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Video rate electrical impedance tomography of vascular changes: preclinical development

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Abstract

Peripheral vasculature disease is strongly correlated with cardiovascular-associated mortality. Monitoring circulation health, especially in the peripheral limbs, is vital to detecting clinically significant disease at a stage when it can still be addressed through medical intervention. Electrical impedance tomography (EIT) maps the electrical properties of tissues within the body and has been used to image dynamically varying physiology, including blood flow. Here, we suggest that peripheral vasculature health can be monitored with EIT by imaging the hemodynamics of peripheral vessels and the surrounding tissues during reactive hyperemia testing. An analysis based on distinguishability theory is presented that indicates that an EIT system capable of making measurements with a precision of $50 \mu\text{V}$ may be able to detect small changes in vessel size associated with variations in blood flow. An EIT system with these precision capabilities is presented that is able to collect data at frame rates exceeding 30 fps over a broad frequency range up to 10 MHz. The system's high speed imaging performance is verified through high contrast phantom experiments and through physiological imaging of induced ischemia with a human forearm. Region of interest analysis of the induced ischemia images shows a marked decrease in conductivity over time, changing at a rate of approximately $-3 \times 10^{-7} \text{ S m}^{-1} \text{ s}^{-1}$, which is the same order of magnitude as reported in the literature. The distinguishability analysis suggests that a system such as the one developed here may provide a means to characterize the hemodynamics associated with blood flow through the peripheral vasculature.

Keywords

electrical impedance tomography; video rate imaging; ischemia and reperfusion; peripheral vascular disease

1. Introduction

Electrical impedance tomography (EIT) is a medical imaging modality that maps the electrical conductivity and relative permittivity of internal tissues (Bayford 2006). In voltage-mode EIT, voltages are applied to a set of electrodes on the skin and the induced currents are measured. These voltages and currents are supplied to an algorithm which solves an estimation problem to recover the interior impedance parameters of the volume probed.

With a single set of measurements collected from an *unchanging* impedance distribution, *static* EIT images of the *absolute* impedance distribution can be recovered. Alternatively, with two sets of measurements collected from a *changing* impedance distribution, *difference* or *dynamic* EIT images can be reconstructed of the *change* in impedance. The static EIT

estimation problem is nonlinear, ill-posed and ill-conditioned making reconstructions difficult due to the small number of measurements, the dependence on high quality data and the use of an approximating numerical model to represent the physical situation. Typically, iterative methods are used to solve the static EIT problem. In dynamic EIT, the problem can be linearized (Cheney *et al* 1990) allowing for single step image estimation. Further, using differences in data collected at different times helps to mitigate errors associated with systemic hardware inaccuracies and model approximations.

A number of investigators have developed static EIT systems in an effort to detect cancerous lesions of the breast (Kerner *et al* 2002b, Malich *et al* 2000, Assenheimer *et al* 2001, Cherepenin *et al* 2001) and lung (Cherepenin *et al* 2002) with mixed results. More successful clinical applications of EIT have primarily focused on interrogating dynamic physiological events including cardiovascular dynamics (Kerrouche *et al* 2001), pulmonary dynamics (Frerichs 2000), gastric emptying (Nakae *et al* 1999, Vaisman *et al* 1999) and evoked neural response (Tidswell *et al* 2001). Reconstructing differences has provided more clinically meaningful images and since EIT systems are capable of collecting data at video frame rates (30 fps) (Edic *et al* 1995, Smith *et al* 1995), imaging changes in physiology has initially proved to be more promising than imaging static conditions.

It is important to continue the development of static EIT for tissue identification in cases where physiological dynamics cannot be imaged; however, here we consider whether it is possible to take advantage of high-speed EIT to image dynamic impedance changes occurring in the peripheral vasculature in order to characterize circulatory health. We explore this question and describe a system that may be used for this purpose. First, we briefly discuss other imaging modalities aiming to characterize cardiovascular and circulatory health based on the hemodynamics of the peripheral vasculature. We then use the theory of distinguishability in EIT (Isaacson 1986) to develop the system requirements necessary to image conductivity changes associated with dilating peripheral blood vessels. This analysis is followed by a description of our high speed EIT system and a suite of phantom imaging experiments that demonstrate its performance. Finally, we present results obtained using this system to image ischemia and reperfusion in the forearm which indicates its ability to image physiological events.

