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■ 論 文 ■

Design of the Detector Head for Single Photon Detection in Breast Cancer Diagnosis and Its Performance Evaluation

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유방암진단에서의 단일광자검출을 위한 검출기 전단부의 설계와 성능평가

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Abstract - Monte Carlo simulation has been performed to induce optimized parameters of the detector head of gamma camera for the diagnosis of breast cancer and to evaluate it under the diagnosis condition of the breast cancer. For the simulation, we used Tungsten collimator, having a lattice structured array with holes of 3 mm x 3 mm and septal thickness of 0.25 mm, which are corresponding to the pixellated photosensor. For driving optimum parameters we used Trade-Offs procedure between the geometric efficiency and the spatial resolution, varying the detector head components. In order to pre-evaluate the performance of the optimized detector head, we assumed diagnosis condition that the breast tumor is located in the middle of phantom with various sizes and its location is 25 mm from the collimator surface, considering background count caused by radiation sources from other organs, . It was shown that the performance of the optimized detector head can be degraded according to the breast cancer size and the background count under real diagnosis conditions of breast cancer. Therefore, it is concluded that the spatial resolution, which is used as an indicator to distinguish the various sizes of breast cancer and is dependent on the characteristic of the detector head, appears to be meaningless in early diagnosis of the breast cancer.

Key wards : Monte Carlo, detector head of gamma camera, breast cancer, Trade-Offs, geometric efficiency and spatial resolution

요약 - 유방암 진단에 필요한 감마카메라 검출전단부의 최적변수 유도와 유방암 진단 조건 하에서의 평가를 위한 몬테카를로 모사를 수행하였다. 모사를 위해 픽셀화된 포토센서에 상응하는 3 mm x 3 mm의 구멍과 0.25 mm의 격막두께를 갖는 격자배열구조의 텅스텐 콜리메이터를 이용하였다. 최적변수를 도출하기 위해 검출전단부의 구성요소를 변화시키면서 기하효율과 공간분해능의 Trade-Offs 절차를 사용하였다. 최적화된 검출전단부의 사전 성능평가를 위해, 펜텀의 중앙부에 크기가 각기 다르며 콜리메이터 표면으로부터 25 mm 떨어져 있는 유방암이 있고 다른 장기들로부터 나오는 방사선원에 의한 백그라운드 계수를 고려하였다. 유방암의 실제 진단 조건 하에서는 최적화된 검출전단부의 성능이 유방암의 크기와 백그라운드 계수에 따라 저하될 수 있음을 보여 주었다. 따라서 유방암 크기를 변별하는 지표로 쓰이며 검출전단부의 특성에 종속적인 공간분해능은 유방암의 조기 진단시에는 의미가 없다는 결론을 얻었다.

중심어 : 몬테카를로, 감마카메라 검출전단부, 유방암, 균형, 기하효율과 공간분해능

Introduction

Functional scintimammography of breast cancer using ^{99m}Tc MIBI and gamma camera of the discrete scintillator/photodiode structure have been reported in others [1]. This technique is able to detect the breast cancer with more than 90% specificity [2] while the specificity of X-ray mammography is much low. The gamma camera of the discrete scintillator/photodiode consists of a lattice structured array collimator, crystals, photodiodes, and electronic circuits as shown in figure 1.

The lattice structured array collimator is able to pass gamma-rays from the sources in breast tumor to the detector surface perpendicularly. It is made of lead or tungsten depending on the gamma-ray energy. Its height and hole shape, for example, round, hexagonal, and square type are chosen according to its application purpose. The crystal which converts single gamma-ray into light is typically CsI(Tl) for 140 keV gamma-ray from ^{99m}Tc. The photodiode that converts light generated in the crystal into charge carries are generally solid-state detector such as silicon PIN photodiode, HgI_2, and etc. Finally, the electronic circuits handle electronic signal from photodiode and produce image data using Winner Take All (WTA) algorithm to search the position in the detector [1]. The specificity of a scintimammography system is governed by the spatial resolution and the geometric efficiency of the collimator based on the total response of a collimator, which are considered to be a combination of geometric, penetration, and scatter components [3].

