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4D X-ray Computed Tomography

Amit Mehndiratta, Soenke H. Bartling, and Rajiv Gupta

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41.1 Introduction

The concept of X-ray computed tomography (CT) was pioneered by Sir Godfrey Hounsfield and Allan McLeod Cormack in the 1970s at the EMI Central Research Laboratories (Middlesex, UK) and Tufts University (Boston, MA), respectively. Their main idea of using multiple projections to create tomographic images formed the basis of the scanner made by Electrical and Musical Industries (EMI), the first clinical brain scanner. Since the days of the first brain scan in the early 1970s, CT has come a long way. Multi-detector computed tomography (MDCT) has seen a steady increase in capabilities, availability, and dedicated protocols for various applications (see Section III, Chapter 32). One such application is imaging of temporally evolving processes (e.g., imaging flow dynamics in blood vessels), an application that is usually referred to as 4D or dynamic imaging.

Traditionally, Digital Subtraction Angiography (DSA) has been the Gold Standard for acquiring dynamic imaging of blood vessels. DSA enables two-dimensional X-ray projections that can be acquired at a fast rate, usually, 30 frames per second (fps) or higher. As such, DSA enables dynamic imaging of the vasculature and other anatomy in the projection domain (i.e., 3D dynamic imaging using time-elapsed 2D projections). This chapter discusses technologies and methods required to add another dimension to this process, viz., 4D dynamic imaging using time-elapsed 3D tomographic views.

In standard computed tomography, it is routine to acquire 3D slice-by-slice datasets. In fact, MDCT has replaced DSA for routine vascular imaging at many centers. CT Angiography (CTA) by using intravenous contrast injection provides quantitative high-resolution 3D images using a minimally invasive technique. Unlike DSA, however, a CTA is a static snapshot of the vascular anatomy under consideration. It is, therefore, limited in its ability to reveal dynamic features of any time-varying phenomena. There are other issues, such as vulnerability to metal artifacts that limit its application in post-intervention imaging after clipping or coiling of aneurysms, or after embolization of arteriovenous malformations (AVMs).

Increasing temporal resolution of MDCT, which is now capable of rotating at three to five revolutions per second, has opened the possibility of 4D imaging via acquisition of successive 3D datasets separated in time. Wide area detector CT scanners are also available for clinical use. Such scanners—by their ability to acquire 3D datasets in one rotation without the need to translate the patient through the bore of the scanner—can also be operated to capture a 4D dynamic CT. Concomitantly, the advent of larger-area flat-panel detectors has enabled a completely new genre of scanners that use these detectors on a C-arm gantry or a standard CT gantry to acquire 2D, 3D, and 4D images. We discuss these technologies next.

41.2 Imaging Using a Conventional MDCT Gantry

41.2.1 Perfusion Imaging

Assessment of tissue perfusion is important for a variety of applications, that include management of ischemic stroke, cardiac perfusion, and other instances where tissue viability is the main clinical question. The basic paradigm consists of repeatedly
scanning a tissue with a slab of CT data and to observe the dynamics of the first-pass of a bolus of contrast. We illustrate this process with an example from neuroimaging in the setting of acute ischemic stroke.

Figure 41.1 shows the prescription of two slabs on a scout CT that will be the target of perfusion assessment. A typical perfusion CT scan targets one or two slabs, in cine acquisition mode. The number of slabs needed depends on the maximum collimation area available; a wide area MDCT (for example, the Toshiba Aquilion with 16 cm coverage) will require only one slab, while a 64-slice scanner (4 cm coverage) will typically need two slabs to cover the full middle cerebral artery distribution. Newer scanners also allow a shuttle mode where a wide slab is prescribed and the scanner shuttles back and forth between the boundaries of this slab. In this scenario, one trades-off the temporal resolution for the slab acquisition with the area covered by the slab: a wider slab results in greater coverage and poorer temporal resolution.