2. Assessing peripheral vasculature health

There are several clinical circumstances that require the assessment of the vasculature status of peripheral limbs, specifically the forearm. These assessments provide diagnostic information on (1) the progression of peripheral vasculature disease which acts as a surrogate marker for cardiovascular health (Katz *et al* 2005, Perticone *et al* 2001), (2) the health of the peripheral limb prior to and following bypass surgery (Sadaba *et al* 2001, Verma *et al* 2004) and (3) the health of peripheral vasculature in determining the need for an aggressive clinical response through surgical revascularization or amputation (Halperin 2002).

Primary methods used to evaluate peripheral vasculature health include (a) assessing regional vasoactivity through blood-pressure monitoring and plethysmography (Creager *et al* 1992), (b) functional and anatomical imaging of the vascular network with B-mode and Doppler ultrasound (US) and (c) contrast-enhanced angiography using both magnetic resonance imaging (MRI) and x-ray computed tomography (CT) (Reimer and Landwehr 1998). Use of near-infrared (NIR) tomography for imaging the forearm has also been recently suggested (Hillman *et al* 2001).

B-mode ultrasound is typically used to image flow-mediated dilation (FMD) of the brachial or radial artery during pressure-cuff-induced reactive hyperemia testing (Agewall *et al*

2001). Many investigations have suggested a link between the changing vessel diameters and cardiovascular disease (Kao *et al* 2003, Stadler *et al* 1998). Doppler ultrasound has also been used to image blood flow velocities within the lumen of the brachial artery. When simultaneously combined with B-mode scanning, a so-called duplex scan has been shown to improve clinical utility (Reimer and Landwehr 1998). Changes in the venous conduits have likewise been imaged with ultrasound and linked to cardiovascular health (Mollison *et al* 2006). Plethysmography assesses the changing volume of the forearm during a similar cuffing procedure and provides an estimate of total forearm blood flow (Mathiassen *et al* 2006). This procedure offers more of a global measure of the microcirculation within the forearm. Together, these modalities produce local diagnostic information (in the case of ultrasound since the imaging field of view is limited) as well as global diagnostic information (in the case of plethysmography since a single point measurement gives an estimate for the entire forearm).

Angiography with CT and MRI provides a more comprehensive evaluation of the vasculature within the forearm. In these methods, the image acquired following intravenous infusion of a contrast agent is subtracted from the image acquired before infusion and gives a detailed view of the larger vasculature structures within the forearm. While these modalities do provide precise anatomic detail of forearm vasculature, they are limited in their ability to generate functional information and do not yield sufficient diagnostic information on the microcirculation within the tissue (Reimer and Landwehr 1998). Further, these devices have a high initial procurement cost, are procedurally expensive and are large in size, limiting routine clinical use.

As blood flows through the major arteries of the forearm, the nutrient content is ultimately distributed to the tissues and individual cells through a microvasculature network. In patients with diseased or otherwise compromised peripheral vasculature, this nutrient transport mechanism is impaired. Blood flow signatures following induced ischemia and rapid reperfusion studies differ for healthy and unhealthy vasculature systems (Stadler *et al* 1998). A portable and high-speed, precise electrical impedance tomography system may provide a tool capable of assessing both local and global tissue health status during cuff testing. Continuous imaging of the forearm during cuff inflation may capture functional information on the metabolic activity of the cells and the nutrient-providing microvasculature since the electrical properties of tissues have been shown to change during ischemic events (Gersing 1998, Casas *et al* 1999). Further, blood has different electrical conductivity than the muscle and surrounding bones in the forearm and high-speed EIT may be able to image vessel dilation during the rapid influx of blood following cuff deflation. Monitoring these vascular changes with EIT may lead to a new clinical tool that is capable of characterizing the endothelial function, metabolic function and microvascular function, which may provide a very low cost means to routinely visualize specific regions of poor circulation within the forearm.

