CEREBRAL INFRARED OXIMETRY IN INTRACRANIAL HEMORRHAGE

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There have been presented up-to-date data on cerebral infrared oximetry application in intracranial hemorrhage. The principles of the technique enabling to perform noninvasive monitoring of cerebral tissue oxygenation have been given. There has been shown the comparability of cerebral oximetry and invasive assessment techniques of cerebral tissue saturation, jugular oximetry and cerebral microcirculation. Some systems for oxygen status determination have been presented. Special attention has been paid to the use of coefficients and indices of cerebral infrared oximetry to assess functional state of cerebral microvasculature and cerebral autoregulation.

There have been described prospects for further development of cerebral oximetry as a part of many-component monitoring in craniocerebral injury and hemorrhagic strokes.

Key words: cerebral infrared oximetry; craniocerebral injury; hemorrhagic stroke.

Infrared oximetry (synonym — infrared spectrometry) is associated with F. Jobsis, who was the first to apply it *in vivo* in 1977. He showed radiation intensity changes to correlate with the concentration of natural chromophores: oxyhemoglobin, deoxyhemoglobin, cytochrome oxydase, melanin, etc [1, 2].

At first, infrared oximetry was not quantitative and showed the tendency for oxygenation change only, and recorded signals had significant fluctuations and tendency to artifacts [3, 4].

Further development of the technique led to tissue oximetry and cerebral infrared oximetry (CO) in particular, as a result of which a routine assessment of oxygen status of cerebral parenchyma [5–9] in various cerebral pathologies became possible. It enabled to use CO as one of neuromonitoring methods.

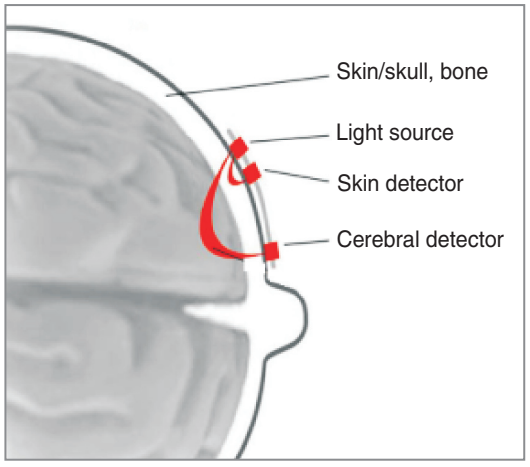

Fig. 1. Cerebral oximetry use scheme

CO technique is based on the effect of light (wavelength from 680 to 1000 nm) penetration through human tissues, and light absorption by natural chromophores: oxyhemoglobin (HbO_2), deoxyhemoglobin (HHb) and cytochrome oxydase. Infrared radiation is delivered from the source via fiber-optic cable (the so called optode) to skin sensors (detectors) consisting of an emitter and a transmitter. These sensors are located symmetrically in relation to midline and spaced 3.5–6 cm apart from one another (Fig. 1) [10].

A light beam from a transmitter penetrates through soft coverings of the head, cranial bones, cerebral parenchyma, and scattering falls on an emitter.

Concentration of chromophores: HbO_2, HHb and cytochrome oxydase — is found to be a variable value and depend on the level of tissue oxygenation and metabolism [11]. Concentration of other light-absorbing substances, such as melanin and bilirubin, and other water-soluble fractions, has a trace character, and can be out of calculations [12, 13].

Bouguer–Beer–Lambert modified formula $I=I_0e^{-k\lambda l}$ is used to calculate the concentration of chromophores [11,14–16].

A monochromic light beam with intensity I_0 passing through an absorbing substance layer, with thickness

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l comes out attenuated up to intensity I determined by the following expression $I=I_0e^{-k\lambda l}$, where $k\lambda$ — index of absorption — coefficient dependent on wavelength λ of absorbed light.

This calculation method is completely applicable in neonatology, since children skull is rather thin to be examined by light from one side to another [15, 16]. However, in adults relative thickness of scalp, cranial bones and brain makes a standard spectroscopy impossible, therefore, CO should be used in a reflection mode, when an emitter and a transmitter are situated on one head side. CO in reflection mode depends on a part of the light, which passes through brain tissue. Human head consists of several layers of different tissues, which exhibit different scattering properties and have different concentrations of compounds absorbing light. As a result, for correct chromophore determination in brain tissues, the introduction of nonlinear coefficients to determine light absorption and scattering was required [16]. Moreover, in order to exclude from calculations the blood in cranial integument, the use of dual receivers situated 2.5–3 cm apart from one another has been suggested recently (See Fig. 1).

