4.1 Introduction

Death or incapacitating disability can be prevented or diminished in some patients who present within 6 h of embolic stroke onset by thrombolytic therapy [1–14]. With the recent advent of the Desmotplase in Acute Ischemic Stroke (DIAS) Trial results, this time window may soon be extended to as long as 9 h [15]. In the treatment of anterior circulation stroke, current guidelines support the use of intravenous (i.v.) recombinant tissue plasminogen activator (rt-PA) administered within 3 h of stroke [12, 13]. Intra-arterial (i.a.) thrombolysis, which has a longer treatment window, can be administered within 6 h of anterior circulation stroke ictus [9]. However, the risk of intracranial hemorrhage from these agents, as well as the probability of treatment failure, increases dramatically with time after stroke onset [6, 13, 16]. In the case of brainstem (posterior circulation) stroke, the treatment window for thrombolysis is greater than 6 h, not only because of the apparent increased resistance of the posterior circulation to hemorrhage, but also because of the uniformly poor outcome of untreated basilar artery strokes and the lack of alternative therapies [10].

Conventional, noncontrast head CT (NCCT) scanning is performed on all patients prior to treatment, in order to exclude hemorrhage or a large [greater than one-third middle cerebral artery (MCA) territory] infarction, both of which are contraindications to treatment [13, 16]. The role of unenhanced CT in the evaluation of acute stroke is fully discussed in the previous chapter. Although NCCT has some value in predicting patients most likely to be harmed by thrombolysis, it is of little value in predicting the
patients – those with large vessel vascular occlusions – most likely to benefit from thrombolysis.

Because NCCT and clinical exam alone are limited in their ability to detect large vessel thrombus, CT angiography (CTA) has been increasingly advocated as a first-line diagnostic test for patients presenting with signs and symptoms of acute stroke. Advancements in scanner technology, novel contrast injection schemes that allow for uniform vascular and tissue enhancement, and the availability of rapid post-processing algorithms have all resulted in the clinically practical detection using CTA of both circle-of-Willis thrombus [17, 18] and parenchymal stroke [19–24].

Due to the narrow time window available for the initiation of thrombolytic treatment, speed is of the essence. The rationale in the workup for acute stroke is therefore to identify as quickly as possible those patients who may benefit from i.a. or i.v. thrombolysis or other available acute stroke therapies. Importantly, CTA excludes from treatment patients with occlusive stroke mimics [e.g., transient ischemic attack (TIA), complex migraine, seizure] who will not benefit from, and may be harmed by, such therapies.

The CTA protocol described in this chapter was designed to provide diagnostically useful information about both the vascular and parenchymal phases of brain enhancement. The administration of contrast serves two purposes: first, to visualize acute clot in the MCA, distal internal carotid artery (ICA), basilar artery, and other major circle of Willis vessels (Fig. 4.1); and second, to better delineate potentially salvageable, underperfused, ischemic areas of the brain that are at risk for full infarction if circulation is not restored. More specifically this includes those areas with a relative lack of contrast enhancement on CTA source images (CTA-SI), consistent with reduced blood volume. Indeed, simultaneous recording and subtraction, on a slice-by-slice basis, of the pre- from the post-contrast axial images results in quantitative maps of cerebral blood volume (CBV) [25]. The reasons for, and clinical significance of, this blood volume weighting of CTA-SIs is fully discussed in the next chapter on CT perfusion. Moreover, CTA-SI can also be of value in suggesting otherwise unsuspected regions of vascular occlusion. For example, CTA-SI “perfusion deficits” present in multiple vascular territories, visible on post- but not pre-contrast images, suggests cardiac emboli as the stroke source.

Helical CT is less expensive and more readily available at most hospital emergency departments compared to MRI; therefore, performing CTA can be a quick and natural extension of the NCCT exam, an exam that is already routinely obtained as part of the pre-thrombolysis workup at most institutions. The addition of CTA seldom adds more than 10 min of scanning and reconstruction time to that of the conventional CT examination. With the speed of the newer generation of multidetector row CT (MDCT) scanners, a complete “arch-to-vertex” CTA can be performed during the dynamic administration of a single bolus of contrast, allowing visualization of the great vessel origins, carotid bifurcations, and Circle
Chapter 4

Stroke CT Angiography (CTA) of Willis – often obviating the need for further evaluation by MR angiography or ultrasound. Although complete post-processing of the entire neurovascular system is typically performed offline (during which time the patient can be prepared for thrombolysis, if necessary), the “critical” maximal intensity projection (MIP) reformatted images of the intracranial circulation can routinely be constructed directly at the CT console in under a minute.

