Quantitation in planar renal scintigraphy: which μ value should be used?

Elif Hindié[1], Irène Buvat[2], Christian Jeanguillaume[1], Alain Prigent[3], Pierre Galle[1]

[1] Laboratoire de Biophysique et Médecine Nucléaire, Faculté de Médecine Paris XII, Créteil, France
[2] U494 INSERM, CHU Pitié-Salpêtrière, Paris, France
[3] Service de Médecine Nucléaire, Hôpital Bicêtre, Assisstance Publique des Hôpitaux de Paris, Le Kremlin Bicêtre, France

Received 15 July and in revised form 25 August 1999

Abstract. The attenuation coefficient value μ used by different authors for quantitation in planar renal scintigraphy varies greatly, from the theoretical value of 0.153 cm^{-1} (appropriate for scatter-free data) down to 0.099 cm^{-1} (empirical value assumed to compensate for both scatter and attenuation). For a 6-cm-deep kidney, such variations introduce up to 30% differences in absolute measurement of kidney activity. Using technetium-99m phantom studies, we determined the μ values that would yield accurate kidney activity quantitation for different energy windows corresponding to different amounts of scatter, and when using different image analysis approaches similar to those used in renal quantitation. With the 20% energy window, it was found that the μ value was strongly dependent on the size of the region of interest (ROI) and on whether background subtraction was performed: the μ value thus varied from 0.119 cm^{-1} (loose ROI, no background subtraction) to 0.150 cm^{-1} (kidney ROI and background subtraction). When using data from an energy window that could be considered scatter-free, the μ value became almost independent of the image analysis scheme. It is concluded that: (1) when performing background subtraction, which implicitly reduces the effect of scatter, the μ value to be used for accurate quantitation is close to the theoretical μ value; (2) if the acquired data were initially corrected for scatter, the appropriate μ value would then be the theoretical μ value, whatever the image analysis scheme.

Key words: Renal quantitation – Attenuation correction – Scatter correction

Eur J Nucl Med (1999) 26:1610–1613

Correspondence to: E. Hindié, Laboratoire de Biophysique, Faculté de Médecine, 8, rue du Général Sarrail, F-94000 Créteil, France

Introduction

Absolute quantitation of kidney tracer uptake using planar scintigraphy requires correction for attenuation due to kidney depth and subtraction of background activity. When using the posterior view method, attenuation correction is performed by multiplying the count density measured in a kidney region of interest (ROI) by the correction factor $e^{\mu d}$, where d is the distance between the kidney centre of gravity and the posterior body outline and μ is an attenuation coefficient. The appropriate μ value to be used for accurate quantitation is not established. Some authors use the theoretical narrow-beam value (0.153 cm^{-1} in water for technetium-99m photons), neglecting the effect of scatter [1, 2]. Others use a broad-beam μ value to compensate for scatter [3]: renal ^{99m}Tc-dimercaptosuccinic acid (DMSA) and ^{99m}Tc-diethylene triamine penta-acetic acid (DTPA) studies have thus been performed with μ values of 0.131 cm^{-1} [4], 0.122 cm^{-1} [5], 0.12 cm^{-1} [6], 0.1 cm^{-1} [7] and 0.099 cm^{-1} [8].

Variations in μ value may introduce major inaccuracies when estimating kidney activity. For a 6-cm-deep kidney (average depth in normal-sized adults), the correction factor $e^{\mu d}$ is 2.5 for μ = 0.153 cm^{-1} and 1.81 for μ = 0.099 cm^{-1}, yielding a difference of up to 30% in measured renal uptake. When using the conjugate view method to estimate kidney activity, the correction factor is $e^{\mu T/2}$, where T is the total body thickness. As $T/2$ is usually greater than d, this method is even more sensitive to variations in μ [5].

When not correcting for scatter, the empirical μ value should compensate for both scatter and attenuation, and the appropriate μ value therefore varies with the amount of scatter in the data. As ROI delineation and background subtraction can also affect the scatter content of the data, these steps, to which little attention has so far been paid, may also affect the appropriate μ value. The purpose of this work was to investigate the role of these image analysis steps and to suggest approaches for improved accuracy.

Material and methods

A 1-cm-thick 8 cm×5 cm Perspex phantom containing 18.5 MBq ^{99m}Tc pertechnetate was used. Acquisitions were performed using a gamma camera (Helix, Elscint) equipped with a low-energy general-purpose collimator.

A 1-min image of the phantom was first acquired with a distance of 15 cm between the phantom surface and the collimator. Then, a 10-cm-thick 20 cm×20 cm Perspex container filled with water was positioned between the phantom and the collimator and a 1-min image was acquired.

These two acquisitions were performed using five different energy windows: the conventional 20% window (140 keV±10%) and four other 20% windows centred on 145, 150, 155 and 160 keV respectively. The offset windows allowed us to obtain images with various proportions of scattered photons: the larger the offset, the lower the proportion of scatter. The window with the largest offset included almost no scatter.

