Reproducibility of parameters of postocclusive reactive hyperemia measured by diffuse optical tomography

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Abstract. The application of near-infrared spectroscopy (NIRS) to assess microvascular function has shown promising results. An important limitation when using a single source-detector pair, however, is the lack of depth sensitivity. Diffuse optical tomography (DOT) overcomes this limitation using an array of sources and detectors that allow the reconstruction of volumetric hemodynamic changes. This study compares the key parameters of postocclusive reactive hyperemia measured in the forearm using standard NIRS and DOT. We show that while the mean parameter values are similar for the two techniques, DOT achieves much better reproducibility, as measured by the intraclass correlation coefficient (ICC). We show that DOT achieves high reproducibility for muscle oxygen consumption (ICC: 0.99), time to maximal HbO\textsubscript{2} (ICC: 0.94), maximal HbO\textsubscript{2} (ICC: 0.99), and time to maximal HbT (ICC: 0.99). Absolute reproducibility as measured by the standard error of measurement is consistently smaller and close to zero (ideal value) across all parameters measured by DOT compared to NIRS. We conclude that DOT provides a more robust characterization of the reactive hyperemic response and show how the availability of volumetric hemodynamic changes allows the identification of areas of temporal consistency, which could help characterize more precisely the microvasculature.

Keywords: diffuse optical tomography; near-infrared spectroscopy; reproducibility.

1 Introduction

Atherosclerosis and resulting cardiovascular diseases such as stroke and myocardial infarction are a major cause of death in developed countries. These account for more than 32% of mortality worldwide\textsuperscript{1} and in England and Wales cardiovascular disease was responsible for almost 30% of deaths in 2011\textsuperscript{2}. Many noninvasive methods have been developed to assess the peripheral vascular system and identify signs of atherosclerosis at an early stage. Optical methods, in particular, have received attention due to their capability of measuring tissue oxygenation and blood perfusion\textsuperscript{3,4} and because they offer several attractive features, such as portability, compactness, fast data acquisition, and noninvasiveness.

Near-infrared spectroscopy (NIRS) can determine changes in tissue hemodynamics and oxygenation\textsuperscript{5} by measuring tissue absorbance at several wavelengths in the near-infrared range of the electromagnetic spectrum (650 to 950 nm). Typically, NIRS employs a few source-detector pairs to carry out measurements. As a consequence, the spatial resolution of NIRS, which is dictated by the optode separation, is relatively low\textsuperscript{6}. Diffuse optical tomography (DOT) overcomes this limitation by employing a larger number of sources and detectors to enable three-dimensional (3-D) volumetric reconstruction\textsuperscript{7} of changes in tissue hemodynamics.

NIRS and DOT have been used in a wide range of applications including functional imaging of the brain\textsuperscript{8} and assessment of muscle oxygenation\textsuperscript{9} and cancer detection. It is widely accepted that DOT outperforms NIRS\textsuperscript{10} for example, localized changes in hemodynamics due to motor tasks in adults and motor-sensory brain activation in neonates were accurately measured using DOT, while NIRS could not even detect relative change\textsuperscript{11} because of low spatial sampling. In another study, DOT could discriminate the somatosensory activation of two fingers, while in the same experiment, a 12-channel NIRS setting failed to resolve the activation\textsuperscript{12}. Among the factors affecting the accuracy of NIRS concentration calculations are the differences in the pathlength factor: location, spatial extent, and heterogeneous distribution of absorption changes, e.g., multiple absorption foci. Although these sources of error could be minimized\textsuperscript{13}, DOT accounts for these problems implicitly.

DOT requires solving two distinct problems: the forward problem and the inverse problem. The forward problem requires solving of the equation governing the photon transport in tissue to predict the detector measurements; typically this involves solving a diffusion equation over a 3-D spatial domain using the finite element method. The inverse problem involves estimating the optical properties of tissue to minimize the difference between experimental and model-predicted measurements; this is achieved by solving a nonlinear optimization problem, which can take hours to complete even on a high-end workstation. However, as we have recently demonstrated\textsuperscript{14}, it is possible to significantly speed up reconstruction of hemodynamic changes in complex tissue structures by using reduced order models of photon transport in tissue. This makes it possible...
to perform real-time monitoring of hemodynamic responses even with relatively modest computing resources.

NIRS, like other methods of measuring microvascular function, can detect significant differences in microvascular function between groups of healthy controls and patients with peripheral vascular diseases, peripheral artery disease (PAD), and coronary heart disease. However, these techniques are highly variable and have not been shown to be helpful in predicting an individual’s risk of future cardiovascular disease. Therefore, from a clinical point of view, refinements that improve reproducibility and reduce variability are highly desirable.

