Title: Antero-posterior (AP) pelvis x-ray imaging on a trolley: impact of trolley design, mattress design and radiographer practice on image quality and radiation dose

Introduction

There are many technical and physical challenges associated with imaging on a trolley which have subsequent impact on image quality and radiation dose. These challenges include: the absence of AEC on a trolley; grid selection; geometric factors; mattress and trolley design.

An antero-posterior (AP) pelvis projection is often performed on trolley bound patients especially in trauma situations because transferring them onto the x-ray tabletop could exacerbate injuries causing further harm [1]. The AP pelvis projection irradiates radiosensitive organs including the gonads and is ranked the third highest radiation dose examination by the Health Protection Agency (HPA) [2]. Lead shielding of the gonads is considered essential when imaging the pelvis *except* for the initial imaging such as for trauma since it might obscure important diagnostic information. Organ dose from a single AP pelvis projection can typically reach 2.1mGy for the testes and 0.52mGy for the ovaries, which are within the primary beam [3.] With the challenges associated with trolley imaging, combined with the radiation implications of AP pelvis projection, it seems to be an important area to explore and subsequently optimise.

The aims of this study were to: 1. explore whether acquisition parameters used for AP pelvis radiography on the x-ray tabletop are transferable to trolley imaging; 2. evaluate different acquisition parameters for trolley imaging in order to optimise image quality and radiation dose for an AP pelvis projection.

Method

This study used an experimental approach by imaging a pelvic anthropomorphic phantom under controlled conditions.

Imaging equipment and technique

A Philips Bucky Diagnost x-ray unit with an Optimus 50kW high frequency generator was used (Philips Healthcare, Netherlands). The same 35 x 43cm Fuji IP HR-V computed radiography image receptor (Barium Flurohalide (BaFX) phosphor) was used for all exposures. This was processed using a Fuji FCR Capsula XII with 50-micron resolution (Fujifilm Medical Systems, Japan). Quality assurance was conducted on all equipment prior to image acquisition in accordance with IPEM 91 [4], which included radiation output reproducibility and sensitometry testing. All test results fell within expected tolerances.

Images were acquired using a Rando SK250 sectional lower torso anthropomorphic pelvis phantom [5]. The phantom was positioned supine on the x-ray tabletop for the acquisition of a reference image which was subsequently used as the optimal comparison image. The acquisition parameters used to acquire the x-ray tabletop reference image were those typically employed in clinical practice and recommended in various published work [6-11]. They included a 110cm source to image distance (SID), the outer chambers of the automatic exposure control, 75 kV, an oscillating grid mounted into the x-ray table Bucky, 3.2mm Al equivalent total filtration and a broad focal spot size (1mm). For all exposures, the collimation was adjusted to the region of clinical interest to include the iliac crests, greater trochanters and proximal one third of the femora.

Experiment technique

The experimental images were acquired on one commercially available trolley (Lifeguard 50 trolley) using two different mattresses (standard 65mm and Bi-Flex 130mm). Images were also acquired with the image receptor holder (platform) elevated and lowered, for comparison. The Lifeguard 50 trolley platform that accommodates the image receptor should be elevated prior to an exposure to reduce object to image distance (OID). However, in clinical practice this elevation may not always be achieved [12]. All images were acquired with a commercially available stationary focused grid (focused to 105cm ±15cm) with a grid ratio of 10:1 and strip density of 40 lines/cm [13]. Initially, images were to be acquired with and without a grid to explore the air gap technique however this idea was eliminated following a preliminary experiment demonstrating significant image quality deterioration without a grid. For each projection on the trolley, the mAs increment was varied from 16mAs (which was the AEC reading derived from the acquisition parameters used to acquire the reference image) to 20mAs, 25mAs and 32mAs. Three different SIDs were also used, with an initial setting of 110cm and then two further distances of 120cm and 130cm. These were to

compensate for the increased OID as a result of trolley design but also to reduce radiation dose as found in previous studies [14-16]. A 130cm SID was considered the maximum practical and achievable SID to be used considering the effective range of the stationary grid and grid cut off. Both Heath et al. and Tugwell et al. also found that image quality deteriorated at higher SID values [14, 16]. SID was measured manually with a tape measure by two radiographers to ensure consistency. All other acquisition parameters remained constant including the use of 75kVp. This resulted in 48 experimental images being produced on the trolley under different conditions.

