CT dose optimisation and reduction in osteoarticular disease

A. Gervaise, P. Teixeira, N. Villani, S. Lecocq, M. Louis, A. Blum

Abstract  With an improvement in the temporal and spatial resolution, computed tomography (CT) is indicated in the evaluation of a great many osteoarticular diseases. New exploration techniques such as the dynamic CT and CT bone perfusion also provide new indications. However, CT is still an irradiating imaging technique and dose optimisation and reduction remains primordial. In this paper, the authors first present the typical doses delivered during CT in osteoarticular disease. They then discuss the different ways to optimise and reduce these doses by distinguishing the behavioural factors from the technical factors. Among the latter, the optimisation of the milliamps and kilovoltage is indispensable and should be adapted to the type of exploration and the morphotype of each individual. These technical factors also benefit from recent technological evolutions with the distribution of iterative reconstructions. In this way, the dose may be divided by two and provide an image of equal quality. With these dose optimisation and reduction techniques, it is now possible, while maintaining an excellent quality of the image, to obtain low-dose or even very low-dose acquisitions with a dose sometimes similar that of a standard X-ray assessment. Nevertheless, although these technical factors provide a major reduction in the dose delivered, behavioural factors, such as compliance with the indications, remain fundamental. Finally, the authors describe how to optimise and reduce the dose with specific applications in musculoskeletal imaging such as the dynamic CT, CT bone perfusion and dual energy CT.

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Since its introduction in the 1970s, computer tomography (CT) has played a major role in the diagnosis of a great many osteoarticular diseases. It has quickly become the choice examination in the diagnosis of traumatic, degenerative or even malformative lesions. Even though image quality is altered by metallic artefacts, CT also found indications in postoperative imaging [1—3]. It is now also used in interventional imaging (injection guidance, bone and soft tissue biopsies, vertebroplasty, etc.) [4]. However, the performance of CT is limited by the inferior analysis of the soft tissue compared with Magnetic Resonance Imaging (MRI).

The CT analysis of intra-articular lesions is also very difficult due to the absence of administration of intra-articular contrast product. CT is also a technique of irradiating imaging. For all of these reasons, the MRI has taken a preponderant role in musculoskeletal imaging.

Nevertheless, the scanner plays an important role in osteoarticular disease with the development of multislice CT, the development of multidetector CT and recent technological evolutions that reduce the dose the patient is exposed to. Over the last years, the speed of acquisition and the temporal and spatial resolution of CT have also considerably improved. Sub-millimetric isotropic acquisitions are the rule and multidetector row and three-dimensional (3D) Volume Rendering (VR) reformations improve the evaluation of bone and soft tissue lesions [5]. The improvement in the speed of acquisition reduces movement artefacts and thereby makes the exploration of large volumes possible. This is, for example, especially adapted for the musculoskeletal assessment of multiple trauma patients (Fig. 1).

Other technological advances in osteoarticular imaging are represented by the development of the dual energy CT and the perfusion CT. The dual energy CT is based on double acquisition with two X-ray beams of different kilovoltage. This better characterises the tissue and also reduces metallic artefacts or even provides access to bone subtraction and iodine contrast product [6]. In addition, the tumour perfusion CT, with the acquisition of successive multiple phases, provides functional information to better analyse bone and soft tissue tumours. In addition the functional analysis is more reproducible than that of the MRI [7,8].

As opposed to the MRI, the other advantages of CT are represented by its lower cost, improved availability, the possibility of use on post-surgical or unstable patients and the absence of contra-indications related to prosthetic materials or pacemakers [4,9].

Finally, CT has benefited from a great many technological innovations over the last few years, thereby considerably reducing the dose delivered. The best example is the recent appearance of iterative reconstructions that reduce the dose by half with an equivalent image quality [10]. With these technological innovations and better control of the optimisation of the acquisition parameters, it is now possible to obtain CT imaging with a dose almost equal that of the standard X-ray assessment while the diagnostic performance of CT is much higher than that of X-rays. With the continued reduction in the dose delivered, the replacement of the X-ray assessment by CT seems to be possible in an increasing number of clinical situations.

After a review of the typical doses delivered during osteoarticular CT, we will in turn discuss the different methods to reduce the dose by emphasising both the behavioural factors and the technical factors.

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Figure 1. Whole-body computed tomography (CT) in a 55-year-old woman for an assessment after falling out of a window. Acquisition with 64-slice CT covering the whole-body at arterial time, that is, an acquisition of 163 cm in 31 s, with 120 kV, automatic modulation of the dose with mAs between 50 and 134, rotation time of 0.5 s, 64 × 0.5 mm, pitch at 0.868, for a DLP of 1428 mGycm. Reformation 3D volume rendering (VR) of the whole-body (a) then centred on the left femoral fracture (b) and sagittal reformation on the whole vertebral column (c). This acquisition is obtained in thin slices, providing reformations in 3D VR in order to obtain a global view of the complex fracture of the left femur and its relationship to the superficial femoral artery and help the surgeon with the presurgical assessment. The fractures of both patellas should be noted. The multidetector row reformations with bone filter help better analyse the whole vertebral column and reveal a fracture of the upper vertebral body T11 without recession of the vertebral body.
Typical doses delivered with osteoarticular computer tomography (CT)

The International Commission on Radiological Protection and the European Commission recommend the establishment of diagnostic reference levels (DRL) [11,12]. The European Commission recommended doses defined by the Weighted-Dose-CT-Index (CTDI(w)) and by the Dose-Length Product (DLP) for several types of CT [12]. For the lumbar vertebrae, the recommended reference levels are a CTDI(w) of 35 mGy and a DLP of 800 mGy cm. For the bony pelvis, (hips, sacroiliacs), the recommended reference levels are a CTDI(w) of 25 mGy and a DLP at 520 mGy cm. For the traumatic spine, the recommended values are a CTDI(w) at 70 mGy and a DLP at 460 mGy cm [12]. However, these doses are based on reports dating from the end of the 1980s and the beginning of the 1990s, before the introduction of spiral and multislice CT [13,14]. Since then, the multislice CT radically changed practices. In 2004, the European Commission published new recommendations taking multislice CT into account. However, they did not recommend new dosimetry references in terms of DLP in the osteoarticular realm [15]. In France, the DRL have recently been up-dated [16]. Among those involving adult computed tomography, only one osteoarticular examination is included and only for the lumbar vertebrae with a DLP of 700 mGy cm. These reference levels are partial and only involve very few explorations in osteoarticular imaging. This is all the more so since, with the improvement in the speed of acquisition of multislice CT, it is now possible to obtain whole spine imaging, leading to new indications such as the possibility of obtaining a whole-body CT in a myeloma assessment [17,18] or even the acquisition of a whole spinal column in osteoporosis [19]. In addition, there is no reference dose for acquisitions of the peripheral joints or for the new perfusion CT or dynamic CT applications.

