Exploration of movement artefacts in handheld laser speckle contrast perfusion imaging

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Abstract: Functional performance of handheld laser speckle contrast imaging (LSCI) is compromised by movement artefacts. Here we quantify the movements of a handheld LSCI system employing electromagnetic (EM) tracking and measure the applied translational, tilt and on-surface laser beam speeds. By observing speckle contrast on static objects, the magnitudes of translation and tilt of wavefronts are explored for various scattering levels of the objects. We conclude that for tissue mimicking static phantoms, on-surface speeds play a dominant role to wavefront tilt speed in creation of movement artefacts. The ratio depends on the optical properties of the phantom. Furthermore, with the same applied speed, the drop in the speckle contrast increases with decreasing reduced scattering coefficient, and hence the related movement artefact increases.

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1. Introduction

As a full-field, non-invasive and affordable imaging tool, laser speckle contrast imaging (LSCI) has been widely investigated for applications such as blood flow studies in brain, neuroscience, dermatology, rheumatology and burns [1,2]. Since the conventional mounted LSCI systems are bulky, development of the technology as a handheld modality has gained attention in the past few years [3–6]. A handheld camera-phone based LSCI has been used to study murine cerebral malaria by visualizing retinal perfusion [7]. In some clinical settings such as neonatal intensive care units, rapid instrumentation and compactness is of paramount interest [8]. Moreover a handheld LSCI can help comfortably monitoring any skin area of a patient, for instance when their knees or feet needed to be observed in case of having Psoriasis, burns or diabetic wounds. Beside the advantages of portability, integrability and affordability, movement artefacts pose challenges in performing an accurate and reliable measurement.

The first study on movement artefacts was carried out by Mahé et al. [9] where the backscattered light from an adjacent opaque surface and skin surface was measured. They found a linear factor of proportionality between the two signals. Then, using point by point subtraction, the movement artefact was shown to be reduced. However, the movements by patients were not applied in a standardized way. The correction should also be optimized per different test subjects. In a later work, this technique, referred to as ‘a specific post-processing procedure’, was employed to examine LSCI during exercise on healthy subjects [10]. Omarjee et al. [11] proposed to use a bi-layer adhesive opaque surface to detect movement-related signal and subtract it from the artefact-induced LSCI signal recorded from skin surface in an attempt to make the artefact removal procedure calibration-free. However, the movement artefacts caused by handheld measurement were not considered in that work. A handheld study by Lertsakdadet et al. [12] sorted the recorded frames based on the speckle contrast value of a fiducial marker used in the imaging protocol to filter the frames exceeding a certain threshold. Then, the selected frames
were aligned based on edge-detection of a fiducial marker to decrease blurring of the resultant time-averaged contrast map. However, selecting only proper frames may cause data loss. They recently showed reduction of movement artefacts during handheld LSCI measurement thanks to the gimbal stabilizer which was at the expense of increasing total weight of the system [13].

Rather than adding an improved method for correcting movement artefacts, here we take a step back to explore the effect of movement artefacts in handheld laser speckle perfusion imaging by investigating the relation between the speckle contrast and applied speed in both motorized and handheld scenarios, for static objects. We measure the movement of a handheld LSCI system using an electromagnetic (EM) tracking system and determine the contribution of tilt and translation of wavefronts on speckle contrast drop. We also investigate to which extent the magnitude of speckle contrast drop depends on the scattering properties of the medium.

