Radiologists are responsible for the manner in which radiologic examinations, including radiographs, mammograms, and CT images, are obtained. Radiographic protocol parameters should be selected to ensure adequate diagnostic performance, and radiologists therefore need to understand the image creation process, in addition to interpreting radiologic examinations [1–4]. The choice of protocol parameter also affects the amount of radiation received by the patient, and radiologists need to ensure that patients are not being subjected to unnecessary radiation exposure.

The most important aspect of the choice of protocol parameters is to ensure that the image quality will be sufficient for a given diagnostic imaging task [5]. It is therefore essential for radiologists to understand how choices for each protocol parameter will affect the resultant image being generated. Key x-ray choices include the voltage across the x-ray tube (kilovoltage), the size of the x-ray tube current (milliamperes), and the imaging exposure time (seconds) [6, 7]. Although artifacts in radiographic imaging are of obvious importance for image quality, these are beyond the scope of this article [8].

**X-Ray Tube Output**

**Quantity**

Important determinants of x-ray beam quantity are the choice of tube current (milliamperes) and the corresponding x-ray beam exposure times (seconds) [9–11]. The product of the tube current and exposure time, known as the milliampere-second value, is the primary indicator of the x-ray beam intensity. The milliampere-second value indicates the relative radiation output for a given x-ray tube when operated at a specified tube voltage, but does not account for differences between x-ray systems using a variety of tubes and filtration. For this reason, it is generally not very helpful or informative to describe any radiographic examination as being performed at a given milliampere-second value.

The intensity of an x-ray beam is quantified by an air kerma, which relates to the kinetic energy released per unit mass when x-rays interact with air [12]. Air kerma is the energy transferred to electrons when normalized by the mass of air (energy/mass) and is measured using grays or milligrays. Air kerma can be thought of as the number of x-rays per unit area, with the photon energies of minimal concern. The air kerma in any x-ray beam is directly proportional to both the tube current and the corresponding exposure time. Because air kerma is well defined and universally understood, it is the metric of choice for specifying x-ray beam intensity. The higher the air kerma, the higher the x-ray beam’s intensity, and the more photons will be incident on the patient.

When air kerma is multiplied by the corresponding beam area, one obtains the kerma-area product in grays times square centimeters (Gy cm²), which is often referred to as the dose-area product [12]. The kerma-area product is the total amount of radiation incident on the patient and will affect the energy deposited into the patient. Air kerma can be thought of as the number of x-rays per unit area, with the photon energies of minimal concern. The air kerma in any x-ray beam is directly proportional to both the tube current and the corresponding exposure time. Because air kerma is well defined and universally understood, it is the metric of choice for specifying x-ray beam intensity. The higher the air kerma, the higher the x-ray beam’s intensity, and the more photons will be incident on the patient.

When air kerma is multiplied by the corresponding beam area, one obtains the kerma-area product in grays times square centimeters (Gy cm²), which is often referred to as the dose-area product [12]. The kerma-area product is the total amount of radiation incident on the patient and will affect the energy deposited into the patient. Air kerma is used to estimate the entrance skin dose and thereby the likelihood of a deterministic (skin) radiation risk. Kerma-area product quantifies the total amount of radiation incident on the patient and is most closely related to the total stochastic patient risk. For most patients, the stochastic risk can be taken as the carcinogenic risk. Table 1 shows representative values of kerma-area product for complete examinations in radiologic imaging [13].

**Quality**

The quality of an x-ray beam refers to the penetrating power of the beam and is normally expressed as a half-value layer (HVL)
of aluminum (in millimeters). When a half-value layer thickness of aluminum is placed into the x-ray beam, it reduces the intensity (air kerma) by 50%. A typical x-ray beam used in abdominal radiography would likely have an HVL of 3 mm of aluminum. Figure 1 shows that 0.3 mm of copper (atomic number, 29), 3 mm of aluminum (atomic number, 13), or 30 mm of soft tissue (atomic number, 7.5) all attenuate half the x-ray beam, illustrating the paramount importance of the atomic number of the attenuator [14]. When the average x-ray beam energy increases, meaning the x-ray beam has increased penetrating power, the HVL also increases.

In radiography, the two important determinants of x-ray beam quality are tube voltage and beam filtration. Increasing the tube voltage or the total amount of filtration or both will increase the average photon energy and thereby the x-ray beam penetrating power. It is important to note that adding filters always reduces the x-ray tube output (quantity) but increases x-ray beam penetrating power (quality). The most common filter material used in radiography is aluminum, with a typical filtration in routine radiography being about 3 mm of aluminum. Copper is also used in modalities for which dose issues are important, specifically in pediatrics and interventional radiology. In mammography, the most common filter materials are molybdenum, rhodium, and silver. CT makes use of relatively heavy filtration, with aluminum, copper, and titanium being the most frequently used materials.