In the literature, few publications have been devoted to CT doses in the realm of osteoarticular imaging and the results vary greatly. In a review of the literature dating 2008, Mettler et al. [20] find a mean effective dose of 6 mSv for the spine CT with values ranging from 1.5 to 10 mSv. In another review of the literature dating 2011 on 19 studies, Pantos et al. [21] find even greater differences in doses, ranging from 0.8 to 15.7 mSv for a lumbar CT, with a median dose of 5.2 mSv (Table 1). In a 2009 study on the analysis of osteoarticular CT doses in their institution, Biswas et al. [22] report of mean dose of 19.15 mSv for the acquisition of a lumbar CT (Table 2). The great variability in doses is mainly due to the difference in the length of acquisition between the studies. For example, based on a single-slice CT, Galanski et al. [23] found a mean dose of 2.7 mSv for a mean length of acquisition of only 5.8 cm. However, with a 16-slice CT, Biswas et al. [22] found a mean dose of 19.15 mSv but for a mean length of acquisition of 25.5 cm. Therefore, more than the difference in terms of number of CT slices, the dose difference is above all now related to their use. In fact, while the passage from the single-slice CT to the multislice CT was accompanied by an increase in the dose delivered to the patient [24], the passage from 4-slice CT to 16 or 64-slice CT is accompanied by technical improvements resulting in a relatively stable dose [25–27]. Therefore, more than the number of slices, the overall increase in the number of CT carried out [28] and the increase in the length of acquisition now result in an increase in the individual and collective exposure [20,29]. Within the same institution, significant variations in terms of ISDP and DLP are also observed (Tables 2–4) [30,31]. This may be accounted for by the adjustment of the acquisition parameters according to the patient morphotype and the indications. The acquisition parameters may be reduced in the exploration of the bone structures, while the milliamps increase when assessment of the soft tissue is required. The establishment of new techniques of dose reduction, such as iterative reconstructions, also influences the dose delivered during CT (Table 4).

Very few studies refer to the peripheral joints. As far as we are aware, Biswas et al. [22] reported the only study presenting a full analysis of all of the doses delivered in osteoarticular imaging, including the peripheral joints. These results show that the farther the anatomic zone is from the trunk, the more the effective dose is minimal or even negligible, as for example for the wrist (Table 2). This is because the peripheral joints are smaller, thereby allowing for a reduction in the acquisition parameters and providing shorter lengths of acquisition. However, this is mainly due to the fact that the tissue-weighting factor used in calculating the effective dose is very small in view of the absence of a radiosensitive organ nearby. Table 5 sums up the values of the tissue-weighting factors used by Biswas et al. [22] in the estimate of the effective dose as a function of the different anatomic locations of osteoarticular CT (the effective dose (E) in mSv is calculated from the Dose-Length Product (DLP) in mGy cm multiplied by a tissue-weighting factor (k) according to the formula: E = DLP × k).

<table>
<thead>
<tr>
<th>Type of CT</th>
<th>CTDI(w) (mGy)</th>
<th>DLP (mGy cm)</th>
<th>Effective dose (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cervical vertebrae</td>
<td>44.3 (5.3–103.2)</td>
<td>324 (56–1275)</td>
<td>2.6 (0.3–7.5)</td>
</tr>
<tr>
<td>Dorsal vertebrae</td>
<td>NA</td>
<td>253 (66–515)</td>
<td>4.6 (1.0–9.8)</td>
</tr>
</tbody>
</table>
| Lumbar vertebrae     | 30.3 (10.6–59.7) | 302 (49–870)
|                      |              |               | 5.2 (0.8–15.7) |

NA: not available; DLP: dose-length product.

The values are indicated as the median and the extreme values in brackets.

Note the difference in the dose of lumbar CT between the review of the literature by Pantos et al. [21] and the values provided by Biswas et al. [22] within their institution (Table 2). This difference is mainly due to an increase in the dose after the passage from the single-slice CT to the multislice CT (Pantos et al. mainly take studies on single-slice CT into account [21] while Biswas et al. use a 16-slice CT [22]) as well as the increase in the acquisition lengths that also accompanied the distribution of multislice CT.
Table 2 Doses of the peripheral joint and spinal computed tomography (CT) according to Biswas et al. [22] (gathered with a 16-slice CT).

<table>
<thead>
<tr>
<th>Joints</th>
<th>CTDI(w)a (mGy)</th>
<th>DLPa (mGy cm)</th>
<th>Effective dosea (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Wrist and hand</td>
<td>14.41 ± 15.52</td>
<td>137 ± 134</td>
<td>0.03 ± 0.03</td>
</tr>
<tr>
<td>Elbowb</td>
<td>21.52 ± 23.83</td>
<td>293 ± 311</td>
<td>0.14 ± 0.22</td>
</tr>
<tr>
<td>Shoulder</td>
<td>19.49 ± 13.77</td>
<td>316 ± 211</td>
<td>2.06 ± 1.52</td>
</tr>
<tr>
<td>Hip</td>
<td>19.83 ± 7.67</td>
<td>422 ± 174</td>
<td>3.09 ± 1.37</td>
</tr>
<tr>
<td>Knee</td>
<td>18.39 ± 14.43</td>
<td>356 ± 289</td>
<td>0.16 ± 0.12</td>
</tr>
<tr>
<td>Ankle and footc</td>
<td>17.88 ± 13.39</td>
<td>310 ± 210</td>
<td>0.07 ± 0.05</td>
</tr>
<tr>
<td>Cervical vertebrae</td>
<td>64.17 ± 29.04</td>
<td>1414 ± 831</td>
<td>4.36 ± 2.03</td>
</tr>
<tr>
<td>Dorsal vertebrae</td>
<td>64.39 ± 22.23</td>
<td>2171 ± 805</td>
<td>17.99 ± 6.12</td>
</tr>
<tr>
<td>Lumbar vertebrae</td>
<td>66.53 ± 21.56</td>
<td>1701 ± 689</td>
<td>19.15 ± 5.63</td>
</tr>
</tbody>
</table>

DLP: dose-length product.

a The values are indicated as the mean ± standard deviation.
b Only elbow (elbow above the head).
c Unilateral.