The geometric component means the passing probability of photons through the collimator hole properly without experiencing any interaction within the patient or the septal material of the collimator. It is the major contribution to the total counts of the detector and has the most influence on the geometric efficiency of the collimator, the spatial resolution, and the field of view. The penetration component through the septa is negligible for low energy photons such as 140 keV in case of high density of tungsten collimator.

The scatter component, which is the number of photons that may experience Compton scatter interaction within the patient body and the septal material of the collimator before reaching the scintillation crystal, causes both poor resolution and high background count.

In this paper, after selecting the geometry and condition of the crystal and the bonding material, the Trade-Offs [4] between the geometric efficiency and the spatial resolution has been performed, which explains the total response of collimator and its performance. Additionally, we compare theoretical expressions with the results of the simulations. Finally, the effects of background counts and various tumor sizes for the various collimator heights are considered to evaluate the designed detector head. From the results of the simulations and analyses, the limits of collimator performance are explained under diagnosis condition.

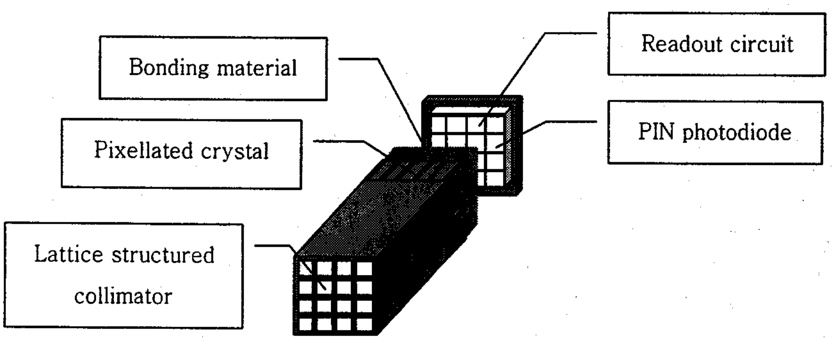

Fig. 1. Detector head and electronics for breast cancer imaging

Materials and Methods

For realistic simulation we divided top area of the collimator and the crystal into 11x11 pixels which has 1,278 mm^2. MCNP4B simulation code was used for the energy deposition in the crystal depending on the thickness of the crystal, the geometric efficiency and the spatial resolution of the collimator. DETECT97 simulation code was used for the light transmission efficiency for the crystal thickness and the surface treatment of the crystal.

Design of Detector Head Components

Three components of the detector head have been considered separately to suggest optimum values for the detector head as follows; the crystal, the optical bonding material between the crystal and the photodiode, and the collimator.

Among the various scintillation materials, CsI (Tl) crystal was chosen because it has larger gamma ray absorption coefficient per unit size and nigh light yield (Photons/MeV) [4]. This CsI (Tl) was taped with Teflon of 0.25 mm thickness to prevent light loss in the crystal and cross talk from each pixellated crystal. Its pixel size also follows as the collimator hole size. The crystal heights varied the range of 0.1 mm to 10 mm for the absorption energy in the crystal and the light transmission efficiency considering the spatial resolution (in unit of mm on FWHM).

Because the condition of treatment for the top and the side surface of crystal effects on the light transmission efficiency, we treated the top and the side surface of the crystal with Ground (harsh surface), Polished (polish surface), Metal-0.95RC (metal coating with 95 % reflection coefficient), Polished-0.98RC (polish surface with 98 % reflection coefficient), and Painted-0.98RC (coloration surface with 98 % reflection coefficient) respectively.

The optical bonding material between the crystal and the photodiode also influences on the light transmission into the photosensor. So as to find out high light transmission efficiency, the refractive index and the thickness of the optical bonding material varied range of 1.6 to 2.0 and 0.1 mm to 0.5 mm, respectively.

The parallel-hole collimator is by far the most common type of collimator used clinically. From the research result [1], the simulated 1 to 1 matched square hole collimator with the pixellated photodiode had demonstrated better spatial resolution than round or hexagonal hole collimator. The walls of the collimator hole, called septa, are made of a material with high atomic number such as lead. However, we used Tungsten (W:Cu= 6:4) material instead of lead for future manufacture with ease to make square hole, and the septal thickness of the collimator was fixed at 0.25 mm, depending on the gap between the pixellated crystals and the photodiode. Also each hole size was 3 mm x 3 mm with the same reason as mentioned above because of the difficulty of wiring at the end of the photodiode connecting the circuits [1].