3. System requirements

The ability to detect a change in conductivity using EIT is a function of the system's precision. The EIT system discussed here is based on that reported in Halter *et al* (2008) and has a maximum voltage range of $2.3 V_{pp}$ and a measured signal-to-noise ratio (SNR) of approximately 100 dB. This voltage range and SNR establish a system precision that allows measurement of voltage changes on the order of $50 \mu V$. Fuks *et al* (1991) have shown that the ability to distinguish two conductivity distributions can be expressed by the following relationship:

$$V_r = |a| \frac{r_o}{2|\sigma_o|} \frac{2|\mu|R^2}{(1+\mu R^2)}, \quad (1)$$

where V_r is the instrument precision, a is the maximum current density applied, r_o is the radius of the body of interest and σ_o is the homogenous conductivity of the body of interest. R is the ratio of the radius of a centrally located conductive ‘variation’ to the full body radius, r_1/r_o , and $\mu = (\sigma_1 - \sigma_o)/(\sigma_1 + \sigma_o)$, where σ_1 is the conductivity of the centrally located ‘variation’. Figure 1 displays the problem configuration.

The system described here applies voltages instead of currents; however, both applied voltages and induced currents are recorded at each electrode. Adaptive (Zhu *et al* 1993) or iterative algorithms (Hartov *et al* 2002, Demidenko *et al* 2005) can be employed to adjust the applied voltages such that specific current patterns are applied. This suggests that a voltage-driven system can be treated as a current-driven system and permits the following analysis based on Fuks *et al* (1991).

The requirement for distinguishing between dynamically changing ‘variations’ can be established by finding the difference in precision required to image an object having a radius ratio of R_1 and of R_2 . For example, given a situation in which a centrally located conductor has conductivity σ_1 and radius r_1 , the instrument precision necessary to sense it, V_{r1} , is

$$V_{r1} = |a| \frac{r_o}{2|\sigma_o|} \frac{2|\mu|R_1^2}{(1+\mu R_1^2)}. \quad (2)$$

If the radius of the conductor increases by Δr from r_1 to r_2 as in figure 1, the new necessary precision, V_{r2} , becomes

$$V_{r2} = |a| \frac{r_o}{2|\sigma_o|} \frac{2|\mu|R_2^2}{(1+\mu R_2^2)}. \quad (3)$$

Similarly, one could determine the necessary precision associated with a change in conductivity, $\Delta\sigma$ from σ_1 to σ_2 instead of a size change.

The system precision necessary to measure changes in radial dimension, $\Delta r = r_2 - r_1$, with respect to a given conductivity change, $\Delta\sigma$, can be calculated by taking the difference between (2) and (3), $\Delta V_r = V_{r2} - V_{r1}$. If we consider the case in which blood pumps into the radial artery of the forearm, we may expect that during a reactive hyperemia test the diameter of the artery will change by approximately 125–325 μm (Agewall *et al* 2001). Blood has a reported conductivity of 0.66 S m^{-1} (Barber and Brown 1984) and the reported values of muscle tissue conductivity range from 0.4 to 0.6 S m^{-1} (Gabriel *et al* 1996b) resulting in values of $\Delta\sigma$ ranging from 0.06 to 0.26 S m^{-1} assuming that the forearm is homogeneously muscle. Figure 2 maps the precision ΔV_r associated with these ranges of Δr and $\Delta\sigma$. The bold yellow line displayed on the figure marks the current precision of our system.

This analysis assumes a two-dimensional domain, which is not the case in practice where induced currents extend outside the plane of the electrode placements. The forearm, however, has a relatively high degree of axial invariance as the bones, muscles and vessels

extend distally toward the hand. While not completely accounting for out-of-plane current flow, this invariance does suggest that the two-dimensional model applied here has sufficient validity for indicating the feasibility of dynamic EIT for peripheral vascular monitoring that warrants further developments in practice.