Radiation dose calculations

Entrance surface dose (ESD) was measured at the surface of the phantom at the centre of the collimation field using the Unfors Mult-O-Meter 407L ionising chamber (Unfors Equipments, Billdal, Sweden). Three repeated exposures were performed and then averaged in order to reduce random error. Effective dose was calculated using Monte Carlo dosimetry simulation software (PCXMC 2.0)(STUK, Helsinki, Finland). This software uses tissue weighting factors from ICRP Publication 103 [17] to estimate effective dose in milliseverts (mSv). Dose area product (DAP) was used in this estimation along with the acquisition parameters.

Assessment of image quality

Following ethical approval from the School of Healthcare Sciences, University of Salford (HSCR14/104), relative visual grading analysis (VGA) with bespoke software to present the images and capture responses from observers [18]. Previous research has reported on the benefits of relative VGA in comparison to an absolute VGA as it allows easier detection of differences in quality as oppose to observers evaluating images utilising criteria without a comparison reference image [19]. The observers consisted of five diagnostic radiographers with more than five years clinical experience who were blinded to the parameters used to acquire all images.

The bespoke software allowed for two images to be presented simultaneously on dual side-by-side 5 megapixel monitors [4,20]; one the reference image (standard practice x-ray tabletop image) which was permanently displayed on the left monitor whilst the experimental images

(acquired on the trolley) were displayed in random order in the right monitor. The display software prohibits post processing capabilities such as zooming and window adjustments and therefore differences detected between images would more likely be the result of acquisition parameters/technique change. The monitors were calibrated for Digital Imaging and Communications in Medicine (DICOM) grayscale standard display function which is to the recommended specification of the Royal College of Radiologists [21]. A visual pattern check (AAPM in report 93) was undertaken prior to each observer undertaking visual evaluation [22]. Room lighting conditions were maintained at a dimmed and consistent level (luminance of >170 cd/m2) in accordance with the European Guidelines on Quality Criteria for Diagnostic Radiographic Images [23].

Observers were required to score the experimental images against the reference image using a visual grading scale which consisted of 15 items [24] (Table 1). The items were scored using a 5-point Likert scale where '1' indicated much worse than the reference image, '2' slightly worse, '3' equal to, '4' better than, and '5' much better than the reference image. Image quality scores for each of the 15 items were totalled; for each image, scores ranged from 15 to 75. An image which scored 45 indicated equal quality to that of the reference image, a score of > 45 was considered an improvement in image quality and anything lower than 45 considered a decrease in image quality. An additional item was also included at the end of the 15 item image criteria scale (Table 1), which involved a binary decision (yes or no answer). For this item, the observers considered the overall diagnostic quality of each experimental image, deciding whether they were acceptable or unacceptable for diagnostic purpose.

The magnification factor was derived for all images. The right femoral head diameter (FHD) was measured in millimetres by one radiographer with experience in pre-operative hip arthroplasty templating. The measurements were carried out using the ruler (callipers) tool in Synapse PACS system (Fujifilm, Japan) using the same workstations as for the visual image quality assessment task. The femoral head of each image was measured eight times and the average, standard deviation (SD), minimum and maximum values were then calculated. No cropping was permitted post processing and therefore the displayed magnification could only be influenced by acquisition parameters used to acquire the images.

Contrast to Noise Ratio (CNR) was calculated as a physical measure of image quality. CNR has been used successfully as a measure of image quality in various optimisation studies [25-27] and in comparison to Signal to Noise Ratio (SNR), CNR takes into consideration the effect of