Using the components of the detector head except the collimator, the spatial resolution and the geometric efficiency can be achieved separately for various collimator heights. However, these two parameters independently do not provide an optimum heights of the collimator for best quality image since these two parameters are so correlated. In order to improve the spatial resolution to obtain the ability to discern the breast cancer in detail, it is necessary to sacrifice the geometric efficiency, resulting in higher image noise and vice versa. It is known for collimator design that the best diagnostic information can be obtained by the "Trade-Offs procedure" between the spatial resolution and the geometric efficiency [4]. We have done also the Trade-Offs procedure after getting the spatial resolution and the geometric efficiency.

Evaluation of Detector Head Performance

From the viewpoint of sensitivity, the worst source or the tumor condition in the breast is at the middle of the breast volume with the lack of activity because that condition, far from the collimator surface, dose not give desirable quantity of gamma ray to detect and also can be interrupted by the unnecessary gamma ray, called background counts, from other organs such as heart, liver, and etc in the patient body. For that reason, in order to simulate and analyze the performance of the detector head, we have located the source at 25 mm in the water phantom from the collimator surface. The background activity per volume in the overall body was assumed 22.5 counts per second in this paper [5].

Results and Discussions

Deposition Energy and Light Transmission

The gamma rays of 140 keV from ^{99m}Tc can penetrate the crystal depositing partial energy if the crystal thickness is not enough. For this reason, the thicker crystal is needed to deposit all gamma energy, but it also influences on light transmission from the top of the crystal to the bottom of the crystal since the thicker crystal will not transmit all light generated in the crystal, but lose partial light caused by escaping from the crystal before reaching the bottom of the crystal. According to that reason, we should also do trade-offs for both the deposition energy and the light transmission. So as to decide the deposition energy or the sensitivity depending on the crystal thickness, we fixed the collimator height of 35 mm and varied the crystal thickness with the range of 0.1 mm to 10 mm.

Figure 2. shows its result and displays that the crystal thickness saturated at 5 ~ 6 mm and 8 ~ 9 mm.

For light transmission, we assumed the light is generated only at the center and just below the top surface of the crystal. Figure 3. shows that the light transmission was degrading as the crystal thickness increase.

As a final step to decide the crystal thickness, we did trade-offs between the energy deposition in the crystal and its light transmission as shown in figure 4. This figure tells us that the optimum thickness of the crystal is about 6 mm, which means the light generation and the light transmission is maximum in this thickness. In this simulation and the result, we did not consider surface treatments of the crystal yet.

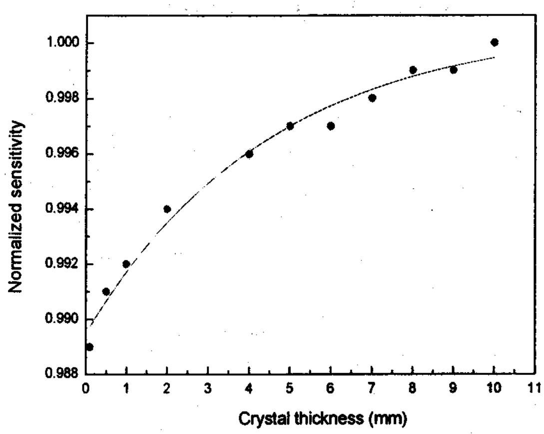

Fig. 2. Normalized deposition energy or sensitivity for each crystal thickness.

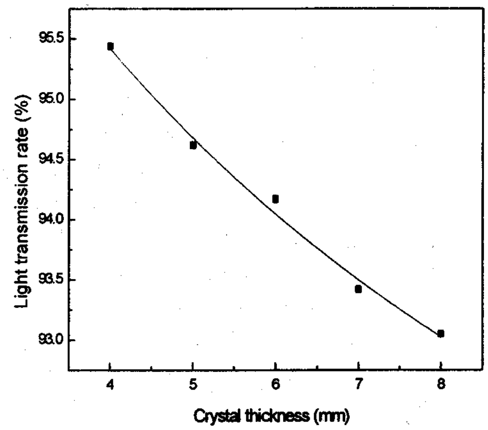

Fig. 3. Light transmission rate or efficiency for each crystal thickness.