Typically, a low tube voltage (e.g., 70–100 kVp) is used to be close to the k-edge of iodine and to increase the conspicuity of iodine. In order to keep the radiation dose as low as possible, the tube current is also kept low (e.g., 200–300 mA). The tube collimation is kept as wide as possible to increase the area of coverage. On a 64-slice MDCT, a typical setting would be 8 × 5 mm slice mode (i.e., 8 CT slices, each 5 mm thick, for a total of 4 cm acquisition per slab). Since the blood flow dynamics varies significantly between the arterial and venous phases, it is advantageous to acquire the arterial phase images at a faster rate than venous phases images. Therefore, one typically divides the entire image acquisition into two phases: (a) Phase I (cine)—fast acquisition (e.g., 1 image every second for 40 seconds, with a 0.5 second reconstruction interval); (b) Phase II (axial)—slow phase (e.g., 1 image every 3 seconds for 27 seconds). This technique allows one to extend the total duration of image acquisition while limiting the radiation dose. In the above example, one can acquire data for 67 seconds, while the X-ray exposure was only for 49 seconds. In this example scenario, the protocol has a CT Dose Index (CTDIvol) of about 470 mGy and a dose-length product (DLP) of 1890 mGy-cm. The overall dose for the example CT perfusion protocol is less than 0.5 Gy CTDI (vol), and is less than the 0.6 Gy recommended by the American Association of Physicists in Medicine (AAPM). One can further reduce the dose by 25% if a tube current of 150 mA is used instead of 200 mA.

Figure 41.2 shows a typical time-density curve for one slice obtained from slab prescription in the middle cerebral artery territory. As can be seen by the perfusion deficit, in this patient the acute ischemic stroke involves the left cerebral hemisphere. The regions of interest (ROI) are color-coded to reflect three different vascular territories: an artery (Medium-gray), non-ischemic brain (Light-gray), and ischemic brain (Dark-gray). As can be seen, the arterial pattern, which forms the reference curve for perfusion analysis, shows a brisk upstroke, followed by washout, followed by recirculation and steady state. These phases are slightly delayed in the normal brain parenchyma, but are significantly delayed in ischemic brain parenchyma.

Perfusion parameters of brain parenchyma and, for that matter, any other tissue, consist of parametrizing the time-density curve for each voxel. One such parameterization is shown in Figure 41.3. The area under the time-density curve

![Figure 41.1](image1.png) **FIGURE 41.1** Prescription of two CT slabs being targeted for dynamic imaging in a patient suspected of acute ischemic stroke on a 64-slice scanner. On a wide area MDCT, a single slab would suffice.

![Figure 41.2](image2.png) **FIGURE 41.2** Time-density curve for three regions of interest (ROI) in a patient with acute ischemic stroke. Each color-coded curve, corresponding to the ROI with the same color, shows the blood flow dynamics of that region. Medium-gray = Arterial input function; Light-gray = Normal non-ischemic brain parenchyma; Dark-gray = Ischemic brain.

![Figure 41.3](image3.png) **FIGURE 41.3** Parametrization of time-density curves to derive perfusion parameters that distinguish ischemic from non-ischemic tissue.
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is a measure of the blood volume in the tissue, usually referred to as the cerebral blood volume (CBV) or tissue blood volume (TBV). The time from the start of the bolus arrival to the end of the first-pass washout is denoted as a marker of the mean transit time (MTT). If the tissue is ischemic, the MTT will be increased. Once can define the cerebral blood flow (CBF) into the voxel as CBV divided by MTT. Of the three perfusion parameters—CBV, CBF, and MTT—only two are independent.