Typically, the evaluation of microvascular function by NIRS relies solely on a few measuring channels, and although repeatability and accuracy of results are promising, there has not been a direct comparison of the parameters obtained during and after arterial occlusion by NIRS versus DOT. Furthermore, while multichannel systems for analysis of microvascular function are available, studies on repeatability are lacking.

This study aims to evaluate and compare, using experimental data collected from healthy volunteers, the intrasubject reproducibility of key parameters of the hemodynamic response during postocclusive reactive hyperemia, obtained using DOT and NIRS, and to highlight the potential advantages of DOT in assessing endothelial function.

2 Methods

2.1 Subjects

This study was approved by the Research Ethics Committee of the University of Sheffield. A group of 17 subjects was recruited, after giving informed consent. The group consisted of 11 men and 6 women. Baseline characteristics are listed in Table 1. Smokers or those with a history of cardiovascular disease were excluded.

It is known that measurement of vascular function is greatly influenced by external factors such as recent activity, diet, time of day, and so on. These influences greatly reduce the applicability of vascular function measurement to clinical practice since in the real world it is difficult or impossible to control these factors. For this reason, we did not ask participants to fast or refrain from physical activity or caffeine for either of their visits.

2.2 Instrumentation

DOT and NIRS measurements were obtained using a dynamic near-infrared optical tomography (DYNOT) instrument (NIRx). The system illuminates the tissue with four laser diodes at wavelengths $\lambda = 725, 760, 810, \text{ and } 830$ nm.

The diodes are modulated at distinct frequencies and then coupled with 30, 1 mm multimode optical fibers—optodes—acting both as sources and detectors. Synchronous detection allows parallel measurement at a sampling frequency of 1.8 Hz. The optodes were organized in a hexagonal pattern with an interoptode spacing of 8 mm as shown in Fig. 1(a) and placed in a solid plastic holder to avoid movement artifacts [Fig. 1(b)]. For simplicity, hemoglobin concentration was calculated using only the wavelengths at 760 and 830 nm, chosen based on their symmetry with respect to the isobestic point of the extinction coefficients of hemoglobin.

2.3 Near-Infrared Spectroscopy

NIRS employs the modified Beer–Lambert law (MBLL)

$$\text{OD} = -\log \frac{I}{I_0} = \varepsilon \text{CLB} + G$$  \hspace{1cm} (1)

to convert from changes in absorption to changes of de/oxy-hemoglobin. In Eq. (1), OD is the optical density and $I_0$ and $I$ are incident and detected light intensities, respectively. $\varepsilon$ represents the extinction coefficient of the tissue, and $C$ is the concentration of the chromophore. $L$ denotes the mean path length of detected photons. $B$ is the path length factor, which accounts for the compensation of the increase of path lengths at various wavelengths caused by the scattering phenomena. $G$ is defined as a geometric factor used to compensate the objective with different geometrical shapes. Typically, $L$, $B$,

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (year)</td>
<td>33 ± 7.5</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>70.5 ± 14.1</td>
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<tr>
<td>Height (cm)</td>
<td>172.1 ± 8.3</td>
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<td>BMI (kg/m$^2$)</td>
<td>23.6 ± 3.4</td>
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<td>Smokers</td>
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</table>

Table 1 Baseline characteristic of study subjects ($n = 17$).

Fig. 1 Experimental setup for NIRS and DOT measurement of reactive hyperemia. (a) Array with 30 optodes placed 8 mm apart in a hexagonal pattern. Optodes in red were used for NIRS measurements. All optodes were used for DOT reconstructions. (b) Placement of optode array on volar forearm of subject. (c) Rectangular finite element mesh used to model forearm tissue with dimensions 92 mm × 35 mm × 24 mm ($W \times L \times D$). Blue circles at $z = 0$ mm indicate the location of the 30 optodes.
and $G$ are constants with monochromatic illumination in a turbid media with unchanging geometry.