Surface Treatments of Crystal Top and Side

For good light transmission from the crystal into the photodiode, it is necessary to decide also crystal surface conditions. With a fixed crystal height of 6 mm, the results in figure 5. show the side and the top surface of the

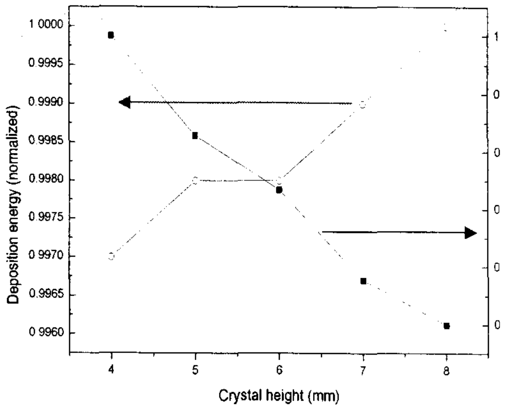

Fig. 4. Trade-off between deposition energy and light transmission in crystal.

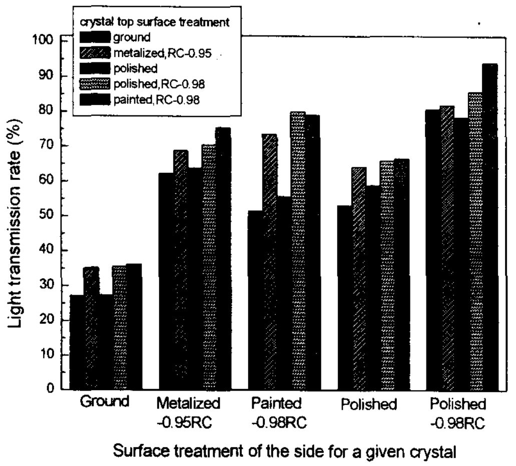

Fig. 5. Light transmission rate for each crystal surface of the top and the side.

crystal are desirable with Polished-0.98RC and Painted-0.98RC, respectively resulting in over 94 % efficiency. In this simulation and the result, we also assumed the light is generated only at the center and just below the top surface of the crystal.

Optical Bonding

The optical bonding material between the bottom of the crystal and the photodiode also was considered including the surface treatment of crystal bottom. As shown in figure 6., the surface treatments of the crystal bottom with

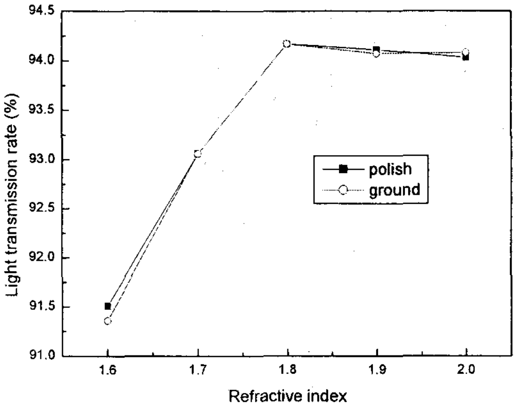

Fig. 6. Light transmission rate depending on the properties of bonding material.

both the Polish and the Ground have no big differences. However, the refractive index of the bonding material, 1.8, was the best resulting over 94 % efficiency.

Geometric Efficiency

The calculation using the theoretical formulation and the simulation for the geometric efficiency were done under the condition of the crystal thickness of 6 mm, a point source, and various collimator heights. Generally, the theoretical formulation of geometric efficiency for parallel hole collimator with square hole is given [6] as follows;

$$G = G_0\left(\frac{A_{open}}{A_{unit}}\right) = \frac{A_{open}}{4\pi l_e^2}\left(\frac{A_{open}}{A_{unit}}\right), A_{open} = a^2, \text{and } A_{unit} = (a+s)^2 \tag{1}$$

where G_0, A_{open}, A_{unit}, and l_e^2 are the geometric factor, the open area of the hole area, the area of a unit cell of the hole array, and the effective hole length of the collimator, respectively. Especially, the effective length, l_e^2 takes into account the penetration of the gamma rays through the septa, but we use just hole length, l, in here since the septa material of the collimator is tungsten which provides no penetration effect. Each parameter used in Eq. (1), displayed in figure 7.

