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Development of Carbon Nanotube Based Reflection Type X-ray Source

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X-ray imaging technology is a useful and leading medical diagnostic tool for health care professionals to diagnose disease in human body. Carbon nanotube based X-ray source, which we have developed in this study, could be also useful and supply integrated diagnostic X-ray imaging tool in diagnosis. Conventionally, thermionic type of tungsten filament X-ray tube is widely employed in the field of biomedical and industrial application fields. However, intrinsic problems, such as poor emission efficiency, low imaging resolution, and high electrical energy consumption etc., may cause the limitation of using the X-ray tube. To fulfill the current market requirement, specifically for medical diagnostic field, we have developed rather a portable and compact carbon nanotube based X-ray device in which microfocus high imaging resolution can be feasible.

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1. Introduction

It is known that today's diagnostic service providers are forced to provide rather a portable X-ray system, which supports the highest radiographic quality with less economical expense. Therefore, in the current study, we report on the development of a new X-ray source by using nanostructured carbon nanotube (CNT) based cold electron emitter. Such an X-ray source provides many advantages over the conventional electron source for many applications [1–5]. Since the CNT based portable X-ray system that we have developed contains a direct digital imaging sensor or detector, real time imaging is feasible and it, therefore, addresses a broad range of clinical and technological applications in general purpose of radiography.

2. CNT emitter synthesis

In order to synthesize double wall carbon nanotubes (DWCNTs) using CH_4, a catalyst was prepared by embedding the Fe–Mo bimetallic catalyst onto MgO powder. Fe–Mo/MgO catalyst was prepared according to the following procedure. A mixture of $Fe(NO_3)_3{\cdot}9H_2O$ (99%, Aldrich) and Mo solution (Aldrich, ICP/DCP standard solution, 10 mg/ml of Mo in H_2O) was dissolved in DI water for 1 h. The mixed Fe–Mo solution was introduced to the solution of MgO powder and DI water followed by sonication for 1 h. The molar ratio of the catalyst was Fe : Mo : MgO = 1 : 0.1 : 12. After drying, the material was baked at 150°C for 15 h in vacuum and then ground in a mortar, and finally calcined in a quartz furnace at 700°C for 2 h in O_2 ambient. The DWCNTs were synthesized in a quartz tube reactor (inner diameter: 70 mm, length: 700 mm). About 200 mg of the supported Fe–Mo catalyst was placed in a quartz boat at the center of the reactor tube. The quartz tube was heated up to 900°C in Ar atmosphere. Subsequently, CH_4 (300 sccm) and a mixture of Ar (500 sccm) and H_2 (100 sccm) were introduced into the reactor. After 20 min, the reactor was cooled to room temperature in Ar atmosphere. The as-synthesized DWCNTs were purified using a two-step purification process. Firstly, as-synthesized DWCNTs were oxidized in a furnace at 350°C in air atmosphere for 30 min to remove the amorphous carbon material on the surface of DWCNTs. Secondly, the oxidized DWCNTs were soaked in a diluted acetic acid solution at room temperature to remove catalyst particles existing in carbon materials. The DWCNT suspension was collected by membrane filtration (pore size: 0.2 μm) and washed with distilled water several times. Finally, we obtained the purified DWCNTs (non-annealed DWCNTs). After collecting the purified DWCNTs, the high temperature thermal annealing was carried out at 1300°C for 1 h in vacuum condition of 1.0×10^{-5} Torr in order to reduce the defects in DWCNTs, resulting in improvement of crystallinity of DWCNTs.

3. Results and discussion

To evaluate the field emission properties of DWCNTs, the planar typed CNT field emitter was fabricated on the circular stainless steel substrate (SUS304) with 5 mm diameter. The Ag paste was coated on the stainless steel

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substrate and then the DWCNTs were spread on the Ag paste/stainless steel substrate using a sieving method. The DWCNTs deposited on the substrates were vertically aligned by a mechanical method containing attaching and peeling off a tape on the surface of DWCNTs. Field emission properties were measured in a vacuum chamber at pressure of less than 2×10^{-7} Torr. The anode was a stainless steel plate with a dimension of 2×2 cm^2 but for the emission pattern, ITO substrate was used to the anode electrode. The gap between the cathode and the anode was 400 μm. To obtain stable field emission properties, an electrical annealing process was performed at the emission current of 1.0 mA/cm^2 several times. After removing some protruding DWCNTs from the DWCNT emitter using the electrical annealing, the field emission properties were evaluated. In order to reduce edge emission from the circular disk-typed emitter, we ground the edge of the circular disk-type metal substrate and maintained the edge angle of the circular disk-typed emitter about 45° before coating the Ag paste. After grinding the edge of the circular disk-typed emitter, we could obtain uniform emission from the emitter. The turn-on field of DWCNTs is about 2.4 V/m at the emission current density of 0.1 μA/cm^2 and the threshold field is about 3.15 V/μm at the emission current density of 1.0 mA/cm^2. Especially, the DWCNT emitter shows much higher emission current density of 87.8 mA/cm^2 at the applied electric field about 5.5 V/m, which indicates a high enough current level (i.e., 17.2 mA for 5 mm CNT emitter) for X-ray source application.