A change in the concentration of the chromophore causes a change in the intensity measured. The parameters $\varepsilon$ and $L$ remain constant, and it is assumed that $B$ and $G$ remain constant. Under this assumption, Eq. (1) can be written as

$$\Delta OD = OD_{\text{Final}} - OD_{\text{Initial}} = -\log \frac{I_f}{I_i} = \varepsilon \Delta CLB,$$  

(2)

where $\Delta OD$ is the change in optical density. $OD_{\text{Final}}$ and $OD_{\text{Initial}}$ are the detected optical density and the optical density of incident light. $I_f$ and $I_i$ are the measured intensities before and after the change in concentration; $\Delta C$ is the change in concentration. In Eq. (2), $B$ is often referred to as the differential pathlength factor (DPF). DPF was determined experimentally for a number of tissues including forearm, calf, adult, and infant head. Changes in detected light are dominated mainly by oxygenated (HbO$_2$) and deoxygenated hemoglobin (HbR) such that

$$\Delta OD^i = (\varepsilon^i_{\text{HbO}_2} \Delta [\text{HbO}_2] + \varepsilon^i_{\text{HbR}} \Delta [\text{HbR}]) B^i L,$$  

(3)

where $\varepsilon^i_{\text{HbO}_2}$, $\varepsilon^i_{\text{HbR}}$, $B^i$ are the extinction coefficients for oxygenated hemoglobin and deoxygenated hemoglobin and DPF at a given wavelength. By measuring the change in intensity at two wavelengths, it is possible to determine the concentration changes in HbO$_2$ and HbR

$$\Delta [\text{HbR}] = \frac{\varepsilon^{32}_{\text{HbO}_2} \Delta OD^{32} - \varepsilon^{31}_{\text{HbO}_2} \Delta OD^{31}}{(\varepsilon^{31}_{\text{HbO}_2} \varepsilon^{32}_{\text{HbR}} - \varepsilon^{32}_{\text{HbO}_2} \varepsilon^{31}_{\text{HbR}}) L},$$

$$\Delta [\text{HbO}_2] = \frac{\varepsilon^{31}_{\text{HbR}} \Delta OD^{32} - \varepsilon^{32}_{\text{HbR}} \Delta OD^{31}}{(\varepsilon^{31}_{\text{HbR}} \varepsilon^{32}_{\text{HbO}_2} - \varepsilon^{32}_{\text{HbR}} \varepsilon^{31}_{\text{HbO}_2}) L}.$$  

Typical source-detector separations in NIRS studies in reflectance mode are in the range 20 to 50 mm. NIRS measurements taken with the DYNOT equipment for separations larger than 45 mm were noisy (CV > 5%), therefore this study was limited to a source-detector separation of $L = 32$ mm obtained by selecting the central fibers indicated with red circles in Fig. 10A. Although DPF depends on inter-optode spacing and wavelength, experiments carried out in the forearm suggest that this parameter becomes constant for distances larger than 25 mm. For this study, DPF = 4.0 was used for the calculation of concentration changes.

### 2.4 Diffuse Optical Tomography

Photon transport in tissue was modeled using the diffusion approximation of the radiative transport equation. This is a more accurate description of photon transport than the MBLT because it takes into account the random scattering of light produced by tissue. Consider the medium $\Omega \subset \mathbb{R}^3$ with boundary $\partial \Omega$, the diffusion equation in the steady-state domain is

$$-\nabla D(r) \nabla \phi_i(r) + \mu_a(r) \phi_i(r) = q_i(r), \quad r \in \Omega,$$  

(5)

where $\phi_i(r)$ is the spatially varying photon fluence at $r$ due to source $q_i$, $\mu_a$ is the absorption coefficient, $D = [3(\mu_a + \mu'_s)]^{-1}$ is the diffusion coefficient, and $\mu'_s$ is the reduced scattering coefficient. The source term represents an isotropic point source $q_i(r) = \delta(r - r_i)$ located at a depth of one scattering length inside the medium ($d = 1/\mu'_s$). The boundary condition is usually of Robin type

$$D(\xi) \frac{\partial \phi_i(\xi)}{\partial n} + \frac{1}{2A} \phi_i(\xi) = 0, \quad \xi \in \partial \Omega,$$  

(6)

where the term $A$ accounts for the refractive index boundary mismatch at the interface. The quantity measured by a detector located at $\xi_j \in \partial \Omega$, given the point source $q_i(r)$, is the outward flux $\Gamma_i(\xi_j)$, and it is calculated from Fick’s law

$$\Gamma_i(\xi_j) = -D(\xi_j) \vec{n}(\xi_j) \cdot \nabla \phi_i(\xi_j), \quad \xi_j \in \partial \Omega,$$  

(7)

where $\vec{n}(\xi_j)$ denotes the direction of the normal vector to the boundary at the detector location $\xi_j$. Equations (6)-(7) constitute the forward problem in DOT, which can also be represented by a scalar operator mapping $\mathcal{B}$ between the space of optical parameters of interest, $\mu_a$ in this case, and the space of measurements as

$$y_{i,j} = \Gamma_i(\xi_j) = P_{i,j}[\mu_a(r)], \quad r \in \Omega,$$  

(8)

where $y_{i,j}$ is the output of the $j$th detector given the source $i$.