noise on our ability to distinguish objects within the image because visibility depends on contrast (the difference between signals). A highly exposed image may have a high SNR but show no useful information on that same image [28]. CNR was calculated by placing a region of interest (ROI) on two homogeneous structures within the anthropomorphic pelvic phantom images in order to sample the mean and standard deviation of the pixel value. The ROI was placed in the same position for the experimental images in accordance with Bloomfield et al. [29] to allow a consistent value for comparison (Figure 1). In order to maintain a consistent ROI, magnification was considered and ROI adjusted to ensure the same anatomy was sampled for all images. This meant that femoral head diameter and thus magnification had to be performed prior to calculating CNR in order to inform the ROI adjustments. This was necessary because using the same size ROI for all images would induce a level of inaccuracy to the CNR measurements since the anatomy sampled within that ROI would vary depending on the magnification level of the image. Image J software (National Institutes of Health, Bethesda, MD) was used to calculate CNR; this software tool is used regularly in literature for similar calculations [11, 30-31]. Using this approach, the mean pixel value and the standard deviation for the ROI was acquired [32]: subsequently the following equation was used to determine CNR:

$$C = \frac{|S_A - S_B|}{\sigma_o}$$

Where S_A and S_B are signal intensities for signal producing structures $A(ROI_1)$ and B $(ROI_2)$and σ_o is the standard deviation (blue ROI) of the pure image noise.

Optimisation score

Many optimisation studies [11, 16, 33] consider radiation dose and image quality data separately; however Williams, Hackney, Hogg and Szczepura [34] proposed a method to combine image quality and radiation dose data where the image quality scores are divided by radiation dose to give a figure of merit. This figure of merit would signify an *optimisation score* (OS) where a high score indicates better image quality at lower dose whereas a low score indicates poorer

image quality at higher radiation dose. This method (Image Quality/Effective dose) has been developed from studies that have used similar calculations but using SNR rather than visual image quality scores [22].

Statistical Analysis

All data were inputted into Excel 2007 (Microsoft Corp, Washington, USA) in order to facilitate descriptive analysis and then transferred to SPSS software package (PASW Statistics 18: version 18.0.2, SPSS Inc., Chicago, IL) for the inferential analyses. For the visual image quality data, intra- and inter-observer variability was evaluated using Intra-Class Correlation Coefficient (ICC) where >0.75 was considered excellent, 0.40-0.75 as fair to good and <0.40 as poor [35].

Image quality and radiation dose data were interpreted using various groupings (e.g. two different mattresses, two different platform positions) and subsequently analysed using an independent t-test with a probability level of $p<0.05$ (95%) regarded as significant. ESD and DAP values were consistently the same when undertaking repeat exposures (x3). Pearson's *r* and scatter plots were used to measure the linear relationship/correlation between visual image quality, CNR and radiation dose. These parametric tests were chosen as all statistical assumptions were met. The Shaprio-Wilk test in SPSS proved that all collected data were normally distributed [36].

Results

Image quality

The ICC value for all five observers was 0.8419 (95% confidence interval 0.8137-0.884) implying a high level of agreement [35]. ICC was also calculated for the last image quality criterion (item 16) in which the five observers had to decide whether the images were diagnostic or not (yes/no). The ICC for this criterion was 0.49 (95% confidence interval 0.22-0.69) which indicated fair to good agreement amongst observers.

From the experimental images, only three (6%) had a mean image quality score equal to or greater than the standard x-ray tabletop acquisition (reference image) (Figure 2 and Table 2 for image coding). Interestingly, for all the experimental images, these three images had the highest level of magnification with an increase of 10.78mm (18%) in femoral head diameter

compared to the reference image (see Table 3 for magnification results).Visual image quality was found to be significantly better when the image receptor platform was lowered ($p<0.02$); no statistically significant difference was found between image quality and the two different mattresses ($p=0.06$)

Of the 48 experimental images, only two were deemed unacceptable by more than half of the observers; these two images were acquired using 16mAs in conjunction with a 130cm SID and an elevated platform.

Image receptor platform position and mattress thickness had a statistically significant impact on femoral head diameter and hence magnification factor of the images ($p<0.01$). As expected, when the platform was lowered, magnification increased by 7% and when the Bi-Flex mattress was used in comparison to the standard mattress, magnification increased by 8%.

No statistically significant difference in CNR ($p>0.05$) was identified between platform position with elevated platform CNR being 7.88(SD =0.42) and lowered CNR being 7.80(SD =0.29). In addition, no statistically significant difference in CNR ($p>0.05$) was identified between the two different mattresses with standard mattress having a CNR of 7.82 (SD = 0.39) and Bi-Flex mattress CNR being 7.87 (SD=0.33).