2. Methods

2.1. Handheld LSCI probe design

The designed handheld probe is illustrated in Fig. 1(a) which has a total weight of approximately 750 g including the attached cables. The light source was chosen to be a continuous wave single longitudinal mode laser (CNI MSL-FN-671) of 671 ± 1 nm wavelength. The output power was 300 mW with a coherence length of longer than 50 m. An absorptive filter with optical density of 0.2 (Thorlabs NE02A) was mounted in front of the laser beam with some angle deviation to prevent direct reflection to the laser source. The laser beam was further directed using broadband dielectric mirrors of wavelength 400 – 750 nm (Thorlabs BB1-E02) to a microscope objective of magnification 20 (Nikon CFI Plan Fluor DIC N2) in order to make a focus to a single mode optical fiber (Thorlabs SM600) with operating wavelength of 633 – 780 nm. The distal end of the optical fiber was mounted on the handheld-LSCI probe followed by a 20° top hat square engineered diffuser (Thorlabs ED1-S20-MD) to form a square uniform light beam. The distance from the fiber tip to the diffuser was set to approximately 2 cm. The backscattered light was recorded using a USB3 monochrome camera (Basler acA2040 55um) with image depth of 8 bits, frame size of 2048 × 1536 pixels, exposure time of 25 ms and operating frame rate of 40 Hz. The camera objective (FUJINON HF16XA-5M) had a focal length of 16 mm and the focus range from 10 cm. The f-number for the experiments was set to F8. Based on our measurement, this is the optimum aperture size for the system in terms of speckle size to meet the Nyquist criterion [14] and in terms of detected light intensity to obtain the required dynamic range for computation of the speckle contrast. To only detect the laser light, a hard coated bandpass interference filter of wavelength 675 ± 12.5 nm (Edmund Optics) was mounted in front of the camera objective. A linear polarizer optimized for the wavelengths 600 – 1100 nm (Thorlabs LPNIRE100-B) was used to suppress detection of surface reflection and increase the speckle contrast. The light beam from the source was linearly polarized and the polarization was partly lost in the single mode optical fiber and engineered diffuser. By rotating the diffuser to a certain angle at which the specular reflection was minimized, we ensured that the direction of laser light polarization is perpendicular to the employed detection polarizer.

2.2. Motorized translational-rotational stage

A translation stage was driven by a DC motor (Faulhaber DC-Minimotor). This system was designed such that a range of continuous speeds of 0 – 10 mm/s can be realized. In our experiments, the distance from the camera to the phantom surfaces was set to 20 cm. A motorized precision rotation stage (Thorlabs PRM1/M28) was connected between the handheld LSCI probe and a vertical bar mounted on the translational panel. This stage was controlled via a brushed DC servo motor unit (Thorlabs KDC101) to enable rotational speeds of up to 25 °/s using the software Kinesis.
Fig. 1. Experimental setup. (a) Handheld LSCI system. (b) Handheld LSCI measurement and positioning. 1: optical fiber; 2: monochromatic camera; 3: color camera; 4: engineered diffuser; 5 and 6: camera objectives; 7: bandpass filter and linear polarizer; 8: panel and grip; 9: handheld LSCI system including the EM-tracker positioning sensor; 10: table-field generator EM-tracker; 11: Delrin plate.

2.3. EM-tracking system

We used an NDI Aurora table top field generator as localization device [15] in order to measure the movements of the LSCI system as translational, tilt and on-surface speeds during the handheld experiments. A six degrees of freedom sensor with root-mean-square accuracy of 0.8 mm and 0.7 ° respectively for position and orientation was installed on the handheld LSCI probe. Location of the probe with a rate of 40 Hz is sensed via inducing small currents in the sensor by altering the electromagnetic field produced by the Field Generator. The positioning accuracy depends on the distance between the sensor and the Field Generator. Based on several experiments in our setup, for a distance of 20 cm with the sensor placed perpendicular to the Field Generator, the highest accuracy was obtained. The output data from the system included quaternion and three-dimensional position matrices. The algorithm to convert quaternion \((Q_0, Q_1, Q_2, Q_3)\) to \((\theta, \phi, \psi)\) was written in a custom-made MATLAB R2017b program.

2.4. Data analysis

2.4.1. Speckle contrast

The speckle contrast is defined as [16];

\[
C = \frac{\sigma_s}{\bar{I}_s},
\]

where \(\sigma_s\) and \(\bar{I}_s\) represent the standard deviation and the mean values of the pixel intensities within a captured speckle frame. The speckle contrast is globally calculated for each frame; therefore, there is one speckle contrast value per frame. To obtain a decent statistical averaging, the region of interest (ROI) was chosen to be 150 × 150 pixels. This ROI was selected from the center part of the camera sensor array. Since imaging and processing time were not in the scope of this work, we directly calculated the speckle contrast values of sequential frames using Eq. (1) and form a profile such as those shown in Fig. 3.

