Published in final edited form as:

Quant Infrared Thermogr J. 2012 January 1; 9(2): . doi:10.1080/17686733.2012.737157.

Reconstruction of Thermographic Signals to Map Perforator Vessels in Humans

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Abstract

Thermal representations on the surface of a human forearm of underlying perforator vessels have previously been mapped via recovery-enhanced infrared imaging, which is performed as skin blood flow recovers to baseline levels following cooling of the forearm. We noted that the same vessels could also be observed during reactive hyperaemia tests after complete 5-min occlusion of the forearm by an inflatable cuff. However, not all subjects showed vessels with acceptable contrast. Therefore, we applied a thermographic signal reconstruction algorithm to reactive hyperaemia testing, which substantially enhanced signal-to-noise ratios between perforator vessels and their surroundings, thereby enabling their mapping with higher accuracy and a shorter occlusion period.

Keywords

thermographic signal reconstruction; infrared imaging; reactive hyperemia; perforator vessels; blood flow regulation; skin microcirculation

1. Introduction

Visualization of the vessels underlying the skin is important for assessment of peripheral vascular disease, the results of skin reconstructive surgery, wound and ulcer management, and (via perforator vessels) aspects of muscular function. Several non-invasive optical imaging methods, such as capillaroscopy, laser Doppler, laser speckle contrast, and orthogonal polarization spectral imaging, have been applied by various groups to quantify functional aspects of the skin vasculature. These methods are sensitive to vasculature plexuses up to ~0.2 mm deep from the skin surface. Because of thermo-conduction of the skin and thermo-convection of the blood via perforator vessels, infrared (IR) imaging, however, can assess vascular plexuses at an estimated depth of 10 mm below the skin surface [1]. Recently IR imaging of a subject's forearm was utilized to visualize skin angioarchitecture in real time in response to infusion of vasodilators[2].

Liu *et al.* [3] showed that, by using IR imaging during 5 minutes of complete arterial occlusion by an inflatable cuff to block blood flow in a human forearm, patterns characteristic to microvascular (MV) perforator vessels, conduit vessels, as well as skin

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areas without IR-detectable vessels (SWV) can be distinguished on the skin. However, 5-min long ischemia brings predictable discomfort and may not always be applicable to subjects with various health conditions or to pediatric patients. In this study, we attempt to map angioarchitecture of the skin reliably by evaluating temperature deviations following 1, 4, and 5 minutes of complete arterial occlusion of the forearm by an inflatable cuff.

To enhance the contrast between MV and SWV, a thermographic signal reconstruction (TSR) approach, commonly used for Non Destructive Evaluation (NDE) of solid materials [4], was adapted. Originally TSR was designed for detecting subsurface defects or fatigue cracks in aerospace structures by applying a short pulse of heat to the test sample externally and then analyzing temperature data recorded during cooling. An observed phenomenon of internal forearm heating immediately after the end of cuff-induced temporary ischemia and subsequent return to baseline temperature was utilized in our procedures to locate perforator vessels in the human forearm. By shortening the occlusion time from 5 minutes to 1 minute, a substantial contrast enhancement between vessels and the surrounding was achieved.

2. Methodology

2.1. Data collection

All subjects (n=5) consented to complete occlusion testing of one of their forearms. Multiple IR images of the dorsal aspects of their forearms were collected during a 10-min baseline (BL) period, 5 minutes of arterial occlusion of the upper arm (OC) with a blood pressure cuff inflated to a suprasystolic level, and a 10-min post occlusion (PO) period. To evaluate the impact of occlusion duration on the contrast between skin surface angioarchitecture and the surrounding tissues during reperfusion of the forearm, the upper arm was also occluded for 1 minute and for 4 minutes in a pilot study (n=1). A calibrated IR camera (Lockheed Martin, USA) with 0.015°C temperature sensitivity (3.0–5.0 μm wavelength, 320 × 256 pixels per image, 14 bits/pixel) was positioned directly above the dorsal aspect of a subject's forearm at a distance of ~90 cm. The fingers of the hand were gently taped down to reduce motion of the forearm, and IR images (~3000 frames) were collected continuously at a rate of 2 Hz.

