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Reconstruction of optical scanned images of inhomogeneities in biological tissues by Monte Carlo simulation

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ABSTRACT

The optical imaging of inhomogeneities located in phantoms of biological tissues, prepared from goat's isolated heart as control tissue and embedded with spleen and adipose tissues representing tumors, by Monte Carlo simulation, is carried out. The proposed scanning probe consists of nine units. Each unit is equipped with one photon injection port and three ports arranged in a straight line to collect backscattered photons emerging from various depths, and one port, placed coaxially to the source on the opposite side of the phantom, to collect the transmitted component. At each position of the grid, superposed on the tissue phantom, photons are introduced through source port into the phantoms and backscattered and transmitted photons are collected by respective ports. Based on the data collected from the entire grid surface the respective gray-level images are reconstructed. The inhomogeneity located at certain depth (2, 4, 6 mm) is visualized in three images formed by the backscattered data collected by three ports. Increase or decrease in normalized backscattered intensity (NBI) observed in their scans corresponds to that of high scattering (adipose) or absorbing (spleen) inhomogeneity compared to that of control tissue and also their location as determined by NBI variation as received at various ports. The images constructed from the transmitted data are associated with decrease in intensity. The scans of these images through their centers show that normalized transmitted intensity (NTI) attains its maximum value when the inhomogeneities are at depth 6 mm. These scans are of higher amplitude for spleen compared to that of adipose tissues. Thus the data received by backscattering and transmission complement each other in identifying the location and type of inhomogeneities.

1. Introduction

The interaction of laser radiations with biological tissues is a complex process due to involvement of several mechanisms. These include backscattering, transmission and absorption. The backscattering and transmission processes are further associated with collimated and multiple scattered photons, respectively [1,2]. Each process involves the interaction of radiations with structural components of tissues, which depend on their optical parameters, absorption and scattering coefficients and anisotropy parameter. Other parameters which influence this process are refractive index of the medium and wavelength of the light source [3,4]. The tissue parameters vary depending on their composition and functional properties between disease and control.

Based on the characteristics of laser radiation, various techniques for clinical applications have been developed. An important application of this is to detect tumor in biological tissues. At present the most prominent screening tool for cancer detection is X-ray mammography, which exposes the patient to ionizing radiation, thus causing risk of cancer induction. The techniques such as positron emission tomography, MRI, ultrasound, and thermography have also contributed significantly in diagnostic medicine. Ultrasound data, based on variation of acoustic impedance, show the presence of tumor at fairly advanced stage after extensive application of computer algorithms. Other techniques require injection of contrast agent and are expensive. In contrast, the optical imaging emerges as an alternative tool in detection of inhomogeneity hidden in soft biological tissues [5,6]. The prominent optical techniques are based on frequency-resolved [6,7], time-resolved [8] and continuous wave procedures, employing different geometries [9,10].

During the past three decades, red and near-infrared spectroscopy and imaging, based on backscattering and transmission of radiation have emerged as powerful research and diagnostics tools for both laboratory studies and clinical trials. Some of the applications include imaging of human tissues, measurements

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of oxygen consumption of muscle, detection of tumor in breast tissues, functional brain mapping and localization of internal organs in thorax region [11,12]. Through his pioneering work on transillumination, Cutler viewed the changes in human breast [13], leading to development of the frequency domain optical mammography, carried out by placing the sandwiched human breast between laser source and set of detectors [14,15]. Recently the transillumination of biological tissues, a time dependent procedure, is carried out by placing the source and detectors on the same surface [16,17].

There are various types of tumors such as carcinoma, subcutaneous fat with melanoma, squamous cell carcinoma, adenocarcinoma, fibrosarcoma with their distinct optical properties [18,19]. The development of tumor may not be confined to single layer of tissues below the surface. The imaging of these, either by application of a single probe directly [20] or indirectly by parameter derived from tissue reflectance, is carried out [21]. The optical coherence tomography (OCT) [22] has extensively been used for imaging of biological tissues to a depth limited to around 1.5 mm. Other techniques include mapping of tissue optical properties using modulated imaging [23], time-resolved diffuse reflectance with null source-detector separation [24] and log-slope analysis [25]. As these are single beam techniques, the backscattered data are confined to single layer of tissues under one measurement condition. In contrast to single transillumination probe, the image reconstructed by optical tomography provides the cross-sectional details of a single layer which could further be extended to multiple images of respective slices [26] for their 3D image reconstruction [27]. But for routine multilayer imaging of biological tissues, an inexpensive spatially resolved optical system, based on backscattering and transmission of radiation, is required.