2.4.2. Motion vector

The algorithm of mapping the six-dimensional displacement of the probe (translations and rotations) into the two-dimensional displacements and tilts of the beam on the reference surface per two consecutive samples is described here. The purpose is to evaluate how much laser beam translated and tilted on the surface due to the applied movements in a typical handheld measurement. We first recorded the location of the sensor tip in three-dimensional space \((t_x, t_y, t_z)\)
in millimeter per acquisition. Its instantaneous rotation along each axis \((r_x, r_y, r_z)\) in degrees was also the input to our algorithm (Fig. 2(a)). The orientation of the rotational vectors along each axis follows the right-hand rule. Assume that at times \(t_0\) and \(t_0 + \Delta T\), the sensor is placed from location \(P_1\) to \(P_2\) with arbitrary rotations. We split the surface displacement of the beam on \(xy\) plane into \(x\) and \(y\) vectors. On the \(x\)-direction (Fig. 2(b)), the pair of \((x_1, x_2)\) represents the locations from a reference point where the pair of \((z_1, z_2)\) corresponds to the heights at \(P_1\) and \(P_2\), respectively. In this case, the displacement of the beam from a perpendicular point due to the rotations can be calculated as:

\[
\begin{align*}
\Delta x_{\phi_1} &= |z_1| \tan \phi_1, \\
\Delta x_{\phi_2} &= |z_2| \tan \phi_2,
\end{align*}
\]  

(2)

where the pair of \((\phi_1, \phi_2)\) are the corresponding pitches. Therefore, the total beam displacement on the \(x\)-direction can be written as:

\[
\Delta T_x = x_2 + \Delta x_{\phi_2} - (x_1 + \Delta x_{\phi_1}).
\]  

(3)

In this way, the projected displacement on the \(x\)-axis is obtained taking to the account the change in (1) height; (2) location on \(x\)-direction; and (3) rotation around \(y\)-axis (pitch). Similarly, on the \(y\)-direction (Fig. 2(c)), one can write:

\[
\begin{align*}
\Delta y_{\theta_1} &= |z_1| \tan \theta_1, \\
\Delta y_{\theta_2} &= |z_2| \tan \theta_2,
\end{align*}
\]  

(4)

where the pair of \((y_1, y_2)\) stands for the associated locations from a reference point and the pair of \((\theta_1, \theta_2)\) are the corresponding rolls. The total beam displacement on the \(y\)-direction is:

\[
\Delta T_y = y_2 - \Delta y_{\theta_2} - (y_1 - \Delta y_{\theta_1}).
\]  

(5)

Here we have taken into account the change in (1) height; (2) location on \(y\)-direction; and (3) rotation around \(x\)-axis (roll). Thus, the magnitude of surface motion vector \((\Delta T_x, \Delta T_y)\) is defined as:

\[
\sqrt{\Delta T_x^2 + \Delta T_y^2}.
\]

This parameter times the sampling rate of the positioning device forms the temporal profile of on-surface beam speed in unit distance per unit time.

This way of calculating on-surface beam speed in a handheld measurement depends on the height and combines the influence of applied translations and rotations. Hence, to make it independent of the distance and distinguish between translational and tilt speeds, we consider the translational speed as:

\[
\frac{\sqrt{(x_2 - x_1)^2 + (y_2 - y_1)^2}}{\Delta t}.
\]  

(6)

And the instantaneous tilt angle (refers to as tilt of wavefronts) is determined as:

\[
\gamma = \tan^{-1} \sqrt{\tan^2 \theta + \tan^2 \phi}.
\]  

(7)

In a similar way, the tilt speed can be calculated by time derivation of the aforementioned tilt angle. The tilt angle is a measure of rotations along \(x\) and \(y\) axes which causes the wavefront tilting. The rotation around the beam axis (i.e. \(z\) axis) is not considered to cause a wavefront tilt. However, it causes a nonuniform translation which is considered negligible in this work.

2.5. Handheld LSCI measurement protocol

Ten healthy subjects with a normal ability of holding the LSCI probe still (e.g. without hand tremor disease) participated in the study. The purpose was to measure the amount of movements
Fig. 2. Mapping three-dimensional (3D) movements of probe to the laser beam displacement on the surface. Schematic diagram of data analysis for six degrees of freedom motion sensor has been shown. (a) 3D coordination system defined as a set of translational and rotational vectors. Solid arrows: translational vectors; dashed curved arrows: rotational vectors; \( t_x \): surge; \( t_y \): sway; \( t_z \): heave; \( r_x \): roll; \( r_y \): pitch; \( r_z \): yaw. Movement of positioning probe during two consecutive data points \((P_1 \text{ and } P_2)\) and the corresponding displacement of laser beam on x-direction (b) and y-direction (c). The pair of \((\Delta T_x, \Delta T_y)\) indicates the total displacement on the xy plane.