This chapter will discuss the role of CTA in the diagnosis and triage of acute stroke patients. First, the general principles of helical CT scanning will be reviewed, including image acquisition and reconstruction techniques. The stroke CTA protocol will then be described, followed by specific issues regarding the accuracy and clinical utility of stroke CTA.

4.2 Background – General Principles of CTA

The development of helical CT in the early 1990s made possible the rapid acquisition of angiographic-type vascular images, with no greater risk of patient complications than that of routine i.v. contrast-enhanced CT scanning. With helical CT scanning, a slipp-ring scanner design developed in the early 1990s, permits the x-ray tube and detectors to freely rotate around the gantry for a full 360°, allowing CT image data to be continuously and rapidly acquired as the scanner table moves uniformly through the gantry. This results in the creation of a three-dimensional helical “ribbon” of data that can be reconstructed at any arbitrary slice increment, and reformatted in any arbitrary plane (Fig. 4.2).

4.2.1 Advantages and Disadvantages of CTA

CTA has both advantages and disadvantages compared to other vascular imaging techniques [26–28]. Increasingly, as scanner speed and spatial resolution continue to improve – for a relatively constant total radiation dose – with the ongoing development of more advanced 16-, 32-, and 64-slice multidetector row CT scanners, these disadvantages are becoming less restrictive, especially with regard to total iodinated contrast dose.

4.2.1.1 Potential Advantages

Speed. The entire length of the ICA can be scanned in under 60 s (the extracranial ICA alone in less than 30 s), minimizing image misregistration from motion and breathing artifacts, and often reducing contrast requirements.

Accuracy. CTA provides truly anatomic, non-flow-dependent data with regard to length of stenoses, residual lumen diameters and areas, and calcifications; flow-dependent techniques such as MR angiography (MRA) and ultrasound (US) are not able to provide these data. Figure 4.3 illustrates the accuracy of CTA as compared to contrast-enhanced MRA.

Low Risk. CTA has a lower rate of patient discomfort, is less expensive, and has considerably lower risk of stroke and other vascular complications compared to conventional catheter arteriography. It is also advantageous in situations when MR is contraindicated or cannot be performed. CTA is typically more readily available than MR, especially in emergency settings. CTA, unlike MRA, lends itself to the imaging of acutely ill patients, as there are no restrictions on the type and quantity of associated support equipment, such as intravenous pumps, ventilators, or monitoring hardware. Because CT scan acquisition is more rapid than that of MRA, CTA is less prone to motion artifact. When CTA is combined with CT perfusion (CTP) for the evaluation of acute stroke, quantitative perfusion data can also be obtained, which is not typically possible with MR perfusion imaging.

4.2.1.2 Potential Disadvantages

Limited Field-of-view and Spatial Resolution. With slower, older generation, single slice helical scanners, CTA is limited in its ability to optimally evaluate tandem or multiple lesions of diverse vessels. This is usually due to restrictions imposed by either x-ray tube heating or the contrast bolus dose. Also, for a given set of imaging parameters (mAs, kV, slice thickness, etc.), single-slice CTA has slightly decreased in-plane resolution compared to conventional axial CT imaging. With the newer, faster, multidetector row scanners, these limitations do not typically exist.
Figure 4.2

Sagittal and coronal MIP images, output from the CT scanner console, in the plane selected by the user. The slice thickness is adjusted to target the vessels of interest and "remove" overlying vessels that would obscure evaluation. 

- **a** Sagittal MIP image shows the course of the anterior cerebral artery with variant additional vessels arising from the anterior communicating artery.
- **b** Coronal MIP image shows the ICA and MCA bifurcations bilaterally.
- **c** Second coronal MIP image adjusted to view the basilar artery in its entirety.