Radiation dose

Forty four of the experimental images (92%) had higher effective dose to that of the reference image with ESD higher for thirty nine of the images (77%). The average ESD and effective dose for the standard mattress at 110cm SID was 1.91mGy and 0.19mSv respectively whereas the average ESD and effective dose for the Bi-Flex mattress at 110cm SID was 2.28 mGy and 0.23mSv respectively. This demonstrated a decrease in ESD and effective dose by 37% and 4% when utilising the standard mattress. However, no statistically significant difference was found between effective dose and ESD for the two different mattresses ($p>0.05$).

When the platform was elevated, the average ESD and effective dose were 1.91mGy and 0.20mSv respectively at a 110cm SID. With the platform lowered, the average ESD and effective dose were 2.3 mGy and 0.22mSv respectively. This demonstrates an increase in

both ESD and effective dose when the platform was lowered. Yet again, no statistically significant difference was found between effective dose and ESD for platform position ($p>0.05$).

A Pearson's *r* correlation identified a low positive relationship between the average visual image quality scores and CNR values (0.35). CNR and effective dose had a moderate positive relationship (0.53), whereas visual image quality and effective dose had a high positive relationship (0.72) [37].

Figure 3 highlights the optimisation scores for the experimental images in comparison to the reference image. The optimisation score for the reference image was 500; none of the experimental images achieved this score with a significant difference observed between the experimental images and the reference image ($p<0.002$). The experimental image with the highest optimisation score was one of the two images deemed non diagnostic by the observers. The subsequent images which had high optimisation scores were those achieved at a 130cm SID and 20mAs. No statistically significant difference was found for optimisation scores between platform position ($p=0.60$) and both mattresses ($p=0.18$)

As demonstrated in Table 4, when comparing the reference image to the experimental images acquired using the same acquisition parameters (16mAs and 110cm SID), image quality for both visual image quality scores and CNR decreased by 13% and 3% respectively; however only the visual image quality score results (13%) had a statistically significant decrease ($p<0.01$), (CNR; $p=.012$). In addition, effective dose, on average, more than doubled (56% average increase) for trolley imaging in comparison to x-ray tabletop using the same acquisition parameters, again demonstrating a significant difference in patient dose ($p<0.01$).

Discussion

The results demonstrate that the acquisition parameters used for the x-ray tabletop need to be adapted when applying to trolley imaging. Radiation dose can significantly increase whereas visual image quality can significantly decrease for trolley imaging when using standard x-ray tabletop aquisition parameters. . As collimation was adjusted to the area of interest for each image acquisition, radiation dose would be influenced by the increased OID at a maintained SID due to beam divergence. This means a larger OID would require collimation to be opened to ensure coverage of the anatomy of interest. The images acquired with a 110cm

SID *and* 16mAs were considered to be non diagnostic by the observers. Nevertheless, the reliability and validity of the sixteenth item (yes/no) is brought into question. For this specific item, the observers had to decide on whether the diagnostic quality of the image was adequate without knowing the clinical indication. This is important because the clinical indication may have influenced observer decision as to the quality of the image because some clinical indications require greater anatomical detail[6]. This may be the reason behind the lower ICC value for the last item when compared to the remaining validated items.

No significant difference was found for visual image quality or effective dose when comparing the standard and Bi-Flex mattresses. On this basis, the Bi-Flex mattress should therefore be considered gold standard when purchasing this specific Lifeguard 50 trolley as it offers more benefits to patients since it is designed to enhance comfort and reduce pressure ulcer incidence. Pressure ulcers remain a major problem in health care and one of the most costly and physically debilitating medical complications in twentieth century care [38- 39]. The only impact that the mattress had on image quality was with regards to magnification. On average, magnification increased by 8% when utilising the Bi-Flex mattress compared with the standard mattress. Magnification may however be an issue that needs attention when imaging AP pelvis because the images might potentially be used for planning orthopaedic surgery without the use of a calibration device. To overcome this problem, specific guidelines need to be established when imaging trolley patients (e.g. maintain constant SID and platform position) in order to minimise variations between different patients and obtain consistent measurements in an individual over time. Otherwise the use of a calibration ball for all AP pelvis projection could be a tool to consider in overcoming this magnification variation. It is accepted that, in some centres, a request for a traditional tabletop examination may follow if the pelvic image is required for detailed surgical planning. This may generate justification issues and therefore if trolley technique can be further standardised this situation may be avoidable.