For each pair of acquisitions performed with and without attenuating medium, the μ value was calculated using $\mu = d^{-1} \times \ln Co/C$, where d is the water thickness (10 cm), and Co and C are the average number of counts measured in a given ROI without and with water attenuation respectively. For measuring Co and C, one of the following ROIs was used:

- An ROI corresponding to the whole camera field of view (FOV) (ROI_0).
- An 11 cm×8 cm rectangular ROI drawn around the phantom (ROI_1).
- ROI_1 from which the average number of counts detected in an outside rim ("background" ROI_2) drawn around ROI_1 was subtracted. The rim thickness was 0.88 cm and the distance between the inner border of the rim and ROI_1 was 1.1 cm.

Results

The influence of ROI delineation and background subtraction on the μ values for the different energy windows is shown in Table 1. In the 20% energy window, the estimated μ values varied between 0.119 cm^{-1} and 0.150 cm^{-1} depending on which data were used for the calculation, the latter value being obtained for the kidney phantom ROI combined with background subtraction.

When using offset energy windows, the μ values became less dependent on the image processing steps: the larger the offset, the smaller the variability in μ values (Table 1).

Table 2 shows the percent variations in absolute kidney uptake estimated for various μ values when using the correction factor $e^{\mu d}$ (posterior view method) for kidney depths encompassing the range of realistic values, from young children to obese adults.

When considering only relative quantitation of kidney uptake (ratio of right to left kidney uptake), the choice of μ value does not affect the result when the two kidneys are at the same depth. It does slightly affect the result if the two kidneys are at different depths: when using a value μ' instead of μ, the uptake ratio is multiplied by a factor $\exp[(\mu' - \mu)\Delta]$ when the difference in kidney depths is Δ. For example, if a kidney is 2 cm deeper than the other, which was the case in 13% of a population of patients [2], a relative uptake of 50:50 (ratio=1) with μ=0.153 cm^{-1} would be 47:53 (ratio=0.89) for a μ value of 0.1 cm^{-1}.

Table 1. Influence of ROI delineation and background subtraction on the μ value: results for five different energy windows

Energy window	Sensitivity loss relative to 140 keV±10%	μ value using counts from			Change in μ due to image processing
		ROI_0	ROI_1	ROI_1 and ROI_2	
140 keV±10% (126–154 keV)	0%	0.119	0.138	0.150	26%
145 keV±10% (130–159 keV)[a]	4%	0.125	0.140	0.150	20%
150 keV±10% (135–165 keV)	16%	0.135	0.145	0.152	13%
155 keV±10% (139–170 keV)	43%	0.143	0.149	0.153	7%
160 keV±10% (144–176 keV)	77%	0.149	0.153	0.154	3%

ROI_0, Whole camera field of view; ROI_1, kidney phantom ROI; ROI_1 and ROI_2, kidney phantom ROI combined with background subtraction

[a] The lower energy limit for this window (130 keV) is close to that (129 keV) of a centred narrow window 140 keV±7.5% which is often used on recent cameras

Table 2. Percent differences in kidney uptake: estimates for various broad-beam values with respect to the uptake estimated when using the narrow-beam value $\mu = 0.153\ cm^{-1}$

μ value (cm^{-1}) [reference]	Kidney depth			
	2 cm	4 cm	6 cm	9 cm
0.153 [1–2]	0	0	0	0
0.131 [4]	–4	–8	–12	–18
0.122 [5]	–6	–12	–17	–24
0.12 [6]	–6	–12	–18	–26
0.1 [7]	–10	–19	–27	–38
0.099 [8]	–10	–19	–28	–38

Discussion

Absolute quantitation of tracer activity in a single kidney provides information regarding kidney function which is more useful than a simple comparison of the right and left kidney counts [9, 10]. To measure kidney activity, most nuclear physicians use manufacturer software. These programs include a correction factor for attenuation correction, using either the narrow-beam μ value for water ($0.153\ cm^{-1}$) or one of the broad-beam μ values found in the renal quantitation literature, which vary from $0.099\ cm^{-1}$ to $0.131\ cm^{-1}$ [4–8]. We showed that for the posterior view method, variations in the μ value introduce large differences in estimated kidney uptake (Table 2). The conjugate view method and single-photon emission tomography studies are even more sensitive to variations in μ [5]. These results could explain in part why reported values for absolute renal uptake of DMSA in normal adults (right + left kidney) vary so widely among authors (e.g. 54% of injected DMSA activity [9], 41.7% [4] and 32% [11]).

