High count rate multichannel TCSPC for optical tomography

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ABSTRACT

An improved Time-Correlated Single Photon Counting (TCSPC) technique features high count rate, low differential nonlinearity and multi-detector capability. The system has four completely parallel TCSPC channels and achieves an effective overall count rate of 20 MHz. By an active routing technique, up to eight detectors can be connected to each of the TCSPC channels. We used the system to record optical mammograms after pulsed laser illumination at different wavelengths and projection angles.

Keywords: (170.3660) light propagation in tissue; (170.6960) tomography; (170.3830) mammography

1. INTRODUCTION

Attempts to image structures in deep tissues by optical tomography based on near-infrared (NIR) light are faced with the problem that details are washed out by the strong scattering of photons. Therefore, optical tomography of thick tissues cannot achieve spatial resolution comparable to that of X ray methods. There are, however, two benefits of optical methods: At sufficiently low power NIR radiation does not harm tissue. From absorption coefficients measured at several wavelengths, physiological quantities can be derived, in particular haemoglobin concentration and blood oxygen saturation [1].

Light propagation in tissue is governed by scattering and absorption of photons. Unfortunately, these effects cannot be reliably distinguished in simple steady state measurements. However, when pulsed light is used, time-resolved detection of diffusely transmitted or reflected intensity yields additional information. Although increased scattering and increased absorption both decrease the output intensity, stronger scattering broadens the transmitted or reflected pulse while increased absorption tends to narrow it. By modelling propagation of light as diffusion of photons and by using appropriate boundary conditions when solving the diffusion equation, the reduced scattering coefficient and the absorption coefficient of a homogeneous medium can be distinguished and quantified using the shape of the broadened pulse only [2].

Optical tomography aims at the detection of inhomogeneities in tissue and relies on measurements at a number of detector positions for each source position. Various mathematical methods have been developed to reconstruct the position and the optical properties of inhomogeneities from measured data [3, 4]. Time-resolved techniques improve localisation and characterisation of inhomogeneities [5], as information on the pathlength of each photon becomes available. Inhomogeneities differing from the surrounding tissue by scattering and absorption have different influence on the distribution of times of flight of photons.

Transillumination scanning optical mammography [6] is performed in slab geometry, i.e. the breast is compressed between two glass plates with source and detector, positioned on opposite sides, scanned synchronously. Detection at different projection angles gives additional information on the internal structure of the tissue. The apparent location of the image of structural details depends on the depth of the corresponding structure in the tissue. By shifting and adding images obtained at different projection angles structural details in selected layers can be enhanced while those of other layers are smeared out (fig.1). The method is known as “digital tomosynthesis” in X ray mammography [7] and requires data taken at a sufficient number of projection angles. This empirical approach yields images containing depth information but cannot compensate for the loss in resolution and contrast due to scattering.

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European Conferences on Biomedical Optics, ECBO 2001, Munich, June 2001
Complete reconstruction of tissue structures and optical properties from time-resolved data is extremely demanding and not entirely solved yet. Nevertheless, it turns out that a large number of time-resolved detection channels is required to obtain meaningful optical tomographic images. For in-vivo applications it is important to keep the measuring time short, i.e. simultaneous recording at different projection angles is highly desired. An additional advantage of a multi-channel measurement is that for a given number of photons injected the total number of detected photons is increased improving signal-to-noise ratio compared to single-channel detection.

![Fig. 1: Appearance of an object under different projection angles (top) and tomosynthesis (bottom)](image)

To record data for optical mammography containing temporal information, modulation techniques and time-correlated single photon counting techniques are used. TCSPC has the benefit of a higher system bandwidth limited by the transit time spread of the detector rather than by the width of its pulse response. Furthermore, TCSPC yields a shot-noise limited signal-to-noise ratio and a near-ideal sensitivity. Conventional TCSPC electronics is, however, too bulky for the required number of channels and poses severe restrictions on the photon count rate. We present improved TCSPC electronics which can be used for up to 32 detector channels with an effective overall count rate of 20 MHz and demonstrate its application to optical mammography.