Three images which had equal or higher visual image quality scores than the reference image were all acquired using the Bi-Flex mattress, platform lowered and an SID of a 110cm. These conditions resulted in the largest image magnification factor with a femoral head diameter of 25cm. This raises the question of whether magnification influenced the visual image quality scores, as the criteria were based upon how well structures are visualised. Manning, Ethell and Donovan [40] suggests that visual image quality is influenced by more than just the sharpness of anatomical outlines and the image noise, the size and complexity of structures

can impact upon observer interpretation too. The principles behind visual acuity and the use of the Snellen chart strengthens this argument that visual perception in radiology may be influenced by the size of the objects observed hence displayed magnification [41-43]. The visibility of an object is proportional to its area with contrast, noise, object size and shape all affecting our ability to extract visual information from an image [28]. The fact that there was no statistical difference identified between CNR and the two variables discussed (mattresses and platform position) also suggests that observer assessment may be influenced by something other than contrast and noise. This was why the resultant air gap from these three images was also disregarded as the potential reason for the increase in visual image quality as noise results from scatter however CNR did not detect this improvement. In addition, a grid was used for all images and the use of the air gap in conjunction with a grid has never been previously explored. The purpose of an air gap is to replace a grid as a method of scatter rejection and therefore it could be assumed that both air gap and grid combined would absorb useful image producing photons.

Lastly, if the optimisation scores are considered for this current study, the optimum acquisition parameters for imaging the AP pelvis on a trolley were 20mAs, 130cm SID, standard mattress and platform lowered. These parameters resulted in an image with the highest optimisation score and also no observers deemed this image to be non diagnostic. See Table 5 for recommended acquisition parameters for trolley imaging based on this study.

Limitations

There are further factors that must be explored before implementing these changes into clinical practice which includes the consideration of the following study limitations. More variables need to be explored such as different grids since only one oscillating and one stationary grid was used. This work was also limited to one type of axial examination, the AP pelvis projection. It would be beneficial for further research to be conducted on other body parts that are imaged on the trolley using the image receptor holder in order to reveal its effects on image quality and radiation dose. In addition, this study used one commercially available trolley to perform the experiment. However there are multiple trolley manufacturers with different trolley designs available suitable for imaging which need to be explored. A single anthropomorphic phantom was used which had no size or pathological variation therefore these findings need to be confirmed using patients in clinical practice. Lastly, this study was conducted using one CR system and therefore it would be advisable to validate the

results on different CR and DDR systems especially when considering the different systems available and the technological advancements over the past 20 years.

Conclusion

The results of this study demonstrate that the acquisition parameters used for AP pelvis x-ray tabletop imaging are not directly transferable to trolley imaging. Consideration should be given to the difference between these two situations, especially the increased OID which would benefit from an increase in SID to a 130cm in order to reduce both magnification and radiation dose. Radiation dose significantly increased for trolley imaging whilst visual image quality decreased. It is therefore important that separate exposure charts or diagnostic reference levels (DRL) are set for trolley imaging to ensure optimal image quality at the lowest possible dose. Lastly, the clinical indication for the AP pelvis on a trolley should be considered when selecting appropriate acquisition parameters because certain exposure factors may be sufficient depending on the objective of the examination.

References

1. Lee C, Porter K. The prehospital managment of pelvic fractures. *Emergency Medicine Journal* 2007; **24**(2): 130-133

2. Hart D, Wall BF, Hillier MC, Shrimpton PC. In: Agency HP, editor. Frequency and collective dose for medical and dental X-ray examinations in the UK, 2008. Oxford, United Kingdom: Health Protection Agency; 2010.

3. Wall BF, Haylock R, Jansen JTM, Hiller MC, Hart D, Shrimpton PC. Radiation risks from medical X-ray examinations as a function of the age and sex of the patient. Oxford, United Kingdom: Health Protection Agency; 2011.