At present, several unique CO indices are offered:

1. rSO_2 (regional saturation O_2) — INVOS and EQUANOX monitors manufactured by Covidien and Nonin Medical (USA), respectively;

2. TOI (Tissue Oxygenation Index) — NIRO monitor manufactured by Hamamatsu Photonics (Japan);

3. $rSctO_2$ (regional cerebral tissue saturation O_2) — Fore-Sight monitor manufactured by Casmed (USA).

A lot of studies have shown high degree of reliability of the indices represented that makes oxygen status monitoring a standard procedure [17].

For additional assessment of cerebral oxygen status there have been offered various coefficients and indices showing functional state of microcirculatory bloodstream and cerebral autoregulation:

1) hemispheric asymmetry coefficient — ratio of saturation difference of both hemispheres to their smaller value, expressed as a percentage [8, 18];

2) hemodynamic compliance index — CO values-to-mean AP ratio [19];

3) cerebrovascular reactivity of cerebral saturation index (TOx);

4) total hemoglobin reactivity index (THx), etc.

CO has found its general use in the assessment of changes in brain regional oxygenation and oxygen status in traumatic brain injury [20–22] and cerebrovascular pathology [3, 6, 23], as well as in patients with pathologies of carotid arteries [24].

Cerebral saturation changes in patients with intracranial hemorrhages are found to correlate significantly with oxygenation changes in the bulb of jugular vein ($SjvO_2$), as well as with brain tissue oxygen tension according to invasive cerebral tissue oximetry ($PbtO_2$) [25, 26]. Cerebral autoregulation in patients with traumatic brain injury and hemorrhagic strokes based on the comparison of reactivity indices has shown that total hemoglobin reactivity index (THx) has highly reliable relationship with intracranial pressure reactivity index (PRx) [19]. In addition, there has been found a direct significant correlation between other cerebral autoregulation indices: cerebral saturation reactivity index (TOx) and linear cerebral blood flow reactivity index (Sxa) indicating a high accuracy and safety of CO findings [27].

The results obtained by THx and TOx in about half of patients enabled to determine "optimal" cerebral perfusion pressure (CPPopt) [19, 28] that made CO indispensible in targeted therapy improvement in patients with intracranial hemorrhages, particularly in cases when intracranial pressure monitoring is impossible for some reason [29–32]. CO application enables to estimate brain perfusion changes non-invasively [33–37].

P. Taussky et al. [38] studied the correlation between the parameters of computed tomography brain perfusion and cerebral oxygenation level in patients with non-traumatic intracranial hemorrhages and revealed a highly reliable correlation between $SctO_2$ and volumetric CBF. Similar data were obtained when comparing CO and positron emission tomography findings [39].

It should be noted that by now no ideal technique has been found to measure the regional total cerebral blood volume (CBV) or cerebral blood flow (CBF) bedside and effectively, quickly, repeatedly and non-invasively. The existing methods applied to measure CBF are technically difficult, laborious, and use radioactive materials or require transportation of patients to perform brain imaging [40].

On the other hand, since CO can measure HHb and HbO_2, it is possible to measure total hemoglobin (THgb) [33, 35, 36, 41]. On the assumption that the concentration of hemoglobins is the combination of a larger number of smaller levels of hematocrit, and this correlation is constant during the investigation, and changes in THgb mean CBV change according to the following equation: $\Delta CBV=[THgb]\cdot(0.89/Hgb)$.

Thus, it became possible to make a noninvasive assessment of brain perfusion, and it was used in the management of patients with non-traumatic subarachnoid hemorrhages and showed high reliability [39, 42–45].

At the same time, the findings of cerebral perfusion computed tomography and CO in patients with traumatic brain injury, on the contrary, showed a reliable correlation between cerebral oxygenation level and regional CBV [46]. The authors explained the peculiarities of brain perfusion and cerebral oxygenation in traumatic and vascular injury of brain by the fact that regional CBF in contrast to regional blood volume can also depend on arterial bed condition, and therefore, can vary significantly in cerebral angiospasm.

