Technical Note: Determining regions of interest for CCD camera-based fiber optic luminescence dosimetry by examining signal-to-noise ratio

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Purpose: The goal of this work was to develop a method for determining regions of interest (ROIs) based on signal-to-noise ratio (SNR) for the analysis of charge-coupled device (CCD) images used in luminescence-based radiation dosimetry.

Methods: The ROI determination method was developed using images containing high-and low-intensity signals taken with a CCD-based, fiber optic plastic scintillation detector system. A series of threshold intensity values was defined for each signal, and ROIs were fitted around the pixels that exceeded each threshold. The SNR for each ROI was calculated and the relationship between SNR and ROI area was examined.

Results: The SNR was found to increase rapidly over small ROIs for both signal levels. After reaching a maximum, the SNR of the low-intensity signal decreased steadily over larger ROIs, but the high-intensity SNR did not decrease appreciably over the ROI sizes studied. The spatial extent of the normalized images showed intensity independence, suggesting that a fixed ROI is useful for varying signal levels.

Conclusions: The method described here constitutes a simple yet effective method for defining ROIs based on SNR that could enhance the low-level detection capabilities of CCD-based luminescence dosimetry systems. © 2011 American Association of Physicists in Medicine.
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Key words: scintillation dosimetry, image analysis, region of interest

I. INTRODUCTION

Much of the current research in *in vivo*, real-time medical radiation dosimetry involves the development and testing of fiber optic systems based on various luminescence detectors including plastic scintillation detectors (PSDs), inorganic scintillators, and optically stimulated luminescence dosimeters.¹⁻⁶ The charge-coupled device (CCD) has be-

come increasingly popular for this application as it can be used to measure multiple signals simultaneously, e.g., to remove the Cerenkov component from a fiber optic signal or to image the output from multiple detectors.^{7,8} Typically, in the quantitative signal analysis of CCD images, a region of interest (ROI) is defined that includes pixels containing the desired signal and excludes other pixels. The choice of

whether to include certain pixels of an image in a specific ROI may change depending on what “looks best” to different observers. While such choices do not necessarily have a negative impact on the final results, they do introduce a level of subjectivity to an important aspect of the analysis and are likely not tailored to acquire dynamic signals most effectively. For example, to perform accurate real-time *in vivo* dosimetry during procedures such as intensity-modulated radiation therapy, a detector must be able to account not only for the easily measured high-intensity signals resulting from direct hits by the radiation beam but also for the accumulation of low-level signals due to scatter or the beam grazing the detector. The goal of the research described here was to develop a technique for defining an ROI so that signal-to-noise ratio (SNR) may be maximized with respect to lower-intensity CCD signals.

II. METHODS AND MATERIALS

The data used for developing the ROI optimization method were acquired using a PSD system based on that described previously by Archambault *et al.*,⁶ with the key difference being that the single PSD used here consisted of a BCF-60 scintillating fiber (Saint-Gobain Crystals, Paris, France). In this system, the spectral band containing most of the scintillation from the BCF-60 and the shorter-wavelength band containing mostly Cerenkov light (resulting from irradiation of the scintillator and the adjacent portion of the fiber optic cable) are discriminated via a dichroic mirror setup per the chromatic removal technique.⁷ The light signals from these two bands are imaged simultaneously but spatially apart by a Luca electron-multiplying CCD camera (Andor Technology, Belfast, N. Ireland). Only scintillation-dominant signals were used in this study.

Two “stacks” of images—one of five high-intensity signals (corresponding to a delivered dose of 0.808 Gy) and one of 25 low-intensity signals (0.008 Gy)—were obtained by exposing the PSD to a collimated ⁶⁰Co beam, and two similar stack of background images were acquired with no radiation delivered. Two mean background images were constructed (one from each background image stack) such that each pixel equaled the mean value of the pixels at the same location in each of the 5 or 25 images of the original background stack. The mean background was subtracted from the signal images, and two mean signal images were formed in the same fashion as the mean background images. The standard deviation corresponding to each of the pixel means (i.e., the standard deviation of the values of a pixel across an image stack) was also calculated. All of the mean pixel values within an ROI were then summed to obtain the integrated signal value, S_{ROI} , and the total standard deviation for an ROI, σ_{ROI} , was calculated by summing all of the individual pixel standard deviations in quadrature. The SNR for each ROI determined using the steps given below was calculated using $\text{SNR} = S_{\text{ROI}} / \sigma_{\text{ROI}}$.

