruments that give NIRS the option to be used even outside the hospital unit.

**Laser Doppler**

Laser Doppler flowmetry is a blood flow measurement method based on the Doppler shift of NIR light scattered by moving red blood cells. It is similar to Doppler ultrasound, which uses the Doppler frequency shift between the sent and detected wavelengths to quantify blood flow speed. However, in contrast to laser Doppler, the resolution of Doppler ultrasound is insufficient for microcirculation monitoring. Laser Doppler provides continuous, noninvasive, real-time, and quantitative information on microvascular perfusion. The wavelengths used are usually 630, 780, or 830 nm selected to target hemoglobin.

The frequency shift of the scattered photons depends on the speed of the scattering object, the red blood cell. This shift is given by the Doppler equation: beam of frequency \( f_1 \) incident at an angle \( \phi \) on a scattering particle moving at speed \( v \) will be scattered with a frequency \( f_2 \) as given by Eq. (23):

\[
f_1 - f_2 = \frac{2
\nu \sin \varphi}{\lambda}
\]

The Doppler shift is very small, around 0–20 kHz for a 780 nm measurement, whereas the frequency of the impeding light is about \( 10^{14} \) Hz. Detectors are not sensitive enough to detect such small shift; thus, another method is used to extract the signal: the shifted signal interferes with the nonshifted light on the photodetectors and causes fluctuations of the detector current, which are measurable.

Laser Doppler can be used either for monitoring or for imaging. A laser Doppler perfusion monitor uses an optical fiber or optode to deliver laser light into tissue and another optode to detect the light. It measures real-time perfusion in a small sampling volume (around 1 mm\(^3\)) whose position depends on the wavelength and fiber separation (for 780 nm a depth of about 0.5–1 cm is achieved at a separation of 0.25 cm). Microcirculation is temporarily and spatially heterogeneous, and a laser Doppler perfusion monitor cannot assess a bigger volume in space and time simultaneously. This leads to the use of a laser Doppler perfusion imager, which utilizes a multichannel array to monitor the perfusion in a bigger area and gives a perfusion map with average perfusion in a heterogeneous medium in a single measurement.

The most common use of laser Doppler in the clinic is the assessment of adequate skin circulation. This is important during skin transfer during transplantation, wound healing in burn victims or the Raynaud phenomenon, a problem with the thermoregulation of extremities, and other connective tissue diseases. It can also be used for monitoring cerebral microcirculation and perfusion.

**Future Clinical Use of Near-Infrared Light**

Monitoring oxygenation, hemoglobin concentrations and blood flow are currently the only applications of NIR light for monitoring in the clinic. However, recent research shows other promising use of NIR light that can be used for assessing metabolism, blood flow, and spatial imaging of hemodynamics.

**Diffuse Optical Tomography**

NIRS can be used to obtain spatial information about the illuminated tissue. This technique is called diffuse optical tomography (DOT). It utilizes the penetration of light through brain tissue and detection of the emerging light to create 3-D quantitative imaging of optical properties, including functional and anatomical information. In DOT, multiple light sources and detectors are placed over the whole surface of the head and image the whole volume of the brain, not only surface areas. It relies on inverse modeling of light transport through the tissue.

DOT has been under development for over 30 years and is still not ready for clinical applications. The main difficulty of it is the strong scattering of light and the use of diffuse photons in image reconstruction. However, the research is driven by the huge potential of this imaging technique; DOT is often applied in brain studies including functional activation, and the principle can be also used in breast cancer; joint, muscle, and thyroid imaging; and neonatal research. In general, the advantage of DOT lies in the ability to detect lesions without exogenous probes and contrast agents.

**Blood Flow Measurements**

Diffuse correlation spectroscopy (DCS) is a novel method of measuring cerebral perfusion. Brain tissue is illuminated with an NIR laser light that is scattered on red blood cells while propagating through the head. The movement of the red blood cells causes temporal fluctuations of the intensity of the scattered light. This can be measured by quantifying the intensity changes of the emerging light and gives information on cortical cerebral blood flow. Research has so far shown very promising results both in
neuroscience and clinical application, especially in neonates as their skulls can be easily penetrated by light. Combining DCS measurements with NIRS yields information on the cerebral metabolic rate of oxygen and could hence be used for a complete assessment of cerebral health and physiology. It is expected that DCS will become a valuable tool in clinics in the near future.