Figure 41.4 shows a non-contrast CT and CT angiogram (CTA) for another patient. As expected, it is very difficult to visualize the early ischemic changes from the acute stroke on a non-contrast CT. The lack of cerebral perfusion due to an acute thrombus in the left middle cerebral artery (MCA) is much better appreciated in the CTA slice. Figure 41.5 shows the perfusion maps for this patient at the level of the insula. In this figure, the diffusion-weighted magnetic resonance (MR) image (DWI) depicts the core of the infarct (i.e., irreversibly infarcted brain tissue). As can be seen, the MTT is increased in the ischemic territory, and there is a concomitant decrease in the CBF. In addition, the MTT and CBF abnormalities are much larger than the core of the infarct.

Given the hyper-acute presentation for this patient, and relatively small perfusion defect given that the entire left cerebral artery is threatened, this patient was taken for catheter angiography and thrombectomy. Figure 41.6 shows four images from the catheter angiography showing progressive recanalization of the left middle cerebral artery. Figure 41.7 shows two non-contrast CT scans and a slice from the CT angiogram of the same vascular territory performed the next day. An intracranial stent was deployed in this case, and can be seen in this image. As can be seen, the left MCA has been completely recanalized. Figure 41.8 shows the CT perfusion maps after thrombectomy and recanalization of the left MCA. As can be seen, the majority of the perfusion defect has been reversed.

CT perfusion (CTP) has been in research and clinical use for some time now; however, its role in the work-up of acute ischemic stroke is a topic of current research. Research in CTP perfusion is challenging secondary to a multiplicity of technologies and processing paradigms, relatively low contrast-to-noise and signal-to-noise ratios, and potentially high radiation dose. One can address the algorithmic complexity and the radiation dose. However, the biology of acute stroke is challenging, and interpretation of CTP maps for acute patient triage remains controversial.
Atherosclerosis of the coronary arteries with its various clinical manifestations is the major cause of morbidity and mortality in industrialized nations. Even though effective treatment strategies to lower coronary event risk, such as statin therapy, are available, there are $\sim 540,000$ myocardial infarctions and $\sim 515,000$ deaths from coronary artery disease (CAD) every year in the United States (AHA 2003). Invasive coronary angiography remains the standard for the detection of coronary artery stenosis in symptomatic patients. In 2004, 1.4 million coronary angiograms were performed for diagnosis only. The overall complication rate was 8% (3.6% of diagnostic procedures and 15.1% of therapeutic procedures). The procedure-related mortality rates were 0.2%, 0.1%, and 0.5%.
A non-invasive method for visualizing coronary stenosis could significantly reduce the number of diagnostic angiograms, with associated reduction of morbidity and annual expense ($9 billion), and provide the opportunity for earlier diagnosis, assessment of the natural history of CAD, and response to therapy. Two non-invasive imaging techniques—magnetic resonance (MR) and CT—can be utilized in image coronary stenosis. MR has the ability to detect coronary stenosis in proximal vessels. However, artifacts due to cardiac and respiratory motion, limited through-plane resolution, and long overall imaging times make coronary MR imaging a significant challenge. In a review of magnetic resonance imaging (MRI) by Kaandorp et al. (2005), both sensitivity and specificity were unsatisfactory for clinical evaluation of coronary stenosis. Future developments may provide necessary speed and resolution for imaging coronary stenosis (Yeh et al. 2005).

Multi-detector CT (MDCT) constitutes a new approach to coronary artery imaging. Simultaneous data acquisition in several parallel slices with sub-second gantry rotation times and data reconstruction with electrocardiogram (ECG)-correlated partial scan algorithms permits high-resolution visualization of coronary arteries in 1 to 2 seconds. Meta-analysis demonstrates that the improvement of scanner technology resulted in a significant increase in diagnostic accuracy for the detection of stenosis (Hoffmann et al. 2004a,b). Most of the progress can be attributed to the improvement in temporal resolution and the corresponding reduction of motion artifacts (Nieman et al. 2002; Ropers et al. 2003; Dewey et al. 2004; Hoffmann et al. 2004a,b; Kuettn er et al. 2004, 2005; Mollet et al. 2004, 2005). With the availability of fast, wide area MDCT scanners and robust ECG gating, cardiac imaging has emerged as a prominent application of MDCT, especially for evaluation of acute chest pain in the Emergency Room.

