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Radiation therapy at compact Compton sources

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ABSTRACT

The principle of the compact Compton source is presented briefly. In collision with an ultrarelativistic electron bunch a laser pulse is back-scattered as hard X-rays. The radiation cone has an opening of a few mrad, and the energy bandwidth is a few percent. The electrons that have an energy of the order of a few tens of MeV either circulate in storage ring, or are injected to a linac at a frequency of 10–100 MHz. At the interaction point the electron bunch collides with the laser pulse that has been amplified in a Fabry-Perot resonator. There are several machines in design or construction phase, and projected fluxes are 10^{12} to 10^{14} photons/s. The flux available at 80 keV from the ThomX machine is compared with that used in the Stereotactic Synchrotron Radiation Therapy clinical trials. It is concluded that ThomX has the potential of serving as the radiation source in future radiation therapy programs, and that ThomX can be integrated in hospital environment.

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Introduction

The use of synchrotron radiation (SR) has opened new avenues in biomedical research [1,2]. 3-dimensional structures of macroscopic objects can be imaged with a resolution of a few microns and elemental distributions in cells can be mapped with spatial resolution better than 100 nm. These are made possible by the high intensity and natural collimation, energy tunability, and partial transverse coherence of SR, which are utilized in phase contrast imaging and in X-ray micro-spectroscopy.

The limited availability of SR has curtailed the applications mostly to basic research, and to methods development with the distant goal of clinical use. Only a few programs have reached the stage of human studies, the first being coronary angiography with intravenous injection of the Iodine contrast agent. For these studies a new method, K-edge subtraction (KES) imaging was developed, and has been applied in studies of circulation and ventilation in animal models. The radiation dose in KES imaging is within clinically acceptable limits, and there are many potential applications in functional imaging of human organs [3]. Phase contrast imaging in the propagation mode has been used in mammography, demonstrating enhanced contrast resolution [4]. Clinical trials for radiation therapy of brain tumors is underway at the European Synchrotron Radiation Facility (ESRF) using the Stereotactic Synchrotron Radiation Therapy (SSRT) method [5].

There have been many projects for transfer of phase contrast imaging methods to clinical environment without the use of large scale synchrotrons see Ref. [2]. In most cases, rotating anode or micro-focus X-ray tubes are used as radiation sources, and new concepts have been introduced to overcome the anode cooling problems. The anode may be replaced by a wire or thin stream of liquid metal, and potentially these sources are at least an order of magnitude brighter than conventional X-ray tubes. However, even with these improvements the exposure times would be prohibitively long for most modalities of human imaging.

Probably the only way to break out of the confinement of scarce synchrotron radiation at large facilities on one hand, and the insufficient X-ray flux from tube sources on the other, is the introduction of compact SR sources for dedicated medical imaging and/or therapy facilities. Most concepts of compact sources are based on the use of very high-field magnets in small storage rings, but still the X-ray energies remain too low for most imaging applications, and totally exclude radiation therapy [6]. A different way of producing X-rays from a high-energy electron beam is to collide the beam with a laser beam, which is scattered back in a process that is usually called inverse Compton scattering [7]. Originally, the method was introduced in high-energy particle physics, but more than a decade ago it was demonstrated that a linac-laser combination produces short, intense X-ray pulses of sufficient intensity for medical imaging [8]. In recent years, many projects have been

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initiated, and the first and the only compact inverse-Compton source on market is the Lynceantech machine, which delivers X-rays in the range 15–35 keV [9]. It is expected that these sources will become standard equipment for clinical imaging in large hospitals.

Inverse Compton scattering and compact Compton sources

The inverse Compton scattering process is shown schematically in Fig. 1(a). In the collision, the energy of the electron is transmitted to the photon. The kinematics of the process can be described as collision of two particles, or using the synchrotron radiation analogy, where the electron travels through the “micro-undulator” of the electromagnetic field of the photon. The energy of the scattered photon in the laboratory frame is [10]:

$$\mathrm{E_X} = 2\gamma^2 \mathrm{E_L}(1+\cos\theta_c)\Big/\left(1+\gamma^2\theta^2\right), \tag{1}$$

where γ is the electron energy in the units of its rest energy, $\gamma = E_e/mc^2$, E_L the energy of the incoming photon, θ_c and θ respectively the incident angle and the scattering angle with respect to the electron trajectory. In derivation of (1), it is assumed that $\gamma \gg 1$ and $E_L \ll mc^2$. Expression (1) shows that there is an univocal relation between the energy and the emission angle of Compton photons, the hardest radiation being concentrated on-axis where $\theta = 0$. So in head-on collision (i.e. when $\theta_c = 0$), the maximum energy of the scattered photons is $\mathrm{E_X^{max}} = 4\gamma^2\mathrm{E_L}$. For instance, 0.1 nm (12.4 keV) X-rays are produced in collision of a laser pulse of 1.0 μm wavelength ($\mathrm{E_L} = 1.24$ eV) with a 25 MeV electron bunch. The normalized spectral cross section $\sigma_\omega(\omega)$, where $\omega = \mathrm{E_X}/\mathrm{E_X^{max}}$, is represented on Fig. 1(b) and can be written as [11]:

$$\sigma(\omega) = 3\Big/2\left(1-2\omega+2\omega^2\right). \tag{2}$$

For electron/laser collisions in the x–z plane, the number of X-rays emitted per second is:

$$\mathrm{F_{tot}} = \frac{\sigma_T N_e N_L f_{rep}\,\cos(\theta_c/2)}{2\pi\,\sqrt{\sigma_{ye}^2+\sigma_{yL}^2}\,\sqrt{\sigma_{xe}^2+\sigma_{xL}^2+\sin^2(\theta_c/2)\left(\sigma_{ze}^2+\sigma_{zL}^2\right)}} \tag{3}$$