#### 2.4.1 Image reconstruction

3-D volumetric reconstruction of the absorption coefficient was carried out using a modified version of the normalized difference approach proposed by Pei et al. At each sampling time $t_k$, the 3-D absorption coefficient map $\mu_a(r, t_k) = \mu_a(r, t_k) + \Delta \mu_a(r, t_k)$, discretized over the 3-D mesh, is determined by minimizing, using a nonlinear conjugated-gradient algorithm, the following cost function

$$F = \sum_{r \in \Omega} \sum_{i=1}^{N_s} \sum_{j=1}^{N_d} \left\{ \frac{y_{i,j}(r)}{\gamma_{i,j}(r)} \tilde{P}_{i,j}[\mu_a(r)] - \tilde{P}_{i,j}[\mu_a(r, t_k)] \right\}^2.$$  

(9)

The first summation in the right side of Eq. (9) is over the volume elements of the 3-D mesh. In Eq. (9), $N_s$ is the total number of sources, $N_d$ is the number of detectors for the $i$th source, $y_{i,j}(r)$ is the flux measured by the $j$th optode given the source $i$ at time $t_k$, $y_{i,j}(t_0)$ is a reference state defined as the mean of the baseline measurements, and $\tilde{P}_{i,j}[\mu_a(r)]$ and $\tilde{P}_{i,j}[\mu_a(r, t_k)]$ are the model predicted measurements for the reference and estimated absorption coefficient at time $t_k$. Essentially, for each time sample, the initial guess in the optimization process is the reference state $\mu_a(r)$. The conjugate gradient descent algorithm is applied iteratively for $n$ steps to compute the estimate $\mu_a(r, t_k)$. Note that the optimized image from a previous time sample can be used also as the initial guess. However, in this paper, concentration calculations were carried out independently of previous or subsequent samples.

#### 2.4.2 Calculation of hemoglobin concentration

The absorption coefficient is related to the extinction coefficient and concentration as $\mu_a = \varepsilon \, c$. Assuming that the primary source of absorption changes is a combination of hemoglobin chromophores.

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J. Biomed. Opt. 21(6), 066012-3 (June 2016)
\[ \Delta \mu_s = \epsilon_{\text{HbO}_2} \Delta [\text{HbO}_2] + \epsilon_{\text{HbR}} \Delta [\text{HbR}], \]

where \( \epsilon_{\text{HbO}_2} \) and \( \epsilon_{\text{HbR}} \) are the extinction coefficients for oxyhemoglobin and deoxyhemoglobin at a given wavelength. Reconstruction of absorption changes at two different wavelengths provides two independent equations, which can be solved simultaneously to calculate deoxyhemoglobin and oxyhemoglobin concentration changes as

\[
\Delta [\text{HbR}] = \frac{\epsilon_{\text{HbO}_2} \Delta [\text{HbO}_2]}{\epsilon_{\text{HbO}_2} \epsilon_{\text{HbR}}} - \frac{\epsilon_{\text{HbR}} \Delta [\text{HbR}]}{\epsilon_{\text{HbO}_2} \epsilon_{\text{HbR}}},
\]

\[
\Delta [\text{HbO}_2] = \frac{\epsilon_{\text{HbR}} \Delta [\text{HbR}]}{\epsilon_{\text{HbO}_2} \epsilon_{\text{HbR}}} - \frac{\epsilon_{\text{HbO}_2} \Delta [\text{HbO}_2]}{\epsilon_{\text{HbO}_2} \epsilon_{\text{HbR}}}.
\]

### 2.4.3 Finite element modeling

A simplified finite element model of the skin and muscle tissue was used to implement the DOT reconstruction algorithms. The tissue was modeled as a rectangular cuboid with two layers [Fig. 1(c)] with dimensions 92 mm \times 35 mm \times 24 mm (length \times width \times depth). A tetrahedral mesh with 6900 elements and 1848 nodes [Fig. 1(c)] was generated for this domain. Each layer was assigned realistic optical parameters within the range of clinically normal tissue: \( \mu_s = 0.01 \text{ mm}^{-1} \) and \( \mu_s' = 1.00 \text{ mm}^{-1} \) for skin, and \( \mu_s = 0.02 \text{ mm}^{-1} \) and \( \mu_s' = 0.50 \text{ mm}^{-1} \) for muscle.

A simulation study was carried out to calculate the photon measurement density functions for different source-detector configurations. PDMFs characterize the regions of the tissue that contribute to the measurement signal and can be used to determine a suitable combination of measurements to reconstruct optical properties of tissue within a region of interest (ROI).