4. Hiles P, Mackenzie A, Scally A, Wall B. Recommended standards for the routine performance testing of diagnostic X-ray imaging systems: Institute of Physics and Engineering in Medicine (IPEM); Report No. 91. York; 2005.

5. The Phantom Laboratory. Sectional lower Torso SK250 [Internet] [cited 1BC Aug 27]. Available from: http://www.phantomlab.com/library/pdf/sectional_SK250DS.pdf; 2013

6. Chan CTP, Fung KKL. Dose optimisation in pelvic radiography by air gap method on CR and DR systems; a phantom study. *Radiography* 2015; **21**(3): 214-223.

7. Carver E, Carver B. *Medical Imaging: Techniques, Reflection and Evaluation.* 2nd ed. Philadelphia: Churchill Livingstone.

8. England A, Evans P, Harding L, Taylor E, Charnock P, Williams G. Increasing source-to-image distance to reduce radiation dose from digital radiography pelvic examinations. *Rad Tech* 2015; **86**(3): 246-56.

9. Harding L, Manning-Stanley A, Evans P, Taylor M, Charnock P, England A. Optimum patient orientation for pevic and hip radiography: a randomised trial. *Radiography*. 2014; **20**: 22-32.

10. Manning-Stanley A, Ward A, England A. Options for radiation dose optimisation in pelvic digital radiography: a phantom study. *Radiography* 2012; **18**: 256-263.

11. Lança L, Franco L, Ahmed A, Harderwijk M, Marti C, Nasir S, et al. 10kVp rule - An anthropomorphic pelvis phantom imaging study using CR system: Impact on image quality and effective dose using AEC and manual mode. *Radiography* 2014; **20**(4): 333-338.

12. Tugwell J. Here comes a trolley: Imaging the trolley bound patient - current working

practices and experience. *Imaging and Therapy Practice* 2014, September.

13. Sandborg M, Dance D, Carlsson G, Persliden J. Selection of anti-scatter grids for different imaging tasks: the advantage of low atomic number cover and interspace materials. *Br J Radiol* 1993; **66**(792): 1151-63.

14. Heath R, England A, Ward A, Charnock P, Ward M, Evans P, et al. Digital pelvic radiography: increasing distance to reduce dose. *Radiologic Technology* 2011; **83**(1): 20-28

15. Woods J, Messer S. Focusing on dose. *Imaging & Therapy Practice* 2009, September:16-20.

16. Tugwell J, Everton C, Kingma A, Oomkens D, Pereira G, Pimentinha D, et al. Increasing source to image distance for AP pelvis imaging - impact on radiation dose and image quality. *Radiography* 2014; **20**(4): 351-355.

17. Protection ICoR. The 2007 recommendations of the ICRP on radiological protection, publication 103. Annals of ICRP. ; 37((2-4)). ICoR Protection. The 2007 recommendations of the ICRP on radiological protection, publication 103. *Annals of ICRP 2007*; **37**(2-4).

18. Hogg P, Blindell P. Software for image quality evaluation using a forced choice method. In Manchester: United Kingdom Radiological Conference; 2012, p.139

19. Almen A, Tingberg A, Mattsson S, Besjakov J, Kheddache S, Lanhede B, et al. The influence of different technique factors on image quality of lumbar spine radiographs as evaluated by established CEC image criteria. *Br J Radiol* 2000; **73**(875): 1192-9.

20. The Royal College of Radiologists. *Picture archiving and communications system (PACS) and guidelines on diagnostic display devices.* London: RCR; 2014.

21. The Royal College of Radiologists. Quality assurance in radiology reporting: peer feedback. London: RCR; 2014.

22. Samei E, Dobbins J, Lo J, Tornai M. A framework for optimising the radiographic technique in digital x-ray imaging. *Radiat Prot Dosimetry* 2005; **114**(1-3): 220-229.

23. Allen E, Hogg P, Ma WK, Szczepura K. Fact or fiction: An analysis of the 10 kVp 'rule' in computed radiography. *Radiography* 2013; 19: 223-227.