Figure 4.3

- **a** Contrast-enhanced MRA: there is apparent nonfilling of the left vertebral artery (arrow).
- **b** Curved reformat from CTA of the vertebrobasilar junction now demonstrates patent vertebral arteries (arrow).
**Stroke CT Angiography (CTA)**

**Long Post-processing.** CTA reconstructions may require long post-processing times. Although axial, coronal, and sagittal MIP images of the circle of Willis can typically be obtained at the CT console by technologists in under a minute, more complex reconstructions, such as curved reformats of the entire length of the carotid or vertebral arteries, can take considerably longer.

**Physiological Data.** Unlike MRA and US, CTA is limited in its ability to provide physiological data, such as flow velocity or direction.

**Accuracy of Reconstructions.** Dense, circumferential calcifications can cause beam hardening and degrade the accuracy of vascular reconstructions. Optimal methods for measuring small (<1–1.5 mm) vascular stenosis are under investigation.

**Iodinated Contrast Risk.** The risks of routine i.v. contrast administration accompany CTA, including the possibility of allergic or idiosyncratic reactions or glomerular injury. Compared with US and MRA, which do not require i.v. contrast, CTA can be inconvenient for routine follow-up studies. Some hospitals commonly use US and MRA for noninvasive vascular screening and follow-up of carotid occlusive disease. In the nonemergency setting, CTA can be reserved for use as a problem-solving tool when the results of US and MRA are inconsistent, or when tandem lesions are present, such as in Fig. 4.4, and can often obviate the need for conventional catheter arteriography.

4.2.2 CTA Scanning Technique: Pearls and Pitfalls

The rapid acquisition of the large digital dataset required for CTA places great demands on CT imaging hardware and software. Because CT image noise is inversely proportional to x-ray photon flux through a given axial slice, optimal helical scanning requires a high x-ray tube heat load capacity.

Certain user-defined imaging parameters are unique to helical CT scanning. “Pitch” refers to the ratio between table increment per gantry rotation and x-ray beam collimation [29].

Depending on the precise vendor and generation of scanner (single-slice, 4-slice, 8-slice, 16-slice, etc.), tube-heating constraints may limit the total deposited tube current [in milliampere-seconds (mAs)]. For a given tube current (in mAs) and voltage (in kV), increasing the pitch, by increasing the table increment per gantry rotation, will allow greater coverage; however, there will be fewer photons per slice, resulting in quantum mottle. When the table travel per rotation is equal to the beam collimation, there is a one-to-one ratio between the column of transmitted x-rays and the detector width; the pitch is “1,” and image quality is improved [30]. When the table travel per rotation is less than the beam collimation, pitch is less than 1, and there is “overlap” of photon flux through the imaged tissue bed, resulting in even greater image quality. Figure 4.5 illustrates the advantage of overlapping the acquisition of helical data. Hence, in all our neuroimaging protocols at Massachusetts General...
In order to both optimize posterior fossa image quality and minimize helical “windmill artifact,” we therefore typically employ a pitch of 0.5–0.6, along with a relatively slow table speed of 5–7 mm/rotation, with the newest generation of multidetector row scanners. Additionally, in an attempt to reduce motion artifact, our protocols typically call for the most rapid achievable gantry rotation rate. This can be as fast as 0.5 s per rotation on the newer 16-slice scanners. Moreover, spatial resolution in the z-direction is optimized by the acquisition of minimally thin slices, although image noise and quantum mottle are reduced by the review of thicker reformatted images (Table 4.1). Finally, we routinely choose our tube current (in mAs) setting to be no higher than that which will permit the use of the small – rather than the large – focal spot size.

By maximizing longitudinal coverage and image quality, the overall time to complete imaging increases. This demands an increased amount of iodinated contrast media, so as to adequately opacify the arterial tree being imaged, while minimizing arterial dilution and wash out, as well as venous opacification. Ideally, the overall dose of contrast agent should also be minimized so as to lessen the risk of contrast-induced nephropathy (CIN), as well as allergic reaction. Variations on the theme of contrast administration include injection strategies such as a saline bolus chaser and multiphasic injection, which are explained below. For a given pitch, an increased table speed permits greater z-direction coverage of the arterial tree during peak, uniform contrast enhancement, whereas a slower table speed, although provid-

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**Table 4.1** Optimization of posterior fossa image quality and minimization of “windmill artifact”

- Minimal pitch (0.5–0.6)
- Slow table speed (5–7 mm/gantry rotation)
- Maximal gantry rotation rate (0.4–0.5 s/rotation)
- Thinnest possible acquisition matrix; thick reformats for image review
Stroke CT Angiography (CTA)

ing greater image quality, may result in imaging portions of the vascular system with less than optimal enhancement, and more venous opacification.