It is well known that the sensitivity in the reflectance mode is higher at the surface and diminishes as function of the depth. Higher sensitivity in the reflectance mode can be used to determine a suitable combination of measurements to reconstruct optical properties of tissue within a region of interest (ROI).

The study confirmed that for the chosen optode arrangement, separations larger than 12 mm contributed less than 5% to the sensitivity function. As a result, the ROI used in the analysis was a rectangular cuboid with dimensions 80 mm \times 20 mm \times 12 mm (length \times width \times depth) as shown in Fig. 1.

![Fig. 2 The ROI is indicated with the volume in green. The location of the optodes in relation to the ROI is indicated with the blue circles. The dimensions of the ROI are 80 mm \times 20 mm \times 12 mm (length \times width \times depth). The ROI is located directly under the array of 30 optodes.](image)

Yu et al. showed through experiments and simulation studies that for source–detector separations larger than 1 cm, the hyperemic response is mainly influenced by the autoregulation of muscle tissue, while smaller separations are primarily influenced by subcutaneous tissue layer. For this reason, first- and second-nearest neighbor source–detector pairs were not used in the reconstruction.

3-D maps of absorption changes were obtained by minimizing the cost function in Eq. (9). To minimize the occurrence of artifacts, a two-step sign constraint algorithm was used during the reconstruction process. Positivity/negativity constraints were imposed after each iteration in separate reconstructions, and then the final absorption value was calculated as the sum of the two partial solutions. Positivity/negativity constraints force the algorithm to seek only positive/ negative changes in order to minimize the cost function given in Eq. (9). In each case, convergence was usually achieved after 5 to 10 iterations. Pei et al. demonstrated theoretically and experimentally that by using sign constraints, the quality and specificity of the recovered images improved significantly.

Finally, Eq. (11) was used to compute \( \text{HbO}_2, \text{HbR}, \) and \( \text{HbT} (\text{HbO}_2 + \text{HbR}) \). Near-infrared parameters of postocclusive reactive hyperemia (PORH) were calculated on a nodal basis and then averaged on the ROI.

### 2.5 Study Protocol

The weight, height, and age for each subject were recorded. The baseline characteristics of the subjects are summarized in Table 1. The parameters measured during and after the arterial occlusion test are (Fig. 3):

(i) mVO2 (mlO2/min/100 g), muscle oxygen consumption: calculated from the gradient of the HbO2 at the beginning of the arterial occlusion and converted into mlO2/min/100 g.

(ii) 1/2 THbO2 (s), 1/2 time of recovery of HbO2: time needed for half recovery of HbO2 from maximum deoxygenation at the end of the occlusion period to maximum reoxygennation during reactive hyperemia.
(iii) THbO₂ (s), time to maximal HbO₂: time interval after cuff release until maximum HbO₂ value is reached.

(iv) MaxHbO₂ (µM), maximal amplitude of HbO₂: maximal hyperemic response reached by HbO₂.

(v) MaxHbT (µM), maximal amplitude of HbT: maximal hyperemic response of HbT.

(vi) THbT (s), time to maximal HbT: time to peak value of HbT after cuff release.

(vii) Increase rate to maximal HbO₂ (µM/s), reoxygenation rate of HbO₂: calculated by dividing Max HbO₂ by the time to maximal HbO₂.

(viii) Increase rate to maximal HbT (µM/s), reoxygenation rate of HbT: calculated by dividing Max HbT by the time and maximal HbT.

(ix) AUC HbO₂ (a.u.), area under the curve of HbO₂ after cuff release.

(x) AUC HbR (a.u.), postocclusion area under the curve of HbR.

(xi) AUC HbT (a.u.), area under the curve of HbT after cuff release.

The reproducibility was assessed using the intraclass correlation coefficient (ICC) and the standard error of measurement (SEM) based on NIRS parameters measured on two separate occasions. ICC provides the degree to which individuals maintain their rank order across repeated tests, and SEM is a measure of absolute repeatability. To interpret repeatability based on ICC values, the following guidelines were considered: ICC ≥ 0.8 indicates “excellent,” 0.6 ≤ ICC < 0.8 is defined as “good,” 0.4 ≤ ICC < 0.6 indicates “moderate,” and ICC < 0.4 indicates “fair.” On the other hand, SEM is expressed as a percentage and the ideal value is zero. In general, the smaller the SEM, the smaller the variability between individual repeated measurements. ICC was calculated for near-infrared parameters derived using DOT and NIRS, and this is denoted by ICC_DOT and ICC_NIRS, respectively (similarly for SEM: SEM_DOT and SEM_NIRS).