Before data processing, image registration with rigid-body transformation algorithms [5] was applied over the collected frames for motion correction, utilizing built-in functions of the Statistical Parametric Mapping Toolbox (<http://www.fil.ion.ucl.ac.uk/spm/>) of MATLAB 7.6 (MathWorks, Inc., USA). The image collected immediately after cuff deflation served as a reference to align the remaining images. As a result, the average misalignment of the forearm was less than one pixel.

2.2. TSR and data processing

The TSR approach [3] assumes that a given thermal energy will follow the one-dimensional heat diffusion equation (left side of Eq. (1)) when traveling inside the test sample, until it encounters a change in thermophysical properties, e.g. a wall or a defect:

$$\frac{\partial^2 T}{\partial z^2} - \frac{1}{\alpha} \frac{\partial T}{\partial t} = 0 \Rightarrow T = \frac{Q}{e \sqrt{\pi t}} \quad (1)$$

$$\Rightarrow \ln(T) = \ln\left(\frac{Q}{e}\right) - \frac{1}{2} \ln(\pi t) \quad (2)$$

where T is the temperature, z is the direction perpendicular to the surface of the test sample, α is the thermal diffusivity, Q is the input energy, and e is the thermal effusivity.

The right side of Eq. (1) is the solution of the diffusion equation. By taking the *log* of both sides of Eq. (1), the time variable t is separated from the energy variable Q and the material characteristic e (Eq. (2)). The left side of Eq. (2), $\ln[T(t)]$, is then approximated by polynomial fitting. Each collected image reflects a combination of IR radiation emitted from the target, from which we can infer its physical properties, as well as IR radiation from background or environmental sources. The fitting calculated for each pixel in TSR includes both contributions. However, the logarithmic derivatives make it possible to separate target emission and background artefacts, allowing our application to distinguish vasculature with uneven, hyperaemia-related heating from other skin tissue. The derivative of $\ln[T(t)]$ can be calculated analytically after the polynomial fitting.

Three distinctions are noteworthy. *First*, the given energy Q in most NDE applications is controllable and can be modelled, while in our application Q is an endogenous variable and depends on vessel size, resistance, and elasticity, as well as the density of microcirculation and vascular autoregulation parameters. *Second*, as described in [3], one plane of the test material was considered to be heated uniformly. The direction of heat wave going through the material is usually simplified as being perpendicular to the incident plane. In our case, after pressure in the forearm cuff was released, the heat wave associated with warm blood traveled via the artery and arterioles towards fingers, via perforators and arterial-venous anastomoses towards skin, and finally, in the latest stage of reperfusion, via the veins from the fingers back to the torso [2]. Since we have to deal with multiple heating sources at different depths, heterogeneous spatial distribution, and heat waves with time-dependent directions, one might conclude that a step-heating approximation can be considered primary for perforator vessels and only for the first moments of reperfusion. *Third*, the thermal effusivity of a test sample that undergoes NDE can be measured prior to the test. For skin temperature measurements, however, this thermal property is difficult to assess due to the complex and heterogeneous structure of human skin. Therefore, a single value cannot define the thermal effusivity of human skin. Although knowledge of Q and e can be utilized to estimate depth for the defects in NDE applications, such information was either missing or not applicable to estimating the depth of vessels in our study.

To adapt the TSR approach, the following steps were implemented during off-line analysis with MATLAB software. The term ‘temporal profile’ was defined as a one-dimensional curve indicating the change of y values along with the time (x -axis).