Moreover, CO introduction in clinical practice enabled to find a number of its use limitations. Soft tissue

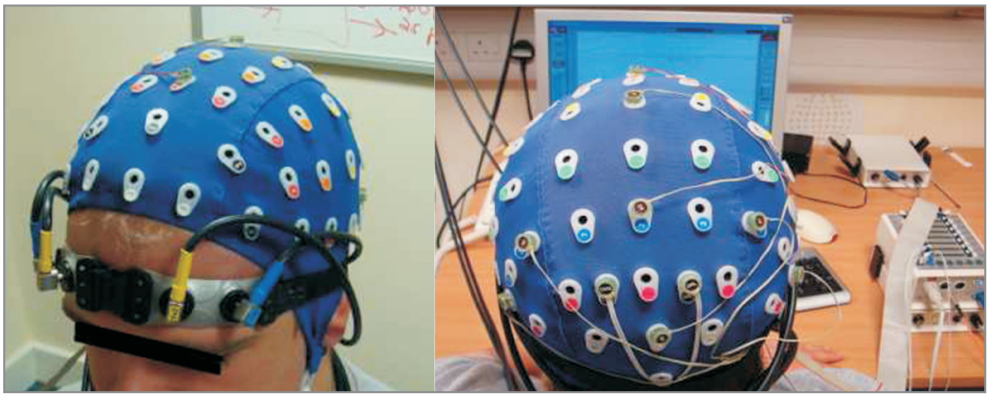

Fig. 2. Appearance of a prospective hybrid device — Brain-Computer Interface [55]

hemorrhages and skull fractures were shown to result in local change of natural chromophore concentration preventing from correct determination of regional saturation in brain parenchyma [47, 48]. Similar errors are described if sensors are located in the areas of high concentration of hair follicles, in sinuses and frontal sinuses [8].

As far as CO measures saturation of brain pooled blood (in arteries, veins and capillaries), it is still impossible to determine separate saturation of gray and white matter [8, 19], as well as to perform CO and magnetic resonance imaging simultaneously [15, 16]. Moreover, the existing cerebral oximeters are incompatible with MPT that limits their application in simultaneous investigations [15, 16].

Some researchers [22] suppose that critical intracranial hypertension can reduce the accuracy of brain saturation measurements related to impaired venous outflow and development of vasogenic brain edema.

Finally, CO limitations are individual changes in chromophore levels that reduce the accuracy of cerebral saturation absolute values, therefore, until recently, only the dynamics of indices was of practical importance. However, developed in recent years cerebral infrared spectroscopy using coherent-light sources (lasers) has improved significantly monitoring results and enabled some researchers to position such devices as “an absolute cerebral oximeter” [26].

It is worth mentioning that cerebral oximetry is a fast-developing technology, which has the potential for technological elaboration. The technique improvement, accuracy and specificity increase will expand its application in clinical practice. CO is not only a promising, cheap, non-invasive bedside technique used for volumetric CBF measuring, but also the basis for brain function and structure mapping [40, 49–52].

There are the prospects of development and introduction into practice of so called hybrid devices combining electroencephalography and CO. In English literature such devices are called “Brain-Computer Interface”. They make it possible to increase resolution of brain functional state mapping (Fig. 2) [53–59].

The same objective is pursued by CO integration into control and imaging systems, such as computed tomography, magnetic resonance imaging, and duplex ultrasonic units [4]. Portable CO devices integrated with wireless telemetry are being tested [42].

High hopes are put on such technologies as cerebral infrared spectroscopy with temporal, phase and spatial resolution [60–63].

Conclusion. Cerebral infrared spectroscopy has a number of advantages over other monitoring techniques. It provides continuous non-invasive control of brain oxygen status, being relatively easy to use and at the same time – rather sensitive to record brain oxygenation changes.

Real time monitoring of brain tissue saturation changes using infrared spectroscopy enables to detect early critical ischemic events, before they are manifested clinically. It makes the technique feasible in modern complex of neuromonitoring.

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Conflict of Interests. The authors have no conflicts of interest to declare.

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