The use of entangled-photon imaging in optical biopsy: a feasibility study

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ABSTRACT

We examine a novel diagnostic method suitable for optical biopsy, i.e., the noninvasive *in vivo* detection of malignant lesions in human tissue. Entangled-photon imaging is an emerging technology based upon the use of non-classical sources of light such as optical parametric oscillators (OPO). These sources generate above-threshold signal and idler beams that have intensity fluctuations highly correlated in space and time (twin beams)[1,2]. It has been shown that low-intensity OPO's make possible high sensitivity absorption measurements of weak targets, below the shot-noise limit[3,4]. The direct use of this technology for optical biopsy is severely restricted by the large amount of scattering noise associated with light-tissue interaction. We report what is, to our knowledge, the first feasibility study on a differential wavelength, OPO-based setup targeted for mammography. Constraints related to the entanglement time, OPO selection and background suppression are analyzed. The paper concludes with a review of future developments and challenges.

SUMMARY

The proposed setup has the following configuration (fig.1):
a) the *primary source unit* including:
-a frequency-doubled Nd:YAG laser which generates a cw pump beam centered in the near IR region .
-a high-finesse mode-cleaning cavity which supplies 75% transmission of the pump beam.
-a tunable OPO with potassium titanyl phosphate (KTP) crystal as a gain medium.
-a half-wave plate and a polarizer cube (CP) which separate the signal and the idler beams in two orthogonal paths as shown.
b) the *auxiliary source unit* including:
-a single-mode laser diode (LD) which is amplitude modulated by an electro-optic modulator (AM) and chopped by a mechanical chopper (CH) at a frequency of f_{CP} = 650 Hz.
-a dichroic mirror (DM_1) on which the modulated beam and the signal beam are superimposed.
c) the *detection and demodulation unit* including:
-a dichroic mirror (DM_2) which filters out the background modulation introduced by the auxiliary source unit.
-a pair of identical photodetectors (D_1 and D_2) capture the exit signal and idler beams which are then subtracted and demodulated at (DEM).
For clarity, the beam coupling components are omitted from the layout.
The twin beams remain correlated within a time window called "entanglement time" (τ_e), typically in the order of picoseconds[5]. It follows that, if τ_t represents the mean time-of-flight of the signal photon in tissue, a necessary imaging condition amounts to:

$$\tau_e \geq \tau_t \tag{1}$$

Let $< Z >$ denote the mean penetration depth of the signal photon and let s be the transverse separation between the input and output fibers. Assuming a random-walk model for photon migration in tissue[6], we show that condition (1) leads to :

$$\tau_e \geq 9.705.[1/(v.\mu_s')].< Z >^2.s^{.2214} \tag{2}$$

in which v represents the speed of light in tissue and μ_s' the reduced scattering coefficient.
The entanglement time depends upon the OPO crystal length and the pump beam waist[7]. In this context, relation (2) serves as a basis for tailoring the selection of the laser pump and OPO modules, given the required mean penetration depth and the relative position of the input and output fibers.
In general, if the laser pump optical frequency is 2ω, the twin beams are centered about ω, which must match the absorption frequency of the target. Efficient suppression of the background noise generated by multiple scattering events in the tissue requires tuning the probing beam at two frequencies $\omega_{1,2}$ and subtracting the corresponding differential signal[3,8]. It follows

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that the OPO tuning bandwidth must supply the range of frequencies that allow the difference between the target absorption signal and the background one to be maximized.
Let $\langle I_{b,s} - I_{b,i} \rangle_{1,2}$ be the weighted fluctuation of the differential signal generated by the background noise and measured between the two probing modes at frequencies $\omega_{1,2}$:

$$\langle I_{b,s} - I_{b,i} \rangle_{1,2} = [(I_{b,s} - I_{b,i})_1.\omega_2 - (I_{b,s} - I_{b,i})_2.\omega_1]/(\omega_2 - \omega_1) \quad (3)$$

Let η represent the quantum efficiency for D_1 and D_2, t_d the mean detection time, A the photodetector area and σ^2 the energy variance associated with statistical fluctuations in $(I_{b,s} - I_{b,i})_{1,2}$. If the twin beams are perfectly entangled, we show that the differential photocount noise assumes the form:

$$N_{12} = 2.\pi.\eta.[(\omega_2 - \omega_1)/(\omega_1.\omega_2)].\ [A.\langle I_{b,s} - I_{b,i} \rangle_{1,2}.t_d + 2.\pi.\eta.(\omega_1 + \omega_2).\sigma^2/(\omega_1.\omega_2)] \quad (4)$$

with Planck's constant normalized to unity. Relation (4) sets the ideal bound for the overall detection capability of the system, below the conventional shot-noise limit.

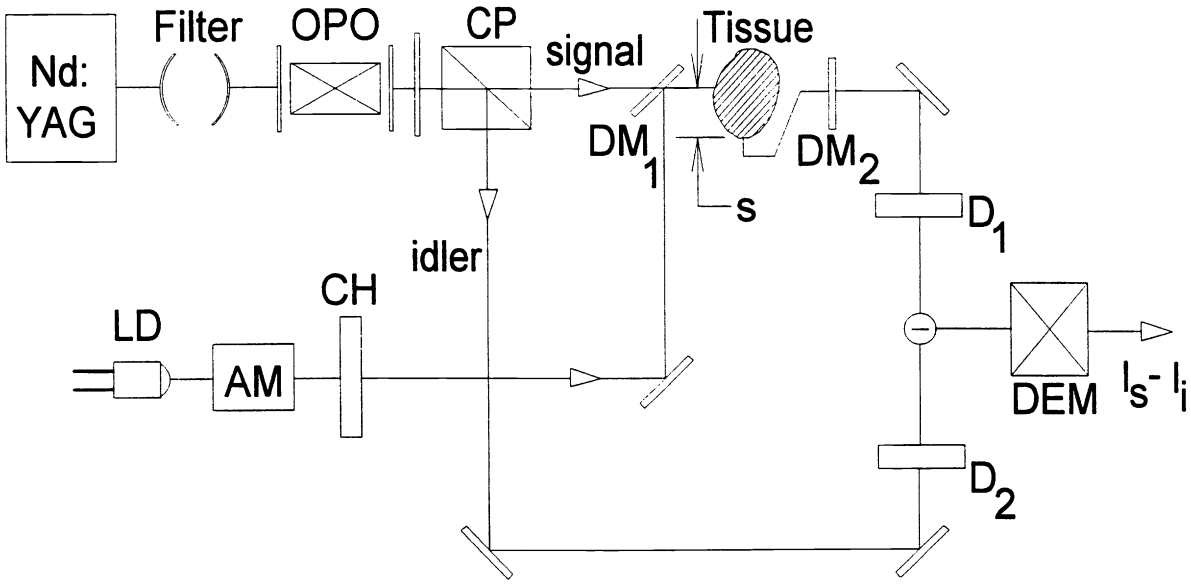

Fig. 1

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