

LASER APPLICATIONS AND OTHER TOPICS IN QUANTUM ELECTRONICS https://doi.org/10.1070/QEL16208

On angiography with a Thomson laser-electron X-ray generator

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Abstract. **We consider a possibility of application of laser-electron X-ray generators for diagnosing the vessel status of internal organs. It is shown that modern lasers and linear accelerators can be used for the development of angiographic instruments of a new type with an increased spatial and temporal resolution while maintaining or reducing the radiation load on the patient and medical staff. Such improvements in diagnostic and ambient factors cannot be achieved with the use of X-ray tubes. All particular estimates and calculations have been performed for a contrast agent based on iodine compounds.**

Keywords: *laser-electron X-ray generator, Thomson X-ray generator, laser medicine, angiography.*

1. Introduction

The idea of the practical use of a laser-electron X-ray generator (LEXG) based on Thomson scattering in medicine was first suggested by F. Carroll, in whose works this idea received experimental development [1, 2]. Currently, LEXGs operate in two research centres [3, 4], and Lyncean Technologies Inc. [5] offers them on a commercial basis. As a rule, we are referring to a multi-purpose installation, which is operated, similar to accelerators or large laser systems, by multiple users. The advantages of LEXGs are a significantly smaller occupied area and cost. This has stimulated works aimed at using LEXGs in traditional X-ray technology, including security systems [6], therapy [7] and medical diagnostics [8]. This paper discusses the requirements to an LEXG and its components, which will allow obtaining the images of human vessels with an exposure of no more than 1 ms, while maintaining the required spatial resolution, and thus surpassing the angiographic standards of modern medicine.

Angiography represents visualisation of the walls and lumen of blood vessels by X-raying, with the administration of a contrast agent into the vessels. Let us estimate the number of X-ray quanta required to produce a single image on the example of an idealised object having thickness L and containing a blood vessel with diameter δ (Fig. 1).

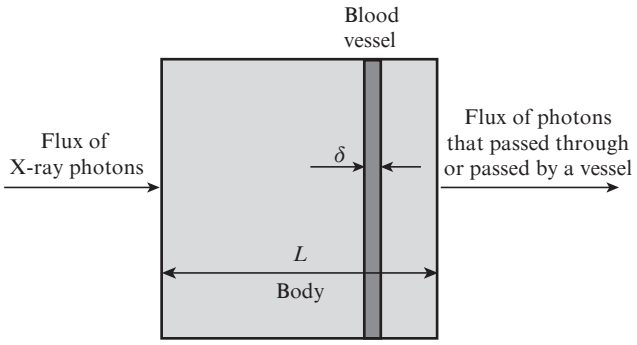

Figure 1. Determination of the exposure dose in angiography.

The condition for vessel recording by a detector against the background of a noncontrasted object can be written in the form

$$\sqrt{N_\delta \exp(-\mu L)} < c\{N_\delta \exp(-\mu L) - N_\delta \exp[-\mu L - (\kappa - \mu)\delta]\}, \quad (1)$$

whence the minimum exposure dose per resolution element appears as

$$N_\delta(E) = \exp(\mu L)\{c[1 - \exp[-(\kappa - \mu)\delta]]\}^{-2}, \quad (2)$$

where the linear absorption coefficients outside (μ) and inside (κ) of a contrasted vessel depend on the energy E of an X-ray photon, and c is the contrast ratio defined by the particular diagnostic task and detector. Hence, we find that for obtaining an image containing $M \times M$ pixels, we need

$$N(E) = M^2 \exp(\mu L)\{c[1 - \exp[-(\kappa - \mu)\delta]]\}^{-2} \quad (3)$$

photons. Since Eqn (3) describes the number of quanta incident to the region under study, it can be applied to any monochromatic source and object containing a contrasted component of cylindrical shape.

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Received 16 September 2016; revision received 5 December 2016
Kvantovaya Elektronika **47** (1) 75–78 (2017)
Translated by M.A. Monastyrskiy