1. All collected IR images (~3000 frames) were assigned to a single 3D image cube and a mask for pixels containing only the human hand was applied to all images.
2. The Temperature vs. time (T vs. t) profiles were extracted from the entire image cube for each pixel related to the human hand.
3. From each T vs. t profile, only the first 50 s for the PO period of the temporal profile (T vs. t_{50}) containing temperature changes from ~ 100 images were segmented as the input for further TSR.
4. T vs. t_{50} profiles were converted to a logarithmic scale, $\ln(T)$ vs. $\ln(t_{50})$.
5. Each $\ln(T)$ vs. $\ln(t_{50})$ profile computed for each pixel was reconstructed by a 5th order polynomial $f\ln[T(t_{50})]$ fitting.
6. The 1st order derivative was calculated from the reconstructed temporal profile, $\exp\{f\ln[T(t_{50})]\}$, and was considered as a “derivative temporal profile”.
7. A total of 100 derivative images of the human hand were synthesized, in which the initial temperature value of each pixel was replaced with a value from the corresponding “derivative temporal profile”.

8. A visual inspection of 100 synthesized images was employed to choose the frame with maximum signal to noise ratio between MV and SWV.

To quantify responsiveness of different vascular territories of human skin under occlusion tests of different durations, a temperature-derived reactive hyperemia index (RHI_T) was calculated for each pixel after TSR as the maximum of the 1st order derivative within the first 50 seconds after the occlusion. To map RHI_T , images for 1, 4, and 5- min occlusions were computed, where each pixel with an initial temperature value was replaced with an RHI_T value within the range of -9.3 and 722.8 arbitrary units.

2.3. Quantification of Contrast Enhancement

In this study, we considered image contrast to be “enhanced” when all vessels in an original image could be discerned more easily after an image enhancement procedure was applied. Such enhancement is due to the larger contrast between the vessels and their surroundings. To quantify the contrast in a way convenient for human visual perception, Beghdadi and Negrate [6] combined calculation of the mean pixel intensities of the edges with the contrast computation. The method was adapted in our study as follows:

1. To enhance the MV and also avoid the presence of hairs being picked up by the following edge detection, we convolved the image I by a virtual disk-averaging filter with radius 7 pixels.
2. Performed edge detection on the image I to emphasize the boundaries. The result is denoted as L .
3. For each pixel L_{ij} , a $m \times m$ window W_{ij} centered at L_{ij} was defined, and the mean edge intensity $\bar{E}_{ij}$ was calculated as

$$\bar{E}_{ij} = \left(\sum_{i,j \in W_{ij}} L_{ij} \cdot I_{ij} \right) / \sum_{i,j \in W_{ij}} L_{ij} \quad (3)$$

1. The contrast of pixel I_{ij} , $C_{ij} \in [0,1]$, was calculated as

$$C_{ij} = \frac{|I_{ij} - \bar{E}_{ij}|}{|I_{ij} + \bar{E}_{ij}|} \quad (4)$$

In the edge detection of step 2, a Sobel operator, and a Laplacian of Gaussian operator were tested. The latter had better detection performance when the size and σ values were set to 7 and 0.5, respectively, based on the tortuosity of the MV patterns. In step 3 the window was set to 41×41 pixels. A larger window size m has been suggested [6] to avoid enhancing the contrast of noise. If the window size is too small, the contrast measurement will be more biased. For example, an improperly sized window can cause an image with a clear vessel pattern to have a lower calculated contrast than another image with a vague vessel pattern.

3. Results

Figure 1 shows unprocessed IR images of a human hand with the clearest MV perforators (bright spots) from two normal subjects (50 and 43 year old females) during reactive hyperemia following 5-min of occlusion. Even though conduit vessels are well visualized in both images, the image in Figure 1b has poor MV visibility compared to the image in Figure 1a.

The TSR approach was applied to subjects with poor MV visibility, such as in Figure 1b, for further MV enhancement. For clarity, only two regions of interest (ROIs) representing conduit vessels (ROI #2) and MV (ROI #1) were selected, as seen in Figure 2a. Note, only the averaged temperature profile for the ROI #2 is plotted in Figure 2b. The temperature during the BL period, the temperature decrease caused by inflation of the occlusion cuff (OC period), and the dramatic temperature increase during the first 50 s of PO period caused by cuff deflation can be recognized in the temporal profile. The image in Figure 2c was selected among 100 synthesized images and corresponds to initial IR images collected at 6 s after the cuff deflation (time point indicated by a black arrow in Figure 2d). Black arrows on Figure 2c point to several enhanced MV spots. Figure 2d shows two derivative temporal profiles for the first 50 s of PO period, reflecting temperature deviations for the two selected ROIs.