In order to understand dynamic imaging of a beating heart, one must consider two unique technical aspects of Cardiac CT: (1) Projection data acquisition and reconstruction, and (2) cardiac synchronization.

### 41.2.2.1 Projection Data Acquisition and Reconstruction

In general, slightly over 180 degrees of projection data—specifically, 180 degrees plus the fan angle—is required to reconstruct a CT slice (see Section III, Chapter 33). This dataset can be acquired in one of two ways. In half-sector reconstruction, the entire dataset for one slab is acquired in a single heartbeat. This requires fast gantry rotation, and the ability to target a cardiac phase where the heart is relatively stationary. Figure 41.9 depicts the projection data acquisition scheme and a comparison between an ungated scan versus ECG-gated half-sector reconstruction.

Half-sector requires a fast gantry rotation such that one is able to acquire the data for one slab (typically, the detector width) in the diastole of one heartbeat. If one assumes a typical heart rate of 60 bpm, each heartbeat is 1-second-long, of which, the useful part of the diastole is about 600 ms. For a gantry rotation time of one rotation per second, this is just sufficient for half-sector reconstruction. Higher heart rates would be challenging for this gantry rotation time. One way around this problem is to acquire the projection data over multiple cardiac beats, as shown in Figure 41.10, in a scheme sometimes referred to as multi-sector reconstruction.

In multi-sector reconstruction, the needed projection data is acquired over multiple heartbeats. The number of heartbeats required for this purpose depends on the heart rate and rotation speed of the gantry. Multi-sector reconstruction assumes that projection data from different beats can be merged for reconstruction. Therefore, it implicitly assumes that there is perfect synchrony between the electrical and mechanical cardiac cycles, and that the heart returns to the same precise location after each heartbeat.

### 41.2.2.2 Cardiac Synchronization

For successful 4D reconstruction, the projection data has to be correlated with the cardiac cycle. To achieve this synchronization, in a scheme called prospective gating, the scanner targets a particular phase in the cardiac cycle and turns on the X-ray source only during that phase. As a result, only one phase of the
One can think of prospective gating as conventional 3D imaging targeted to one cardiac phase. For acquiring 4D imaging, the scanner acquires projection data continuously and time stamps each projection by the phase of the cardiac cycle, based on the concurrently acquired ECG. All projection data is then retrospectively sorted and reconstructed, as shown in Figure 41.12. This scheme, referred to as retrospective ECG-gating, is capable of making a 4D map of the heart. One can use such a reconstructed image to assess valvular motion, ejection fraction, and other parameters of the cardiac function.

### 41.2.3 Wide Area MDCT Systems and Shuttle Modes

Starting from a single slice scanner, the number of slices and the detector width of MDCT systems have progressively increased. Wide area MDCT systems with 256 or 320 detector rows...
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(e.g., Philips ICT and Toshiba Aquilion 320) have been commercially available for some time. These scanners, by virtue of their wide detector size, are able to acquire a 3D dataset in one rotation measuring as much as 12 to 16 cm along the Z dimension. By rotating the gantry continuously, one can acquire a time-resolved volumetric dataset. Using this functionality, one can visualize dynamic processes such as evolution of intravenously injected bolus of contrast over time. One can also use this functionality to assess cardiac function or brain perfusion.

Such 4D imaging may also be accomplished using smaller detector size—for example, a 64 slice scanner with a detector array that is only 4 cm wide—by using a feature known as the shuttle mode. Shuttle modes are now available on nearly all modern scanners. They function to increase the Z-coverage of dynamic imaging, albeit at the expense of temporal resolution. In the shuttle mode, the patient table rocks back and forth between two pre-defined Z locations, as the scanner continuously rotates, as shown schematically in Figure 41.13. Each forward excursion of the patient table captures the anatomy at one point in time, and each backward excursion captures the same anatomy at a slightly later point in time. Each slab of data may be acquired in the axial or helical mode. A temporal sequence of such a scan constitutes a 4D map of any portion of patient anatomy. Clearly, the wider the separation between the two Z extents of the scanned volume, the poorer the temporal resolution. However, clinically meaningful 4D imaging is possible using such as scan mode.