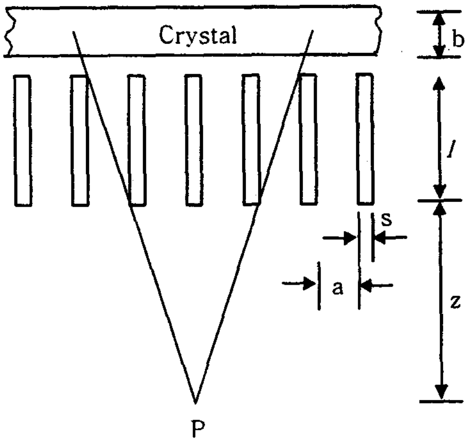

Fig. 7. Schematic diagram of collimator, crystal, and source for calculation of theoretical geometric efficiency.

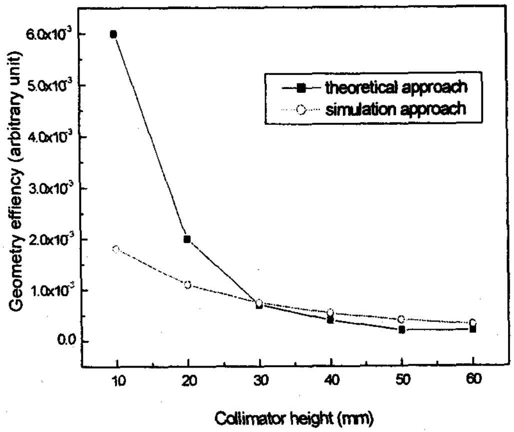

Fig. 8. Geometric efficiency for both theoretical and simulation approaches.

From this equation we can estimate and compare to the simulation result showing in figure 8. From figure 8, even if the tendency that increasing the collimator heights decreases their geometric efficiency was equal each other, there is a somewhat differences except longer collimator height over 30mm.

Spatial Resolution

The spatial resolution for both the theoretical formulation and the simulation was considered also with the same conditions as used in the geometric efficiency. The spatial resolution for parallel hole collimator with square hole was

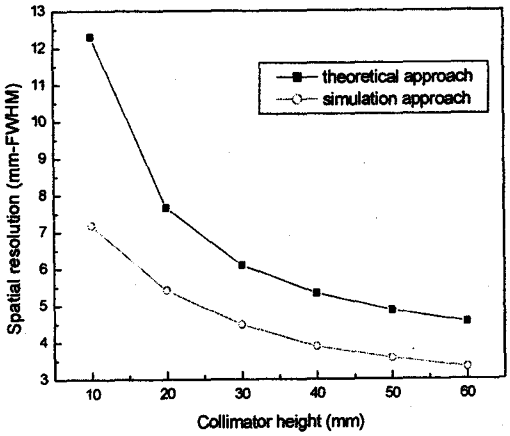

Fig. 9. Spatial resolution for both theoretical and simulation approaches.

given as by Anger [7] follows;

$$R = \left[\frac{a\left(l_e + z + b\right)}{l_e} \right] \tag{2}$$

where we used also hole length l instead of l_e, with the same reason as in equation (1).

Figure 9. shows the tendency also that increasing collimator heights increases their spatial resolution. The difference between theoretical and simulation was appeared since the theoretical approach is based on the planar crystal, not the pixellated crystal, which means the planar crystal has wider light spreading than the pixellated as it is well known [8]. However, the spatial resolution in the simulation results can not be small below 3 mm since the minimum size of the collimator and the pixellated crystal is 3 mm.

Trade-Offs

With the above results from the geometric efficiency and the spatial resolution, both are conflicting parameters in the collimator design as shown in figure 10.

The former requires short collimator height, while the latter requires the opposite. According to the purpose of collimator for the diagnosis

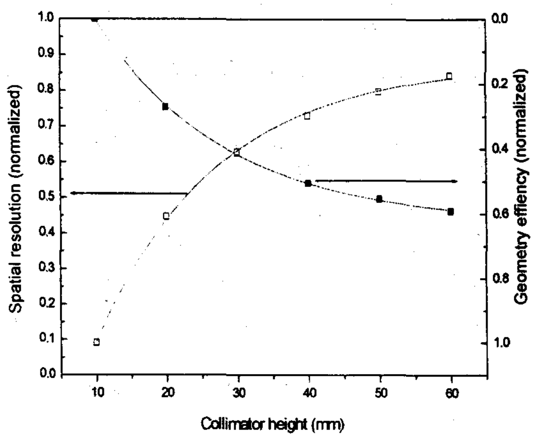

Fig. 10. Trade-off between spatial resolution and geometric efficiency.