3 Results

3.1 Microvascular Function Reproducibility

Table 2 lists the mean and standard deviation of near-infrared parameters, calculated using DOT, during and after the arterial occlusion test, together with ICC and SEM repeatability measures. Similarly, Table 2 displays mean and standard deviation of near-infrared parameters measured using NIRS. Mean values in both tables are very similar; however, the standard deviation for near-infrared parameters derived using DOT is smaller in most cases. More importantly, the ICC_DOT is higher in all but one case, and the repeatability is extremely high (~1) for mVO₂, THbO₂, Max. HbO₂, and THbT. In comparison, only one parameter (mVO₂) measured with NIRS reached the “excellent” ICC category. The least repeatable parameter is AUC HbR, and this is consistent in the two measuring approaches.

SEM calculated for tomographic imaging showed consistently smaller values across all parameters. This suggests that region-wise averaged measurements provide more robust and repeatable measurements than point-wise measurements, as typically measured in NIRS.

4 Discussion

Our results are in agreement with similar studies using NIRS; however, the main finding is that tomographic imaging provides more robust and repeatable results. Figure 3 shows the mean and SEM of de/oxyhemoglobin and total hemoglobin across all
Table 3 Intrasubject reproducibility of PORH parameters obtained using NIRS spectroscopy.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Test 1</th>
<th>Test 2</th>
<th>ICC</th>
<th>SEM</th>
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<tbody>
<tr>
<td>mVO₂</td>
<td>0.08 ± 0.06</td>
<td>0.08 ± 0.04</td>
<td>0.97</td>
<td>10.98</td>
</tr>
<tr>
<td>1/2THbO₂</td>
<td>16.57 ± 11.18</td>
<td>15.64 ± 3.29</td>
<td>0.64</td>
<td>26.82</td>
</tr>
<tr>
<td>THbO₂</td>
<td>45.90 ± 18.63</td>
<td>47.79 ± 17.56</td>
<td>0.77</td>
<td>18.55</td>
</tr>
<tr>
<td>MaxHbO₂</td>
<td>16.56 ± 4.66</td>
<td>15.85 ± 4.46</td>
<td>0.67</td>
<td>16.29</td>
</tr>
<tr>
<td>MaxHbT</td>
<td>17.72 ± 6.08</td>
<td>16.01 ± 4.42</td>
<td>0.37</td>
<td>24.70</td>
</tr>
<tr>
<td>THbT</td>
<td>19.90 ± 2.88</td>
<td>20.69 ± 6.38</td>
<td>0.71</td>
<td>12.25</td>
</tr>
<tr>
<td>Inc. rate to max HbO₂</td>
<td>1.06 ± 0.52</td>
<td>0.85 ± 0.38</td>
<td>0.21</td>
<td>41.94</td>
</tr>
<tr>
<td>Inc. rate to max HbT</td>
<td>0.73 ± 0.40</td>
<td>0.60 ± 0.45</td>
<td>0.40</td>
<td>50.17</td>
</tr>
<tr>
<td>AUC HbO₂</td>
<td>1053.21 ± 740.53</td>
<td>1131.97 ± 838.85</td>
<td>0.78</td>
<td>33.89</td>
</tr>
<tr>
<td>AUC HbR</td>
<td>810.67 ± 657.29</td>
<td>1016.41 ± 840.48</td>
<td>0.45</td>
<td>61.01</td>
</tr>
<tr>
<td>AUC HbT</td>
<td>1385.63 ± 929.66</td>
<td>1897.82 ± 836.57</td>
<td>0.19</td>
<td>48.47</td>
</tr>
</tbody>
</table>

subjects computed using NIRS and DOT. NIRS calculations are noisier than DOT reconstructions, despite using the same filtering techniques in both methods. One reason might be that single-channel measurements are more susceptible to systematic errors, such as position-dependent differences in muscle oxygenation or probe localization and interobserver reproducibility. On the other hand, DOT samples a larger tissue volume. Averaging over a large number of voxels also smooths the response. Furthermore, the image reconstruction method for DOT implements regularization that further improves the SNR.

Figure 5(a) shows the volume averaged HbO₂, HbR, and HbT for a representative subject while Fig. 5(b) shows a tomographic reconstruction for HbO₂ at its maximum value (t = 516 s). The heterogeneity of tissue is evident, and this has been reported in similar studies involving two-dimensional images of venous occlusion tests. Furthermore, spatial dependence of the hemodynamic responses in the forearm has also been reported in nerve stimulation studies and in exercising hand in the reflectance and transmittance modes. This variability that occurs in all three spatial directions may contribute to diminished repeatability if the probe location is not consistent across examinations.