Table 3 Osteoarticular computed tomography (CT) doses within our institution with our previous 16-slice CT (Sensation 16, Siemens) [30].

<table>
<thead>
<tr>
<th>Type of CT</th>
<th>CTDI(w)a (mGy)</th>
<th>DLPa (mGy cm)</th>
<th>Effective dosea (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cervical vertebrae</td>
<td>21 (18.5–45.2)</td>
<td>411 (321–766)</td>
<td>1.3 (1–2.4)</td>
</tr>
<tr>
<td>Lumbar vertebrae</td>
<td>32 (23.4–56.4)</td>
<td>782 (399–1527)</td>
<td>8.8 (4.5–17.2)</td>
</tr>
<tr>
<td>Pelvic bone</td>
<td>21 (15.6–33.4)</td>
<td>602 (366–1359)</td>
<td>4.4 (2.7–9.9)</td>
</tr>
<tr>
<td>Shoulders</td>
<td>25 (23.4–35.0)</td>
<td>332 (253–688)</td>
<td>2.2 (1.6–4.5)</td>
</tr>
<tr>
<td>Knee</td>
<td>18 (10.9–31.2)</td>
<td>425 (195–757)</td>
<td>0.2 (0.1–0.3)</td>
</tr>
</tbody>
</table>

DLP: dose-length product.

a The values are indicated as the median and the extreme values in brackets.

Ways to reduce the dose in osteoarticular computer tomography (CT)

The ways to reduce the CT dose are based on the three main principles of radioprotection: justification, optimisation and substitution [32]. They have been adopted by the Euratom 97/43 European Community Directive [33] and by the ALARA (As Low As Reasonably Achievable) principle of precaution. All of these ways have been extensively detailed in the literature [9,34–38]. We will discuss them, in turn distinguishing the behaviour from the technical factors and focusing on their applications in the realm of osteoarticular CT.

Table 4 Doses with the lumbar vertebrae computed tomography (CT) and shoulder arthro-CT within our institution. Gathered with a 320-slice CT (Aquilion One, Toshiba) before and after implant of AIDR 3D iterative reconstructions (Adaptive Iterative Dose Reduction 3D, second version of the Toshiba iterative reconstructions).

<table>
<thead>
<tr>
<th></th>
<th>CTDI(w)a (mGy)</th>
<th>DLPa (mGy cm)</th>
<th>Effective dosea (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lumber vertebrae CT</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Before implant of iterative reconstructionsb</td>
<td>40.2 ± 11.4</td>
<td>1094 ± 309</td>
<td>12.32 ± 3.5</td>
</tr>
<tr>
<td>With AIDR 3D</td>
<td>25.5 ± 11.9</td>
<td>695 ± 338</td>
<td>7.83 ± 3.8</td>
</tr>
<tr>
<td>Shoulder arthro-CT</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Before implant of iterative reconstructionsb</td>
<td>43.9 ± 15.9</td>
<td>611 ± 259</td>
<td>3.98 ± 1.7</td>
</tr>
<tr>
<td>With AIDR 3D</td>
<td>16.1 ± 4.3</td>
<td>205 ± 82</td>
<td>1.34 ± 0.5</td>
</tr>
</tbody>
</table>

DLP: dose-length product.

a The values are indicated as the mean ± standard deviation.
b CT imaging acquired by filtered back projection with QDS (Quantum Denoising System, Toshiba).
Table 5 Tissue-weighting factors used to calculate the effective dose for different anatomic locations in osteoarticular disease, calculated according to Biswas et al. [22].

<table>
<thead>
<tr>
<th>Computed tomography (CT)</th>
<th>Tissue-weighting factor(^a) ((\mu)Sv/mGy cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder</td>
<td>6.52</td>
</tr>
<tr>
<td>Elbow(^b)</td>
<td>0.48</td>
</tr>
<tr>
<td>Wrist and hand</td>
<td>0.22</td>
</tr>
<tr>
<td>Hip</td>
<td>7.31</td>
</tr>
<tr>
<td>Knee</td>
<td>0.44</td>
</tr>
<tr>
<td>Ankle and foot(^c)</td>
<td>0.23</td>
</tr>
<tr>
<td>Cervical vertebrae</td>
<td>3.08</td>
</tr>
<tr>
<td>Dorsal vertebrae</td>
<td>8.29</td>
</tr>
<tr>
<td>Lumbar vertebrae</td>
<td>11.26</td>
</tr>
</tbody>
</table>

\(^a\) These factors were calculated from the relationship between the effective dose and the dose-length product by Biswas et al. [22]. Note that Biswas et al. calculated these factors using IMPACT dosimetry software based on the ICRP 60 data [11]. New factors have to be re-calculated to take into account the new ICRP 103 data.

\(^b\) Only elbow (elbow above the head).

\(^c\) Unilateral.