To avoid the strong contrast from conduit vessels, the enhancement was quantified only for the area with MV, as indicated by black arrows in Figure 2c. Contrasts calculated by Eq. 4 for the same areas in Figure 2a and 2c were 0.366 and 0.761, respectively. Thus, adaptation of the TSR approach to collected IR images of a human forearm almost doubled the contrast for MV during reperfusion.

The duration of complete occlusion had an impact on contrast and MV appearance. The MV patterns on IR images with the highest contrast appeared at 27, 21.5, and 31.5 seconds within the first 50 s of PO, corresponding to 1, 4, and 5-min occlusions, respectively (Figure 3a–c). For this subject, an image collected at 27 s corresponding to 1-min occlusion did not reveal any MV patterns, whereas images corresponding to 4-min and 5-min occlusions revealed identical MV patterns. To reveal finer details on IR images, the same grey rectangle area was selected and enlarged (Figure 3d–f) with linear contrast rescaling. No MV patterns were visually detected on these images.

After TSR processing the frames at 4, 3, and 8 seconds within the t_{50} period, corresponding to 1, 4, and 5 minutes of occlusion, respectively, showed the highest contrasts of 0.613, 0.729, and 0.814 (Figures 4a–c). The same MV was recognizable in all 3 images as a bright vascular pattern seen to the right of the grey rectangle, probably reflecting arteriole involvement during hyperemic vasodilatation. The relatively dark conduit vessel pattern probably reflects venous return with a bolus of cold blood pushed from the hand digits by fresh arterial blood. After 1 minute of occlusion, a MV pattern, though of less intensity compared to after 4 and 5-min occlusions, was still detected (Figure 4a).

For the enlarged region of interest (Figures 4d–f) a pattern with three dots was identified after 5-min occlusion (Figure 4f). A pattern of two dots was recognizable in Figure 4e. Figure 4d only showed one dot at most, which was also a very weak signal. The calculated contrast values for Figs. 4d–f were 0.482, 0.591, and 0.793, respectively.

The unbalanced temperature distribution between the arm and hand (Figures 3a–c) was also alleviated after TSR processing (Figures 4a–c).

To demonstrate the effectiveness of contrast enhancement, the contrast defined in Eq. (4) was calculated for each of the images collected during the t_{50} period for both the original and the TSR-processed imaging sets. The area used for contrast calculation is marked by grey rectangles in Figure 5a and 5b. Temporal deviation of the contrast calculated for the first 50 s after 1, 4, and 5 minutes of occlusion was observed (Figures 5c–e). Overall, contrast improvement was evident for the TSR-processed imaging set (dash-dot line) compared to the original imaging set (solid line). The highest contrast, with values of 0.613, 0.729, and 0.814, occurred 4, 3, and 8 s after cuff deflation, respectively. The lowest

contrast, with values of 0.422, 0.461, 0.463 for respective 1, 4, and 5-min occlusions, occurred at 2 s. The highest contrast for the TSR-processed imaging set was observed 3 – 9 s after cuff deflation, which was much earlier than the 21 – 32 s latencies seen in the original imaging set.

Without TSR processing, a significant image enhancement (more distinction between MV and surrounding skin) was observed for 4-min and 5-min occlusions only. Differences between the max and min contrast values were 0.032, 0.172, and 0.154 for the 1, 4, and 5-min occlusion tests, respectively. After TSR processing, the max-min differences of the three tests were 0.192, 0.268, and 0.351. Although both parameters – contrast value and max-min difference – for the TSR-processed imaging set increased with duration of occlusion, the relationship between the latency to the contrast max (4th, 3rd, and, 8th seconds, according to Figure 5c~e) and occlusion duration (0.192, 0.268, and 0.351) was not linear.

Depending on the image selected within the first 50 s after cuff deflation, the contrast between vasculature and the surrounding (i.e. vascular features) was enhanced or diminished (Figure 6).