The method used to determine the ROIs for a particular signal level is described as follows. A range of n threshold values A was calculated such that for $1 \leq i \leq n$,

FIG. 1. Two- and three-dimensional depictions of the high-intensity signal image data. Multiple ROIs are plotted around the 3D surface at the threshold pixel intensities by which they were constructed. Rounded threshold values for $i=5, 18, 32$, and 50 [see Eq. (1)] are shown for illustrative purposes. The ROI in bold shown on both depictions matches the ROI that produces the highest value of SNR_{low} .

$$A_i = s_{\text{peak}} \left(1 - \frac{i}{n} \right) + s_{\text{base}} \left(\frac{i}{n} \right), \quad (1)$$

where s_{peak} is equal to the maximum pixel intensity of the background-subtracted mean signal image and s_{base} is equal to three times the standard deviation of the background ($3\sigma_{\text{bkgd}}$). The value for s_{base} was chosen to increase the statistical likelihood that the ROI determined from the lowest threshold would encompass the entire imaged scintillation signal. For each of the thresholds $\{A_1, A_2, A_3, \dots, A_n\}$, a rectangular ROI was computed to fit around the region containing pixels with values exceeding the threshold. Figure 1 illustrates the ROIs corresponding to $n=50$ threshold values obtained for the high-intensity signal. SNR values (obtained using all of the pixels within each ROI) were plotted vs the corresponding ROI areas.

FIG. 2. SNR vs ROI area for both low- and high-intensity signals. The left vertical axis corresponds to SNR_{low} data (solid), while the right vertical axis corresponds to SNR_{high} data (hollow). Two data points ($[\text{ROI area}, \text{SNR}_{\text{high}}] = [361, 1129], [576, 1116]$) are not shown in order to maintain an abscissa scale that presents the smaller ROI data in a clear fashion.

III. RESULTS

Figure 2 shows the SNR vs ROI area results for both high- and low-intensity signals. The plots reveal an initial rapid increase in SNR with ROI area toward a maximum value. The SNR for the low-intensity signal (SNR_{low}) reached a maximum of 50 at an ROI area of 121 pixels and showed a steady decrease as the ROI area increased further. The high-intensity SNR (SNR_{high}) begins to level off at an ROI area of 100 pixels and reaches a maximum of 1129 at 272 pixels. This level is maintained until reaching the largest calculated ROI size of 576 pixels, where SNR_{high} decreases to 1116 (data not shown, see caption of Fig. 2). Identical ROIs—and thus, identical SNR values—were often produced over several adjacent threshold values. This explains why Fig. 2 shows fewer than $n=50$ data points for each signal level.

An *ad hoc* investigation into the constancy of the relative intensity profiles of the high- and low-level signals was performed by normalizing the pixels in each image by the image's maximum pixel value, applying a smoothing filter to reduce the influence of pixel noise in the low-intensity signal, and taking the ratio of the signal contained within the ROI that corresponded to maximum SNR_{low} to the signal contained within the entire image (for reference, see two-dimensional depiction in Fig. 1). This ratio equaled 0.835 and 0.826 for the normalized low- and high-intensity images, respectively, giving a percent difference of 1.1%, which is below the 2% measurement uncertainty (1σ , from the reciprocal of the maximum SNR_{low}) associated with that particular ROI.

IV. DISCUSSION AND CONCLUSION

We found that the integrated signal increases faster than pixel noise over small ROIs. However, over larger ROIs, the increased amount of pixel noise and diminished signal increase caused SNR_{high} to plateau and caused SNR_{low} to decrease steadily. The absence of any substantial decrease in SNR_{high} for larger ROIs may be due to our chosen range of

ROI areas, as the amount of pixel noise necessary to cause a significant decrease in SNR_{high} might only be found using ROIs much larger than the largest used here (576 pixel area). While Fig. 2 shows that the ROI corresponding to the maximum SNR depends on the intensity of the signal for which the ROI is being determined, the normalized intensity profile comparison shows that the relative signal within a fixed ROI is well preserved regardless of the original signal intensity. We therefore suggest that a fixed ROI chosen according to the highest SNR corresponding to the signal with the lowest intensity of interest may be used to effectively measure signals with a large dynamic range. This approach would guarantee greater SNR (i.e., greater reproducibility) for low-signal measurements with only a small sacrifice of SNR at higher signal intensities, where having the highest possible SNR may not be crucial. It is important to note that the reliability of ROIs constructed via this method may depend heavily on the preservation of the mechanical alignment of the CCD camera and associated optics, as well as the constancy of the intensity pattern of the light output from the fiber optic cable.