**Behavioural factors**

**Awareness and education**

First, like in other domains, the education and awareness of radiologists and radiology technicians is an important element in the reduction of the dose in osteoarticular disease. Wallace et al. [39] have shown that, after doctor education, it was possible to obtain a 29% reduction in the dose of lumbar CT at several institutions. This education also emphasises the situations in which the reduction in the dose is especially important. For example, the patient’s age is a major factor since the potential risk of radio-induced cancer due to low-doses of X-rays decreases with age [40]. Special care is the rule for young subjects. In addition, the anatomic location of the CT is important. The effective dose of an acquisition at a distance from radiosensitive organs, as is the case for the peripheral joints, will be negligible (with effective doses sometimes lower than those of a chest X-ray) as opposed to lumbar CT imaging or proximal joint CT imaging (however, it should be noted that the calculation of the effective dose does not take into account the radiosensitivity of tissue related to age). However, the awareness of radiologists and operators requires knowledge of the doses delivered. From this point of view, the display of the DLP on the CT console before the acquisition is indispensable and currently systematically available for all manufacturers. This awareness is also increasingly guaranteed by the software used in the gathering and analysis of the doses delivered. This software allows for dosimetric monitoring per patient and detects the cumulative dose, which is sometimes high. The software also includes dosimetric warnings that help optimise the protocols and help monitor the overall reduction in the doses during optimisation [41]. More globally, national or international dose registers are also being created, such as, for example, the CT Dose Index Registry [42]. Under the initiative of the American College of Radiology, this dose register aims at obtaining the CT radiation doses from a great many American and foreign radiology departments in order to compare the doses and harmonise practices.

**Justification and substitution**

Justification and substitution are also two important elements in osteoarticular imaging where substitution by non-irradiating imaging techniques such as sonography or MRI is often possible [4,9,43]. For example, Oikarinen et al. [44], in their study on 30 lumbar CT carried out in patients under the age of 35 years, demonstrated that only seven of them (23%) were indicated. Of the 23 lumbar CT not indicated, 20 of them would have been able to benefit from an MRI whereas imaging was not indicated for the other three patients. Clarke et al. [45] also demonstrated that 90% of the lumbar CT could be replaced by an MRI. However, the MRI is not always possible due to problems of claustrophobia, non-compatible implants, pacemakers or even precarious medical conditions [9]. In addition, in certain diseases, the performance of CT is superior than that of MRI [4]. For example, in the study on the spine, the CT has proven to be more sensitive in the detection of bone changes following an infection [46]. CT is also better than MRI in the characterisation of certain structures such as gas and calcifications. With its good spatial resolution, CT also provides a better visualisation of scaphoid fractures [47] or even the detection of certain osteoid ostomas that remain invisible in MRI [48]. The angio-CT is sometimes more efficient than the angio-MRI in the assessment of vascular invasion of musculoskeletal tumours [49,50]. In our institution, CT imaging is indicated in the following situations: complex fractures, fractures with suspicion of vascular impairment, fractures-dislocations, initial assessment of musculoskeletal tumours, post-surgical monitoring, bone dysplasia and congenital malformations, dis-co-radicular spinal disease and assessment of joint impairment. In addition, arthro-CT may be carried out for almost all joints. It provides a better visualisation of superficial cartilaginous lesions than the MRI. It is also possible to carry our useful multislice reformations with the CT in a postsurgical assessment [51,52]. CT arthrography of the wrist also enables a better analysis of ligament lesions than the MRI or MR arthrography [53] while CT arthrography of the shoulder is a very efficient imaging technique for the detection of SLAP lesions [54].

**Length of exploration and number of acquisitions**

During CT imaging, the dose can be controlled by reducing the number of acquisitions (that is, the number of phases) and the length of the zone explored [55]. The length should be restricted to the zone of interest, previously detected by the CT topogram(s). As indicated above, this is one of the main reasons in the differences in doses between the different examinations. As regards the number of acquisition phases, CT often only includes one phase without injection in osteoarticular disease. However, with the development of interventional CT imaging and dynamic and perfusion CT, the limitation in the number of phases is of prime importance.
Position and centering

Exact centering of the anatomic zone for the CT imaging at the centre of the ring provides optimum image quality and dose delivered. The spatial resolution is actually better at the centre of the ring since more data is obtained there than at the periphery [56]. Moreover, good centring is especially required with the use of milliampere automatic modulation since this modulation considers that the patient is at the centre of the ring [57]. In case of poor centring, the automatic modulation significantly increases the dose [58]. The patient’s position also has an effect on the dose and quality of the image. The volume explored should be as thin as possible to limit the artefacts of beam hardening and reconstruction. This is why the shoulders are placed at a different height during exploration of the pectoral girdle (Fig. 2). During the imaging of a leg joint (foot, ankle, knee), the volume explored should be reduced by raising the opposite leg. Similarly, the peripheral joints should be acquired as far as possible from the patient’s trunk in order to reduce the dose received by radiosensitive organs. Biswas et al. [22] demonstrated that the acquisition of an elbow along the body compared with a position above the head was responsible for a considerable increase in the effective dose (8.35 versus 0.14 mSv).

Technical factors

Type of computer tomography (CT) acquisition

With the development of multislice CT, the spiral mode has extensively replaced sequential axial acquisition. However, the appearance of wide-area detector CT has enabled its return. The 320-slice CT has, with a single rotation, helped acquire a volume of up to 16 cm in length, thereby covering most joints (shoulder, wrist, hand, hip, sacroiliac, knee, ankle and foot). The advantage of this type of volume acquisition is that it considerably reduces the time of acquisition (up to 0.175 s for the acquisition of a volume of 16 cm, without shift between the first and last slice) and therefore patient movement artefacts. In addition, this type of volume acquisition reduces the irradiation compared with the spiral mode. In fact, with wide-area detector CT, the shadow phenomenon (overbeaming) is proportionally smaller than that with 16 or 64-slice CT [59,60]. It should be noted that this shadow phenomenon is independent of the collimation and is therefore relatively greater in case of narrower collimation. Therefore, the use of a reduced number of slices should be avoided with a multislice CT. The use of the volume mode also eliminates pre and postspiral irradiation (or over-ranging), characteristic of the spiral mode [61]. The dose of additional irradiation due to pre and postspiral irradiation is higher with an increase in the number of detectors and is also proportionally higher for acquisitions of smaller length [62] as is the case for acquisitions of the peripheral joints. With the acquisition of shorter anatomic zones with a 16 or 64-slice CT, certain authors recommend the use of the axial and non-spiral mode to eliminate the dose due to over-ranging [34,63].