To explore how the temporal change of the 1st order derivative for different ROIs affects vascular imaging, 4 ROIs representing perforator vessel (small bright dots), MV (irregular larger bright area), conduit vessel (stripe), and skin tissue were selected (Figure 7a). Their derivative responses under different occlusion times were plotted in Figure 7b–d. For 1-min occlusion, ROIs for the perforator vessel and skin tissue had similar temporal changes of their derivatives. However, these signals diverged after longer occlusion times (Figure 7c,d). The derivatives for MV and conduit vessel ROIs showed more dramatic temporal changes compared to the other two ROIs. According to Figure 7b–d, the derivative value for MV increased immediately after occlusion. After reaching the maximum at 7, 10, and 16 seconds after respective 1, 4, 5-min occlusions, the derivative value started to decrease. Change of derivative values over time, particularly for conduit vessels, were noticeably different among occlusion durations.

To demonstrate that RHI_T varies in different skin areas, maps were generated for the 3 occlusion durations. Figure 8 shows the images computed for 1, 4, and 5-min occlusions. Durations of occlusion showed substantial impact on the heterogeneity of RHI_T . The conduit vessel and some MV (indicated by white arrows) showed relatively higher RHI_T than the surrounding.

RHI_T images for the 4 and 5-min occlusions (Figure 8b,c) showed much more details of vasculature than the 1-min occlusion RHI_T image (Figure 8a). According to the intensity bar scale and the corresponding values, MV had higher RHI_T than the skin tissue. Without the interference of arm hair, this difference might be more pronounced. Only these 2 classes were distinguishable in Figure 8b and 8c. Unlike Figure 4, the conduit vessels were not predominantly darker than the skin tissue. Rather, they were generally thinner and could only be recognized based on their diameter. Part of the skin was even darker than the conduit vessels.

4. Discussion and Conclusion

Skin vasculature is of increasing interest to clinicians and researchers seeking a window to assess microvasculature resistance vessels in the skin, subcutaneous microcirculation in fat depots, and (via perforator vessels) muscle microcirculation.

Visualization of individual angioarchitectures within the skin's microvascular bed and their functional characterization are the motivation of this work. Because arterial blood at core temperature is warmer than the exposed skin surface, which has been cooled by room air, local blood flow can be used as an endogenous, natural thermal contrast agent for IR monitoring. Examples of such images were previously shown for exposed brain [7, 8] and kidneys during surgical procedures [9], as well as for human forearm [2].

In this pilot study, we demonstrated that enhancement of subcutaneous functional angioarchitecture, acquired by IR imaging, requires the following two conceptual modifications of the original TSR method:

- The heat source is internal and localized. In our case, the step-heating approximation can be applied primary for perforator vessels, but only for the first moments of reperfusion.
- The TSR algorithms should be applied to a selected time window in the rewarming process. In our case, we chose the 50 s time window during the PO period in which the temperature of the hand was sharply increasing.

Application of TSR to IR imaging revealed functional angioarchitecture of subcutaneous MV after reperfusion of the human forearm. Functional irregularities (and autoregulation) of blood supply--such as differences in local blood pressure, vascular resistance, tone, and elasticity--for different vascular territories of the forearm, and structural factors--such as differences between vascular patterns--might have caused the observed variation in temporal delays for vessel reactivity to reperfusion.

A substantial improvement in signal-to-noise ratio between the perforator vessels and the surrounding tissues was achieved (Figure 4f) by using the TSR approach. In addition, shortening the occlusion time was found to be desirable, since 5-min occlusion causes numbness and might be unacceptable for children and several other patient groups. As shown in this article, without TSR visibility of MV on IR images was diminished as the duration of occlusion (OC period) was shortened. Application of the TSR algorithm, however, to a short-duration occlusion test yielded acceptable contrast between vasculature and the surroundings.