2. Coronary angiography. Exposure and radiation dose

Before evaluating, according to Eqn (3), the number of photons required for obtaining a single image of coronary heart vessels, we give general information about the methods of coronary angiography. In one form or another, this procedure is annually assigned to millions of patients. In recent decades, selective coronary angiography is deemed as a 'gold standard' of diagnostics of heart vessels. It represents an invasive procedure requiring administration of a contrast agent through a catheter, which is held through a large artery directly into the mouth of coronary vessels, which in itself is associated with the risks of dangerous consequences for health. The rapid development and expansion over the past 15 years of several noninvasive methods [9], in which the contrast agent is introduced by means of common intravenous administration, has allowed reducing the probability of undesirable consequences of the 'gold standard'. In parallel, the possibilities of improving the quality of angio-images through the use of tunable monochromatic synchrotron radiation sources [10], X-ray tubes with a rotating anode [11], and LEXGs [12] have been investigated. The effect here is achieved by subtracting the digital images obtained on both sides of the K-edge of photoabsorption of the contrast agent, commonly iodine. However, in terms of a number of indicators, including temporal and spatial resolution, as well as cost, noninvasive methods are inferior to invasive methods and require further development [13–15]. Below we define the LEXG parameters that offer a new angiography method which is superior to the 'gold standard' method with respect to a number of key parameters.

We first define the exposure dose, i.e. the number $N(E)$ of X-ray quanta needed to obtain a single angiogram. To this end, it is necessary to substitute in Eqn (3) the parameters adopted in medical radiography: patient body thickness L = 20 cm, vessel diameter δ = 1 mm. The tissue absorption coefficient μ is taken from the tabular data for water [16], the absorption coefficient κ of blood vessels, contrasted by an iodine-containing dye, is expressed through the mass concentration of iodine η or the Hounsfield number N_{CT}, which is often used in X-ray diagnostics. The relationship between them is determined by the equations [17]:

$$\kappa - \mu = \left(\frac{\mu}{\rho}\right)_I \rho_I = \eta\left(\frac{\mu}{\rho}\right)_I \rho_{H_2O}, \quad \eta = \frac{\rho_I}{\rho_{H_2O}}, \tag{4}$$

$$N_{CT} = \frac{\kappa - \mu}{\mu} 1000, \quad \frac{N_{CT}}{1000} = \eta\left(\frac{\mu}{\rho}\right)_I \left(\frac{\mu}{\rho}\right)^{-1}_{H_2O}, \tag{5}$$

where ρ is the tabulated density of water and iodine; and μ/ρ are their mass absorption coefficients tabulated in [16].

The exposure dose $N(E)$ is shown in Fig. 2 for various mass concentrations η of iodine in a contrasted vessel in the energy range of quanta of 30–120 KeV; it is assumed that the image contains 0.06 megapixels (M = 256), while the contrast ratio is c = 0.6. Figure 3 allows an easy transition from η to the Hounsfield number N_{CT}. The results of its measurements using the data of angiographic examinations in various medical centres vary: N_{CT} = 50–500 [18, 19]. This range corresponds to the mass concentrations shown in Fig. 2.

As can be seen from Fig. 2, the exposure dose is minimal in the energy range of quanta from 35 to 45 Kev. At reasonable concentrations of iodine, it may constitute 10^{11}–10^{13} photons. We present here such a large interval since the experimental data on the iodine concentration have been obtained on serial angiographs using nonmonochromatic radiation. This exposure dose does not exceed the radiation dose that is applied in modern angiography and coronarography, and amounts to a few millisieverts [20].

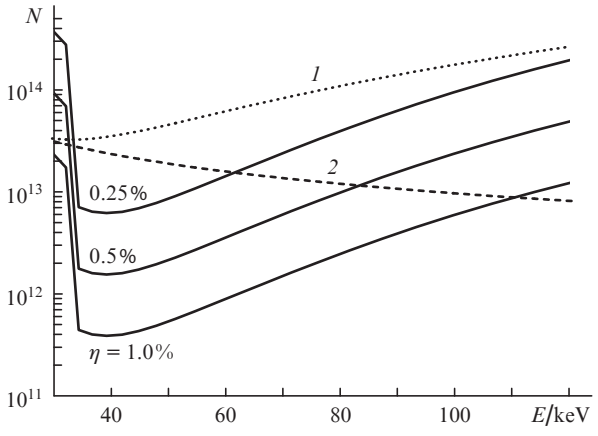

Figure 2. Number N of X-ray photons needed to produce a single angio-image for 1 ms [see Eqn (3)] as a function of photon energy E at δ = 1 mm, L = 20 cm, c = 0.6, and the number of pixels 256×256 (M = 256). Dependences (*1*) and (*2*) have been obtained at a radiation load of 2 mSv (the object weight is 75 kg) for two models of radiation transfer in tissues: (*1*) with neglect of multiple scattering of photons and (*2*) with total absorption in the process of multiple scattering.