### 2. HIGH COUNT RATE MULTICHANNEL TCSPC

The TCSPC device is shown in fig. 2, left. It is a package of four completely parallel TCSPC channels. Each channel can be expanded to record the signals of eight detectors simultaneously and has a maximum useful count rate of 5 MHz. The package is operated in a standard PC. The block diagram of one TCSPC channel is shown in fig. 2, right.

![Fig. 2, left: The complete 4-channel device ready to be inserted in a PC right: Block diagram of one TCSPC channel. CFD - constant fraction discriminator, TAC - time-to-amplitude converter, DAC - digital-to-analog converter, ADC - analog-to-digital converter, SUB - subtraction circuit](image)

Basically, the device uses the well-known reversed start-stop principle with constant fraction discriminators, CFD, time-to-amplitude converter, TAC, and analog-to-digital-converter, ADC, that addresses a memory in which the histogram of the photon density over time is built up.
In contrast to the conventional setup, we use an ultra-fast ADC with an error reduction circuitry and an extended memory structure for multidetector operation [8]. The applied principle of analog to digital conversion is shown in fig. 2. The ADC is supplemented by a up-down counter which counts the incoming photon pulses. The counter output data is fed to a DAC generating a triangle signal which is added to the output of the TAC. The ADC converts the sum of both signals, giving the sum of the TAC value and the counter data. At the output of the ADC, the counter data is subtracted from the ADC result. This restores the original detection time measured by the TAC. Compared to a direct ADC conversion, this principle has a striking benefit: It converts the time of each photon in a different place on the ADC characteristics and therefore smoothes out the ADC errors. This smoothing works so efficiently, that we can use an ultra-fast ADC with only 12 bit non-missing code accuracy. Together with a speed-optimised TAC, we achieve a signal processing time of only 125 ns per photon.

The effect of the ADC error reduction is shown in fig. 3, left. An unmodulated light signal was recorded without error correction, and for a 7 bit and 9 bit width of the counter. Fig. 3, right, shows that the instrument response function is not substantially broadened by the error reduction.

Fig. 3: left: Unmodulated light recorded without error reduction, and with a counter data width of 7 bit and 9 bit right: Corresponding instrument response function for an electrical test signal

Fig. 4 shows how one TCSPC channel is expanded for operation with eight detectors. The expansion is based on the fact that it is unlikely to detect several photons in the same laser pulse period. This is a general condition for any TCSPC measurement and therefore not a restriction for the multi-detector configuration.

The single photon pulse from each detector is fed to a discriminator that responds when the detector has seen a photon. The discriminator outputs are encoded to generate a 3 bit channel number. The output signals of all detectors are merged into one output line by a summing amplifier. Thus, on the detection of each photon we get a detector pulse and a detector number.

The detector pulse is connected in the usual way to the CFD input of one TCSPC channel (fig. 2). When the TCSPC channel detects this pulse, it writes the ‘channel number’ bits into a data latch that controls the memory segment in which the photon is stored. Thus, in the TCSPC memory eight histograms corresponding to the individual detectors are built up. In the unlikely case that several detectors respond in the same laser pulse period, the encoder delivers an ‘invalid’ signal which inhibits the storing of the current photon. Therefore, the well-known pile-up distortion of the histogram is even smaller than with a single detector operated at the overall count rate of the eight detectors.

### 3. APPLICATION TO LASER-PULSE MAMMOGRAPHY

The TCSPC device was tested in the laser-pulse mammograph described in [6]. For the experiments described below, the mammograph was upgraded to four detection channels. When recording mammograms, the source-detector arrangement is scanned across the slightly compressed breast in two dimensions. Time-resolved transmittance is measured within 100 ms at each of 1000-2000 scan positions, 2.5 mm apart. Mammograms are recorded within 3-6 min. The diameter of the illuminated spot at the upper surface is about 3 mm, the detector fibre bundles are 4 mm in diameter.

All detectors were Hamamatsu R7400U-02 photomultipliers. Preamplifiers (50db) were used to compensate for the relatively low gain of the R7400U-02 detectors. The arrangement of the detector fibre bundles D1 to D4 is shown in
fig 5 (upper right). D1 was the direct channel opposite to the source fibre, D2, D3 and D4 were offset by 2 cm. Picosecond laser pulses from a 670 nm and a 785 nm diode laser (PicoQuant, Berlin) were multiplexed.

