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BRIEF COMMUNICATIONS

Feasibility of Using Laser Ion Accelerators in Proton Therapy

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Abstract—The feasibility of using laser plasma as a source of high-energy ions for the purposes of proton therapy is discussed. The proposal is based on the efficient ion acceleration observed in recent laboratory and numerical experiments on the interaction of high-power laser radiation with gaseous and solid targets. The specific dependence of proton energy losses in biological tissues (the Bragg peak) promotes the solution of one of the main problems of radiation therapy, namely, the irradiation of a malignant tumor with a sufficiently strong and homogeneous dose, ensuring that the irradiation of the surrounding healthy tissues and organs is minimal. In the scheme proposed, a beam of fast ions accelerated by a laser pulse can be integrated in the installations intended for proton therapy. *© 2002 MAIK "Nauka/Interperiodica".*

Recently developed compact lasers capable of generating ultrashort pulses in the multi-terawatt and petawatt power range have found progressively wider applications [1]. Thus, it was proposed to use these lasers to create new types of charged particle accelerators [2], sources of hard X-ray and gamma radiation [3], and charged particle injectors [4], as well as to apply them to the problem of inertial confinement fusion (in the framework of the concept of fast ignition of fusion targets with the use of laser-accelerated electron [5] or ion [6] beams). The above applications are based on the fact that the nonlinear interaction of high-power laser radiation with matter is accompanied by the efficient conversion of laser energy into the energy of fast particles. The generation of collimated fast ion beams was observed in many experiments on the interaction of ultrashort laser pulses with gaseous and solid targets [7] and was thoroughly investigated using multidimensional particle-in-cell computer simulations [8]. In present-day experiments [7], the energy of fast electrons (protons) attains several hundreds (tens) of MeV. Computer simulations show that, by optimizing the parameters of a petawatt laser pulse and a target, it is possible to achieve a protons energy of about several hundreds of MeV [8].

In this paper, we discuss the feasibility of using laser plasma as a source of high-energy ions for the purposes of hadron therapy. Hadron therapy is a constituent part of radiation therapy, which makes use not only of high-energy ion beams, but also of pi mesons, neutrons, electron beams, and X-ray and gamma radiation to irradiate cancer tumors (for details, see [9] and the literature cited therein). Generally, surgical removal, chemotherapy, and radiation therapy are applied in parallel to treat cancers. In developed countries, the radiation therapy is applied to more than one-half of oncological patients. After more than 40 years of experimental research, clinical centers of proton therapy (PT), intended for the treatment of up to 1000 patients per year, are now being actively developed. Three such centers have already been put into operation; in the nearest future, there will be nine PT centers [10]. Each of them is equipped with a special medical proton accelerator, from where proton beams are delivered to three to five treatment rooms. A mandatory and the most expensive attribute of such centers is the Gantry system for the multiple-field irradiation of the lying patient (Fig. 1a).

Proton therapy has a number of advantages. First, a proton beam is insignificantly scattered by atomic electrons, which reduces the irradiation of healthy tissues located on the side of the tumor. Second, the deceleration length of protons with a given energy is fixed, which allows one to avoid undesirable irradiation of healthy tissues behind the tumor. Third, the presence of a sharp maximum of proton energy losses in tissues (the Bragg peak) provides a substantial increase in the radiation dose in the vicinity of the beam stopping point (see, e.g., [9]). Up to the present time, conventional particle accelerators (synchrotrons, cyclotrons, and linacs) have been used to produce proton beams with the required parameters. The use of laser accelerators seems to be very promising because of their compactness and additional capabilities of controlling the proton beam parameters.

There are two versions of using laser accelerators. In the first version, a conventional accelerator is replaced with a laser one. In the second version, which seems to be more attractive, laser radiation is delivered to a target, where it is converted into fast ions. The target is situated directly at the entrance to the treatment room, which is especially important for the design of the Gan-