Finally, although this analysis considers the background conductivity to be homogenous, the geometry to be two-dimensional and the object of interest centrally located, it suggests that imaging the forearm during reactive hyperemia testing may capture significant conductivity changes associated with the dilating peripheral vasculature. Analysis of these dynamic impedance signatures may provide means for characterizing parameters such as FMD. With a heterogeneous background encompassing a three-dimensional volume, more stringent requirements on system precision may be required; however, as the object is moved away from the center and closer to the electrodes, these precision requirements also become more relaxed.

4. System design

This system was initially designed for breast imaging and its hardware is described in Halter *et al* (2004), Halter (2006) with the salient features presented in table 1. Here we briefly discuss how this system is configured for video rate imaging. It is a 64 channel distributed system with channel-dedicated digital signal processors (DSPs) used for signal generation and demodulation. Signal generation is implemented by stepping through a 256-word wavetable at an adjustable clocking frequency, f_{clk} . Demodulation is performed by sampling all locations within the waveform and applying a matched filter (Smith *et al* 1992) to recover the signal magnitude and phase.

The system has been optimized to collect 15 different spatial patterns for a single image frame. The data collection procedure consists of (1) loading the particular spatial pattern into each channel's wavetable, (2) sampling the voltage and current waveforms and (3) processing these samples using a matched filter (Smith *et al* 1992) to recover the magnitude and phase of the voltages and currents. Each spatial pattern is collected in this way and the process is repeated for multiple frames as shown in figure 3. A system command requesting the data be transferred from the individual channels is invoked when all data have been collected and stored within DSP memory. Currently, the system is limited to collecting a maximum of 40 frames due to data transmission and memory limitations.

The setup time, $T_{\text{SETUP}} = 77.6 \mu\text{s}$, and matched filter processing time, $T_{\text{MF}} = 31.3 \mu\text{s}$, are independent of the signal frequency. The time to sample the waveform, T_{SAMPLE} , depends on the sampling rate, f_s , of analog-to-digital conversion (maximum 1 MHz in this system) and the number of times each full waveform is sampled for averaging, N_{avg} . The sampling rate is signal frequency dependent (Halter 2006) and chosen to approach 1 MHz for each signal frequency. For the higher signal frequencies ($>312.5 \text{ kHz}$), multi-period under-sampling (Halter 2006) is employed in order to take full advantage of the 1 MHz ADC sampling rate. With $N_{\text{avg}} = 1$, each pattern is sampled in approximately $256 \mu\text{s}$. More

averaging increases the SNR; however, the improvement is only proportional to $\sqrt{N_{\text{avg}}}$.

Video rate imaging requires that individual data frames must be collected in 33.3 ms or less. Each frame's measurement time is a product of the number of spatial patterns, P , being collected and the sum of the setup, processing and sampling time for a single pattern's worth of data collection,

$$T_{\text{Frame}} = P \times (T_{\text{SETUP}} + T_{\text{SAMPLE}} + T_{\text{MF}}). \quad (4)$$

Table 2 shows the achievable frame rates ($1/T_{\text{Frame}}$) for a number of different averages and signal frequencies. These rates assume that 15 spatial patterns are collected per frame.

5. Phantom imaging

A set of dynamic phantom experiments were conducted to demonstrate video rate imaging over the full bandwidth of the system. An 8.5 cm diameter cylindrical water-tight tank was filled with saline ($\sigma = 0.1 \text{ S m}^{-1}$) to a height of 2 cm. Sixteen stainless steel electrodes (1 cm diameter), spaced 22.5° apart, were embedded flush with the tank's inner surface. The electrode centers were 1 cm above the base of the tank, and conductive polymer plugs were used as a point of contact between the system electrodes (Ag/AgCl) and the tank electrodes. A 1.3 cm diameter plastic rod was submerged approximately 0.5 cm from the edge of the tank and was connected to a motor spinning at 240 rpm (4 Hz) (see figure 4). Full 40-frame dynamic data sets were recorded at approximately 30 fps for signal frequencies of 10 kHz, 127 kHz, 1.129 MHz and 10 MHz. N_{avg} was set to 1 and 15 spatial patterns were collected for each frame.