41.3 Imaging Using a Flat-Panel Detector-Based System

41.3.1 Dynamic Imaging Using C-Arm Systems

The flat-panel (FP) detector technology was originally developed to improve the standard radiography by providing a higher absorption coefficient and a wider dynamic range than available with X-ray film. Digital readout and higher frame rates per second offered an additional practical advantage and the possibility for fluoroscopic examination. This new FP detector technology has been under investigation for standard X-ray computed tomography applications and under evaluation for a new dedicated scanner design. Currently, FP detector technology is widely available.
in clinical practice as a C-arm system built for radiography and fluoroscopy imaging. These systems can rotate and can take projection data over an angular range of 180° or more (Kalender and Kyriakou 2007).

The idea to acquire projection data over a fan angle of 180° or more was investigated in the 1990s (Fahrig et al. 1997; Fahrig and Holdsworth 2000; Linsenmaier et al. 2002; Grass et al. 1999); the early work was conducted using a C-arm system with a conventional image intensifier. Image intensifier-based C-arm systems were typically used only for high-contrast vessels, imaging with an intra-arterial injection of contrast agent. They provided high-resolution imaging at contrast levels exceeding 1000 HU (Kalender and Kyriakou 2007).

Modern C-arm systems based on FP detector technology offer higher dose efficiency, better image quality, and dedicated applications for planning, guiding, monitoring, and assessing interventional procedures (Jaffray and Siewerdsen 2000; Ning et al. 2000; Groh et al. 2002; Jaffray et al. 2002; Kalender 2003; Holdsworth et al. 2005; Siewerdsen et al. 2005; Fahrig et al. 2006). C-arm systems are generally used in fluoroscopy mode; the source and detector are parked at one specific location; multiple X-ray projections are acquired in cine mode to capture the time series information during the intervention. The C-arm system can also be operated with the CT option, reconstructing the multiple slices and, thus, displaying the volume in a 3D view to the intervention expert during the procedure. CT fluoroscopy provides near-real-time feedback to the operator. The key advantage of CT fluoroscopy over X-ray fluoroscopy is the invaluable anatomical information that 3D CT can provide about the structures within the area of interest in complex procedures (Hsieh 2009). This information helps clinicians understand the best approach to introduce a percutaneous needle or a catheter.

Flat-panel C-arm systems differ from conventional MDCT in its performance parameters, focal spot size, and power levels. Typical parameters for C-arm system and clinical MDCT are given in Table 41.1 (Kalender and Kyriakou 2007).

Some common interventional procedures that can be performed with a C-arm system are transthoracic needle biopsy (Jiao et al. 2015), neuroendovascular interventions (Benndorf et al. 2005; Heran et al. 2006; Wallace et al. 2008; Hausegger et al. 2011), hepatic vascular interventions including arterial infusions, embolization, chemoembolization, radioembolization (Liu et al. 2005; Wallace et al. 2007; Meyer et al. 2007), and lower thoracic and lumber spine intervention procedures like vertebroplasty (Hodek-Wuerz et al. 2006).

### 41.3.2 Continuously Rotating Flat-Panel Detector-Based CT Gantries

Using a C-arm system, tomographic acquisition has hardware limitations; the system cannot do full circular rotation. A C-arm gantry is equipped with an X-ray source and a FP detector that can acquire projections for slightly more than a 180° angular span. Such a limited angular span limits the good signal to noise ratio for a CT like image reconstruction. The system also cannot rotate continuously acquiring multiple projections for temporally evolving 3D reconstructions (i.e., for capturing dynamic 4D acquisitions). Motivated by the need to capture resolution of a C-arm system and, simultaneously, the ability to perform continuous gantry rotation like MDCT, flat-panel Volumetric CT (fpVCT) systems were designed.