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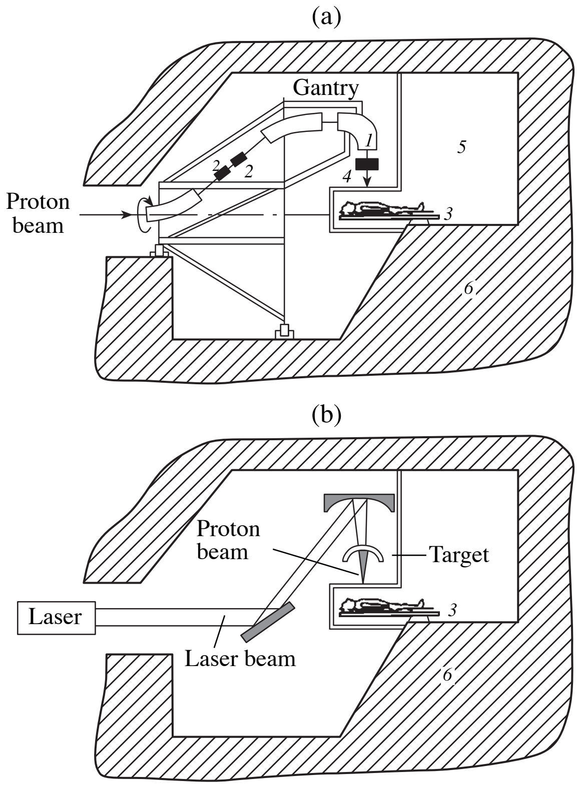

Fig. 1. (a) Conventional Gantry system: (*1*) deflecting magnets, (*2*) quadrupole lenses, (*3*) positioner, (*4*) dose delivery system and dose monitoring, (*5*) treatment room, and (*6*) concrete protection; (b) personal laser accelerator.

try system. Note that the Gantry system is not only expensive, but also very large (6–8 m in diameter and 10–12 m in length) and heavy (100 t or higher) facility. The weight and cost of the system are mainly determined by a powerful and very precisely made magneto-optical system rotating as a single entity. The use of the second version will allow one to both simplify the system and substantially reduce its cost. In this case, the central accelerator, the beam transportation channels, and the larger part of the Gantry magnetic system become unnecessary. One of the possible engineering solutions is shown in Fig. 1b.

The requirements for the main parameters of a medical proton beam (a beam intensity of $(1–5) \times 10^{10}$ proton/s and a maximum proton energy of 230–250 MeV) can easily be satisfied with the use of modern acceleration technique. At the same time, the other two requirements—the beam must be highly monoenergetic, $\Delta\mathcal{E}/\mathcal{E} \leq 10^{-2}$, and the duty-factor (the useful time fraction of a pulsed beam) should be no worse than 0.3 (otherwise, the repetition rate must be no lower than several Hz)—are rather difficult to satisfy.

A comparison of the ion energy spectrum required for medical applications ($\Delta\mathcal{E}/\mathcal{E} < 10^{-2}$) with that observed in laboratory and numerical experiments on the interaction of laser radiation with matter shows that the available ion spectra are far from being monoenergetic. It follows from [7, 8] that, at energies lower than the maximum energy ($\mathcal{E} < \mathcal{E}_{max}$), the energy distribution is quasi-thermal with the effective temperature T several times lower than $\mathcal{E}_{max}$. Such a spectrum is unacceptable for medical purposes, because it does not ensure an essential increase in the radiation dose in a local area and will cause an unacceptably intense irradiation of healthy tissues.

In order to improve the quality of a proton beam, it can be "cut" into narrow beams in energy space. However, in this case, the conversion efficiency of laser energy into the energy of fast particles is significantly reduced and, what is more important, the number of the beam particles decreases. A more promising approach is related to the use of multilayer targets. Thus, it was proposed to use a foil target consisting of high-Z atoms and covered with a thin hydrogen-containing film (see Fig. 2a). When the target is irradiated with an ultrashort laser pulse, heavy atoms are partially ionized and their electrons escape from the foil, thus creating the charge-separation electric field. Because of their large inertia, the heavy ions remain at rest, while the more mobile protons are involved in the process of acceleration. The energy spectrum of protons accelerated in the charge-separation field near the target surface can be found from the continuity condition for the particle flux in energy space, namely, $N(\mathcal{E}) = n_0(x_0)|dx_0/d\mathcal{E}|$. Here, the energy $\mathcal{E}$ is a function of the Lagrangian coordinate x_0, $N(\mathcal{E})$ is the differential particle energy spectrum, and $n_0(x_0)$ is the initial spatial distribution of the particles. We can see that, in order to produce a highly monoenergetic proton beam, it is necessary that the hydrogen layer thickness Δx_0 be small (i.e., the function $n_0(x_0)$ should be highly localized in x_0 space) and/or the derivative $|d\mathcal{E}/dx_0|$ should vanish at a certain point x_m. In the vicinity of this point, the function $\mathcal{E}(x_0)$ can be represented as $\mathcal{E}(x_0) = \mathcal{E}_{max} - \alpha(x_0 - x_m)^2/4$. It follows from here that, in the vicinity of the maximum energy, the particle energy spectrum has the form $N(\mathcal{E}) = n_0/\alpha^{1/2}(\mathcal{E}_{max} - \mathcal{E})^{1/2}$. The necessary conditions for this acceleration regime can be ensured by using a two-layer target with a hydrogen-containing film deposited on the front side of the foil, as is shown in Fig. 2b. Note that, when preparing this paper, we become acquainted with the recently published paper [10], in which the increase in the efficiency of ion acceleration in the interaction of laser radiation with two-layer targets was demonstrated experimentally.

