Time-gated x-ray tomography

Gratz, M; Kiernan, L; Wahlstrom, C. G; Svanberg, Sune; Herrlin, K

Published in:
Applied Physics Letters

DOI:
10.1063/1.122623

1998

Time-gated x-ray tomography

M. Grätz,[a] L. Kiernan, C.-G. Wahlström, and S. Svanberg
Department of Physics, Lund Institute of Technology, P.O. Box 118, S-221 00 Lund, Sweden

K. Herrlin
Department of Diagnostic Radiology, Lund University Hospital, S-221 85 Lund, Sweden

(Received 16 June 1998; accepted for publication 18 September 1998)

Time-gated x-ray tomography with scatter reduction is demonstrated using a laser-produced plasma as an ultrashort-pulse x-ray source in combination with a time-resolving streak-camera detector. Backprojections of a phantom imbedded in 9 cm of water show an effective 50% increase in contrast when scattered x-ray quanta (being delayed in time) are suppressed by gating on the prompt, nonscattered photons. Implications for future volumetric tomography, in particular concerning possible dose reductions, are discussed. © *1998 American Institute of Physics.* [S0003-6951(98)02846-0]

Time-gated x-ray tomography, using ultrashort x-ray pulses from a laser-produced plasma in combination with time-resolved detection, is demonstrated in the present letter as a possible means of reducing deposited dose in future volumetric tomography.

X-ray projection radiography is the most common technique in clinical radiology,[1] and has been refined over the last 100 years. Through the development of computerized tomography (CT), the normal line-of-sight shadow images can be replaced by cross-sectional data of high quality.[2] In spite of the obvious success of current x-ray diagnostics, there is a concern regarding x-ray-induced malignant tumors. In fact, a recent study suggests[3] that, for example, at least 150 million dollars could be saved annually in Europe, not to mention human suffering, if the collective patient dose could be reduced by a factor 2.

Different methods for dose reduction do exist: the use of sensitive digital image plates instead of conventional x-ray film has the potential of reducing patient dose, if the dynamic range and the sensitivity of the plates are fully utilized. Monochromatic x rays bridging the K absorption edge of iodine, used as a contrast agent in angiography, also have the potential to reduce patient dose. This has led to an interest in developing synchrotron-radiation-based differential radiography.[4] Magnetic resonance imaging, ultrasound and optical transillumination are alternative diagnostic techniques, where the use of ionizing radiation has been completely eliminated.

As first pointed out by Gordon *et al.*[5] and further elaborated theoretically and experimentally by Grätz *et al.*,[6] x-ray time-gated imaging using ultrashort laser-produced x-ray pulses in combination with a time-resolving x-ray detector has the potential to reduce the dose in radiography. This technique relies on the temporal suppression of scattered x-ray radiation, which can constitute a large part of the signal reaching the image detector. For example, in whole-body x-ray radiography about 90% of the detected photons are scattered and have lost the image information. Time-gated imaging reduces the statistical noise level due to the strong scattered-radiation background. Therefore, a lower total radiation dose is required to achieve a given image quality. A similar principle has already earlier been investigated in the optical domain (see, e.g., Ref. 7).

Time-gated x-ray imaging became possible after the development of compact ultrashort-pulse terawatt laser systems. By focusing such pulses on a high nuclear-charge metallic target such as tantalum, intense bursts of x rays with photon energies extending up into the MeV range can be produced.[8] Biological imaging using such radiation was first demonstrated in our group,[9] including single-shot recordings (exposure time a few ps) and magnification radiography, taking advantage of the very small source size.[10] Further, it was demonstrated that differential imaging of a radiographic contrast agent in principle could be achieved using the characteristic radiation in different target materials.[11] In the present letter we give a proof of principle that, using this short-pulse source, high-quality tomography with scatter reduction can be achieved in time-gated imaging, possessing a potential for the development of a viable new technique for radiography, in particular volumetric tomography.