NIRS and DOT calculations have clear methodological differences, both prone to systematic errors. For NIRS, the key parameter is the DPF, which corrects for the effect of scattering in tissue and therefore is wavelength and tissue dependent. This variable has been the subject of several investigations, and there is an agreement on its main characteristics at population levels. However, in most of the cases, DPF is not calculated as part of the arterial/venous occlusion protocol, and calculations are based on data and tables available in the literature.

The techniques used in DOT are more complex and also susceptible to systematic errors such as parameterization of optical parameters, initial guess, or reconstruction convergence criteria. In this study, a differential approach was followed, which has demonstrated to be insensitive to boundary effects and the medium’s initial guess. On the other hand, the major disadvantage is that it is not possible to determine the absolute distribution of optical parameters, but only the change of absorption or diffusion from a given baseline.

In general, our results show time-based parameters have excellent repeatability. These parameters have shown great promise in distinguishing between healthy volunteers and patients with PAD and diabetes. For these patient groups, the reactive hyperemic response showed delayed times of recovery. Parameters derived from HbO₂ are also the most repeatable, and this has been observed by different groups.

4.1 Regions of Hemodynamic Consistence

The availability of 3-D maps allows the selection of more specific areas of interest. Selection techniques, either manual or automatic, are available and have been shown to enhance the response by improving the signal-to-noise ratio. Wang used...
Fig. 5 (a) Typical hemodynamic responses of oxygenated (HbO$_2$), de-oxygenated (HbR) and total hemoglobin (HbT) and (b) HbO$_2$ ($x, y, z, t$) at $t = 516$ s, i.e., the time average HbO$_2$ reaches maximum.

Fig. 6 (a) Total hemoglobin curve (HbT) for a healthy subject; the point at maximal HbT ($t = 467$ s) is marked with a point. (b) 3-D map of HbT($x, y, z$) at the time of maximal HbT. The node with maximum contribution to the HbT response, HbT($x_M, y_M, z_M$), is indicated by the arrow and its time response is denoted with HbT$_{max}$($t$). (c) 3-D map of correlation coefficients $\rho(x, y, z)$ computed between HbT$_i(t)$ at each node location ($x_i, y_i, z_i$) and HbT$_{max}(t)$. (d) 3-D map of correlation coefficients $\rho(x, y, z)$ computed between HbT$_i(t)$ at each node location ($x_i, y_i, z_i$) and HbT$_{mean}(t)$. 
Step 1. Compute the average total hemoglobin time series over the ROI, denoted by $H_{bTROI}(t)$, and determine the time point satisfying $H_{bTROI}(t_{\text{max}}) = \max[H_{bTROI}(t)]$.

This point is located at $t_{\text{max}} = 467$ s [Fig. 6(a)].

Step 2. Determine the node $(x_M, y_M, z_M)$ satisfying $H_{bT}(x_M, y_M, z_M, t) = H_{bT_{\text{max}}}(t)$ and $H_{bT}(x, y, z, t) = H_{bT}(t)$ using $H_{bT_{\text{ROI}}}(t)$.

Step 3. For each node $(x, y, z)$ within the ROI, compute the cross-correlation between $H_{bT}(x_M, y_M, z_M, t)$ and $H_{bT}(x, y, z, t)$ for $t = 467$ s [Fig. 6(b)].

where $N$ is the total number of samples and the bar over each variable denotes the time average over the duration of the experiment. The analysis was done on the entire experiment to illustrate the different dynamic signatures that can be obtained when different reference signals are used. Note also that $H_{bT_{\text{max}}}(t)$ is not necessarily the time series with the highest $H_{bT}$ across the experiment.

The resulting 3-D correlation map $\rho(x, y, z)$, shown in Fig. 6(c), indicates very strong correlation ($\rho > 0.9$) with the reactive hyperemic response $H_{bT_{\text{max}}}(t)$ for nodes located at a depth $\sim 2$ mm, which largely correspond to muscle tissue. In contrast, the correlation coefficients for nodes located in the superficial layer (skin) are very low ($\rho \leq 0.4$). A different correlation map was computed using as a reference the averaged $H_{bT}$ signal over the ROI, denoted $H_{bTROI}(t)$.

The corresponding 3-D correlation map, shown in Fig. 6(d), also exhibits strong correlation for nodes ($\rho > 0.9$) located below the skin layer. The correlation maps can be used to define tissue compartments with similar hemodynamic profiles. For example, regions of hemodynamic consistency (RHC) can be defined based on the correlation maps shown in Fig. 6(c) and 6(d) by selecting all nodes with correlation coefficient $\rho \geq 0.9$.