The characteristic energy of protons accelerated in the charge-separation electric field $E \approx 2\pi n_0 Zel$ in the region of size $R_\perp$, equal to the laser spot radius, can be estimated in the order of magnitude as $\mathcal{E}_{max} = 2\pi n_0 Ze^2 l R_\perp$ and the energy spectrum width as $\Delta\mathcal{E} =$

$2\pi n_0 e^2 \Delta x_0 R_\perp$. Here, n_0 is the density of high-Z ions in the foil, Ze is their electric charge, l is the foil thickness, and Δx_0 is the thickness of the hydrogen-containing film. It is assumed that the intensity of laser radiation is sufficiently high for all the electrons to be blown out of the laser-irradiated region under the action of the ponderomotive pressure. For this purpose, it is necessary that the dimensionless amplitude of the laser field $a = eE/m_e\omega c$ be larger than $a_c = l/d_e$, where E is the laser field strength, ω is the laser frequency, e and m_e are the charge and mass of an electron, and $d_e = c/\omega_{pe}$ is the collisionless skin depth. For a laser wavelength of about 1 μm and a foil thickness of 5 μm, the threshold field amplitude is $a_c = 50$; i.e., the laser intensity is $I = 5 \times 10^{21}$ W/cm^2, which corresponds to the petawatt power range [1]. It follows from [8] that, in this case, the energy of fast protons attains several hundreds of MeV, which is sufficient for PT. As was noted above, the flux of accelerated protons in a beam should be no less than $(1–5) \times 10^{10}$ proton/s, which is achievable, because, in experiments of [7, 8], the number of accelerated protons attained a value of 10^{12}–10^{13} proton/pulse. To ensure such a number of fast particles, it is sufficient that the thickness of the solid-state hydrogen-containing film on the target surface be about 1 μm at a laser spot diameter of about 10 μm. In addition, the parameters of a proton beam can be controlled by varying the target geometry (as was shown in [8], the protons accelerated from the surface of a convex foil target are focused into its center of curvature). Note that the high quality of the proton beam (the smallness of the ratio $\Delta\mathcal{E}/\mathcal{E}$) is of fundamental importance not only for medical applications, but also for charged particle injectors [4] and the problem of fast ignition of fusion targets with the use of laser-accelerated ion beams [6].

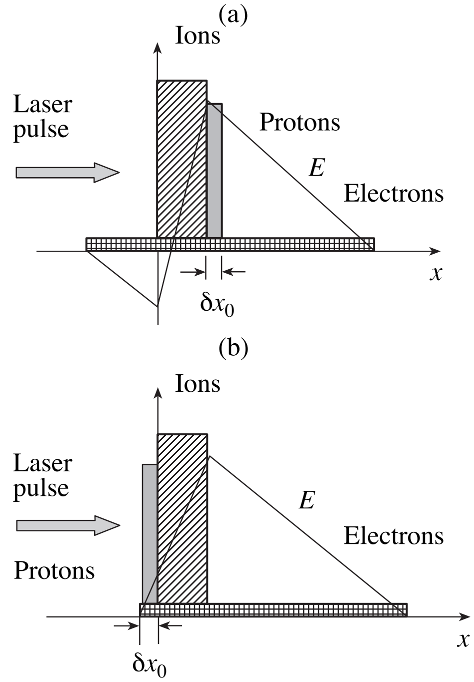

Fig. 2. Two-layer target: a thin hydrogen-containing film is deposited on (a) the rear side and (b) the front side of a foil consisting of high-Z atoms.