in a typical handheld operation. Therefore, subjects were asked to avoid any over-concentration for reducing the movements. After the start of each measurement, the probe was mounted on the table for 15 seconds to make the baseline. Then, it was lifted slowly and kept still for 45 s. To make sure that the time interval during which the probe was lifted was not included in the data analysis, the last 40 s of each measurement was accounted for the effective handheld measurement. During the handheld operation, subjects stood in front of the table top Field Generator in a relaxed manner with arm bent at elbow at 90° (see Fig. 1(b)). To prevent metal artefacts, subjects were asked to remove any metal-made wearables. In addition, the approximate distance from the front side of the handheld probe to the Delrin’s surface was kept at 20 cm to minimize the noise level of the EM-tracker’s signal. Here, the metal artefact is referred to as interference of the electromagnetic fields caused by metal objects located close to the positioning sensor [17].

2.6. In-vitro static phantoms

We made four 3D printed molds with Polylactic Acid (PLA) material, each of dimensions 195 × 60 × 14 mm³ in which agar phantoms were cast. To make the phantom static (reduce the Brownian motion) a stock solution of demi-water with 1% agar powder (Sigma A7921-500G) was prepared. Using the spectrophotometer of wavelength [300 – 1100 nm] (Shimadzu UV-2600) the absorption coefficient of Ecoline 700 ink (Talens) was measured as 24.6 mm⁻¹ at the operating wavelength of 671 nm. The Ecoline 700 was added when the stock solution had cooled down to around 60 °C. We used Intralipid 20% (Fresenius Kabi Nederland BV) for making the phantoms optically scattering. Assuming the reduced scattering coefficient of Intralipid as 26 mm⁻¹ at the operating wavelength of 671 nm[18], the molds were poured with 3.7, 7.7, 11.5 and 15.4 vol%. Then, the phantoms were kept for two hours to reach room temperature. To realize a high scattering media, a black metal plate was painted with Chalk spray (Vintage) of color ultra matte. The sample called Delrin was of Polyoxymethylene material. The absorption coefficients of the agar phantoms were chosen to be the same and equal to 0.01 mm⁻¹ while each having different reduced scattering coefficient, namely 1, 2, 3 and 4 mm⁻¹ to cover the scattering properties of human tissue. These optical property values were adapted from Lister et al.[19] for the operating wavelength of 671 nm.
2.7. In-vivo measurements

Translational speeds from 0 to 10 mm/s in 3.3 s were applied to the LSCI system and the speckle contrast was measured on a window of 150 × 150 pixels with the system facing on the forearms of test subjects. The distance from the camera sensor to the skin surface was set to 20 cm. The camera operated at 40 Hz acquisition rate with an exposure time of 25 ms and 0 dB gain. Two healthy test subjects participated in the study including 2 phases of (1) measuring on a skin area on the forearm with normal perfusion level and (2) 15 minutes after application of 0.2 ml vasodilating cream (60 gr Midalgan cream extra warm, Qualiphar, Meppel, The Netherlands) on an area of 20 × 5 cm². Each phase consisted of a 3 s baseline measurement during which the LSCI system was kept still. Then the system started to move along the forearm. Subjects were asked to breath normally and to keep their arm still during the measurements.

3. Results

3.1. Speckle contrast due to controlled translation and tilting for various scattering properties

To investigate the contribution of translation and tilt of wavefronts on the speckle contrast drop, two independent experiments were designed. For each experiment, static phantoms of various optical properties were used as diffuse media. The speckle size for this dataset was measured as 3 × 3 pixels. With a choice of ROI size of 150 × 150 pixels, there will be around 2500 samples per frame which leads to a statistically reliable calculation of speckle contrast.