Kilovoltage

The reduction in kilovoltage is the source of a major reduction in the dose. However, it is also the cause of an increase in noise (for example, by maintaining the other parameters constant, a reduction in kilovoltage from 120 to 80 kV reduces the dose delivered by a factor of 2.2 [58] but also increases the noise by a factor of 2 [58,64]). In practice, the increase in noise results in a deterioration in the quality of the image that becomes grainy. This appearance is harmful during the analysis of structure with small differences in density (as is the case for the analysis of soft tissue) in view of an alteration in the contrast to noise ratio. However, it is not harmful for the analysis of bone structures due to a high natural contrast. It is therefore possible to acquire peripheral joints (wrist, knee, ankle, foot) at 100 or even 80 kV (Fig. 3). For example, for a CT of the wrist with a centred acquisition of 6 cm, with 80 kV and 50 mAs, the quality of the image is satisfactory for the bone analysis, including a cast immobilisation (Fig. 4). This acquisition provides a total DLP total of 20.9 mGy cm, corresponding to an effective dose of 0.0046 mSv (with a tissue conversion factor of 0.22 μSv/mGy cm according to Biswas et al. [22]). By comparison, this effective dose is only 3.3 times as high as that of an X-ray assessment comprising five wrist incidences (4.6 versus 1.38 μSv) [65] and is less irradiating than a front chest X-ray (about 0.07 mSv) [66]. For thicker proximal joints (shoulder, hip, sacroiliac, spine), the kilovoltage should be adapted to the patient’s morphotype: 120 kV in a patient with a standard morphotype, 100 kV in thin patients while in overweight patients, a kilovoltage at 135—140 kV is sometimes required in order to maintain a satisfactory quality of image. Given that the iodine attenuation value increases with a decrease in kilovoltage [67], during arthro-CT of the proximal joints, it is preferable to use a maximum kilovoltage at 120 in order to improve the contrast to noise ratio. For the same reasons, peripheral arthro-CT (wrist, knee, ankle) may be carried out at 80 kV (Fig. 5). During vascular or perfusion exploration, a reduction in the kV is also possible at 100 or even 80 kV according to the thickness of the anatomic zone to cover [68]. Certain teams have also proposed low-dose acquisition protocols at 100 kV for the assessment of spinal trauma [69], myeloma [70] or even at 80 kV for the assessment of scoliosis [71] or even osteoporosis [72].

Milliamps

A milliamp reduction induces a proportional reduction in the dose delivered as well as an increase in image noise (the
noise value in inversely proportional to the square root of the milliamps). This may be harmful for the interpretation of the examinations requiring a good contrast to noise ratio, such as for the analysis of disco-radicular disease. In their study on lumbar CT, Bohy et al. [73] demonstrated that not more than a 35% reduction in milliamps was possible in the standard protocol since the diagnostic performance deteriorated beyond this point. In this study, Bohy et al. [73] used constant milliamps but adapted the body mass index for each patient. The development of the automatic modulation in the milliamps in the three planes allowed for the automation of the adaption of the milliamps to the patient’s morphotype [74]. Van Straten et al. [75] demonstrated that this modulation was especially useful in the shoulder and pelvic regions where it reduced the effective dose by 11 and 17% respectively. Its use is also of interest in adapting the milliamps to the patient’s morphology during the acquisition of lumbar CT imaging, while maintaining the same

Figure 3. Computed tomography (CT) of the right knee in a 41-year-old woman for a knee trauma. Acquisition with a 320-slice CT and volume rendering (VR) with 100 kV, 100 mAs, rotation time of 0.5 s, slice thickness 0.5 mm and 16 cm coverage for a total DLP of 93.5 mGy cm, corresponding to an effective dose of about 0.04 mSv. Axial plane of 0.5 mm passing through the anterior tibial tubercle (ATT) (a) and Reformation 3D VR (b): non-displaced fracture of the ATT with irradiating secondary articular fracture between the tibial spine and the medial tibial plateau.

Figure 4. Computed tomography (CT) with cast immobilisation of the right wrist in a 19-year-old man for an assessment after 4 months of a Schernberg type III scaphoid burst fracture. Acquisition with a 320-slice CT with volume rendering (VR) and a height of 6 cm with 80 kV, 50 mAs, rotation time of 0.5 s, slice thickness 0.5 mm and AIDR 3D iterative reconstruction for a dose-length product (DLP) of 20.9 mGy cm, corresponding to an effective dose of 0.0046 mSv. Axial plane of 0.5 mm (a) and front reformation of 1.5 mm (b) revealing the persistence of the fracture and absence of signs of consolidation. Note the good quality of the image in spite of the presence of cast immobilisation and the major reduction in the acquisition parameters and the DLP.
image quality. Mulkens et al. [76] also demonstrated that the use of the automatic milliamp modulation in the three planes reduced the dose of a lumbar CT by 37%. Mastora et al. [77] also demonstrated that, during the exploration of the thoracic outlet syndrome, the use of automatic milliamp modulation allowed for a 35% reduction in the dose without a loss of image quality. Moreover, certain authors proposed carrying out a low-dose protocol with low milliamperes. For example, Horger et al. [17] demonstrated that the low-dose acquisition of a whole-body CT was possible in the diagnosis of lytic lesions and the assessment of the fracture risk in patients monitored for multiple myeloma. A collimation of 16 $\times$ 1.5 mm was used with 120 kV and between 40 and 70 mAs milliamps per second. The effective dose of the CT carried out with 40 mAs was only 1.7 times higher than the dose of a standard whole-body X-ray assessment (4.1 mSv versus 2.4 mSv) [17].

Pitch

With certain current multislice CT comprising modern techniques of automatic milliamp modulation, the change in the pitch does not change the dose since it results in an automatic adaption of the milliamps [78]. A high pitch, of about 1.5, is preferable to reduce the time of acquisition and the movement artefacts (for example, during the exploration of a multi trauma patient). Nevertheless, the pitch should remain under two in order to maintain the optimum quality of the multiplane reformations [78] and avoid the appearance of spiral artefacts [34]. On the other hand, a low pitch is preferred for the reduction of metallic artefacts related to osteosynthesis materials [79].