A typical radiographic x-ray beam in abdominal radiography will have an x-ray tube voltage of 80 kV and a total filtration of about 3 mm of aluminum. The x-ray beam quality is likely to also be 3 mm of aluminum, which is easily confused with the filtration because they are often quantitatively very similar.

Beam qualities in mammography are very low and of the order of 0.4 mm of aluminum. A typical CT x-ray tube voltage is 120 kV, and the filtration can consist of 2 mm of aluminum with an additional 0.1 of copper or 1 mm of titanium. The resultant HVLs will be of the order of 6 mm of aluminum or even higher, markedly greater than those encountered in radiography.

**Technique Selection**

When the amount of radiation needed to generate an image is to be increased, this will generally be achieved by increasing the milliampere-second value, as shown in Figure 2, but will also increase the beam energy (quality/HVL). In radiography, the x-ray tube output (air kerma) is generally taken to be proportional to kilovoltage squared, whereas in CT the x-ray tube output is taken to be proportional to kilovoltage to the 2.6th power [15]. However, the choice of x-ray tube voltage is generally determined by considering patient penetration, image contrast, and dynamic range requirements. Only very rarely, if ever, is the kilovoltage adjusted simply to change the amount of radiation in the x-ray examination. When kilovoltage is modified, it is always necessary to evaluate other factors, including patient penetration, scattered radiation intensities, x-ray transmission through a grid, and changes in photon absorption efficiencies in the x-ray detectors.

**Protocol Design**

Protocol design requires the initial selection of an appropriate x-ray tube voltage and a subsequent adjustment of the x-ray tube milliampere-second to ensure that the radiation intensity at the image receptor is the one that is desired for image quality. Figure 3 (left panel) shows a spine that is invisible (too white on the image), requiring the tube voltage to be increased to improve beam penetration (Fig. 3, middle panel). The increase in kilovoltage achieves the correct amount of penetration, but now the image is too dark (Fig. 3, middle panel), so the milliampere-second now needs to be reduced to ensure the appropriate amount of radiation.

### TABLE I: Representative Values of Kerma-Area Product in Radiologic Imaging

<table>
<thead>
<tr>
<th>Type of Radiologic Examination</th>
<th>Kerma-Area Product (Gy · cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Radiography</td>
<td>1</td>
</tr>
<tr>
<td>Fluoroscopy (gastrointestinal and urologic)</td>
<td>20</td>
</tr>
<tr>
<td>Interventional radiology</td>
<td>100–300</td>
</tr>
</tbody>
</table>

---

Fig. 1—Material thickness (atomic number, Z) needed to attenuate x-ray beam (80 kV + 3 mm aluminum filtration) by 50%. Notice dramatic reduction in material thickness as atomic number increases. (Illustrations by Abrahams RB)

Fig. 2—Increasing milliampere-second (mAs) value proportionally increases number of x-rays produced with no change in average photon energy (left). When tube voltage (kV) increases, more photons are produced, and average photon energy also increases (right). Increasing milliampere-second value only increases quantity, whereas increasing kilovoltage increases quantity and quality. (Illustrations by Abrahams RB)
(air kerma) at the image receptor to achieve a satisfactory image density (Fig. 3, right panel). Because tube voltage affects output and penetration, as depicted in Figure 2, the x-ray tube current in any x-ray examination the x-ray intensity (milliampere-second) must always be adjusted when kilovoltage has been modified to ensure the appropriate amount of radiation at the image receptor.

### Image Quality

#### Contrast

Consider a lesion that appears in any radiologic image, where the lesion intensity is given by \( I_{\text{lesion}} \). If the intensity of the surrounding normal tissues is \( I_{\text{tissue}} \), then the lesion contrast is \( I_{\text{lesion}} - I_{\text{tissue}} \). Whether the contrast is positive or negative is determined by the lesion’s characteristics and is not important from an image quality perspective. For example, adding iodine or carbon dioxide to a blood vessel will make the vasculature visible but of differing polarity because iodine absorbs more x-rays, whereas carbon dioxide will absorb fewer x-rays than adjacent tissues. Contrast is independent of the amount of radiation used to generate an image, as shown in Figure 4. If a lesion transmits 5% more photons than the surrounding normal tissues, this is always true regardless of whether the x-ray tube setting (and corresponding air kerma output) was 1, 10, or 100 mAs. Images shown in Figure 4 illustrate the only effect of reducing the milliampere-second, is to increase the amount of image mottle.

