Computer Methods and Programs in Biomedicine 141 (2017) 11–17

Contents lists available at ScienceDirect

Computer Methods and Programs in Biomedicine

journal homepage: www.elsevier.com/locate/cmpb

Simulation of laser backscattering system for imaging of inhomogeneity/tumor in biological tissues

J.B. Jeeva[a], Megha Singh[a,b,*]

[a] Department of Sensor and Biomedical Technology, School of Electronics Engineering, V.I.T. University, Vellore 632014, India
[b] Center for Biomedical Engineering, S.G.N. Educational Foundation, #12, III Street, Park Avenue, Velachery, Chennai 600042, India

ARTICLE INFO

Article history:
Received 31 January 2016
Revised 13 January 2017
Accepted 17 January 2017

Keywords:
Monte Carlo simulation
Tissues phantoms
Inhomogeneities
Normalized backscattered intensity
Reflectance imaging system

ABSTRACT

Background and objectives: The optical characteristics of biological tissues vary in health and diseases. By analysis of photons scattering process by Monte Carlo simulation (MCS) the inhomogeneities in tissues are to be identified and their images reconstructed.
Methods: Digital phantoms with goat's heart as a control tissue embedded with inhomogeneities adipose (high scattering) and spleen (high absorption) are simulated. The phantoms considered are – (a) simulation of the developed stage of inhomogeneity by inclusion of adipose and spleen tissues in control and (b) its onset stage by increasing the optical parameters by 10% at fixed locations in control tissue. These phantoms are scanned by simulated system, consisting of nine ports for photon injection and backscattered photons from each port are received by three ports located at 2, 4 and 6 mm from the injecting port, placed in the direction of *x*-axis. By the data collected from the entire surface, by processing, three grey-scale images are constructed. For localization of inhomogeneities these images are scanned in terms of normalized backscattered intensity (NBI).
Results: The images obtained by MCS with 1 million photons, with error minimized, at respective ports, show the presence of inhomogeneities at various depths, which is further supported by the increase or decrease in the NBI compared to that of control for adipose or spleen, respectively. The increase or decrease is more at first port compared to others. The inhomogeneities located at 2 mm below the surface are better identified by the receiving port located at 2 mm on the surface. The same applies to inhomogeneities located at 4 and 6 mm, respectively.
Conclusion: The present simulated system not only shows the presence of inhomogeneties at various depths in tissue phantom but also presents their characteristics.

1. Introduction

The interaction of laser radiation with biological tissues within the optical therapeutic window (600-1300 nm) is associated with enhanced scattering and less absorption, thus allowing the laser beam to penetrate deeper into the tissues. A fraction of the incident beam after multiple interactions with tissues emerges as backscattered away from the beam entry position, a part of the beam is transmitted, and the remaining part is absorbed within the medium [1,2]. During the development of cancer many processes get activated inside the cells, leading to accelerated cell proliferation, reduced apoptosis and neovascularization of the interstitial tissues. These processes modify the biochemical content of the intracellular and extracellular environment [3], leading to changes in their optical parameters.

The detection of these by x-rays and ultrasound as attenuation coefficient and acoustic impedance is carried out at fairly developed stage of tissue changes. In contrast, the optical techniques are capable of measuring these changes through variation in refractive index and other optical parameters at the onset stage [4,5]. Some of the prominent procedures, based on this principle, are the development of temporal analysis of reflected signal [6], multichannel analysis of temporal data [7], frequency domain optical mammography (FDOM) [8] and CW techniques based on different geometries [9,10]. Their optical characterization is further carried out by the change in optical absorption and reduced scattering coefficients [11], measurement of diffuse reflectance [5], the backscattering at forty-five degree [12], the light transmitted between two points on the surface [13], and multispectral imaging [14].

Some of the advanced optical imaging techniques include the optical coherence tomography, providing the information on tissue

* Corresponding author.
E-mail addresses: jbjeeva@vit.ac.in (J.B. Jeeva), msingh_iitm@yahoo.com (M. Singh).