Slice thickness

In general, the acquisitions are carried out in thin slices (0.5 to 1 mm), required for the analysis of bone structures and reconstructed in thicker slices (2 to 5 mm) for the analysis of soft tissue. The sub-millimetric slices improve the spatial resolution, reduce the effects of partial volume and allow for multiplane reformations [80]. However, at constant noise, the acquisition in thin slices is the cause of an increase in the irradiation [81]. If there is an excess reduction in the milliamps, the acquisition in thin slices leads to a major increase in the image noise. Therefore, whereas the acquisition is obtained in sub-millimetric slices, during the interpretation of the images, the thickening of the slices helps increase the signal to noise ratio [80] and improve the analysis of the soft tissue [82,83].

Field collimator

Field collimators are placed at the outlet from the tube and help limit the beam of exposure at the field chosen, thereby allowing for a reduction in the dose. The smallest possible field collimation should be used, in particular with the exploration of the small joints.

Iterative reconstructions

The use of CT iterative reconstructions represents major progress in the reduction of the dose (Table 4). The first results show that they allow for a reduction of up to 50% of the dose, while maintaining the same image quality [84,85]. Until now, few studies have assessed the value of iterative reconstructions in musculoskeletal imaging. In our institution, we carried out a study on 15 lumbar CT acquired with volume mode on a 320-slice CT with Adaptive Iterative Dose Reduction (AIDR), the first version of the iterative reconstruction by Toshiba and by comparing them with standard reconstructions in Filtered Back Projection (FBP). Our results found a mean reduction of 31% in the image noise with AIDR compared with the FBP images [10], without an alteration in the spatial resolution. This noise reduction corresponds to a potential reduction in the dose of 52%. These initial results are promising, especially since the versions of iterative reconstruction are quickly evolving. In our institution, the new version of AIDR 3D iterative reconstructions was recently installed and new studies are required to assess their impact on the reduction in the dose and the improvement of image quality (Fig. 6). While these iterative reconstructions are of particular interest in reducing the dose of CT imaging requiring a good contrast to noise ratio, their performance on examinations where bone analysis is of major importance, such as for example in the search for a fracture, is not as important. In fact, the high natural contrast of the bone structures allows for low-dose noise acquisitions without this affecting the interpretation in a significant manner [86]. Nevertheless, one of the benefits of iterative reconstructions is the possibility of reducing the artefacts related to FBP reconstructions and, in particular, the beam hardening artefacts [87] (Fig. 7). This is of particular interest in the analysis of the soft tissue and bone structures in contact with osteosynthesis material.
Traditionally, the best visualisation of metal materials requires an increase in the acquisition parameters such as the kilovoltage and milliamps as well as a low pitch and a thin collimation. All of these parameters result in an increase in the dose [88]. Iterative reconstructions allow for a reduction in these artefacts while avoiding an increase in the dose by the optimisation of other parameters (Fig. 8).

Noise reduction filter
The improvement in the contrast to noise ratio required for the analysis of the soft tissue, in particular for disco-radiculopathy, may also be obtained by the use of noise reduction filters by post-treatment software. These filters are applied on already reconstructed images. This allows them to be used with any CT image, including 3D reformations. As opposed to the filters used during the reconstruction of images, some of these noise reduction filters seem able to smooth the image without altering the spatial resolution. However, studies have to be carried out in order to confirm the value of this new post-treatment hardware.

Active collimation
Although not accessible in the choice of parameters, this technology is worth describing. Shields help reduce pre- and postspiral irradiation phenomena by using active collimation in the z-axis at the beginning and end of the spiral [89]. These shields are of particular interest when the pre- and postspiral irradiation accounts for a large share of the dose of patient irradiation, that is, during short acquisition with a 16 or 64-slice CT. Christner et al. [90] have shown that with a 64-slice CT, for a pitch of 1, a beam collimation of 38.4 mm and an acquisition length of 15 cm, the shields provide a 16% reduction in the total dose delivered to the patient. However, with an acquisition greater than 300 mm with a 64-slice CT, the pre- and postspiral irradiation accounts for less than 3% of the total dose, irrespective of the pitch [90]. In osteoarticular disease, this active collimation is therefore of great interest in reducing the dose during acquisition with 16 or 64-slice CT of the shoulders and hips considering the low coverage and the proximity of radiosensitive organs (thyroid and gonads).

In practice
The justification and substitution of examinations by CTg-raphy remains fundamental: "a scan that is not carried out is one that irradiates the least". The limitation of the coverage of CT imaging is also a simple way to reduce the irradiation: "the smaller the acquisition coverage, the smaller the dose". The volume acquired should be as thin as possible and a small field collimation should be used for the exploration of small joints. The kilovoltage and milliamps should be adapted to the type of acquisition (reduction of these acquisition parameters for the exploration of a peripheral joint, reduction of the kilovoltage for the acquisition of an arthro-CT), the indication (analysis of the soft tissue requiring higher milliamps than the analysis of bone structure) and the morphotype of each individual. The new techniques to reduce the dose should also be used as soon as possible (automatic modulation of the milliamps and the kilovoltage, iterative reconstructions, active collimation). Finally, with dynamic or perfusion acquisition, the dose should be reduced by limiting the number of acquisition phases (Boxed text 1).

Nevertheless, even if the reduction of the dose is primordial, in particular in young subjects, it should not be carried
out at the expense of the quality of the image and especially of the diagnostic performance: CT imaging with a reduced dose that provides poor image quality and does not allow for a diagnosis is more harmful than CT imaging with a normal dose that allows for a proper diagnosis.

**Dynamic computed tomography (CT) imaging of the joints**

Joint kinematics may be examined by a static study in different positions or by a continuous dynamic study. The latter should be privileged during study of the joint kinematics [91–93] since the constraints differ between a system in action and a static system [94,95]. The improvement in the temporal resolution of multislice CT and the development of multidetector CT now allows for dynamic studies to be carried out on the peripheral joints [96,97]. The adaptation of the acquisition parameters as well as the application of recent methods of dose reduction help maintain a low-dose delivered to the patient, often lower than that with traditional acquisition on a conventional CT. Therefore, CT imaging is a tool of functional analysis improving knowledge of the joint kinematics and its dysfunctions.