For the translation study, the probe was faced perpendicular to the object surface and linear translational speed was applied to the system. Figure 3(a) depicts the measured speckle contrast in terms of the applied translational speeds. For the same applied speed, the less scattering media tend to cause larger drop in the speckle contrast. Visualization 1, Visualization 2, and Visualization 3 illustrate speckle patterns and the corresponding contrast versus applied speed for matte, Delrin and the phantom of $\mu'_s = 1$ mm⁻¹, respectively. For rotation, the LSCI system was mounted still and the phantom was rotated around a vertical axis with a constant acceleration (Fig. 3(b)). Since the center of rotation was aligned to the phantom surface, the effect of tilt of wavefronts was taken into account without translation. Similarly, for the same applied tilt speed, the less scattering media tend to cause larger drop in the speckle contrast. Visualization 4, Visualization 5, and Visualization 6 illustrate speckle patterns and the corresponding contrast versus applied tilt speed for matte, Delrin and the phantom of $\mu'_s = 1$ mm⁻¹, respectively.

Fig. 3. Dependence of speckle contrast on the applied translations and tilts for various levels of scattering. (a) Speckle contrast vs. applied translational speeds. (b) Speckle contrast vs. applied tilt speeds. $\mu'_s$: reduced scattering coefficient. Solid curves: second order exponential fit functions.
3.2. Characterization of movements of the handheld LSCI system

During a handheld measurement, the movements of the LSCI system can be described as a combination of pure translations and pure rotations which causes translations and tilts of wavefronts with respect to the scattering surface. The translation of the beam on the level of the medium’s surface will be referred to as on-surface speed. The on-surface speed is calculated by time derivation of beam positions on the scattering surface in which the system translation, rotation, and distance of the system to the medium play a role. The tilt speed accounts for tilting of wavefronts which is calculated as time derivation of the angle at which the handheld system is pointing with respect to the normal to the surface. To obtain an estimation of these parameters, several handheld measurements by various healthy test subjects were carried out.

Figure 4(a) is a so-called Lissajous graph of a representative handheld measurement on the $xy$ plane. This is an estimation for displacement of the light beam on a surface in which three episodes are of interest. (1) The baseline episode around the origin during which the system is mounted (red circle); (2) the episode where the system is being lifted (red arrow); and (3) the effective handheld measurement episode (black square). This graph shows an almost 15 mm and 20 mm total displacements on horizontal and vertical directions, respectively. Visualization 7 demonstrates a representative handheld measurement including a progressive plot of on-surface speed and speckle contrast. For the baseline area, the standard deviation values of the location signal fluctuations are $\sigma_x = 7.6 \ \mu m$, $\sigma_y = 8.7 \ \mu m$ and $\sigma_z = 9.9 \ \mu m$. Figure 4(b) depicts the total displacement of on-surface locations in time domain relative to the starting position. The absolute values of the Fourier transform of on-surface locations for baseline (noise) and effective handheld (signal) is shown in Fig. 4(c).

The rotations of the system along $x$ and $y$ axes are shown as a Lissajous plot in Fig. 4(d) which is color coded with time. Based on Eq. (7) the instantaneous tilt angle is calculated and depicted in Fig. 4(e). The absolute values of the Fourier transform of tilt angle for baseline and effective handheld measurements is also illustrated in Fig. 4(f). For the baseline area, the standard deviation of three dimensional angles are $\sigma_\theta = 6.5 \ ^\circ$, $\sigma_\phi = 13.1 \ ^\circ$ and $\sigma_\psi = 12.6 \ ^\circ$.

The absolute on-surface speed is defined as the absolute value of time derivative of on-surface location vectors and is shown in Fig. 4(g). The root mean square (RMS) value of signal to noise ratio (SNR) for this measurement is 21.5 dB, with the signals in time intervals 0 – 10 s and 20 – 60 s are considered as noise and signal, respectively. The absolute tilt speed is also obtained by time derivation of tilt angles with an SNR of 14.5 dB (Fig. 4(h)). The observed speckle contrast as a function of on-surface and tilt speeds for a sample handheld measurement is shown in Fig. 4(i) (see Visualization 8 for a better view on this graph). The average values of extracted speed elements have been summarized in Fig. 5 where the mean values for all 10 operators for translational, tilt and on-surface speeds are 0.6 cm/s, 1.1 $^\circ$/s and 0.9 cm/s, respectively.