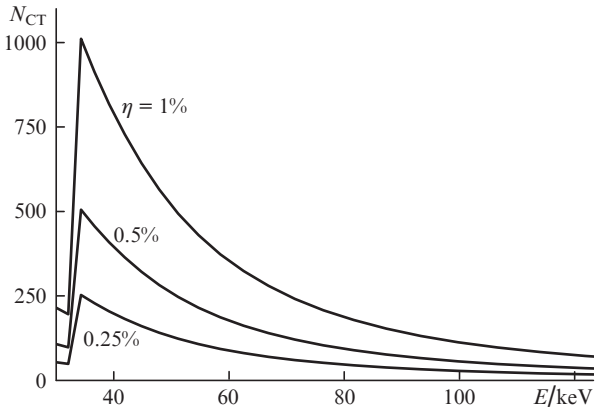

Figure 3. Dependences of the Hounsfield number N_{CT} on the photon energy at different mass concentrations of iodine (see also [17]).

Thus, the minimum X-ray source output for millisecond coronary angiography constitutes 10^{11}–10^{13} photons, and the optimal energy exceeds the K-edge of iodine photoabsorption by several keV. The latter is qualitatively confirmed by the measurements [21] implemented in different operation regimes of serial angiographs.

Below we consider the requirements for LEXGs, which allow obtaining the required exposure dose for 1 ms, which, under the condition of high spatial resolution, will significantly improve the quality of vascular system visualisation.

3. General information about LEXGs

The schematic diagram of an LEXG is shown in Fig. 4. X-ray radiation is generated in the accumulator of photons in the

process of Thomson scattering of laser pulses on relativistic electron bunches generated in the form of trains by a linear accelerator. The trains of pulses follow with a frequency ν_e and contain n_e bunches with N_e electrons in each bunch. The photon accumulator enhances (in the resonator [22]) or accumulates (in the circulator [23]) the pump laser pulses (the driver). The number of accumulated quanta is $N^s_{las} = \xi N_{las}$, where N_{las} is the number of quanta in the driver pulse; and ξ is the coupling coefficient. Below in our estimates, we use $\xi = 10^3$ for the resonator, and $\xi = 0.25$ for the circulator, which currently may be considered as quite realistic values [24, 25].

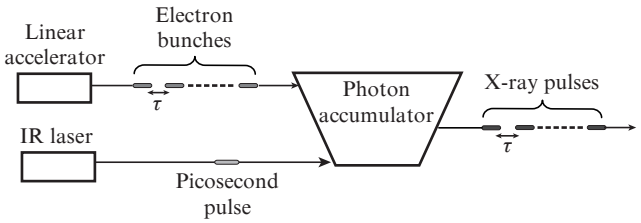

Figure 4. Schematic diagram of LEXG operation.

The energy E and the flux Φ of X-ray quanta emitted in the Thomson scattering of laser photons on electrons depend on the geometry of colliding electron and laser bunches. For colliding bunches, these parameters can be represented in the form [26]

$$E \approx 4\gamma^2 E_{las}, \ \Phi = n_e \nu_e N = n_e \nu_e N_e N^s_{las} W = \frac{1}{eE_{las}} I \Sigma^s W, \quad (6)$$

where $N = N_e N^s_{las} W$ is the number of X-ray quanta per pair of colliding particles; Σ^s is the laser pulse energy stored in the optical accumulator; e is the electron charge; I is the accelerator current; E_{las} is the laser photon energy; W is the normalised X-ray radiation output that is used in calculating both the total output of X-ray radiation, and spectral and angular distributions. Since the point in question is radiography, the total yield of X-rays is of interest. In this case, $W \approx \sigma_T/s$ by the order of magnitude, where $\sigma_T = 6.6 \times 10^{-25}$ cm^2 is the Thomson cross section, and s is the area onto which the colliding beams are focused. For example, if the area diameter is 30 μm,

$$W \approx \sigma_T/s = 0.93 \times 10^{-19}. \quad (7)$$

The overall picture of Thomson scattering of a laser pulse on a relativistic bunch differs both from scattering on a stationary electron bunch and from scattering of a photon on a relativistic electron. In the conditions close to frontal collision, the scattering diagram is elongated along the direction of the electron velocity and is concentrated in a narrow angular range $\delta\theta \approx 1/\gamma$, where $\gamma = E_e/(m_e c^2)$ is the relativistic factor, and the spectrum width constitutes 15%–20%. The energies of electrons, laser and X-ray photons are related by $E = 4\gamma^2 E_{las}$. If E is close to the K-edge energy of iodine (33.2 Kev), and E_{las} – to 1 eV, then $\gamma \approx 100$ and $\delta\theta \approx 1/\gamma = 10$ mrad. The details of angular and spectral distributions are determined by the distribution functions of colliding particles in momentum. Derivation of exact formulas for W and also for the LEXG spectrum is available in paper [26] and references therein.