The dynamic study of joints is possible in spiral mode with a 64-slice CT. Tay et al. [98] have shown, in an experimental study, that it is possible to obtain the dynamic acquisition of a wrist in four phases with a very low pitch (0.1) by using a protocol with retrospective synchronisation of the movement. However, this technique induces a great many movement and "step" artefacts and a major increase in irradiation [98] rendering the efficacy much lower than that of volume acquisitions with multidetector CT.
In our institution, we study joint kinematics with a 320-slice CT. This allows for the acquisition of a volume up to 16 cm long. A rotation time of 0.35 s combined with a technique of partial data reconstruction (reconstruction of data over 180° by rotation) provides a temporal resolution of 0.175 s per rotation. Volume acquisition also has several advantages: reduction of the dose compared with the spiral mode by elimination of pre- and postspiral irradiation [61] and acquisition of the entire volume at one time, without shift between the first and last slices or shift due to the movement of the table. With this technique, it is possible to study the kinematics of the peripheral joints and provide diagnostic information in several clinical indications: occult instabilities of the wrist, femoro-patellar syndrome, posterior conflict of the ankle, thoracic outlet syndrome. Nevertheless, these indications are new and an assessment of the diagnostic performance of this type of examination is required in order to justify their use, especially in view of their irradiating nature. In fact, these dynamic studies require the repetition of several acquisitions leading to an increase in the irradiation when compared with a single acquisition. Nevertheless, on a peripheral joint, with the possibility of obtaining low-dose acquisitions without disturbing the interpretation of the movement [99], it is possible to obtain dynamic acquisitions with an effective dose under 1 mSv. For example, for the study of the kinematics of radio-ulnar deviation of the wrist with 12-phase volume rendering (80 kV, 11 mA, rotation time of 0.35 s, acquisition coverage of 6 cm), the DLP is only 91.2 mGy cm, corresponding to an effective dose of 0.02 mSv (Fig. 9). With this low effective dose, it is possible to study several types of movement (for example for the wrist: flexion/extension, tightening, ulnar and radial deviation) while maintaining a total effective dose much lower than 1 mSv. In the same way, for a dynamic exploration of the ankle, the DLP is under 200 mGy cm, corresponding to an effective dose of 0.046 mSv.

For dynamic explorations of the hips and shoulders, it is especially important to reduce the dose due to the optimisation of the acquisition parameters. If the dynamic exploration only involves bone segments, the high natural contrast of the bone allows for a considerable reduction in the kilovoltage and milliamps [100]. It is also important to reduce and centre the acquisition coverage of the zone of interest. In addition, even if Hristrova et al. [96] demonstrated that image quality is improved by the continuous acquisition of data, the additional irradiation makes acquisition in intermittent mode preferable, with a number of phases in general limited to 12. On the pelvis, this allows the dose of irradiation to be kept at less than 10 mSv, corresponding to a standard multiphase abdominal and pelvic examination.

Perfusion computed tomography (CT)

Perfusion CT was first described a great many years ago [101]. Like with dynamic examinations, the CT perfusion of musculoskeletal tumours is possible due to an improvement in the temporal resolution of multislice CT imaging and the development of multidetector CT. The study of perfusion by CT combines the advantages of perfusion imaging, providing data similar to that provided by the MRI but with better visualisation of the bone invasion (periosteal reaction, cortical break, osteolysis), the femoral necrosis and the neovascularisation. The quantification of perfusion curves by CT is as easy as with MRI [102]. This tumoral perfusion may be carried out in spiral mode with a multislice CT due to a 2D scanning movement [103] or in "step-and-shoot" volume rendering with a multidetector CT. The advantage of tumoral perfusion with volume rendering is that it is obtained without moving the table and therefore there are fewer movement artefacts, helping improve the quality of the reconstructions and perfusion curves. This technique

Boxed text 1 The seven gold rules to reduce radiation by osteoarticular computed tomography (CT).

- Comply with the indications.
- Whenever possible, prefer a non-irradiating method of imaging.
- Limit the CT coverage.
- Limit the number of acquisition phases during dynamic and perfusion examinations.
- Adapt the kilovoltage and milliamps to the indications and morphology of each patient.
- Reduce the kilovoltage during the exploration of the peripheral joints and arthro-CT.
- Use modern methods to reduce the dose (iterative reconstructions, automatic modulation of the milliamps...).
Figure 9. Dynamic arthro-computed tomography (CT) of the left wrist in a 37-year-old man for an assessment of persistent pain at the wrist following a trauma. Acquisition with a 320-slice CT with static acquisition of the arthro-CT with volume rendering (VR) with 80 kV, 50 mAs, rotation time at 0.5 s, thickness of the section 0.5 mm, 8 cm coverage for a dose-length product (DLP) of 25.7 mGy cm, then dynamic acquisition during a movement of radio-ulnar deviation comprising 12 volume acquisitions with 80 kV, 11 mAs, rotation time of 0.35 s, 0.5 mm section thickness and 6 cm coverage for a total DLP of 91.2 mGy cm. Frontal reformation of the arthro-CT in 1.5 mm section (a) revealing a transfixing rupture of the lunotriquetral ligament. The 3D VR reformations of the dynamic CT (a: radial deviation; b: neutral position; c: ulnar deviation) do not reveal instability of the carpus with dynamic movements of the radio-ulnar deviation of the wrist.
Figure 10. Bone perfusion computed tomography (CT) in an 18-year-old man for an assessment of osteoid osteoma of the distal femoral metaphysis of the left knee. Volume acquisition with a 320-slice CT of 15 phases (acquisition without injection then acquisition after injection with nine phases every 5 s then five phases every 10 s) with 100 kV, 50 mAs, rotation time of 0.5 s, section thickness of 0.5 mm and 4 cm coverage for a total dose-length product (DLP) total of 123 mGy cm. 0.5 mm axial sections in filtered back projection (a), with AIDR iterative reconstruction (b) and after temporal fusion of the different phases (c), 0.5 mm axial sections at arterial time (phase 4) without (d) and with bone subtraction (e) and perfusion curve (f). Note the deterioration in the quality of the native images (a) due to the major reduction in the acquisition parameters and the improvement in image quality due to iterative reconstructions (b) and the temporal fusion technique (c). Also note the hypervascularisation of the nidus of the osteoid osteoma with intense enhancement at arterial time, fully visible due to the reconstructed images with bone subtraction (d). The hypervascular aspect of the enhancement of the nidus is also confirmed by the perfusion curve (f) of the nidus (green curve) when compared with the perfusion curve of a region of interest placed in the popliteal artery (purple curve).
also allows for use of the fist acquisition as a bone subtraction mask, improving the detection and characterisation of bone anomalies. However, these perfusion studies give rise to a high increase in the dose delivered [103]. The protocol is optimised by reducing the coverage of the CT imaging, by limiting the number of acquisition phases and by adapting the acquisition parameters (reduction in the kilovoltage and milliamps). However, this reduction in the acquisition parameters leads to the deterioration of image quality. This may be compensated by carrying out a temporal fusion of the different phases. This technique allows for a summation of several images derived from different acquisition phases in order to obtain an image with less noise and of better quality (Fig. 10).