4.2.2.1 Single-slice Protocols

In order to achieve optimal arterial image contrast and resolution, single-slice CTA protocols must therefore be tailored to the region being studied, the clinical question being asked, and the heat load capability of the scanner being used. In general, values for slice acquisition, thickness, scan field of view (FOV), and pitch should be as small as possible such that the entire region of interest can be covered in a single scan. Of note, for both single and multislice scanners, the diameter of the body part being scanned should never exceed the scan FOV, or beam-hardening artifact from outside the FOV will result. Conversely, tube current should be as large as possible, within tube heating constraints, so as to maximize photon flux and therefore minimize quantum mottle (image graininess). The image reconstruction interval should also be minimized in order to reduce the step or “zipper” artifact. For example, optimal characterization of a known, focal, carotid bifurcation stenosis can be achieved using the following single-slice CT parameters: 1 mm collimation, pitch of 1, 0.5 mm reconstruction interval, 12 cm FOV, 140 kV, and 250 mA, scanned narrowly around the level of maximal stenosis. Screening of the entire carotid artery, however, requires 3 mm collimation, a pitch >1, and typically a reduced tube current and voltage, to permit greater z-direction coverage [27, 28, 31].

4.2.2.2 Multi-slice Protocols

With multidetector row CT (MDCT), the choice of optimal scan parameters becomes more complex. Armed with newer x-ray tubes capable of higher heat capacity, as well as novel detector designs, multislice scanners can comfortably image entire vascular territories without the aforementioned constraints. The more rapid MDCT affords imaging of the arterial phase during more uniform contrast enhancement, more closely mimicking a catheter arteriogram [32]. Early MDCT scanners, circa 1994, had a detector ar-ray of four rows along the z-axis, with four data channels to record the x-ray attenuation. Newer MDCT scanners employ 16 or more detector rows with varying combinations of individual detector widths, data channels, maximal z-direction coverage, and minimal slice thickness. Some vendors build detector arrays with narrower central rows than outer rows – so-called adaptive array detectors [33]. At the time of writing, 64-slice scanners will soon be commercially available, which will be capable of up to 4 cm of z-direction coverage and 0.6-mm-thick slices per gantry rotation. Such scanners will be ideally suited to coronary artery imaging. Overall, when compared with single-slice scanners, multislice scanners allow for reduced contrast dose, shorter scan duration, and thinner slices.

4.2.3 Radiation Dose Considerations

Interestingly, as the number of detector rows and channels increases in the evolution of MDCT scanners, allowing more coverage and better resolution, overall radiation dose to the patient has not significantly increased for a given level of image quality. At most institutions, the radiation dose for a typical head CT remains in the range of 0.06–0.12 Gy or 6–12 rad. To put this in context, the limits for occupational exposure to radiation (as set by the Nuclear Radiology Commission), for negligible likelihood of harmful effect, is no greater than 50 milli-sievert (mSv) per year (or 5 rem/year) [34]. In comparison, environmental exposure to radiation, such as from radon and cosmic x-rays, is approximately 3.5 mSv/year, and that of a single chest radiograph is roughly 0.05–0.1 mSv [35, 36]. It is noteworthy that these limits exist for occupational exposure, but not for patients, as it is assumed that patient exposure is diagnostically required.

To compare exposure between different CT protocols, one can use the absorbed dose, usually reported as the CT dose index (CTDI) in mGy. This concept was introduced by Jucius and Kambic to predict the multiple slice average dose at the center of a set of axial scans [37]. The product of the CTDI_{vol} and the z-axis coverage for the particular scan then yields the “dose-length product” (DLP), where CTDI_{vol} is the
average dose in the standard head CT dosimetry phantom [38, 39].