In simple terms, one can think of a flat-panel volume CT as a conventional multi-detector CT (MDCT), in which the rows of detector elements have been replaced by an area detector (Grasrauck et al. 2005; Nikolau et al. 2005; Popescu et al. 2005; Gupta et al. 2006, 2011). For example, a current fpVCT prototype uses a PaxScan 4030CB (Varian Medical Systems) CsI-amorphous silicon flat-panel detector (Gupta et al. 2006). The flat-panel detector in this system has an active area of 40 cm × 30 cm, that gives a 25 cm in-plane field-of-view and 18 cm coverage in the z-axis. The detector consists of a 2048 × 1536 matrix of elements, each with a dimension of 194 μm². The scanner can operate in 1 × 1 binning mode, giving the ultra-high, isotropic spatial resolution of 150 × 150 × 150 μm³; in a 2 × 2 binning mode the resolution is 200 × 200 × 200 μm³. These innovative systems thus bring into focus anatomy that heretofore has been in the domain of microscopy.

In the current fpVCT scanners, while a readout rate of 100 fps is theoretically feasible, the current implementations are limited to 30 fps. To optimize the stability of the machine and to acquire enough projection data enabling it to reconstruct a clinically useful CT image, the scanner can operate at a user-selectable rotation time that can be varied from 2 to 20 seconds per rotation. The gantry can also continuously rotate for 80 seconds while acquiring the projection data. This mode is suitable for dynamic imaging of an evolving scene. The Z-coverage afforded by these scanners is large enough to image an entire organ such as the brain, heart, liver, or kidneys in one axial scan. Unlike micro-CT, VCT is suitable for in vivo imaging of large animals, or for human studies.

The wide area detector used by the fpVCT prototype enables the operator to scan from a fixed angular position with a high image frame rate as well as scan continuously while the gantry is rotating. This flexibility gives rise to the following scanning modes.

1. **Ultra-high-resolution mode**: The fpVCT system can be operated in the standard CT scanning mode by acquiring projections during a full rotation and reconstructing

### Table 41.1

<table>
<thead>
<tr>
<th>Detector-Based CT Gantries</th>
<th>Typical Parameters of FP-C-Arm and MDCT</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>FP-C-Arm</strong></td>
<td><strong>MDCT</strong></td>
</tr>
<tr>
<td>Tube voltage</td>
<td>80–140 kVp</td>
</tr>
<tr>
<td>Tube current</td>
<td>10–600 mAs</td>
</tr>
<tr>
<td>X-ray power</td>
<td>20–100 kW</td>
</tr>
<tr>
<td>Focal spot size</td>
<td>0.6–1.2 mm</td>
</tr>
<tr>
<td>Rotation time</td>
<td>0.33–1 s</td>
</tr>
<tr>
<td><strong>Detector elements</strong></td>
<td></td>
</tr>
<tr>
<td>XY plane</td>
<td>512–1024</td>
</tr>
<tr>
<td>Z plane</td>
<td>16–256</td>
</tr>
<tr>
<td><strong>Field of view</strong></td>
<td></td>
</tr>
<tr>
<td>XY plane</td>
<td>500–700 mm</td>
</tr>
<tr>
<td>Z plane</td>
<td>2–40 mm</td>
</tr>
<tr>
<td>Minimum slice thickness</td>
<td>0.6 mm</td>
</tr>
<tr>
<td>Typ. scintillator/thickness</td>
<td>GdO₂S/1.0–1.4 mm</td>
</tr>
<tr>
<td>Data rate</td>
<td>≤1000 MB/s</td>
</tr>
</tbody>
</table>