A linearized difference algorithm was used to estimate the changing conductivity distribution between frames. A two-dimensional circular mesh with 1345 nodes was generated to model the experimental geometry. The change in conductivity at each of the mesh nodes, $\Delta\sigma$ was calculated using

$$\Delta\sigma = (J^T J + \lambda I)^{-1} J^T \{\Phi^{\text{ref}} - \Phi^{\text{frame}}\}, \quad (5)$$

where J is a Jacobian matrix representing the sensitivity of changes in conductivity to changes in boundary voltages, I is the identity matrix and λ is a regularization parameter used to stabilize the inversion. Φ^{ref} is a set of boundary voltages recorded from a reference conductivity distribution, while Φ^{frame} is the set of boundary voltages collected from a different conductivity distribution.

For these experiments, a data set collected with only saline in the tank was used as the reference, with a separate reference set collected for each of the frequencies considered. The reconstructed images displayed in figure 4 correspond to data collected at frequencies of 10 kHz and 10 MHz where the precise frame rates were 30 fps and 32 fps, respectively. The expected drop in conductivity associated with the plastic inclusion is visible for all frames at both frequencies and verifies that the system is capable of recording video rate images over the full frequency spectrum of operation. The slightly different frame rates and the fact that the inclusion was not at the same angle of rotation when data acquisition was initiated for the different frequency data sets explain the small phase shift visible between the two figures. Additionally, the observed region of increased conductivity proceeding the inclusion and that of decreased conductivity following the inclusion stem from the hydrodynamics of system. The rotating rod perturbs the saline in its path, enlarging the cross-sectional area available for current flow. The 2D reconstruction assumes a planar geometry and does not account for changing out-of-plane current flow. Instead, this out-of-plane disturbance manifests itself as the observed increase in conductivity. Similar effects are observed in the wake of the rod's trajectory since there is a displaced volume of solution that generates a small void following the rod. The reconstructed inclusion follows the circular trajectory of the spinning insulating rod and the y-location of the maximum conductivity drop was

computed for each frame and frequency and plotted in figure 5(a). Note that the phase shift due to the rod being at a different initial angular position was removed so that at Frame 0 the inclusion is at the 12 o'clock location (maximum y-location). Further, the 2 fps differences in frame rates between 10 kHz and 10 MHz are observed at the end of the frame sequence where the peak at 10 MHz has not yet reached the 12 o'clock location. Figure 5(b) shows the associated spatial frequency spectrum of this trajectory with the expected peak at 4 Hz for each signal frequency used.

6. Imaging physiology

High speed data collection for capturing dynamically changing impedances in a physiological setting was demonstrated through an experiment imaging ischemic effects in a human forearm. The forearm consists primarily of blood, muscle, adipose and bone tissues having conductivities of approximately 0.66 S m^{-1} , $0.4\text{--}0.6 \text{ S m}^{-1}$, 0.05 S m^{-1} and 0.02 S m^{-1} , respectively (Gabriel *et al* 1996a, 1996b).

A set of 16 1 cm diameter Ag/AgCl electrodes spaced 1.6 cm apart were placed around a volunteer's arm approximately 5 cm below the elbow. The electrodes were fixed in place with a surgical tape and connected to 16 system channels as shown in figure 6. The system was programmed to collect data at 10 fps over 4 s intervals at a signal frequency of 127 kHz with $N_{\text{avg}} = 1$. This slower frame rate enabled capturing a wider temporal period of the physiological change since faster dynamic effects were not expected under the experimental conditions. A frequency of 127 kHz was chosen to specifically interrogate conductivity differences occurring between blood and muscle since the conductivity contrast is larger at this mid-band frequency than at the system's top frequencies (Gabriel *et al* 1996a, 1996b). Further, use of this frequency reduced the effects of contact impedance occurring at the lower frequencies ($<50 \text{ kHz}$). After the data were collected, it was offloaded to the system computer followed by a 26 s wait. Then another 4 s series of data was collected. This process was repeated throughout the duration of the experiment. An occluding cuff was placed on the volunteer's upper arm, and by inflating the cuff an ischemic environment was established within the forearm. Prior to high speed data collection, the effective electrode contact impedance was continuously monitored by making impedance measurements between adjacent pairs of electrodes until no significant effective impedance changes ($<1\%$) were observed. This ensured that experimental measurement changes were due to the effects of the upper arm being occluded and not to the changing contact impedances.