To test the system we used the phantom shown in fig. 5 (upper left). It consisted of a rectangular cuvette with several black wires of 1.7 mm diameter and one transparent and three black spheres of 8 mm diameter. The wires were arranged at different depths and orientations. The two longest wires ran parallel to and 0.6 cm apart from the front and rear surfaces of the scattering liquid, respectively. The black spheres were positioned 1.2 cm from the front plane, at the centre plane and 1.2 cm from the rear plane. The transparent (glass) sphere was placed at the centre plane. The inner thickness of the cuvette was 6.8 cm. As scattering liquid a mixture of whole milk and water with addition of a small amount of black ink was used. At 670 nm the reduced scattering and absorption coefficients were about 10 cm$^{-1}$ and 0.04 cm$^{-1}$, respectively, and thus typical of the optical properties of breast tissue.

Fig. 5 shows the result of a 65 by 53 pixel scan with a step size of 2.5 mm. The images were created from the time-of-flight distributions of the photons integrated over time and normalised at a common reference position. The different appearance of the phantom under different projection angles is clearly visible. The image coordinates are referenced to the coordinates of the source. The objects close to the source (lower sphere and wire running from upper left to lower middle of the image, s. fig. 5) do not change their position at different viewing angles. The other objects (e.g. upper sphere and wire running from lower left to upper middle) appear shifted depending on their depth and on the detector position.

Figs. 6b and c show the results of tomosynthesis [7, 6] based on the images corresponding to detector positions D1 to D3. For comparison, the image corresponding to D1 (s. fig. 5) is also included (fig. 6a). The raw images were added with appropriate offsets to synthesise images corresponding to a plane 0.8 cm below the source (fig. 6b) and a plane at the centre of the phantom (fig. 6c). The images show the black objects (wires and spheres) of the phantom more or less clearly. The contrast of objects in the synthesised plane remains unchanged while the contrast of objects in other planes is smeared out. There is, however, no indication of the glass sphere at the centre plane, not even in the corresponding tomosynthesis image.

European Conferences on Biomedical Optics, ECBO 2001, Munic, June 2001
Fig. 6: Tomosynthesis for localisation of objects at various depth using images representing total photon counts (s. fig. 5): Raw image derived from the central detector D1 (a), tomosynthesis of images recorded by detectors D1, D2 and D3 for a plane close to the source (b) and for the central plane (c).

Fig. 7: Tomosynthesis of images generated from photons in an early time window: Raw image derived from the central detector D1 (a), tomosynthesis of images recorded by detectors D1, D2 and D3 for the same planes as in fig. 6 (b, c).

Fig. 7 was obtained from photon counts in an early time window of the times-of-flight distribution of photons. The window was adjusted to contain 10% of all photons detected at a reference position. Although the images contain some noise due to the smaller number of photons, the glass sphere shows up. Adding images appropriately shifted for tomosynthesis enhances not only the objects in the synthesised plane but also reduces the noise. It should be noted that three projection angles are sufficient to triangulate the depth of an inhomogeneity of any shape. However, to obtain images of selected planes a much larger number of projection angles must be used.

Fig. 8: Mammograms of a volunteer recorded simultaneously at four projection angles. The images were generated from photon counts in a late time window. The arrangement of the mammograms corresponds to that of the detectors D1-D4 (s. fig. 5).
Fig. 8 shows 51 by 21 pixel mammograms of a breast of a volunteer recorded at 785 nm by the four-channel setup. The mammograms were obtained by using photon counts in the 8th of 10 consecutive time windows each containing 10% of all photons detected at a reference position. It was shown previously that mammograms based on this time window are essentially free of edge effects without requiring correction algorithms to be applied [6]. The mammograms of fig. 8 corresponding to the different projection angles show slight differences. Since the pattern of the superficial blood vessels does not change significantly with the projection angle it can be concluded that these vessels were close to the source plane.

4. CONCLUSIONS
The TCSPC device described above has successfully been used to record time-resolved transmittance of phantoms and a compressed breast at high speed, high accuracy and high time resolution.

5. REFERENCES