Table 1 summarizes the relative drop in speckle contrast due to translation and tilt of wavefronts. The values are extracted from Fig. 3 with two data points from each phantom: the speckle contrast at speed zero and the speckle contrast at on-surface and tilt speeds of 0.9 cm/s and 1.1 $^\circ$/s, respectively. These on-surface and tilt speed values are the averaged values of 10 handheld measurements estimated by the EM-tracker (Fig. 5).

| $\mu'_s$ (mm$^{-1}$) | >4 (Matte) | 4 | 3 | 2 | $\approx 2$ (Delrin) | 1 |
|----------------------|-----------|---|---|---|-----------------|---|
| $\Delta C_{\text{on-surf.}}$ (%) | 63.4 | 63.9 | 67.1 | 71.9 | 72.6 | 78.2 |
| $\Delta C_{\text{tilt.}}$ (%) | 5.5 | 46.3 | 53.6 | 59.6 | 63.2 | 70.3 |
| $\Delta C_{\text{on-surf.}}/\Delta C_{\text{tilt.}}$ | 11.6 | 1.4 | 1.2 | 1.2 | 1.1 | 1.1 |
Fig. 4. Analysis of movement and speed of handheld LSCI system. Representative data of a handheld operation is shown. (a) Lissajous plot indicating the locations of the light beam on a scattering surface. Red circle: baseline measurement while the system is mounted; Red arrow: the episode during which the system is lifted; Black square: the effective handheld measurement. (b) Temporal fluctuations of on-surface locations. (c) Absolute Fourier transform of on-surface locations. Signal: effective handheld measurement; Noise: baseline measurement. (d) Lissajous plot of rotations along x and y axes shown as $\theta$ and $\phi$, respectively. (e) Temporal fluctuations of tilt angle. (f) Absolute Fourier transform of tilt angle. Temporal profiles of absolute on-surface and tilt speeds. $v$: on-surface speed; $\gamma$: tilt speed. (i) Observed speckle contrast on a Delrin plate as a function of on-surface and tilt speeds. $C$: spatial speckle contrast. (g) and (h) are still images of Visualization 7. (i) is still image of Visualization 8.
3.3. In-vivo measurements

Temporal fluctuation of speckle contrast measured on the forearm of the first test subject after the application of Midalgan is shown in Fig. 6(a). The sudden drops in the speckle contrast are due to the heartbeats which occur approximately every second. The average heart rate based on these time intervals for this test subject is calculated as 64 beats per minute (bpm). In this graph, the system starts to move at the time 4.8 s. A comparison between the observed speckle contrast vs the applied translational speed before and after application of Midalgan is shown in Fig. 6(b-c) for the first and the second test subjects, respectively. Here the skin area was approximately the same for each test subject and also the perfusion of the measured area is approximately the same at the start and the end locations. The speckle contrast level at zero speed after application of Midalgan is lower than that of normal perfusion level and this is the case for both test subjects. Moreover, the two graphs are not identical to each other.

4. Discussion

Movement artefacts during a handheld LSCI measurement are caused by tissue motions [9,11,20–23] and motions of the LSCI system [5,12,13]. The former can be caused by breathing or patient movements while the latter are generated in the wrist, elbow and shoulder, and motions due to heartbeat and breathing of the operator. In this work, we focused on the movement of the handheld LSCI system, and the overall motions which cause movement artefacts are considered as on-surface translational motions of the laser beam and tilt of wavefronts.

To generate controlled motions, the LSCI probe was installed on a motorized stage and the speckle contrast was measured during the applied movements. The experiments were carried out on static objects in order to exclude any additional speckle decorrelation source other than the applied external motions. The influence of optical properties of the medium on the speckle contrast was investigated for translation and tilting. For a given absorption level, we observed (Fig. 3) that both for translation and tilt, the medium of less scattering properties causes greater drop in the measured speckle contrast values. Hence, tissues with lower scattering levels will generate higher movement artefacts than tissues with higher scattering levels. These findings can be explained with the use of the optical Doppler shifts on a single position on the imaging plane (the camera array sensor). Here we use the fact that intensity fluctuations have frequencies equal to the differences of Doppler shifts of the incoming light, since two light beams with Doppler frequencies \( \omega_1 \) and \( \omega_2 \) on interference give an intensity fluctuation at beat frequency \( |\omega_1 - \omega_2| \).