After the effective contact impedance was stable, two 4 s data sets were collected prior to inflating the cuff. After this acquisition, the cuff was inflated to 200 mmHg to ensure complete occlusion of both the venous and arterial vessels feeding the forearm. 4 s data sets were continuously collected followed by 26 s waits and subsequently offloaded for 10 min while the cuff remained inflated. After 10 min, the cuff was deflated immediately ($\sim 1 \text{ s}$) after a 4 s data acquisition was initiated to ensure that the impedance dynamics associated with the rapid influx of blood was measured.

Stopping blood flow to tissue induces an ischemic environment where neither oxygen nor nutrients are provided to the cells. As the cuff remains inflated, cellular metabolism increases in order to sustain a homeostatic environment and since this energetic process receives no new oxygen and nutrients, the cells exhaust their energy reserves. The dominant morphological response is ultimately that cellular edema results, where the intracellular fluid volume increases while the extracellular fluid decreases. The extracellular fluid is the primary conduit for current flow and as its volume decreases, conductivity is expected to decrease. Likewise, upon cuff deflation, blood flow rapidly returns and the conductivity is expected to increase to its initial value. Other physiological responses to cuff inflation/

deflation include changes in pH and temperature. Changes in impedance associated with these parameters have been noted (Kun *et al* 2003, Jaspard and Nadi 2002); however, their effects typically occur over time periods longer than the ones employed here.

The data collected were reconstructed by the same linearized difference algorithm used in the phantom studies with the initial frame obtained during the un-occluded arm serving as the reference data set. Additionally, the same reconstruction mesh was used, but scaled to match the diameter of the forearm (6.5 cm). Figure 7 shows a sequence of $\Delta\sigma$ images taken at different times throughout the experimental procedure. The circled areas of conductivity within the first image change considerably during the course of the experiment and are used as regions of interest (ROIs) for further analysis. The center of ROI₁ ($x = 0$ cm, $y = -1$ cm) was qualitatively selected at a centrally located area of conductivity change, while the center of ROI₂ ($x = 2.19$ cm, $y = -0.82$ cm) was quantitatively selected as the point of maximum conductivity change occurring in the frame immediately prior to cuff deflation ($t = 645$ s). Each of the ROI diameters was 1.3 cm which was 20% of the full 6.5 cm mesh diameter. Mean conductivities were calculated for the entire image and for the ROIs at each time instance. The mean conductivities, shown in figure 8, drop linearly with time while the cuff is inflated and at the point in time when the cuff is deflated the impedances within each region immediately begin to increase.

Conductivity changes associated with ischemia were analyzed first. A first-order polynomial fit was established for the linear region occurring between 200 s and 615 s and the rates of conductivity change for the full image, ROI₁, and ROI₂, were determined to be $-2.77 \times 10^{-7} \text{ S m}^{-1} \text{ s}^{-1}$, $-4.63 \times 10^{-7} \text{ S m}^{-1} \text{ s}^{-1}$, $-3.78 \times 10^{-7} \text{ S m}^{-1} \text{ s}^{-1}$, respectively. Other authors have also reported that conductivity decreases with time during ischemia in different organs including heart and liver (Gersing 1998, Casas *et al* 1999). Schafer *et al* (1999) have specifically reported on the dielectric properties of skeletal muscle during ischemia and have found that the dielectric loss factor ($\alpha\sigma$) decreases with time. The rate of conductivity change occurring due to ischemia can be extrapolated from one of their graphs (figure 1 in Schafer *et al* (1999)) to be approximately $-5 \times 10^{-7} \text{ S m}^{-1} \text{ s}^{-1}$ which is in close agreement with the rates reported here. Through the linear portion of the time sequence in figure 8, the change in conductivity occurring during individual 4 s intervals is relatively small because of the slow overall rate of change in conductivity ($\sim -1 \times 10^{-7} \text{ S m}^{-1} \text{ s}^{-1}$). This rate of conductivity change was determined for each of these 4 s intervals using the same fitting approach applied to the entire linear portion of the above data set. The mean of these rates is close to those reported above, with a mean rate of change occurring during each 4 s interval of $-2.88 \times 10^{-7} \text{ S m}^{-1} \text{ s}^{-1}$, $-3.66 \times 10^{-7} \text{ S m}^{-1} \text{ s}^{-1}$, $-4.8644 \times 10^{-7} \text{ S m}^{-1} \text{ s}^{-1}$, for the full image, ROI₁ and ROI₂, respectively.