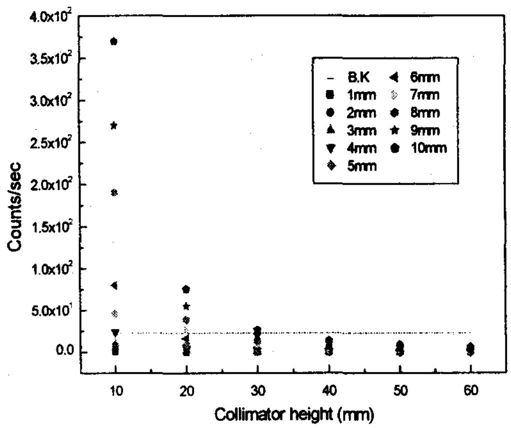

Fig. 11. Counts limit depending on the collimator heights and tumor sizes

of the breast tumor, it is useful to choose desirable collimator for the expected tumor size through the Trade-Offs. For example, the collimator heights of 10 ~ 20 mm for high sensitivity, 20 ~ 40 mm for all-purpose, and 40 ~ 60 mm for high resolution are generally recommended.

Evaluation of Detector Head Performance

From the simulations and their Trade-Offs, it is recommendable to use the high sensitivity collimator having short collimator to detect small cancer with early. However, we should also consider whether it is possible to detect small size tumor with the recommended high sensitivity collimator.

In the diagnosis of the breast cancer, considering the background count from non-tumor volume, there is a restriction to use recommended collimator. To give more detail information on diagnosis, we calculated detectable counts, *C*, depending on the tumor size and compared with the background count. The calculation based on simulation with derived parameter values is given in equation (3).

$$C = \sum \varepsilon_{photopeak} \times RBT \times Time \times Volume \qquad (3)$$

where $\sum \varepsilon_{photopeak}$ is the summation of geometric efficiency of total photopeak counts using simulation. The *RBT* (Radioactivity in Breast Tissue per volume) is assumed 14800 cps, equal to 400 nCi/cm^3 [5]. The *Time* is detecting time for diagnosis, and the *Volume* is the various tumor volume. The background count per volume was assumed as 22.5 [5] from non-tumor volume. Figure 11. shows the limits of each collimator according to the background count.

This figure tells us that although the collimator of 30 mm has a capability to detect breast cancer below 5 mm size as shown in figure 11., it is impossible to detect that size of breast cancer if we consider background count from non-tumor organ.

Conclusion

In this paper, we derived the optimum parameters of the detector head of gamma camera for breast cancer detection. From the research results we proposed that the optimum crystal height is 5 ~ 6 mm and the top and the side surface condition of the crystal should be painted 0.98RC and polished 0.98RC respectively. Under these conditions, the light transmission is about 94 ~ 94.5 %. In the case of the septal thickness of 0.25 mm with the hole size of 3 mm x 3 mm, the collimator heights of 10 ~ 20 mm for high sensitivity, 20 ~ 40 mm for all-purpose, and 40 ~ 60 mm for high resolution are generally recommended. For the optical bonding material the refractive index of 1.8 gives the best light transmission rate and the effect of its thickness is negligible.

Generally, the Trade-Offs procedure between the geometric efficiency and the spatial resolution has been recommended to drive optimum parameters of detector head of gamma camera and select for diagnosis purpose. However, when the selection of the collimator is concerned, the results of the Trade-Offs give only a reference even though this Trade-Offs shows specific value such as the efficiency or sensitivity and the spatial resolution. The background level in practical use condition is a limit factor in selection of best collimator particularly in early detection of small tumor.

Therefore, in early diagnosis of breast cancer when the size of breast cancer is relatively small, it is concluded that the spatial resolution, which is used as an indicator to distinguish the various sizes of breast cancer and is dependent on the characteristic of the detector head, appears to be meaningless.

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