http://dx.doi.org/10.1016/j.cmpb.2017.01.010
0169-2607/

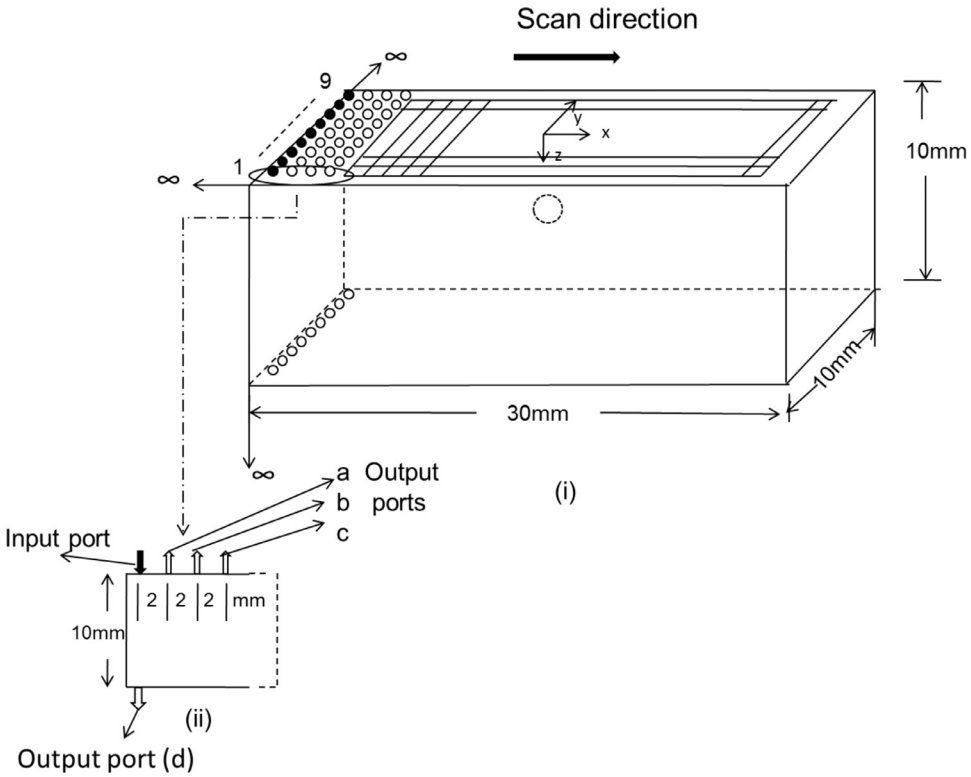

Fig. 1. (i) Schematic of the optical scanning head, for measurement of backscattered photons after interacting with tissues, consists of nine units. (ii) Each unit consists of four ports, one for photon injection and other three to collect backscattered photons at various distances from beam entry port.

changes through high resolution images of retina and tissue structure, [15–18], non-contact Doppler scanner, showing flow changes in tissues [19] and development of high resolution images through non-contact optical modulating technique [20]. These techniques are sensitive to tissue changes but the depth of imaging is limited to 1-2 mm. The basic requirement for any tumour detection system is not only to detect this but also to characterize in terms of its type and depth of penetration in healthy tissues. This could not be achieved by single source –detector system as output measured by this is limited to single location in the tissues. We have shown that the backscattered fractions from various tissues layers are emerging at different locations away from the beam entry point [10]. By application of this, the backscattered signal data are obtained along a section of tissues and by multiple scans the data from entire surface are collected. By application of this, the images of three layers of complex tissue structure are obtained and in terms of their optical parameters are characterized [21,22,23]. The backscattered photons received by a detector are not only emerging from a selected layer but are also contributed, to a lesser extent, by photons emerging from layers above and beneath the selected one [21,24,25]. But by this manual system the scanning over a tissue surface is time consuming and is limited to number of layers attributed to locations of photodetectors. Further theoretical analysis is required for the development of a rapid scanning system to obtain reflectance data from the entire tissue.

There are several theoretical procedures for analysis of tissue-radiation interaction, such as numerical algorithm to solve the time-dependent diffusion equation [26], combination of the discrete ordinate method with a pseudo-spectral/finite difference method [27] and Monte Carlo simulation (MCS) [28,29,30]. In contrast to other methods the MCS provides accurate results for measurement of backscattering component even close to beam entry point [14].The uncertainty associated with the process through the variation in input parameters (as mean ± standard deviation D) introduces uncertainty in the final results but recently we have shown that the convergence associated with mean standard error is improved with number of input photons to 1 million. The amplitude of the backscattered component decreases as the depth of location of inhomogeneity increases, attributed to multiple scattering in the medium [22,31].

In contrast to existing techniques [21–22], for characterization of an inhomogeneity, the present system requires simultaneous collection of data from the entire tissue surface by a specific arrangement of multiple input and output ports as units. A combination of various parameters, such as axial distance of the output port from beam entry point, depth of occurrence of interaction, changes at onset and advance stage of tumour in terms of their optical parameters, are further required. These are taken as inputs to Monte Carlo simulation of photon scattering process and based on output data the respective images of tissue sections are reconstructed. To authors knowledge the simulation of such a system, which is very essential for tumour localization and characterization in wide variety of tissues, has not been reported to date.

2. Materials and methods

2.1. Simulated scanning system

Fig. 1(i) shows the schematic of the simulated scanning system, consisting of a slab of homogeneous heart tissue (control) of infinite dimensions with selected size $30 \times 10 \times 10$ mm around its centre. Each inhomogeneity was placed below its centre (0,0,0) at various depths. The scanning head consisted of nine units and each unit consisting of one photon injection port and three output ports located in a straight line along *x*-axis direction. Initially photons of wavelength 632.8 nm were injected at input port 1, which after interaction within the medium emerged at various distances on the surface and were collected by three output ports, located at dis-