The most obvious changes in conductivity occurred immediately after cuff release when the tissue was reperfused. There are two time scales that can be considered here. The first is the rapid response occurring during the 4 s interval in which the cuff was deflated (circled data in figure 8), while the second takes place over the 160 s following cuff release. Figure 9 shows the set of frames collected during the 4 s interval in which the cuff was deflated, where the first frame in this interval was used as the reference. The graph in figure 10 shows the mean increase in conductivity of the full image, ROI₁ and ROI₂ occurring during this interval. Clearly, the point at which deflation began is marked with the rapid increase in σ . The forearm was oriented with the venous and arterial vessels situated near ROI₁ and ROI₂ (see figure 6) and the blood rushing through these vessels at the onset of cuff deflation increases blood vessel volume and may give rise to the rapid change in conductivity observed. Although the physiological response to cuff deflation is complex, the immediate influx of blood has been shown to increase vessel volume beyond its normal resting value (Agewall *et al* 2001). The drop in conductivity observed at the end of the 4 s interval may

correspond to the blood volume returning to the resting state. This, however, requires further investigation to verify.

With respect to the longer time scale, previous electrical impedance spectroscopy (EIS) studies of ischemia have primarily been interested in developing EIS as a tool for monitoring the state of excised tissue used in transplant surgery and have not investigated reperfusion of the tissues (Gabriel *et al* 1996a, 1996b). Wascher *et al* (1998) have, however, performed an experiment similar to ours where they generated ischemia within the forearm by placing a cuff at the upper arm and left it inflated at 200 mmHg for 10 min. Instead of making measurements of electrical impedance, they recorded blood flow every 10 s after the cuff was released. They report a rapid increase in blood flow immediately following cuff release, with a rise in blood flow from 5 mL min⁻¹ per 100 mL to 35 mL min⁻¹ per 100 mL occurring within the first 20 s, and then a slowly decreasing blood flow dropping to 10 mL min⁻¹ per 100 mL over the next 100 s. The change in conductivity, shown in figure 11, that we measured over the 120 s following cuff deflation exhibited similar characteristics. There is a rapid increase in conductivity occurring immediately at cuff release due to the rapid influx of blood (rapid increase in blood flow), which is subsequently followed by a slower increase in conductivity associated with the decreasing blood flow and the re-establishment of cellular O₂. Dzwonczyk *et al* (2004) observed a similar conductivity increase in ischemic myocardial tissue when reperfused following bypass surgery. Similar to our findings, they suggest that the measured change in conductivity occurring during reperfusion may be a direct measure of blood flow through an organ's vasculature.

7. Conclusions

To the best of our knowledge, this is the first EIT system that is capable of collecting video rate impedance images from a dynamically changing environment at signal frequencies up to 10 MHz. Collecting high frame rate data sets at multiple frequencies may provide useful clinical information for monitoring the health of the peripheral vasculature and should be explored. We have demonstrated the system's ability to collect high frame rate impedance images from an *in vivo* system and plan to use it to visualize changes in a larger cohort of patients in order to establish the typical temporal conductivity signatures for both healthy and diseased patients.

Acknowledgments

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Figure 1.

Arrangement for the detectability problem. (a) Initial homogenous case. (b) Inhomogenous case 1—centered circular conductor (c) Inhomogenous case 2—centered circular conductor with increased radius.

Figure 2. Precision maps for dynamic volume changes. The contour labels display the required resolution in mV for distinguishing between two conductivity distributions as their radii change by a factor of Δr . The yellow line denotes the current system's precision.