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**Note**: The data provided in Table 41.1 is indicative and may vary depending on specific model configurations and manufacturer specifications.
them as a volumetric stack. The only difference is that the entire volume is acquired in one rotation, and images have a much higher resolution because of the inherent spatial resolution of the flat-panel detector. For ultra-high spatial resolution scanning the detector is read out in the $1 \times 1$ binning mode. In this mode every pixel is read out separately, and a spatial resolution of approximately $150 \mu m$ is achieved. In order to improve the signal-to-noise ratio and increase the frame readout rate, a $2 \times 2$ binning mode is used. In this mode, four neighboring pixels are averaged to make one effective pixel with outstanding spatial resolution of approximately $200 \mu m$. Despite this quantum improvement in spatial resolution compared with MDCT, the low contrast resolution is quite reasonable and approaches 5 HU in slices 10 mm thick.

2. Dynamic CT mode: fpVCT has the ability to monitor a volume of interest continuously over a period of time. The current prototype can rotate continuously for 80 seconds while acquiring projection data. The rotation time can be varied from 2 to 20 seconds. This enables observation of time-evolving processes such as first-pass dynamics of a contrast bolus, aneurysmal pulsations, blood flow pattern through an AVM, and tumor vascularity. This feature is further enhanced by the fact that fpVCT can cover a large volume in each rotation. If the temporal resolution is short enough and the imaging is conducted for an appropriate length of time, depending on the tissue type being studied, the evolution of a contrast bolus can be followed through the arteries, soft tissue, viscera, and veins. Using such a dataset, a perfusion study of these tissues can be performed, making it possible to combine fpVCT angiography and fpVCT perfusion in one dynamic imaging process. The temporal resolution of such a dataset will be equal to the rotation time of the gantry. In general, a higher rotation speed improves the temporal resolution at the expense of image noise and spatial resolution, because of the fixed frame rate of the detector. By increasing the rotation speed, the number of frames used in reconstruction is proportionately decreased.

3. Fluoroscopy mode: The imaging chain of the fpVCT can be “parked” in one angular position and real-time projection data acquired. Such a fluoroscopy mode enables visualization and intervention from any user-selectable angular position.

Below, a few clinical applications of fpVCT are shown, using one of the three operating modes singly, or in combination with 4D dynamic CT (Mehndiratta et al. 2015). Figure 41.14 shows the representative case with bilateral internal carotid

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**FIGURE 41.14** Four phases from a dynamic CTA of an internal carotid stent (a–d) showing 3D rendered view and (e–h) showing axial view. (a, e) Non-contrast phase, (b, f) arterial phase (Dark-gray arrows), (c, g) mixed phase, and (d, h) venous phases (Light-gray arrows). (Modified from Mehndiratta, A. et al. 2015. *European Radiology* 25:1901–10. With permission.)
artery (ICA) stents (only one is visible in this image). Four selected phases from a 16-phase dynamic CTA show the temporal evolution from non-contrast to early arterial, parenchymal, and late venous phases. The image quality, without any significant beam hardening, is satisfactory to evaluate the stented lumen. The figure has been adapted and modified from Mehndiratta et al. (2015).

**Figure 41.15** 4D dynamic images of traumatic lower limb injury after free-flap reconstruction. (a, b) The posterior tibial artery supplying the free flap with a patent arterial anastomosis and venous return. (c–e) Zoomed-in views at the level of the arterial and venous anastomosis, showing (c) arterial, (d) mixed, and (e) venous phases of the blood flow to the muscle free-flap. The venous coupler and the clips in the figure are approximately 1 to 4 mm in size. (Modified from Mehndiratta, A. et al. 2015. *European Radiology* 25:1901–10. With permission.)