24. Mraity H, England A, Cassidy S, Eachus P, Dominguez A, Hogg P. Development and validation of a visual grading scale for assessing image quality of AP pelvis radiographic images. *Br J Radiol* 2016; **89**(1061). doi: 10.1259/bjr.20150430

25. Hess R, Neitzel U. Optimizing image quality and dose for digital radiography of distal pediatric extremities using the contrast-to-noise ratio. *Rofo* 2012; **184**(7): 643-9.

26. Mori M, Imai K, Ikeda M, Iida Y, Ito F, Yoneda K, et al. Method of measuring contrast-to-noise ratio (CNR) in nonuniform image area in digital radiography. *Electronics and Communications in Japan* 2013; **96**(7): 32-41. DOI. 10.1002/ecj.11416

27. Martin C. Optimisation in general radiography. *Biomed Imaging Interv J* 2007; **3**(2). doi: 10.2349/biij.3.2.e18

28. Vladimirov A. Comparison of image quality test methods in computed radiography. MSc Thesis. Univeristy of Tratu, Estonia. Available from: http://dspace.ut.ee/bitstream/handle/10062/15191/Vladimirov_Anatoli.pdf?sequence=1 [last accessed 18.05.16].

29. Bloomfield C, Boavida F, Chabloz D, Crausaz E, Huizinga E, Hustveit H, et al. Experimental article - Reducing effective dose to a paediatric phantom by using different combinations of kVp, mAs and additional filtration whilst maintaining image quality. *In Hogg P, Lanca L, editors. Erasmus Intensive Programme OPTIMAX;* 2014; Lisbon, Portugal.

30. Desai N, Singh A, Valentino D. Practical Evaluation of Image Quality in Computed Radiographic (CR) Imaging Systems. In Proceedings of SPIE, Medical Imaging: Physics of Medical Imaging; San Diego: The International Society of Optical Engineering; 2010. doi:10.1117/12.844640

31. Jang K, Kweon D, Lee J, Choi J, Goo E, Dong K, et al. Measurment of Image Quality in CT Images Reconstructed with Different Kernels. *Journal of the Korean Physical Society* 2011; **58**(2): 334-342.

32. Sun Z, Lin C, Tyan Y, Ng K. Optimization of chest radiographic imaging parameters: a comparison of image quality and entrance skin dose for digital chest radiography systems. *Clin Imaging* 2012; **36**(4): 279-86.

33. Ma W , Hogg P, Norton S. Effects of kilovoltage, milliampere seconds, and focal spot size on image quality. *Rad Tech* 2014; **85**(5): 479-485.

34. Williams S, Hackney L, Hogg P, Szczepura K. Breast tissue bulge and lesion visability during stereotactic biopsy - A phantom study. *Radiography* 2014; 20: 271-276.

35. Rosner B. Fundementals of biostatistics. 7th ed. Boston: Cengage Learning; 2010.

36. Ghasemi A, Zahediasl S. Normality tests for statistical analysis: a guide for non-

statisticians. *Int J Endocrinol Metab* 2012; **10**(2): 486-9.

37. Hinkle D, Wiersma W, Jurs S. *Applied Statistics for the Behavioural Science*. 5th ed. Boston: Houghton Mifflin; 2003.

38. Agrawal K, Chauhan N. Pressure ulcers:back to the basics. *Indian J Plast Surg* 2012; **45**: 244-54.

39. Angmorterh SK. An investigation into interface pressure (IP) risk of healthy volunteers on modern medical imaging and radiotherapy tables. Degree of Doctor of Philosophy (PhD). Univeristy of Salford, Manchester, United Kingdom; 2016.

40. Manning D, Ethell S, Donovan T. Detection or decision error? Missed lung cancer from posteroanterior chest radiographs. *Br J Radiol* 2004; **77**(915): 231-235.

41. Alexander K. Reducing error in radiographic interpretation. *Can Vet J* 2010; **51**(5): 533-536.

42. Colenbrander A. *Measuring vision and vision loss*. In Tasman W, Jaeger E. Duane's Ophthalmology. Philadelphia: Lippincott Williams&Wilkins; 2013.

43. Marchiori D. *Clinical Imaging: With Skeletal, Chest, & Abdominal Pattern Differential.* 2nd ed. St Louis: Elsevier Health Sciences; 2004.