And higher frequencies of the intensity fluctuations lead to more speckle blurring on integration.
And higher frequencies of the intensity fluctuations lead to more speckle blurring on integration (the camera array sensor). Here we use the fact that intensity fluctuations have frequencies equal (Fig. 3) that both for translation and tilt, the medium of less scattering properties causes greater detected at the center pixel for medium 1 (yellow fluence distribution) is smaller than that of greater than that of medium 2, that is of translation where the light source and the imaging system move at a direction parallel to the increase with decreasing scattering level of the medium. Figure 7(a) shows schematically the case of translation where the light source and the imaging system move at a direction parallel to the object plane with speed 

\[ \mathbf{v} = (v_x, v_y, v_z) \]

The reduced scattering coefficient of medium 1 is greater than that of medium 2, that is \( \mu'_s > \mu'_s \). Therefore, the size of the diffuse cloud of light detected at the center pixel for medium 1 (yellow fluence distribution) is smaller than that of medium 2 (red fluence distribution, partly overlapping with the yellow fluence distribution). In other words, for a given position of the diffuser in the illumination system, medium 2 causes larger variation of the incoming wave vectors \( \mathbf{k}_i = (k_{i_x}, k_{i_y}, k_{i_z}) \) since light imaged to a single point originates from light injected into the medium over a larger area. For both medium 1 and medium 2, the wave vectors of the detected light \( \mathbf{k}_d = (k_{d_x}, k_{d_y}, k_{d_z}) \) are the same since they are defined by the detection lens aperture. For solid body translation the Doppler shift is [24]

\[ \omega = \mathbf{v} \cdot (\mathbf{k}_d - \mathbf{k}_i), \]

and hence does not depend on the photon paths inside the medium: it is only the incoming and outgoing wave vectors that contribute to the Doppler shift. The greater the variation of incoming wave vectors \( \mathbf{k}_i \) of the light that is eventually detected, the greater the frequencies of

![Fig. 6. In-vivo measurements of speckle contrast. (a) Temporal fluctuation of speckle contrast driven by heartbeat pattern. Red open circles indicate manually chosen sudden drops. During the first 4.8 s, the LSCI system is still. The vertical dashed line indicates the moment the system starts to move. (b-c) Speckle contrast vs translational speeds for test subjects 1 and 2, respectively. ‘Normal’ refers to the skin area with normal perfusion level. ‘Midalgan’ refers to same skin area after 15 minutes application of Midalgan. Solid curves in (b-c): second order exponential fit functions.](image-url)
intensity fluctuations, leading to a lower speckle contrast. And this is the case for the lower scattering level, since the variation in $\vec{k}_i$ is larger for the wider diffuse light distribution associated with lower scattering media. We explore the consequences for a diffuser at 200 mm from the medium’s surface. For the difference in $\vec{k}_i$ for light originating from a single point in the diffuser but entering the medium at 0.5 mm left or right from the point on the tissue surface conjugated with the point on the imaging plane, using Eq. (8) we find a difference in Doppler shifts of approximately 10 Hz per mm/s of translational speed, leading to intensity fluctuations of 10 Hz per mm/s of speed. For the applied integration time of 25 ms an intensity fluctuation at this will lead to a significant drop in speckle contrast. Hence, our explanation makes it credible that, for a given translational speed, scattering level variations leading to variations in the dimensions of the ‘diffuse cloud’ in the millimeter range lead to significant variations of the speckle contrast. This makes our Doppler based explanation a plausible one.

Fig. 7. Schematic diagram of wave vectors in a reflection geometry. (a) Translational displacement. (b) Object rotation. 1: Diffuse distribution of light eventually imaged in a single point on the image plane, for medium of higher scattering level; 2: Idem, medium of lower scattering level; $\vec{k}_1$ and $\vec{k}_2$ illustrate wave vectors of incoming beams in 1 and 2, respectively. $\vec{k}_s$: wave vectors of light to be imaged on a single point of the image plane. The curved solid lines in the fluence distributions show random photon paths. The dashed arrows show the movement directions.