**41.4 Advanced 4D Reconstruction for Image-Guided Interventions**

Similar to perfusion imaging, the same volume needs to be scanned repetitively in interventional radiology. Ideally, guidance information would be available continuously, and in a 4D (three spatial dimensions plus time) manner. This is currently not possible with MDCT scanners, because the CT fluoroscopy mode provided by them employs a prohibitively high amount of radiation dose and has low temporal resolution. Recent research in advanced reconstruction techniques, however, has revealed that, by using a certain kind of reconstruction algorithm and prior knowledge, 4D intervention guidance is feasible (Flach et al. 2012; Kuntz et al. 2013a,b). This can be done using flat-panel
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The requirements for tomographic imaging in intervention guidance are different from tomographic imaging in diagnostic imaging:

- Same or very similar objects are being repeatedly scanned, and the changes over time are small.
- Attention focused on high-contrast structures are of interest (wires, catheters, contrast media filled vessels).
- Absolute or relative consistency of the CT values is not necessary.
- Certain types of artefacts can be tolerated without losing the ability to provide sufficient information for intervention guidance.
- Prior knowledge (e.g., anatomy, structure of the instruments, etc.) specific to intervention guidance is available.

All of these characteristics can be advantageously used for tomographic scanning and reconstruction in intervention guidance to decrease radiation dose to enable continuous, tolerable 4D imaging.

Standard tomographic reconstruction algorithms treat every volumetric dataset as an independent volume. Therefore, they do not make use of the first condition (i.e., that changes over time are usually small) in the list of special conditions that pertain to interventional procedures, mentioned above. By defining the right constraints and by identifying useful prior knowledge to be incorporated into the reconstruction algorithm, the amount of necessary projection can be reduced. Figure 41.16 shows an example of real-time 4D imaging for intervention guidance.

To enable low dose 4D imaging, a prior 3D image is continuously updated with the help of low dose current projections as they are continuously acquired as the gantry rotates around the patient. The difference between the prior view and the updated view is reconstructed using an adapted compressed sensing scheme (Kuntz et al. 2013a,b). Two variations of this concept have been described in the literature (Flach et al. 2013; Kuntz et al. 2013a,b). Many more are feasible, each with interesting trade-offs between dose and image quality, and possible, as indicated in Figure 41.17.

The increasing use of cone beam CT for image guidance is very much a reality in daily clinical practice. However, 4D intervention guidance is not. At the current time, the developmental status of this modality is restricted to proof-of-concept implementations in phantom and animal studies. While many technical issues still need to be addressed, they will be overcome with advances in computing speed and power. None of the current limitations require unforeseeable discoveries to be made in order to overcome them. In fact, the trend suggests that incorporation of even more complex post-processing algorithms and prior knowledge into tomographic reconstruction can afford even greater savings in radiation dose. Other fields that depend on processing vast quantities of graphical as well as structural data (e.g., computer games, autonomous driving, etc.) are much more advanced than medical imaging, and gains made in those fields will directly benefit medical imaging.

Catheter procedures in the head and neck region seem to be best suited for 4D intervention guidance. These interventions are highly complex, and the need for an accurate depiction of a 3D relationship between the interventional instrumentation (e.g., stent struts and coils) and the underlying anatomy (e.g., the aneurysm sack) (Kuriyama et al. 2016). Head and neck interventions are also attractive because of their relatively small scan field-of-view and the lack of moving organs. In addition, the operational setup requires no major alteration in the workflow, because the imaging space (head and neck) in these interventions is far away from the working space (groin). Specialized neuro-interventional C-arm setups are being marketed to the neuro-interventional

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**FIGURE 41.16** An example of 4D intervention guidance from a living pig experiment. The guidewire (white) can be assessed within the carotid artery (red) from all potential directions during the procedure. Three selected views (anterior, lateral, and inferior) are depicted here. The surrounding anatomy (e.g., vertically oriented pig skull base) is viewed from the anterior direction to exemplify the versatility of 4D visualization.
community, so a niche market already exists. This market segment could see the first use of 4D intervention guidance.

REFERENCES