Another way of assessing blood flow is by monitoring a different chromophore than hemoglobin using NIRS; indocyanine green dye. It is injected into the circulatory system, and it is possible to quantify blood flow based on the rate of accumulation of the dye in the tissue of interest over measurement time. It is more common to use it in monitoring blood flow in muscle, but recent research has also applied this method in neonatal brain studies.

**Mitochondrial Metabolism Monitoring**

Cytochrome c oxidase (CCO) is another chromophore that can be monitored with NIRS. CCO is the terminal electron acceptor of the respiratory chain and is essential for the generation of adenosine triphosphate. The absorption spectrum of the redox state of CCO has a broad peak in the NIR region and thus could potentially be monitored similarly to hemoglobin. The contribution of measuring redox CCO is that it is an indicator of oxygen metabolism—hemoglobin can only inform on the presence of oxygen, whereas CCO can prove that oxygen is being utilized in the mitochondria. Additionally, the concentration of CCO in the brain is much higher than in extracerebral tissue; thus, a more brain-specific signal, without scalp and skull contamination, can be obtained. The research is driven by the huge potential of CCO—it could provide a cellular indicator of cerebral oxygenation and could inform on physiological processes that affect metabolism earlier than hemodynamics, for example, in neonatal brain injury. It could also help in interpreting the TOI signal. One of the difficulties of monitoring the redox CCO signal is that the concentration of CCO is 10 times lower than hemoglobin. This also requires the use of several wavelengths (this is called broadband NIRS) to diminish noise and cross talk with the much stronger hemoglobin signal. The delay of clinical validation of the CCO signal is also given by the nonavailability of a golden standard to which the signal could be compared. This is not an issue for hemoglobin as it is not difficult to isolate the molecule and acquire in vitro spectra.

**Functional imaging**

Functional NIRS (fNIRS) is currently a useful cognitive neuroscience tool tracking hemodynamic changes. NIRS is sensitive to cerebral hemodynamic fluctuations following functional activation. The origin of this NIRS signal is neurovascular coupling. During the activation of a specific brain region, CBF increases temporarily in that location. This rise lasts over the period of neuronal activity and decreases at the end of the stimulus. The result can be measured as a local increase in \( \text{HbO}_2 \) concentration and decrease in \( \text{HHb} \). With many NIRS channels, it is possible to produce a real-time map of cortical hemodynamic responses. Other modalities are available for measuring functional activation: functional MRI, EEG, positron emission tomography, and others. Each of these methods has its strengths and limitations, and the selection of the imaging modality depends on the research objective. The popularity of fNIRS in research is growing, and it has been applied in research in social sciences, neuroimaging, and medicine. fNIRS has the advantage of being noninvasive and real-time, has quite high temporal resolution (up to 100 Hz, usually 1–10 Hz), is low cost, can be portable, and has very high experimental flexibility. Further, it is particular useful in neonatal and infant experiments as it is relatively robust to movement and allows a natural environment for study. It can be integrated with other neuronal activation imaging tools. The disadvantages include the following: it has the inability to inform about brain structure for anatomical reference, it cannot observe changes in deep brain regions, it relies on good skin-detector contact, and the spatial resolution is quite low (about 1 cm). Similarly to NIRS, fNIRS lacks standards for instrumentation, signal processing, and data analysis (Fig. 11).

![fNIRS study with a baby conducted at Baby and Child Research Center Nijmegen. The used instrument is a UCL frequency-multiplexed near-infrared topography system with a custom-built headgear.](image-url)
Summary

The relative transparency of biological tissue in the NIR region enables using light for assessing hemodynamics. The most common clinical application is pulse oximetry and NIRS, which are based on the differences of light absorption by oxygenated and deoxygenated hemoglobin. Pulse oximetry measured oxygenation of arterial blood; NIRS is usually focused on monitoring cerebral microcirculation and is influenced by both the arterial and venous compartment of the circulatory system. Even though both methods have shown their benefit in clinical care and are broadly used, they cannot be used as stand-alone indicators of sufficient oxygen supply. NIR light can also be used for assessing blood flow; in laser Doppler, which measures Doppler shift by scattering on red blood cells, it can track diffusion patterns in diffusion correlation spectroscopy or measure the rate of tracer accumulation. Imaging using NIRS could be soon be used for assessing cerebral health or detecting breast cancer. Current NIRS research is focused on studying the potential use of oximetry in other clinical situations, on making the methods more reliable and improving the clinical significance of the signal.

Further Reading