In the present experiments, intense and ultrashort hard x-ray pulses were produced by focusing pulses from the Lund chirped-pulse-amplification Ti:sapphire terawatt laser[12] onto a rotating tantalum target. The pulse energy on target was about 150 mJ delivered in about 100 fs. The effective x-ray source size was measured by pinhole imaging to be about 40 μm. The x-ray emission was filtered with a 0.15 mm copper foil, so that radiation with photon energies lower than 25 keV was effectively reduced. After traversing the object under study the radiation was incident on the photocathode (CsI, 5 μm thick) of a Kentech x-ray streak camera, producing temporally resolved one-dimensional (1D) images with a time resolution of about 50 ps. The cathode efficiency is noticeably lower than for conventional x-ray films or image plates. The streak camera was read out using an intensified charge coupled device (CCD) camera (La Vision Flame-Star IIF). Further details of the experimental setup are given in Ref. 6.

The phantom used in the tomographic imaging was a plastic cuvette with a 10×10 mm^2 cross section, containing

[a]Electronic mail: matthias.graetz@fysik.lth.se

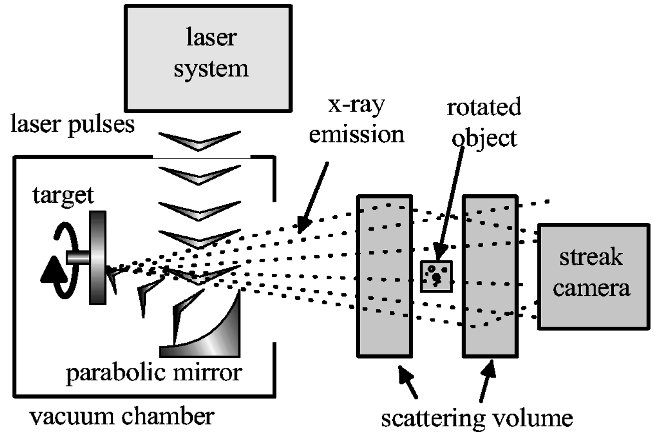

FIG. 1. Schematical experimental arrangement used in time-gated x-ray tomography (not to scale).

several copper wires (diameters 1.9, 0.8, 0.6 and 0.3 mm) and a porcelain rod (diameter 1.7 mm) embedded in a silicon gel for mechanical stability. The phantom was placed vertically on a rotary stage 38 cm from the laser-produced plasma and 10 cm away from the horizontally orientated x-ray cathode. The cathode was ~1 mm high, adopted to a usual slice thickness in tomography. The cathode width of about 23 mm was limiting the maximum object size. Two 4.5 cm thick, water-filled, brick-shaped plastic containers ($17\times13\ \mathrm{cm}^2$) could be placed around the phantom in order to simulate tissue attenuation and scattering. The experimental arrangement is illustrated in Fig. 1.

A total of 36 projections separated by 5° phantom rotation were recorded and used for the subsequent tomographic backprojection. Time-resolved data were taken both with and without the water-filled containers (using 4200 and 1200 laser shots, respectively) and evaluated in a 45 ps wide gate window centered 15 ps before the maximum of the unscattered peak. Time-integrated data were recorded with the

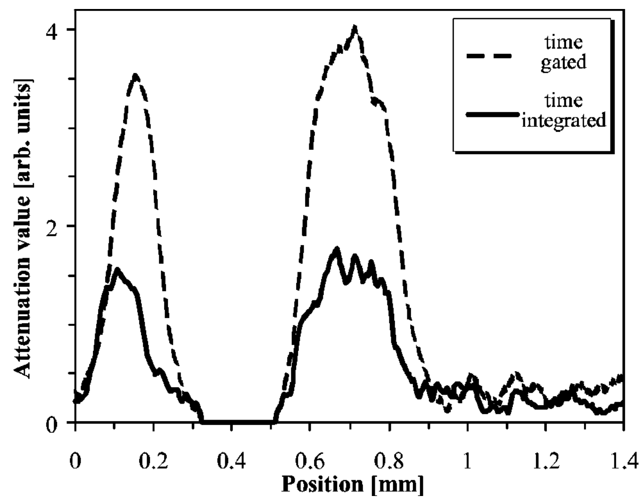

FIG. 3. Profiles along the lines indicated in Fig. 2. The increase in contrast due to time gating is visible. The low values in-between the peaks are due to reconstruction artifacts.

same setup by disabling the temporal sweep of the streak camera (using 3000 laser shots with water and 1200 laser shots without water).