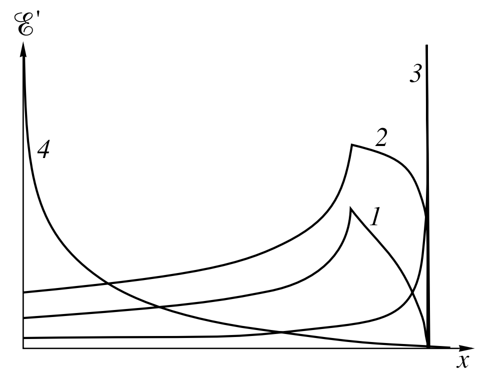

Fig. 3. Energy deposition profiles for beams with different energy distributions.

As the fast ion beam propagates through a tissue, it loses energy. The energy loss rate is equal to $d\mathcal{E}/dx = K/\mathcal{E}^2$ (see, e.g., [11]), where the K factor logarithmically depends on the proton energy. Figure 3 shows the profiles of the energy deposited by proton beams with different energy distributions. Curve *1* corresponds to the distribution $N(\mathcal{E}) = n_0\theta(\mathcal{E}_{\max} - \mathcal{E})\theta(\mathcal{E} - \mathcal{E}_{\min})$, where $\mathcal{E}_{\max}$ and $\mathcal{E}_{\min}$ are the maximum and minimum proton energies in the beam, $\theta(\xi) = 1$ at $\xi > 0$, and $\theta(\xi) = 0$ at $\xi < 0$ (for the given distribution, $(\mathcal{E}_{\max} - \mathcal{E}_{\min})/\mathcal{E}_{\max} = 0.1$). In curve *2*, which corresponds to the energy spectrum $N(\mathcal{E}) = n_0\theta(\mathcal{E}_{\max} - \mathcal{E})\theta(\mathcal{E} - \mathcal{E}_{\min})/(\mathcal{E}_{\max} - \mathcal{E})^{1/2}$, we can see a pronounced peak of energy losses. For comparison, Fig. 3 also shows the profiles of the energy deposited by monoenergetic and quasi-thermal proton beams. Curve *3* corresponds to a monoenergetic beam: $N(\mathcal{E}) = n_0\delta(\mathcal{E}_{\max} - \mathcal{E})$. We can see the Bragg peak of energy losses in the vicinity of the beam stopping point. Curve *4* corresponds to the energy distribution of the form $N(\mathcal{E}) = n_0\theta(\mathcal{E}_{\max} - \mathcal{E})\exp(-\mathcal{E}/2T)^{1/2}$, where $T = \mathcal{E}_{\max}/2$ is the effective temperature. It is this distribution that is usually used to approximate the fast proton energy spectra observed in laboratory and numerical experiments on the interaction of laser radiation with non-optimized targets. It is clear that this distribution is unacceptable and it is necessary to use optimized multilayer targets.

The future of hadron therapy is related to the creation of specialized oncological centers equipped with modern diagnostics and medical accelerators. Cities with a population of two to three million inhabitants need centers with four treatment rooms. For smaller cities, centers with one treatment room are required; however, to create such centers based on conventional accelerators and existing Gantry systems is inexpedient from the economical standpoint. At the same time, the creation of such centers in medium-sized hospitals would expand the range of PT applications and, a very important point, would made it possible to bring medical centers closer to patients. One of the solutions to this problem is the creation of relatively inexpensive specialized medical laser proton accelerators having relatively small dimensions (2–3 m) and weighting no more than several tons. In fact, this can substantially simplify the design and reduce the cost of PT centers. The use of multilayer targets with various shapes and structures provides additional possibilities for controlling the parameters of fast ion beams, such as the energy spectrum, the number of particles, the focal length, and the size of the region where the proton energy is deposited.

As concerns the low repetition rate of the proton pulses, there are two possible solutions to this problem. At present, the repetition rate of the required ultra-high-power (petawatt) laser pulses is rather low. The repetition rate of proton pulses can be increased by implementing an assembly of lasers shooting at a single target in succession. It is also possible to accelerate the bunch of protons repeatedly, each time increasing the energy of protons by several MeV. This may be achieved with sequentially positioned targets and lower power lasers, which nowadays have a sufficiently high repetition rate.

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