With regard to MDCT radiation exposure, several factors must be weighed. There is up to a 4.5% loss of efficiency versus single-detector row scanners due to absorption of radiation in the z-axis by septa between the detector rows [33]. However, with MDCT, more of the x-ray beam is utilized per rotation due to the increased number of detector rows, with less of a penumbra [33]. Subsequently, these effects balance, and the dose efficiency of MDCT has been shown to be comparable to that of single-slice helical CT [33].

A related issue is the so-called cone beam effect, which to date has been a rate-limiting step in the development of larger (32-, 64-slice) detector arrays. This refers to the fact that the further a given detector row is from the center of the x-ray beam, the more angled the beam is through that detector, introducing nonlinearities to the image reconstruction algorithm.

There are several strategies to reduce radiation exposure in MDCT. Reducing photon flux is the primary means; however, decreasing tube current (mAs) results in increased image noise, as there are fewer x-rays per slice interrogated. This “quantum mottle” therefore becomes exaggerated with thinner slices. As already noted, the noise associated with thin slices can be compensated for by review of thicker slab reformatted images. Indeed, it has been shown that increasing depth from 4 to 8 cm requires doubling the tube current (mAs) to maintain the same amount of noise [38, 39]. Again, to lower the tube current (mAs) and maintain diagnostic image quality with MDCT, thinner slices with decreased tube current (mAs) can be acquired and reconstructed at thicker, less noisy slice intervals. Table speed and pitch can also be increased, thereby decreasing exposure, so long as image quality is not degraded [40].

Lowering kilovoltage peak (kVp) is another strategy to reduce radiation exposure. The kVp setting reflects both photon number and photon energy. With higher kVp values, although fewer photons are absorbed (due to less interaction with biological tissue, which typically has a relatively low “k-edge”), each photon has higher energy. Hence, calculation of actual absorbed radiation dose becomes complex for low kVp scanning, in which tube current (mAs) is often increased to maintain image quality. Typical kVp for head CT scanning is 140 or 120 kVp [41]. A notable exception to this is CTP, which employs 80 kVp, so as to take advantage of the low k-edge of iodine, in order to increase conspicuity of the relatively small amounts of contrast reaching the brain parenchyma (see Chapter 5).

Specific advances in MDCT have been directed towards lowering radiation dose. Among these, automatic tube current modulation is more important for body than for head imaging. With this technique, the tube current is adjusted according to body diameter on the scout image, so as to maintain the same total amount of noise in the image (proportional to photon flux) for any given slice (Table 4.2) [42].

### 4.3 CTA Protocol for Acute Stroke

#### 4.3.1 General Considerations

A successful CTA protocol requires balancing multiple scanning, contrast administration, reformating, and reconstruction parameters, in order to obtain diagnostically useful images of the enhanced vascular tree. Reformating here refers to the immediate post-process-
Stroke CT Angiography (CTA)

Table 4.3  General principles of single-slice CT and multidetector row CT (MDCT)

<table>
<thead>
<tr>
<th>Single Slice CT</th>
<th>MDCT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard contrast dose</td>
<td>Reduced contrast dose</td>
</tr>
<tr>
<td>Longer imaging time</td>
<td>Shorter imaging time</td>
</tr>
<tr>
<td>Thicker slices</td>
<td>Thinner slices</td>
</tr>
</tbody>
</table>

more overlap of the helical dataset, raw images benefit from increased density of the x-ray attenuation data, resulting in less noise.

After reformating, MDCT data are displayed as both axial images (usually with a slice thickness that is a multiple of the acquisition slice thickness) and reconstructed images (discussed below). With MDCT, the interslice gap can be arbitrarily selected, so the subsequent 2D and 3D reconstructed images will not be subject to “zipper” or “stair step” artifact.

4.3.2 Contrast Considerations

Optimal CTA imaging is dependent on the concentration of intravascular iodine, which in turn is dependent on both choice of contrast and injection strategy. Nonionic CT contrast agents have been shown to be safe in an animal model of MCA stroke, without significant neuronal toxicity, even to already ischemic neurons [43, 44]. There are several forms of nonionic contrast, with varying concentrations of iodine. The relationship between concentration and enhancement is demonstrated graphically in Fig. 4.6 [45, 46].