The analysis of radiation interaction with multilayer inhomogeneous tissues requires not only the optical parameters of individual tissues but also their refractive indices [28]. Experimentally this has been shown that the backscattered photons emerging closer to beam entry point originates from the tissue layers close to surface, whereas, photons received at farther distances emerge from the deeper layers [29]. The surface profile constructed from these backscattered signals is a unique feature of biological tissues and by best-fit of this profile with that as obtained by Monte Carlo simulation, the optical parameters of various tissues are also determined [30,31]. In contrast to this the transmitted signal emerging out on opposite surface carries information on overall composition of tissue structure, which may be valuable for parametric analysis and reconstruction of respective images. For this purpose a detailed analysis of photons interaction in tissue phantoms, with and without inhomogeneities is required. Monte Carlo simulation (MCS), which provides data on scattered photons even close to beam entry point on individual histories, is ideal for this purpose [32]. Thus the objective of the present work, based on the MCS, is to develop an imaging system, which establishes relationship between various optical and tissue phantom related parameters to detect the inhomogeneities in various layers, which are either highly backscattering or absorbing compared to that of control tissues.

2. Materials and methods

2.1. Simulated multilayer system

Fig. 1(a) shows the schematic of the system to measure backscattering and transmission components of radiation after multiple scattering from the control tissue phantom of infinite length and width and thickness 10 mm. For imaging purpose a grid of size 30×9 mm^2 was superposed on the phantom. The origin of the coordinates system of phantom coincided with the center of the grid. The resolution of the grid was 1 mm^2, the same as size of injection

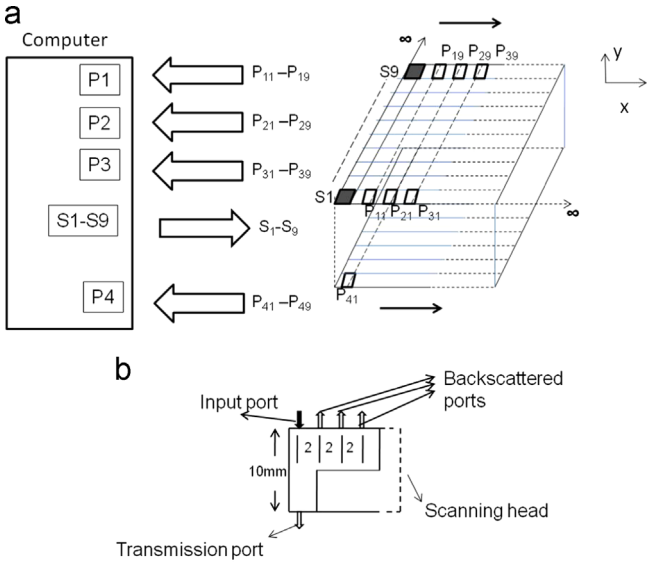

Fig. 1. (a) Schematic of the optical scanning system for measurement of backscattered and transmitted photons after interacting with tissues. (b) Each unit is consisting of five ports, one for photon injection and three to collect backscattered photons at various distances from beam entry port, and one to collect transmitted photons located coaxial to the input port.

and collection ports. The scanning head, consisting of nine units to inject and receive backscattered and transmitted photons, was initially placed at the left side of the grid. Each unit contained one input port for injection of infinitely narrow beam of photons and three receiving ports located at distance 2, 4 and 6 mm from the input port to collect backscattered radiations, placed at the corresponding grid element. Our preliminary studies had shown that the separation between photons injection and collection ports placed in the probe is sensitive to the corresponding depth of tissue layers. Accordingly the nearest receiving port at 2 mm received maximum number of backscattered photons from tissue layer located at depth 2 mm approximately, whereas, the ports at 4 and 6 mm collected photons from deeper layers. The transmitted component in each unit was received by an exit port of area 1 mm^2, placed coaxially with input port of the phantom.

Initially 10 million photons of wavelength 632.8 nm were injected at each input port, which after interaction with medium emerged as backscattered and transmitted components. The outline of the ports is shown in Fig. 1(b). The center to center separation between any pair of ports was 2 mm. After collection of data at first grid line, the scanning head was moved 1 mm away along x-axis. By this process the data from the entire grid of size 30×9 mm^2 were collected with final position of photon entry port at 24 mm.

From the number of backscattered photons collected within 1 mm^2 the normalized backscattered intensity (NBI) was calculated by

$$\text{NBI}\ (\%) = \frac{N_{i,j}}{N_0} \times 100$$

N_0 is the total number of incident photons and $N_{i,j}$ is the total number of photons collected at any port.

The NBI values as obtained at various locations of grid, along with their positions (i,j), were stored in the computer for further processing [33]. Similarly the number of transmitted photons collected within 1 mm^2 grid (i,j) was represented as the normalized transmitted intensity, given by

$$\text{NTI}\ (\%) = \frac{T_{i,j}}{N_0} \times 100$$