The most important factor affecting contrast is the average photon energy used to generate the image, which is determined by the choice of x-ray tube voltage and the amount of x-ray beam filtration. As the photon energy decreases, the differential attenuation between the lesion and surrounding tissues increases. Accordingly, low energy is usually associated with higher contrast and vice versa, as shown in Figure 5. Although this is generally true, it is important to note that the improvement in lesion contrast at low energies also depends on the lesion’s atomic number. As the lesion’s atomic number diverges from that of soft tissue (atomic number, \( \approx 7.5 \)), the contrast is affected much more. Reducing photon energy improves the contrast (i.e., visibility) of calcifications and tissues containing iodine much more than it improves the contrast of soft-tissue lesions.

Two additional factors need to be considered regarding contrast relating to scatter and digital image display. Increased scatter will generally reduce contrast, which can be appreciated by looking at an abdominal radiograph obtained without the use of a scatter removal grid. Image manipulation in digital radiography can be used to modify both the brightness and contrast of any given feature. When the window width is increased, display contrast is reduced, and vice versa. Accordingly, it is possible to increase display contrast by use of narrow window widths. However, when narrow window widths are used, the range of pixel values (tissues) that are visible is also reduced. Therefore, there is a trade-off between display contrast and type of anatomy that is visible in the displayed image. A good example is the use of lung and mediastinum display settings while reading a chest CT.

#### Noise

Consider a digital x-ray detector that is exposed to a uniform x-ray beam, with the aver-
age number of photons detected in a pixel being 100. The statistical nature of x-ray emission (and detection) means that not every pixel will detect exactly 100 photons. Some pixels have more x-rays and appear darker, whereas other pixels have fewer than 100 photons and will appear lighter. The distribution of these darker and lighter pixels is random, and the image will have a mottled appearance (salt-and-pepper distribution). Imaging scientists called this grainy appearance “noise,” which is synonymous with “mottle.” Figure 6 shows examples of chest x-ray images obtained at low and high x-ray tube output values, illustrating how image noise (mottle) decreases as the x-ray tube output (milliamperere-second) increases.

Because noise is random, the amount of noise does not increase in a linear manner when the amount of radiation used to generate an image increases. If the average number of photons per pixel is 100, the SD of this average value is 10, which is normally expressed as a relative SD (fluctuation) of 10%. When the number of photons increases by a factor of 4, the relative SD will be halved (i.e., $4^{1/2}$) to only 5%. It is thus useful to always remember that quadrupling of the number of photons used to generate any radiographic image will halve the amount of image noise, and vice versa.

Noise in virtually all x-ray imaging modalities is dominated by quantum mottle, where the latter relates to the total number of x-rays used to generate an image. Quantum mottle is dominant in all radiography, mammography, fluoroscopy, and CT examinations. This is important because the only technical way to reduce noise using any x-ray imaging modality is to use more x-rays. X-ray detection efficiencies in most x-ray imaging systems are relatively high, so improvements are not likely. In CT for example, current detectors generally absorb well over 90% of all incident photons. A corollary of the fact that virtually all of medical imaging is quantum mottle limited is that replacing components in the imaging chain, such as charge coupled device (CCD) for television camera in fluoroscopy, cannot result in dose reductions if image quality is to be maintained.

Contrast-to-Noise Ratio

When the contrast alone increases or noise alone is reduced, a lesion will clearly become more visible. However, the visibility of a lesion is not determined by just the noise or the lesion contrast alone. The visibility of any lesion in a radiologic image needs to take into account both the amount of lesion contrast and the corresponding amount of noise.

The amount of lesion contrast relative to the amount of noise (mottle) is the key determinant of the visibility of a given lesion. The ratio of lesion contrast to image mottle is known as the contrast-to-noise ratio (CNR). This ratio is an indicator of the relative image quality of this lesion (i.e., visibility), as depicted by the images in Figure 7. Improving the visibility of any lesion will always require the lesion’s CNR to be increased. This can be achieved by increasing lesion contrast or reducing the amount of noise or by a combination of these two methods.

In digital subtraction angiography (DSA) imaging, one way to improve the contrast of an aneurysm (and thereby CNR) is to replace venous administration of iodinated contrast material with arterial administration of iodine. Similar results could be obtained by using more x-ray photons when creating an image, which would reduce the amount of image noise (mottle). From a lesion visibility perspective, it is only the CNR that has to be increased, and increasing contrast is equivalent to reducing noise. The optimal strategy, however, will generally depend on external factors, such as whether an x-ray system can offer more x-rays or whether the patient can tolerate such an increase in dose.