Recorded data from all projections were normalized and backprojected using software specially adapted to our experimental conditions, based on a standard convolution backprojection algorithm.[2] The cross-sectional images, which scale with the object's attenuation coefficient, are shown in Fig. 2. Time-gated as well as time-integrated recordings, with and without 9 cm of scattering water, are shown. Profiles along the indicated lines are shown in Fig. 3. The wires appear to be less attenuating due to the presence of scattered radiation. It can be clearly seen that time-gating provides a substantial contrast increase when an object imbedded in a volume of scattering material is imaged. The contrast, here defined as the difference in attenuation coefficient between a

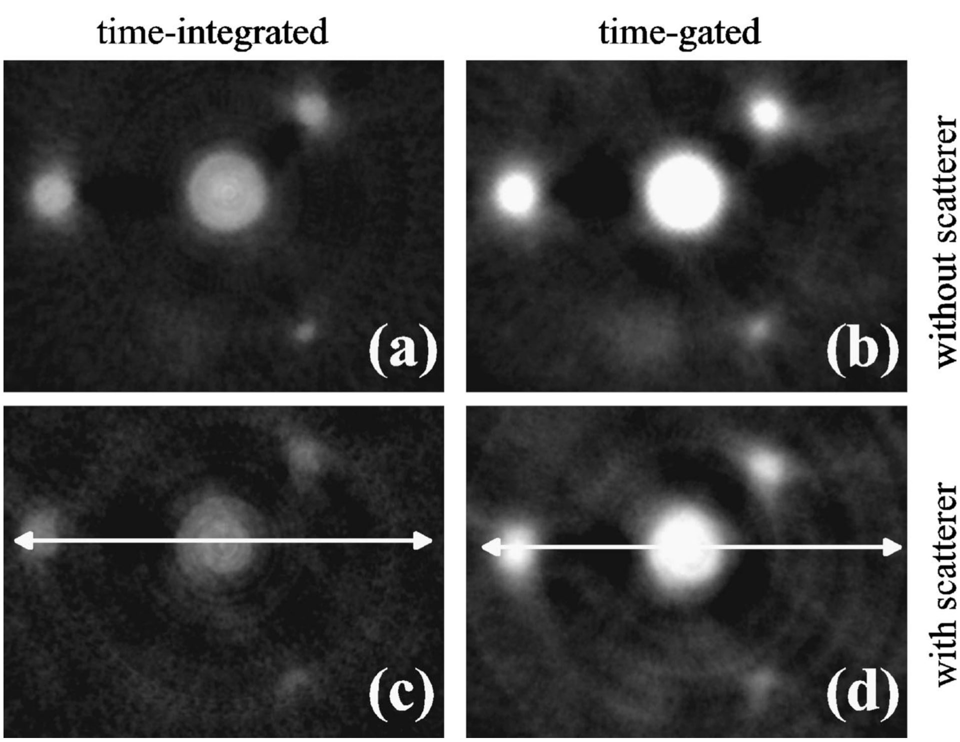

FIG. 2. Backprojected images of the phantom. The brightness is proportional to the attenuation coefficient (only the central part of the phantom is shown).

wire and the plastic gel background, is improved by ∼170% when averaging over the four wires and the rod. This value has to be corrected for the presence of inherent scattered radiation from the x-ray source. The effective contrast improvement, attributed to suppression of scattered radiation from the water, is then ∼50%. This scatter suppression compares well to theoretically obtained values from our Monte Carlo simulations. Significantly higher contrast improvements are expected for thicker, more realistic scattering volumes.[13]