Figure 3.

Dynamic EIT data collection flow diagram for multiple patterns. P is the number of spatial patterns and N_{frames} is the number of frames to collect. Timing equations are shown beside each block of the process. $NOPS$ is the number of DSP operations necessary to complete a block and are performed at a rate ($1/T_{DSP}$) of 66 MHz.

Figure 4. Video rate imaging at 10 kHz (above left) and 10 MHz (above right). Change in conductivity images of a sequence of 40 frames collected at 30 fps (10 kHz) and 32 fps (10 MHz) while an insulating inclusion was spinning in a saline tank at a rate 240 rpm as shown on left schematic. Difference reconstructions were obtained with a blank tank used as a reference data set. Colormap corresponds to changes in conductivity and is in units of S m^{-1} .

Figure 5.

(a) Insulating rod trajectory as a function of time. The y-axis is the y-location of the largest change in conductivity found in each frame. (b) Frequency spectrum of the insulating rods' trajectory. All frequencies have their peaks occurring at 4 Hz corresponding to the rate at which the insulating rod is spinning.

Figure 6.

Experimental configuration for imaging ischemia in a human forearm. The image on the left shows the experimental configuration with the electrode array (top right) fixed around the volunteers' forearm and the occluding cuff positioned around the upper arm. An MRI (bottom right) of the volunteers' forearm is shown in the same orientation as the conductivity images appearing in figure 7.

Figure 7.

Change in conductivity images of an occluded forearm. The time between each image is 30 s, where each image is the first frame of the 4 s data collection window. The cuff was inflated just prior to the image with the solid square border and the cuff was deflated during the 4 s data collection window of the image with the dashed square border. A decrease in conductivity is visible in the central part of the tissue during occlusion and the conductivity returns to its initial state after the cuff has been deflated. The circles in the first image are ROI₁ and ROI₂ used for data analysis. Colorbar is in units of S m^{-1} .

Figure 8.

Change in ROI mean conductivity throughout the duration of the induced ischemia experiment. Each marker represents a single frame of data and the groups of markers are the frames collected during the 4 s high speed acquisition periods. The gap between these is the 26 s allotted to offloading the data.

Figure 9. Change in conductivity images of forearm during cuff deflation. The time between each image is 0.1 s. An increase in conductivity is visible in the central part of the tissue as fluid returns to the tissue. Colorbar is in units of S m⁻¹.

Figure 10.

Change in ROI mean conductivity during cuff deflation with respect to the conductivity at the initial frame collected immediately prior to deflation. This is an expanded view of the circled data in figure 8 except that an offset corresponding to the initial conductivity has been removed. The onset of impedance changes corresponding to cuff deflation occurs at the point where the impedance begins to increase rapidly.

Figure 11. ROI₁ conductivity change during reperfusion. Note the high rate of conductivity change occurring immediately after cuff release which corresponds to the rapid increase in blood flow.

Table 1

Measured system specifications (Halter *et al* 2008).

Specification	Units	
Number of channels	64 (expandable)	
Frequency of operation	10–10000	kHz
Output scaling	14 bits	Full scale (2.3 V _{pp})
SNR	100	dB
Digital precision	15	Bits
Phase coherence	0.1	Degrees
Frame rate	>30	Fps
Minimum frequency scaling	125	mHz
Channel-to-channel variations	0.05	%
Accuracy	99.7	%
Data acquisition bandwidth	12.5	MHz

Maximum achievable frame rates assuming 15 spatial patterns are being used. Frame rates vary between frequencies due to T_{SAMPLE} 's dependence on f_{clk} .

Table 2

Freq (kHz)	Frame rates with a different number of averages (N_{avg}) in frames s^{-1}							
	1	2	4	8	16	32	64	128
10	163.04	94.04	50.93	26.57	13.58	6.87	3.45	1.73
127	181.22	106.35	58.24	30.57	15.68	7.94	4.00	2.00
1129	181.37	106.45	58.30	30.61	15.69	7.95	4.00	2.01
10 000	181.11	106.28	58.19	30.55	15.66	7.93	3.99	2.00