RHI has been widely used as a measure to quantify peripheral endothelial function [10]. We showed that RHI_T varies by site and different occlusion times. Utilizing occlusion testing paradigms and temperature measurements during forearm reperfusion, we attempted to define a new marker of endothelial dysfunction -- RHI_T . As shown in this article, MV had higher RHI_T compared to other areas of the skin. The MV pattern in the calculated maps of RHI_T was shown to be relatively invariant to occlusion durations and can be utilized to identify vasculature territories with low resistance to blood flow during hyperaemia. These maps may also be used to guide the placement of detectors for blood flow and oxygenation.

Overall, the application of both algorithms --TSR and RHI_T --to original IR imaging sets was capable of enhancing image contrast, allowing for non-invasive visualization of subcutaneous functional angioarchitecture in the human forearm.

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Fig. 1.
Vasculature maps of two subjects with (a) higher MV contrast, and (b) lower MV contrast.

Fig. 2.

(a) Unprocessed, low MV visibility IR image (see Figure 1b) with two selected ROIs representing conduit vessels (ROI #2) and MV perforator vessels (ROI #1); (b) Temperature vs. time (T vs. t) profile extracted from the entire imaging cube and averaged over the ROI #2; (c) one of 100 synthesized TSR-based images for the first 50 s of PO period; and (d) two derivative temporal profiles for the first 50 s of PO period (T vs. t₅₀) from TSR approach, reflecting deviations of temperature change for two selected ROIs. The notation 'AU' in y-axis is the acronym of 'arbitrary units' from the direct reading of the IR camera.

Fig. 3.

(a) MV pattern with the highest contrast (0.313) appeared at 27 s after 1-min occlusion; (b) MV pattern with the highest contrast (0.461) appeared at 21.5 s after 4-min occlusion; (c) MV pattern with the highest contrast (0.386) appeared at 31.5 s after 5-min occlusion; (d), (e), and (f) Enlarged area marked by the grey rectangle in (a), (b), and (c), which showed no sign of a MV pattern. Note, contrast values were calculated using the ROI marked in Figure 3a.

Fig. 4.

(a) MV pattern with the highest contrast of 0.613 appeared 4 s after 1-min occlusion; (b) MV pattern with the highest contrast of 0.729 appeared 3 s after 4-min occlusion; (c) MV pattern with the highest contrast of 0.814 appeared 8 s after 5-min occlusion; (d), (e), and (f) Rescaled ROIs (grey rectangles in a, b, and c) showed a distinguishable MV pattern as dots. Note, the contrast values were calculated using the ROI defined in Figure 3a.

Fig. 5.

(a) IR image and (b) TSR-processed IR image with the maximum contrast collected after 4-min occlusion. The same ROI (grey rectangle on upper left) on both images was chosen to calculate the contrasts (Eq. (4)) for original and TSR-processed imaging sets. Temporal deviation of the contrast during the first 50 s after cuff deflation was extracted from original (solid line) and TSR-processed (dash-dot line) post-occlusion imaging sets for (c) 1-min, (d) 4-min, and (e) 5-min occlusions.

Fig. 6. Post-occlusion images of a human forearm calculated as a 1st derivative for the 4-min occlusion test. Representative set of six images (a–f) collected at 0.5, 2.5, 7, 19.5, 32.5, and 50 seconds after cuff deflation. Values for the 1st derivative are shown on a horizontal bar scale for each image.

Fig. 7.

(a) The location of 4 ROIs; (b–d) The derivative temporal profiles calculated for 4 ROIs after 1, 4, and 5-min occlusion (from left to right). The black solid line (curve 1) represents conduit vessel; the grey dash-dot line (curve 2) represents MV; the grey solid line (curve 3) represents perforator vessel; and the grey dashed line (curve 4) represents skin tissue.

Fig. 8.

The RHI maps derived from (a) 1-min occlusion experiment; (b) 4-min occlusion experiment; and (c) 5-min occlusion experiment. Because the absolute unit of the 3 data sets was used for analysis, the difference between the maximum and minimum in Figure 8 was large (around 500~700). The range of the derivative in Figure 7b–d was only $[-100, 100]$ due to the selection of small ROIs.