In a conventional CT machine, in contrast to our experimental setup, the radiation is limited to a thin slice of tissue to reduce scattering from neighboring tissue. Additionally, anti-scatter grids are used, imposing an unwanted dose penalty. Using time-gated imaging as a scatter-reduction method could enable the simultaneous sampling of a whole volume with a large-solid-angle x-ray beam and a two-dimensional time-gated detector. However, efficient two-dimensional time-resolving detectors would need to be used. An interesting development in this field is on its way.[14]

The competitiveness of laser-based x-ray radiography would have to rely on a suitable combination of attractive features such as scatter reduction, demonstrated in this letter, differential imaging with x rays of different spectral content,[11] and ultrasharp magnification imaging.[9,10] In addition, it has been tentatively demonstrated that a certain dose exposure to short x-ray bursts does not induce more cell damage than exposure to a corresponding continuous-radiation dose.[15] Therefore, with realistic kHz repetition-rate terawatt lasers capable of delivering the required integrated dose coming of age, basic requirements for interesting applications of laser-produced x rays in medicine seem to be fulfilled.

This work was supported by the Swedish Natural Science Research Council, the Medical Research Council and the European Commission, under Grant. No. ERB FMRXCT 960080.

[1] T. S. Curry III, J. E. Dowdey, and R. C. Murry, Jr., *Christensen's Physics of Diagnostic Radiology*, 4th ed. (Lee and Febiger, Philadelphia, 1990).
[2] *Technical Aspects of Computed Tomography*, edited by T. H. Newton and D. G. Potts (Mosby, 1981).
[3] B. M. Moores, to appear in the proceedings of the Luxemburg workshop on reference dose, Luxemburg, 1998 (unpublished).
[4] W. R. Dix, Prog. Biophys. Mol. Biol. **63**, 159 (1995).
[5] C. L. Gordon III, G. Y. Yin, B. E. Lemoff, P. B. Bell, and C. P. J. Barty, Opt. Lett. **20**, 1056 (1995).
[6] M. Grätz, A. Pifferi, C.-G. Wahlström, and S. Svanberg, IEEE J. Sel. Top. Quantum Electron. **2**, 1041 (1996).
[7] *Medical Optical Tomography: Functional Imaging and Monitoring*, edited by G. Müller *et al.*, Proceedings SPIE **IS11** (Bellingham, 1993).
[8] J. D. Kmetec, C. L. Gordon III, J. J. Macklin, B. E. Lemoff, G. S. Brown, and S. E. Harris, Phys. Rev. Lett. **68**, 1527 (1992).
[9] K. Herrlin, G. Svahn, C. Olsson, H. Pettersson, C. Tillman, A. Persson, C.-G. Wahlström, and S. Svanberg, Radiology **189**, 65 (1993).
[10] C. Tillman, A. Persson, C.-G. Wahlström, S. Svanberg, and K. Herrlin, Appl. Phys. B: Lasers Opt. **B61**, 333 (1995).
[11] C. Tillman, I. Mercer, and S. Svanberg, J. Opt. Soc. Am. B **13**, 209 (1996).
[12] S. Svanberg, J. Larsson, A. Persson, and C.-G. Wahlström, Phys. Scr. **49**, 187 (1994).
[13] M. Grätz, L. Kiernan, and K. Herrlin, Med. Phys. (to be published).
[14] P. B. Bell, J. D. Kilkenny, O. L. Landen, R. L. Hanks, and J. D. Wiedwald, Proc. SPIE **2519**, 165 (1995).
[15] C. Tillman, G. Grafström, A.-C. Jonsson, I. Mercer, S. Svanberg, S.-E. Strand, S. Mattson, B.-A. Jönsson, and S. Svanberg (to be published in Radiology).