4. Laser-electron generator for pulsed angiography

Owing to beam collimation and not large LEXG spectrum width, virtually entire LEXG radiation may be directed onto the body surface under investigation and used for diagnostics. Next, similarly to (6), the total number of X-ray quanta emitted per LEXG pulse appears as

$$N = \frac{Q}{e} \frac{\Sigma^s}{E_{las}} W, \quad (8)$$

where Q is the total charge per pulse. In accordance with the principal scheme (see Fig. 4), a single LEXG pulse contains n_e X-ray pulses, which correspond to n_e electron bunches generated within a single cycle of accelerator operation, the duration of which is typically less than 1 ms. Since, in our case, virtually entire LEXG radiation is used, Eqn (7) is valid for W estimation. Finally, given the area of interacting beams, the total number of emitted photons is

$$N = \sigma_T \frac{Q}{es} \frac{\Sigma^s}{E_{las}}. \quad (9)$$

Thus, the X-ray output of a pulsed LEXG is equal to the product of two factors: first of them only refers to the electron beam, the second – only to the laser (in fact, it is the number of laser photons stored within an optical accumulator). Equation (9) may be used in the LEXG design to optimise the parameters of the accelerator, laser and angiographic installation as a whole in order to achieve the exposure dose of $N \sim 10^{11}$–10^{13} that is necessary for coronary angiography (see Section 2). Since that problem has never been addressed previously, we have analysed from this point of view the experimental works and LEXG projects oriented to the energy of 12–20 keV, published in recent years. Estimates have been made for IR pump lasers with a wavelength of 1.03 μm, i.e. in Eqn (9) $E_{las} = 1.2$ eV, with the laser pulse energy in the optical accumulator $\Sigma^s = 10$ mJ. Two systems turned out the closest to the target. The basic parameters of these systems – the charge and the cross-sectional area of electron beam focus – are given in Table 1.

Table 1. Expected X-ray output of an LEXG during 1 ms for accelerators developed in some laboratories [25, 27].

Linear accelerator from papers	s/μm^2	Q/μC	N	
			calculated [see (9)]	required (see Section 2)
[25]	3.6	0.01	6×10^{10}	10^{11} – 10^{13}
[27]	100	10	2×10^{12}	10^{11} – 10^{13}

The represented data show that the parameters of the accelerators being currently under design are close to those required for millisecond angiography of coronary vessels. At the same time, it is easy to verify that the required millisecond exposure dose cannot be achieved using the X-ray tube. This is a direct consequence of the directivity of LEXG radiation.

5. Discussion and conclusions

Modern angiographs, in which the radiation sources are X-ray tubes, allow observation of the status of heart blood

vessels in dynamics with a resolution in time of ~0.01 s, and in space of 0.15–0.2 mm [14]. As shown above, the proposed approach employing LEXGs allows these parameters to be surpassed. In particular, the expected exposure does not exceed 1 ms, while the transverse resolution is significantly smaller than the electron beam diameter (10–100 μm) at a resolution according to the contrast (depth) of 1 mm. Going beyond these limits requires an increase in iodine concentration or LEXG intensity, which also seems feasible with the use, for example, of a storage ring [8]. We also note the possibility of spectral retuning of the LEXG beam. Such a retuning allows preservation of LEXG advantages when switching from iodine to gadolinium and other contrast agents. Table 2 compares the main characteristics of X-ray tubes and potential possibilities of LEXGs as applied to coronary angiography. Given the considerable cost of the LEXG-based angiograph prototype, we deem it advisable to conduct, before its manufacturing, model experiments on the channels of synchrotron sources or X-ray tubes. For subtraction angiography, similar studies have been performed in [10, 11].

Table 2. Characteristics of radiation sources for coronary angiography.

Characteristics	Source	
	X-ray tubes	LEXG
Radiation directivity	–	10 mrad
Temporal resolution	10 ms	<1 ms
Spatial resolution	0.15–0.20 mm	<0.1mm (lateral) 0.15 mm (in depth)
Videorecording	Possible	Impossible
Safety of background X-ray radiation	Dangerous	Safe
Spectrum optimisation for contrast agents	Impossible	Possible

Acknowledgements. The authors are grateful to I.A. Artyukov, A.P. Bogatov, A.E. Drakin, Yu.A. Mikhailov, V.L. Polunin and R.M. Feshchenko for discussions.

The work was supported by the Fundamental Research Programme of the Presidium of the Russian Academy of Sciences 'Fundamental and applied problems of photonics and physics of new optical materials'.

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