Techniques and Image Quality

Output

For a single radiographic image, it is the choices of tube current (milliamperes) and exposure time (seconds) that determines the
total x-ray beam intensity (i.e., milliampere-second). The maximum tube current for most radiographic units, when operated using a large focal spot, will be about 1000 mA, and the power loading will likely be about 100 kW. Small focal spot sizes may be used (e.g., to reduce blur in radiographs of the cervical spine and extremities). In these cases, the tube current and focal spot loadings are typically reduced fourfold because the focal spot area is now only a quarter of the size. In practice, an imaging system will select the largest possible tube current that can be tolerated by tube heating requirements because this will minimize the radiographic exposure time.

For a given radiographic examination, the selected milliampere-second value will directly affect the amount of radiation transmitted through the patient and used to produce the resultant radiograph. For a chest x-ray, the required air kerma at the image receptor is currently about 3 μGy. If the amount of radiation at the image receptor were to be quadrupled (milliampere-second value), then the image noise (mottle) would be halved (Fig. 5). Conversely, using a quarter of the radiation (milliampere-second value) would double the amount of image noise (mottle).

Patient dose is directly proportional to the milliampere-second value, and the radiologist needs to select this parameter with great care. When the choice of milliampere-second value is too low, diagnostic performance could be compromised. Conversely, selecting a milliampere-second value that is too high results in patients being unnecessarily exposed to radiation and thus subjected to unnecessary radiation risks.

Voltage

When selecting the x-ray tube voltage in any radiographic examination, the first consideration is the issue of patient penetration (Fig. 3). Low x-ray tube voltages mean low photon energies that may not penetrate the patient and would be of little value in radiographic imaging. X-ray penetration and the appropriate x-ray tube voltage are also related to patient size. Data in Table 2 show how x-ray beam penetration varies with average photon energy for adult patients, and Table 3 shows similar data for pediatric patients. In general, reduced voltages can be used for small patients, whereas for larger patients, it is essential to increase patient penetration by increasing the tube voltage.

Once adequate penetration has been achieved, it may nonetheless be advantageous to use even higher photon energies (i.e., increased kilovoltage). Although image contrast is generally reduced at high photon energies (as already discussed), so is the dynamic range of the resultant image data. The dynamic range is obtained by focusing on the air kerma values in the patient anatomic region and comparing the highest receptor air kerma to the lowest receptor air kerma. For examinations such as chest radiographs, where the intrinsic contrast is generally satisfactory, it is essential to reduce the image dynamic range. On radiographs, low voltages that produce very large dynamic ranges will result in the higher exposures in the lung appearing all black and the lower exposures in the mediastinum appearing all white. High voltages that reduce the im-

---

**TABLE 2: Penetration of Monoenergetic Photons Through Varying Patient Thicknesses (Water or Soft Tissue) Simulating Adult Patients**

<table>
<thead>
<tr>
<th>Patient Tissue Thickness (cm)</th>
<th>Patient Penetration (%)</th>
<th>40 keV</th>
<th>50 keV</th>
<th>60 keV</th>
</tr>
</thead>
<tbody>
<tr>
<td>18</td>
<td>0.82</td>
<td>1.7</td>
<td>2.5</td>
<td></td>
</tr>
<tr>
<td>23</td>
<td>0.22</td>
<td>0.54</td>
<td>0.88</td>
<td></td>
</tr>
<tr>
<td>28</td>
<td>0.06</td>
<td>0.17</td>
<td>0.31</td>
<td></td>
</tr>
<tr>
<td>33</td>
<td>0.01</td>
<td>0.06</td>
<td>0.11</td>
<td></td>
</tr>
</tbody>
</table>

**TABLE 3: Penetration of Monoenergetic Photons Through Varying Patient Thicknesses (Water or Soft Tissue) Simulating Pediatric Patients Undergoing Imaging**

<table>
<thead>
<tr>
<th>Patient Tissue Thickness (cm)</th>
<th>Patient Penetration (%)</th>
<th>30 keV</th>
<th>40 keV</th>
</tr>
</thead>
<tbody>
<tr>
<td>8</td>
<td>5.2</td>
<td>12</td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>2.5</td>
<td>6.9</td>
<td></td>
</tr>
<tr>
<td>12</td>
<td>1.2</td>
<td>4.1</td>
<td></td>
</tr>
<tr>
<td>15</td>
<td>0.39</td>
<td>1.8</td>
<td></td>
</tr>
<tr>
<td>18</td>
<td>0.13</td>
<td>0.82</td>
<td></td>
</tr>
</tbody>
</table>
Techniques, Contrast, and Noise in Radiography

The two parameters that are easy to adjust in any radiographic imaging protocol are the x-ray tube voltage (kilovoltage) and the total radiation intensity (milliampere-second value) used to obtain the image. This is true in all x-ray-based imaging modalities, including radiography, mammography, fluoroscopy, and CT [18, 19]. It is the responsibility of the radiologist to adjust these key parameters, which affect image quality and the corresponding patient dose. The overall objective is to ensure that images are of diagnostic quality without exposing patients to unnecessary radiation.