It is difficult to provide an estimate of the improvement, if any, that the method described here grants to the outcome of real-world measurements. One could subjectively define ROIs that are smaller, larger, or (accidentally) exactly the same as those determined using our method, and therefore little to no improvement may be seen. However, in Fig. 2, a simple comparison can be made between our optimal SNR method and a statistical approach that is based on pixel intensities above 3σ of the background. The latter was included in this study and is represented by the ROI constructed from the lowest intensity threshold, A_{50} , which also has the largest pixel area. The SNR_{low} for the ROI corresponding to A_{50} ($3\sigma_{\text{bgd}}$) is $\sim 25\%$ lower than the maximum SNR, while no substantial difference is seen between the analogous SNR_{high} values. Although this data set is rather small, this shows that a nontrivial improvement in the reproducibility (i.e., reduction in uncertainty) of low-intensity sig-

nals is obtained with our method in this case.

By virtue of its general importance in signal measurement devices, the SNR represents a useful benchmark by which to base the choice of ROI(s) for CCD image analysis. The specific procedures used here to compute the SNR are not essential to the method. Any reasonable approach for calculating SNR pertaining to particular ROIs should suffice. Furthermore, this method could easily be adapted for ROIs having elliptical or perhaps even more complex shapes, although the absolute SNR value(s) would, in general, depend on the ROI size and shape. In conclusion, the presented technique could be useful for enhancing the analysis of CCD images acquired with any luminescence dosimetry system having a basic configuration similar to the one used here.

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- ¹A. S. Beddar, T. R. Mackie, and F. H. Attix, "Water-equivalent plastic scintillation detectors for high-energy beam dosimetry: II. Properties and measurements," *Phys. Med. Biol.* **37**, 1901–1913 (1992).
- ²M. A. Clift, R. A. Sutton, and D. V. Webb, "Water equivalence of plastic organic scintillators in megavoltage radiotherapy bremsstrahlung beams," *Phys. Med. Biol.* **45**, 1885–1895 (2000).
- ³A. L. Huston, B. L. Justus, P. L. Falkenstein, R. W. Miller, H. Ning, and R. Altemus, "Remote optical fiber dosimetry," *Nucl. Instrum. Methods Phys. Res. B* **184**, 55–67 (2001).
- ⁴J. C. Polf, E. G. Yukihiro, M. S. Akselrod, and S. W. S. McKeever, "Real-time luminescence from Al₂O₃ fiber dosimeters," *Radiat. Meas.* **38**, 227–240 (2004).
- ⁵C. E. Andersen, S. K. Nielsen, S. Greulich, J. Helt-Hansen, J. C. Lindegaard, and K. Tanderup, "Characterization of a fiber-coupled Al₂O₃:C luminescence dosimetry system for online in vivo dose verification during ¹⁹²Ir brachytherapy," *Med. Phys.* **36**, 708–718 (2009).
- ⁶L. Archambault, T. M. Briere, F. Pönisch, L. Beaulieu, D. A. Kuban, A. K. Lee, and A. S. Beddar, "Toward a true real-time in-vivo dosimetry system using plastic scintillation detectors," *Int. J. Radiat. Oncol., Biol., Phys.* **78**, 280–287 (2010).
- ⁷J. M. Fontbonne, G. Iltis, G. Ban, A. Battala, J. C. Vernhes, J. Tillier, N. Bellaize, C. Le Brun, B. Tamain, K. Mercier, and J. C. Motin, "Scintillating fiber dosimeter for radiation therapy accelerator," *IEEE Trans. Nucl. Sci.* **49**, 2223–2227 (2002).
- ⁸L. Archambault, A. S. Beddar, L. Gingras, F. Lacroix, R. Roy, and L. Beaulieu, "Water-equivalent dosimeter array for small-field external beam radiotherapy," *Med. Phys.* **34**, 1583–1592 (2007).