An effective broad-beam μ value should match the scatter content of the image, which depends on the acquisition parameters. We demonstrated that the data analysis method used to measure kidney uptake should also be taken into account because it affects the scatter content of the data. Using the whole FOV to estimate μ (ROI_0) includes many scattered events, and hence results in a large underestimation of attenuation. Using a smaller ROI to estimate μ (ROI_1) excludes some scattered photons and thus causes a smaller underestimation. Finally, as there was no background activity in the experimental set-up, background subtraction removed scatter. Indeed, assuming that the average scatter content per pixel did not vary rapidly with location, subtracting the average counts in ROI_2 from the average counts in ROI_1 was approximately equivalent to subtracting scatter from ROI_1. The resulting μ value was therefore close to the narrow-beam value. Note that discussion of the μ value here and in other studies [4–8] concerns the variation in its estimation under variable measurement conditions with a gamma camera and not the attenuation coefficient itself.

The 10-cm water attenuation that we used exceeds the mean kidney depth in adults for a posterior view but is less than the average depth from an anterior view used in the conjugate-view method. Varying the thickness of the attenuating medium (results not shown) mainly affected the μ estimate when using ROI_0 (whole camera FOV). For example, for a 14-cm-thick attenuating medium the μ value was $0.115\ cm^{-1}$ for ROI_0 and $0.149\ cm^{-1}$ when using ROI_1 and background subtraction.

Findings from this work suggest that if ^{99m}Tc-DMSA renal activity is estimated using background subtraction, most of the scatter will be removed by the subtraction, implying that the narrow-beam μ value should then be used. Combining background subtraction and use of the narrow-beam μ value could help reduce the variability of ^{99m}Tc-DMSA uptake measurements in different nuclear medicine units. Whether these findings could also be applied to studies involving other renal tracers such as mercaptoacetyltriglycine and DTPA, for which the ratio of kidney activity to that of neigbouring tissues is lower, needs further investigation.

An even better approach to avoid choosing an empirical μ value would be to correct the data for scatter. So doing, the narrow beam μ value could then be applied whatever the image analysis protocol: we demonstrated that for data that could be considered scatter-free, the appropriate μ value became almost independent of the data analysis method (Table 1). Although we used offset windows to obtain scatter-free data, use of such windows is not recommended because of the sensitivity loss and potential uniformity artefacts it can cause. However, many scatter correction methods are available [12] and their value for improving renal quantitation should be investigated. Absolute quantitation might also benefit from measurement of patient real tissue attenuation properties using transmission acquisition devices.

References

1. Maneval DC, Magill HL, Cypess AM, Rodman JH. Measurement of skin-to-kidney distance in children: implications for quantitative renography. *J Nucl Med* 1990; 31: 287–291.
2. Gruenewald SM, Collins LT, Fawdry RM. Kidney depth measurement and its influence on quantitation of function from gamma camera renography. *Clin Nucl Med* 1985; 10: 398–401.
3. Jaszczak RJ, Coleman RE, Whitehead FR, Physical factors affecting quantitative measurements using camera-based single photon emission computed tomography (SPECT). *IEEE Trans Nucl Sci* 1981; 28: 69–80.
4. Zananiri MC, Jarritt PH, Sarfarazi M, Ell PJ. Relative and absolute ^{99m}Tc DMSA uptake measurements in normal and obstructed kidneys. *Nucl Med Commun* 1987; 8: 869–880.
5. Murase K, Tanada S, Ishine M, Yokoyama M, Hamamoto K. Methods for measuring the renal uptake rate of ^{99m}Tc dimercaptosuccinic acid (DMSA): a comparative study. *Eur J Nucl Med* 1990; 16: 725–731.

6. Fleming J, Keast CM, Waller DG, Ackery D. Measurement of glomerular filtration rate with ^{99m}Tc DTPA: a comparison of gamma camera methods. *Eur J Nucl Med* 1987; 13: 250–253.
7. Russel CD, Dubovsky EV. Gates method for GFR measurement. *J Nucl Med* 1986; 27: 1373–1374.
8. Gordon I, Evans K, Peters AM, Kelly J, Morales BN, Goldraich N, Yau A. The quantitation of ^{99m}Tc DMSA in paediatrics. *Nucl Med Commun* 1987; 8: 661–670.
9. Kawamura J, Hosokawa S, Yoshida O, Fujita T, Yasushi I, Torizuka K. Validity of ^{99m}Tc dimercaptosuccinic acid renal uptake for an assessment of individual kidney function. *J Urol* 1978; 119: 305–309.
10. Taylor A. Quantitation of renal function with static imaging agents. *Semin Nucl Med* 1982; 4: 330–343.
11. Groshar D, Moskovitz B, Issaq E, Nativ O. Quantitative SPECT of DMSA uptake by the kidneys: assessment of reproducibility. *Kidney Int* 1997; 52: 817–820.
12. Buvat I, Benali H, Todd-Pokropek A, Di Paola R. Scatter correction in scintigraphy: the state of the art. *Eur J Nucl Med* 1994; 21: 675–694.