When a patient is exposed to an x-ray beam, there is a well-defined and quantifiable patient dose. The resultant image, however, does not have a corresponding well-defined image quality. The reason for this is that image quality is always task dependent. When a radiograph is obtained to detect a subtle soft-tissue lesion, minimizing the amount of mottle will be very important because the intrinsic lesion contrast is relatively low. On the other hand, a radiograph to identify the location of a swallowed coin would need much less radiation because the intrinsic contrast of the coin is so high.

The amount of acceptable mottle in a diagnostic chest x-ray examination is likely to be low and requires an image receptor air kerma of about 5 μGy. A follow-up scoliosis examination, where the clinical question relates to any changes in spine curvature, could likely be obtained using 10 times less radiation [20]. Although the image mottle at such low doses would obviously be relatively high, the radiograph would nonetheless be adequate for answering the posed clinical question. The fact that image quality is always task dependent is perhaps the most important lesson that imaging scientists have to teach the radiologic imaging community.

Conclusion

Understanding the image creation process as it relates to the interplay of contrast, noise, patient dose, and diagnostic performance is vital in the practice of modern radiology. Tube voltage selection affects the amount of contrast in the resultant image, and the selected milliampere-second value affects the corresponding image mottle. The resultant CNR may be taken as a relative indicator of the visibility of a given lesion. When a lesion needs to be made more visible, this may be achieved by increasing the contrast, reducing the noise, or by a judicious combination of both aspects. The radiographic techniques that may be used to adjust a lesion’s CNR are thus the kilovoltage and milliampere-second value. As in all radiographic imaging, the optimal values depend on the physical characteristics of the patient and the specific diagnostic imaging task.

Contrast-To-Noise Ratio

The two parameters that are easy to adjust in any radiographic imaging protocol are the x-ray tube voltage (kilovoltage) and the total radiation intensity (milliampere-second value) used to obtain the image. This is true in all x-ray-based imaging modalities, including radiography, mammography, fluoroscopy, and CT [18, 19]. It is the responsibility of the radiologist to adjust these key parameters, which affect image quality and the corresponding patient dose. The overall objective is to ensure that images are of diagnostic quality without exposing patients to unnecessary radiation.

When a patient is exposed to an x-ray beam, there is a well-defined and quantifiable patient dose. The resultant image, however, does not have a corresponding well-defined image quality. The reason for this is that image quality is always task dependent. When a radiograph is obtained to detect a subtle soft-tissue lesion, minimizing the amount of mottle will be very important because the intrinsic lesion contrast is relatively low. On the other hand, a radiograph to identify the location of a swallowed coin would need much less radiation because the intrinsic contrast of the coin is so high.

The amount of acceptable mottle in a diagnostic chest x-ray examination is likely to be low and requires an image receptor air kerma of about 5 μGy. A follow-up scoliosis examination, where the clinical question relates to any changes in spine curvature, could likely be obtained using 10 times less radiation [20]. Although the image mottle at such low doses would obviously be relatively high, the radiograph would nonetheless be adequate for answering the posed clinical question. The fact that image quality is always task dependent is perhaps the most important lesson that imaging scientists have to teach the radiologic imaging community.

Conclusion

Understanding the image creation process as it relates to the interplay of contrast, noise, patient dose, and diagnostic performance is vital in the practice of modern radiology. Tube voltage selection affects the amount of contrast in the resultant image, and the selected milliampere-second value affects the corresponding image mottle. The resultant CNR may be taken as a relative indicator of the visibility of a given lesion. When a lesion needs to be made more visible, this may be achieved by increasing the contrast, reducing the noise, or by a judicious combination of both aspects. The radiographic techniques that may be used to adjust a lesion’s CNR are thus the kilovoltage and milliampere-second value. As in all radiographic imaging, the optimal values depend on the physical characteristics of the patient and the specific diagnostic imaging task.

References

15. Elojeimy S, Tipnis S, Huda W. Relationship between radiographic techniques (kilovolt and milliampere-second) and CTDI_{vol}. Radiat Prot Dosimetry 2010; 141:43–49