In the simulation, a typical triode model of a CNT emitter based X-ray source was employed. To fabricate rather a flawless CNT emitter X-ray source it is essential to perform computer simulation in advance. Basically, the triode model of the X-ray source consists of a CNT emitter as a cathode, a metal grid mesh as a gate, which functions to extract electrons from the tip of the CNT emitter and to modulate the electron beam somewhat, and finally an anode. But, in this model, electrostatic lenses are included to focus the electron beam with a microlevel onto the anode target. The important CNT X-ray source parameters, such as the diameter of the CNT emitter and the distance between the anode target to the CNT emitter source were 5 mm and 23 mm, respectively. Although it is not shown here, the typical kVp of the fabricated CNT X-ray source was measured with 40 kVp.

Figure 1 shows the typical simulation results of the time of flight (TOF) of the electron beam trajectory exiting from the CNT emitter. The diameter of the CNT emitter and the distance between the CNT emitter to the anode target was fixed to 5 mm and 23 mm, respectively. As shown in Fig. 1, the focal point of the electron beam (i.e., specific distance at which the smallest spot size of the electron beam found) tends to increase with increasing the focusing lens voltage. In addition, at higher lens voltage, some fraction of the electron beam did not focus but loose instead as seen in Fig. 1b. In the simulation,

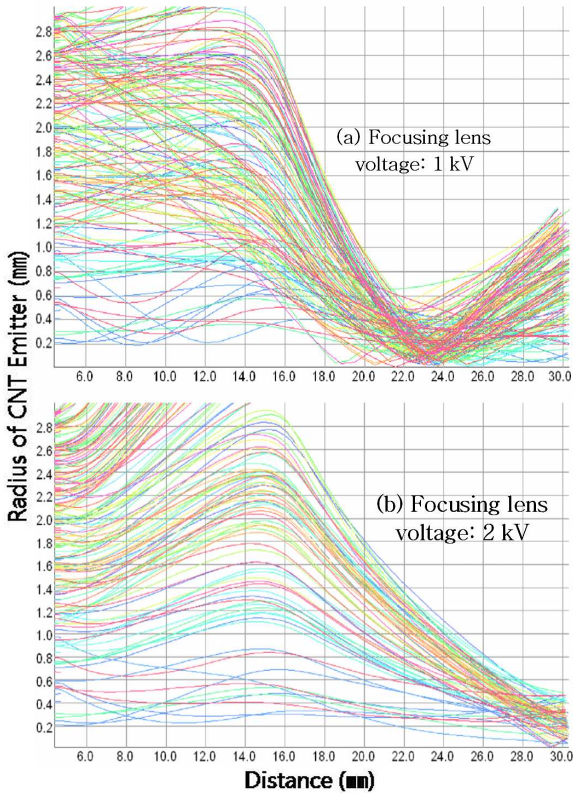

Fig. 1. Examples of typical simulation results of the time of flight (TOF) trajectory of electron beam using the triode model. In the simulation, the diameter of the CNT emitter was 5 mm and the distance between the anode target to the CNT emitter source was 23 mm.

the calculated electron beam spot size is approximately less than 700 μm at the anode target.

The triode model simulated in this study is applied to design and fabricate a prototype of a CNT-emitter based X-ray source. Although it is not shown, here it was housed in the vacuum chamber, which is sustained with low pressure of 10^{-7} Torr. And also, it is automatically operated by computer (i.e., Lab View program) to supply and maintain voltages and currents and also to obtain X-ray images.

4. Conclusions

Using the CNT X-ray source, X-ray images of a finger bone and teeth in human body were obtained. The trabeculation shape in finger bone is clearly observed in Fig. 2a. To obtain the finger bone image, tube currents of 250 μA at 42 kV tube voltage was applied. The human tooth image (Fig. 2b), however, is somewhat unclear because the supplied voltage was limited to maximally 50 kV in the system. It should be noted that normally 60–70 kV of tube voltage is required in dental imaging. Considering this, if the tube voltage is over 60 kV then clearer image can be feasible. Based on these preliminary results, it can be concluded that the CNT based

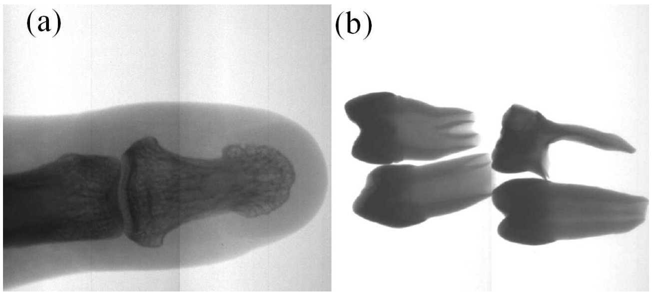

Fig. 2. Imaging of human finger (a) and teeth (b) using the fabricated CNT X-ray source.

X-ray source may be applicable to dental diagnosis and orthopedics in near future.

Acknowledgments

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