In our institution, we have, for example, studied the value of tumoral perfusion in the diagnosis and monitoring of osteoid osteomas [104]. In this disease, the MRI may be faulty [48] and the use of CT imaging more easily indicates the diagnosis by detecting the bone reaction around a small nidus. In addition to this anatomic data, the perfusion CT detects the hypervascularisation of the nidus (Fig. 10). The subtraction of the phases after injection with the first acquisition without injection allows for a sequence of bone subtraction revealing the bone marrow oedema around the nidus. All of this information, usually provided by the MRI, is now accessible with CT imaging. Nevertheless, the value and role as compared with the MRI has to be assessed, since the indications for perfusion CT remain limited to when there is a doubt as to the diagnosis in view of the irradiant nature of this technique and the youth of the patients monitored. To control the dose of irradiation, we target the zone of CT coverage at the zone of interest (only 4 to 8 cm suffices). Moreover, we choose a kV and mAs adapted to the morphotype of the patient and the anatomic zone. The number of phases is also limited to 15 or 16, with an acquisition every 5 s for the first nine phases (arterial phases) and then every 10 s. All of these measures allow for a perfusion CT with a total DLP generally between 100 and 500 mGy cm (Figs. 10 and 11).

**Dual energy computed tomography (CT)**

Dual energy CT is based on the acquisition of two superimposable images with two different kilovoltages. Based on these native images, it is possible to reconstruct a virtual image corresponding to any voltage of the x tube [6]. Each manufacturer proposes dual energy acquisitions on their CT. However, the techniques used often differ. This accounts for the differences in terms of performance and clinical applications between these techniques. For Siemens, dual energy acquisition is obtained from a bi-tube CT. With each rotation, this allows for an image to be obtained with one tube set at 80 kV and the other at 140 kV. For the other manufacturers, only one tube is used to create the dual energy. General Electrics uses a generator that allows for a switch in 0.5 ms between high and low voltage. It is thereby possible, during a single rotation, to switch 500 times and obtain two series of raw data in order to reconstruct two images, one at 80 kV and the other at 140 kV. As for Toshiba, it benefits from the wide coverage of its system of detection (16 cm per revolution) to propose two consecutive revolutions with a change in voltage between both revolutions. Finally, Philips uses a double layer of detector, the first measuring all of the rays transmitted and the second only measuring the hardest beams [105]. Several applications of the dual energy CT in the osteoarticular domain are now clinically available although still in the assessment phase [6]: improvement in tissue characterisation, bone subtraction, differentiation of bone and iodine contrast product or even reduction of metallic artefacts. The improvement in tissue characterisation was first used for the detection and characterisation of urate deposits in gout [106,107]. An initial study by Nicolaou et al. [108] demonstrated that the dual energy acquisition of all peripheral joints (elbows, wrists, hands, knees, ankles and feet) provides good sensitivity and good specificity in the detection of locations of tophaceous gout while the total effective dose ranged from only 2 to 3 mSv. Another dual energy application is the possibility of obtaining reconstructions in bone subtraction. It is then
possible to detect the bone marrow oedema. Pache et al. [109] demonstrated that it was possible to detect post-traumatic bone marrow oedema in knee CT imaging, with an increase in irradiation of about 28% compared with a conventional CT. The dual energy bone CT may also become an alternative in case of counter-indication for an MRI or if it is unavailable. However, clinical studies are required to specify the role of this technique compared with the MRI.

Subhas et al. [67] demonstrated that the use of dual energy on a shoulder arthro-CT provides a better signal to noise ratio with an equivalent dose. Finally, with a dose equivalent that of a standard single energy acquisition, Bamberg et al. [110] demonstrated that the use of dual energy allowed for a reduction in prosthesis-related metallic artefacts.

Conclusion

Computed tomography is an imaging technique that continues to benefit from a great deal of technological progress. These technological developments, in particular with the development of iterative reconstructions, provide a considerable reduction in the dose delivered to the patient. They also make it possible to access new applications such as perfusion CT or dynamic CT. The modern techniques for dose reduction are of special interest in applications that involve repeated acquisition phases. Nevertheless, the optimisation of the milliamps and kilovoltage, the limitation in the coverage of CT as well as compliance with the indications are still the main ways to limit the doses delivered to patients. These
new indications and the possibility of obtaining low-dose or even very low-dose acquisitions, while maintaining excellent image quality, give CT a major role in musculoskeletal disease. While CT has already replaced the standard X-ray assessment in certain indications, its role with respect to non-irradiating imaging techniques, such as the MRI, still has to be defined.

Disclosure of interest

The authors declare that they have no conflicts of interest concerning this article.

References

CT dose optimisation and reduction in ostearticular disease