A similar statement can be made for the case of tilt of wavefronts in the experiment of object rotation demonstrated in Fig. 7(b), although in that case Eq. (8) does not hold, and the Doppler shift will depend on the details of the photon paths. The Doppler shift will be larger for light entering the medium further away from the point of detection, and for light penetrating deeper into the medium before returning. For tilting, the Doppler shift of a given photon path now is the result of a variation of the optical path length through air plus the Doppler shift built up on scattering events in the medium. Also here the argument holds that a lower scattering leads to a larger variation of Doppler shifts, hence higher intensity fluctuation frequencies, and hence a lower speckle contrast. An extreme case is that of a very high scattering medium such as a thin layer of matte paint. Here the speckle contrast tends to be considerably less influenced by the applied tilt speed (see Fig. 3(b)) in comparison to the other phantoms. In this case, the speckle patterns are mainly formed by reflection of a rough surface due to the fact that the thickness of the painted surface is a couple of microns which is comparable with the wavelength of the light source. Hence light falling on a certain location of the camera sensor is originating from locally impinging light in the conjugated point on the tissue, without sideward diffusion.
We employed an EM-tracker to estimate movements of a handheld LSCI system. During the handheld measurements, the metal objects were avoided to be in the range of the electromagnetic field in order to prevent metal artefacts. To assess the accuracy and precision of this system, we first measured the discretized errors of position and angle dataset as 10 µm and 11 m°, respectively. The discretized error is referred to as uniform steps we observed in our raw position and angle dataset which is not necessarily the minimum nonzero difference between two consecutive data points. One of the sources of this error is analog to digital conversion of the acquired signals. Then, to check the stability of raw signals, the standard deviations of location ($\sigma_x = 7.6$ µm, $\sigma_y = 8.7$ µm and $\sigma_z = 9.9$ µm) and angle ($\sigma_\theta = 6.5$ m°, $\sigma_\phi = 13.1$ m° and $\sigma_\psi = 12.6$ m°) data for the baseline episode were determined. Second, we presented the spectra of estimated on-surface locations and tilt angles (Fig. 4(c,f)). In both graphs, the effective handheld data have greater values than the baselines. The spectra of on-surface locations gradually drops to approximately 20 dB of its maximum value at the frequency of 10 Hz and then levels off. The frequency at which the tilt angles reach a 20 dB decay and remain at the same level is approximately 15 Hz. Third, we calculated the SNR of RMS on-surface (21.5 dB) and tilt (14.5 dB) speeds (Fig. 4(g,h)). Note that the time derivative per two consecutive samples increases the noise of speed data rather than that of instantaneous locations and angles.

Worth noting that speckle contrasts at speed zero per each phantom shown in Fig. 3 are not exactly the same. This might be due to mode-hopping of the light source, external sources of vibrations or the fact that observed speckle patterns are not fully developed in practice. However, this issue does not affect the results since the relative speckle contrast is considered. The ratios of two contrast drop percentages suggest that for typical handheld motions (1) translation of the beam on the surface of the medium plays a dominant role over wavefront tilting in decreasing speckle contrast, and (2) the dominance of translation over tilt tends to decrease as the medium becomes less scattering.

We have included the in-vivo measurements in order to have a practical view on the issue of movement artefacts. Due to the movements of the red blood cells (RBCs) in the microcirculatory blood vessels, the speckle contrast values at zero speed are lower than that of static objects. Figure 6(a) shows a temporal speckle contrast profile driven by heart beat pattern. The dynamic range of the fluctuations tends to decrease as the speed increases demonstrating the underlying non-linear response of the speckle contrast to the applied speed depicted in Fig. 3(a). The average relative speckle contrast dips at systole when the system is still is calculated as 0.07, whereas after the system starts to move, they become 0.06, 0.03, 0.02 and 0.02, where the lower values correspond to the higher speeds. The speckle contrast drop percentages from the zero speed to that of at 0.9 cm/s are 54.2% and 36% before and after application of Midalgan, respectively, for the first test subject shown in Fig. 6(b). These values for the second test subject shown in Fig. 6(c) are respectively 55% and 42.1%. This means that for the same applied speed, the relative change in the speckle contrast decrease as the skin perfusion increases and these relative changes vary per subject due to the difference in perfusion levels.

In conclusion, the magnitude of movement artefacts depends on the optical properties of the medium. We have explicitly shown that movement artefacts increase with decreasing scattering coefficient and based on our explanation it is to be expected that movement artefacts will also increase with decreasing absorption. This means that in a handheld LSCI measurement, movement artefact correction based on a simple look-up table approach is not feasible without knowledge of the applied translations and tilts and also optical properties of the tissue. The results of this study can be used for future development of strategies to suppress movement artefacts. Our findings enable the formulation of specifications on the performance of suppression or correction methods, be it based on hardware or software methods, since this study provides values of wavefront tilt and beam translation speeds and their temporal dynamics, and their consequences for various types of tissue.
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