where σ_T is the Thomson cross-section, N_e the number of electrons per bunch, N_L that of photons per laser pulse, and f_{rep} the repetition rate of collisions. The transverse and longitudinal sizes of the (Gaussian) electron bunches and laser pulses at the interaction point are given by $\sigma_{xe,ye} = \sigma_e$, $\sigma_{xL,yL} = \sigma_L$, σ_{ze} and σ_{zL} respectively. The charge of an electron bunch may be 0.1–1.0 nC, so N_e is about 10^9, and the repetition rate is between 10^7 1/s and 10^8 1/s. To give a concrete example with feasible values of $f_{rep} = 20$ MHz, 1 nC electron bunches, and 10 mJ laser pulses of 1 μm wavelength, F~10^{13} ph/s is expected when $\sigma_e = \sigma_L = 40$ μm.

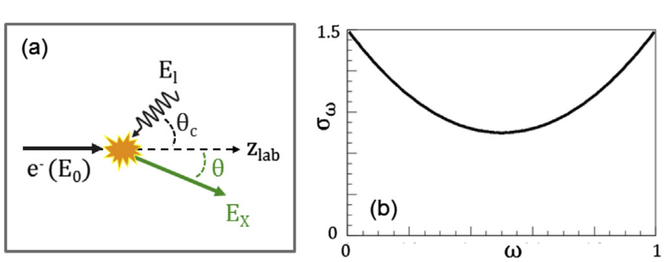

Figure 1. Inverse Compton scattering process (a) and normalized Compton cross section as a function of $\omega = \mathrm{E_X}/\mathrm{E_X^{max}}$ (b).

There are basically two machine designs: the storage ring scheme and the linac scheme, illustrated in Fig. 2. Each one has advantages and technical difficulties. In the storage ring scheme each stored electron bunch interacts many times with a laser pulse. The bunch is blurred by intra-beam scattering and Compton scattering, so that it is ejected and replaced by a new bunch. The repetition frequency is determined by the ring circumference, for instance f_{rep} is 20 MHz when the circumference of 15 m. In the linac scheme each electron bunch interacts only once, and f_{rep} is determined by the electron gun, which is necessarily a superconducting device. The advantage of the linac scheme is the small transverse size of the electron bunch, so that the transverse coherence of the X-ray beam is high. However, continuous cryogenic operation of the electron gun with high repetition rate entails many technical and operational challenges, particularly in clinical environment. In both schemes, intense laser pulses at frequency f_{rep} are created by the amplification of pulses from a low average power pulsed laser which are then sent in a high gain optical resonator for a second amplification stage. For instance, a 10 mJ laser pulse circulating at 20 MHz frequency corresponds to 0.2 MW stored in the optical cavity.

The properties of the X-ray beam depend on the energy spreads of the electron and laser beams, $d\gamma$ and dE_L, respectively, and on the divergences of the beams. The relative bandwidth of on-axis photons due to energy spreads of the beams is approximately $dE_L/E_L{+}2d\gamma/\gamma$, which is typically of the order of 10^{-3}. More important are the effects of beam divergences, and the dominant factor is the electron beam divergence $\sigma'_{xe,ye} = \sigma'_e$, yielding $dE_X/E_X = 2(\gamma\sigma'_e)^2$. With typical values of $\gamma = 100$ (50 MeV electron beam) and $\sigma'_e = 1$ mrad, the relative energy spread of the X-ray beam is 2%.

The brilliance B of the source is given as the number of X-ray photons per second, per mm^2 of source, in solid angle of 1 mrad^2, and in 0.1% spectral bandwidth. Considering only the dominant term due to the electron beam divergence, on-axis B is proportional to $F\gamma^2/\varepsilon_N$, where $\varepsilon_N = \gamma\sigma_e\sigma'_e$ is the normalized (tranverse) emittance of the electron beam [10]. In the same way as for the degree of transverse coherence, the electron beam emittance is the crucial factor for the X-ray beam brilliance.

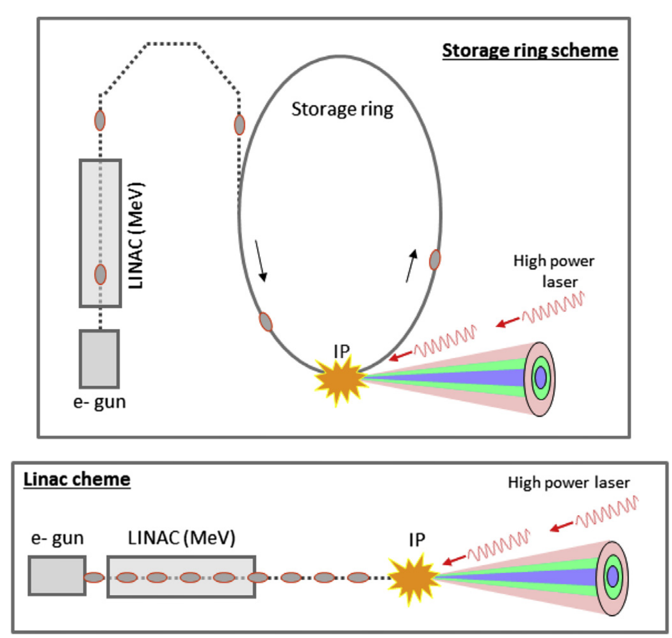

Figure 2. Basic schemes of high-flux compact Compton sources. IP is the interaction point of the laser pulse with the electron bunch.

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