The resulting RHCs are shown in Fig. 7(a); the red and green volumes indicate the regions with high correlation ($\rho \geq 0.9$) with the $H_{bT_{\text{mean}}}$ and $H_{bT_{\text{max}}}(t)$, respectively. The blue volume represents the intersection between the two RHC. The average $H_{bT}$ time series for each RHC are shown in Fig. 7(b) together with the averaged $H_{bT}$ over the entire ROI $H_{bT_{ROI}}(t)$.

Interestingly, while the three signals are different before the end of the occlusion period, after the cuff is released only the region correlated to $H_{bT_{\text{max}}}(t)$ is still distinct compared with the average signal over the entire ROI. This tissue “compartment” identified by our analysis [Fig. 7(a)] appears to exhibit extensive reactive hyperemia both in terms of significantly increased blood flow relative to the baseline as well as duration of hyperemia.

The DOT-estimated $MaxHbT$ and $MaxHbO_2$ also have good reproducibility ($\text{ICC} = 0.69$ and $\text{ICC} = 0.99$, respectively), suggesting that an RHC identified based on $H_{bT_{\text{max}}}(t)$ could be used to define regions of hemodynamic consistency (RHC). The average $H_{bT}$ time series for each RHC are shown in Fig. 7(b) together with the averaged $H_{bT}$ over the entire ROI $H_{bT_{ROI}}(t)$.

Fig. 7 (a) Compartments defined by the two RHC: the A and B volumes indicate the regions with high correlation ($\rho > 0.9$) with the $H_{bT_{\text{mean}}}(t)$ and $H_{bT_{\text{max}}}(t)$, respectively. The AB represent the intersection between the A and B regions. The boundaries of the ROI are denoted with the black box. (b) Total hemoglobin averaged over the entire ROI $H_{bT_{ROI}}(t)$ is indicated with the solid line. The dashed and dotted lines denote the average $H_{bT}$ response of all the nodes with high correlation ($\rho > 0.9$) with $H_{bT_{\text{mean}}}(t)$ and $H_{bT_{\text{max}}}(t)$, respectively.
be a good candidate to focus on in future studies that aim to distinguish between healthy volunteers and patients with vascular disorder.

In contrast, the relatively poor reproducibility of MaxHbT measured using NIRS (ICC = 0.37), which clearly limits its diagnostic potential, highlights the challenges posed by the spatial and temporal heterogeneity of tissue oxygenation and hemodynamics when assessing endothelial function.

To further illustrate the advantage of using DOT to resolve the spatiotemporal properties of the hemodynamic response, we created a short movie (Fig. 8) showing the reconstructed HbT changes in different slice planes along the x axis.

5 Conclusions

This study shows that DOT achieves excellent reproducibility of key PORH parameters. These parameters include muscle oxygen consumption (mVO2), time to maximal HbO2 (THbO2), maximal HbO2 (MaxHbO2), and time to maximal HbT (THbT). Although a direct comparison of these parameters may be enough to distinguish between healthy volunteers and patients with vascular disease, our analysis suggests that obtaining reliable signatures of vascular disease may require first the identification of an appropriate RHC. This is only possible if a full volumetric reconstruction of hemodynamics, as that provided by DOT, is available for the particular ROI. The choice of an appropriate RHC is beyond the scope of this study, and it is the subject of future research.

The availability of volumetric hemodynamic parameters clearly offers more opportunities for the analysis and characterization of PORH and warrants further efforts to evaluate DOT’s potential to measure endothelial function in a clinical environment.

Despite the increased complexity of the instrumentation and reconstruction algorithms used to implement DOT, there are many examples of inexpensive portable and wearable DOT instruments. The availability of freely available software such as NIRFAST and TOAST also encourages the development of DOT-based technology. The use of reduced-order models to speed up the reconstruction process makes it now possible to perform analyses in real-time using mobile devices with modest computational resources.

Overall, the inherent advantages of DOT compared with other imaging modalities, combined with the availability of algorithms and portability of the instrumentation, makes DOT an ideal method for routinely and noninvasively assessing the cardiovascular function, inside and outside the hospital.

Acknowledgments

The authors gratefully acknowledge that this work was supported by the University of Sheffield as part of the “Engineering for Life” 2022 Futures Project and by EPSRC grants EP/H00453X/1 (S.A.B. and D.C.) and GR/T01006/01 (D.C.), by BBRC grant BB/K010123/1 (S.A.B. and D.C.) and by an ERC Advanced Investigator Grant (S.A.B.). E.E.V. R. gratefully acknowledges the support from a grant from the Mexican National Research Council for Science and Technology (CONACYT).

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