Some patients are at higher risk for CIN, most notably those who are diabetic, who have pre-existing renal dysfunction, or who have a combination of the two. Nephrotoxicity from contrast media is dose dependent [47, 48]. Recently, Aspelin et al. [49] demonstrated the benefit of using iodixanol, an iso-osmolar agent, in patients with diabetes and borderline renal function, although this agent has both increased expense, and increased viscosity at room temperature, compared to other low osmolar (but not iso-osmolar) agents [49, 50]. In their randomized prospective multi-center study, diabetic patients with renal impairment undergoing angiography had a 3% incidence of CIN. To counter the increased viscosity of iodixanol, one can pre-warm the contrast media. It is noteworthy, however, that some studies have suggested that other low-osmolar agents, with lower cost and viscosity, may have similarly good safety profiles with regard to CIN.

For those patients who are allergic to iodinated contrast media, premedication with anti-histamines and steroids can blunt the anaphylactic response.
Although a full discussion of the management of iodinated contrast allergy is beyond the scope of this text, it is noteworthy that, in the setting of an acute stroke, there is insufficient time to complete a course of steroid administration, which requires many hours. For such patients, gadolinium may be used as an alternative CT contrast agent for the evaluation of acute stroke [51]. Importantly, some researchers have suggested that – at the doses required for CTA – any potential advantage of gadolinium over iodinated contrast, for use in patients with high risk of CIN, is negated. Figure 4.7 demonstrates the clinical utility of gadolinium for emergent evaluation of the Circle of Willis by CTA.

Yet another strategy to decrease total contrast load is the use of a so-called bolus chaser – saline rapidly injected immediately following the contrast bolus. This has the advantage of clearing the “dead space” of contrast in the i.v. tubing and brachiocephalic/subclavian veins that does not typically contribute to imaging (thereby also reducing streak artifact at the thoracic inlet). Bolus chasing, which requires the use of a dual-head CT power injector, has been shown to be effective in maintaining peak maximum enhancement, with a reduced total contrast dose [52–54]. The addition of a chaser results in a similar initial rate of enhancement, and can additionally contribute to a longer overall duration of enhancement [55], shown graphically by Fig. 4.8. It is estimated that bolus chasing can reduce total contrast load by an average of 25% or more [56, 57].

4.3.2.1 Contrast Timing Strategies

Achieving adequate arterial opacification depends on the volume, rate, and duration of contrast administration, graphically depicted in Fig. 4.9. Ideally, imag-
ing should begin as the arterial tree enhances, and be completed prior to significant venous opacification—something not routinely achievable due to the relatively short mean transit time (2–4 s) of contrast through the capillary bed. This is not typically problematic unless the cavernous sinus is being evaluated, although, as noted earlier, the extremely rapid imaging achievable with the newest generation of 64-slice helical scanners lends new importance to various contrast-timing strategies (Table 4.4).

Indeed, with up to 16-slice scanners, our acute stroke protocol typically calls for a fixed 25-s prep delay between the onset of contrast administration and the onset of scanning, except for patients with atrial fibrillation (or other causes of significantly decreased cardiac output), who require a longer, 35- to 40-s delay.

Scanners equipped with a “bolus tracking” function can use a variable prep delay, in order to minimize venous enhancement; however, this adds complexity to scanning protocols, and is seldom clinically important unless scanning is very rapid. With bolus tracking, scanning starts once a preset Hounsfield opacification is reached in a vessel of interest. As already noted, venous opacification does not routinely interfere with diagnostic evaluation, with the possible exception of cavernous sinus evaluation. The inherent lag within the system, between the desired time to start imaging and the actual image acquisition, remains the main disadvantage with this technique. This can range from 5 to 15 s, as reported in several studies [58–60]. Modern scanners have a delay of as long as 4 s with the bolus tracking function.

As an alternative to bolus tracking, a “smartscan” type function can be used, in which a test bolus is administered to determine the scan delay [61]. A region of interest is selected, typically in the proximal ICA, and 10 ml of contrast is injected. This region is scanned continuously using the “low mAs/kVp”
technique, and the prep delay is chosen as the time corresponding to 50% of maximal test vessel opacification. As with bolus tracking, however, a test bolus is seldom clinically necessary. Of the thousands of CTAs performed at MGH, scanning began too early with the use of a 25-s delay for only a handful of patients with low cardiac output secondary to atrial fibrillation. Figure 4.10 demonstrates the limited utility of a scan when the delay is not chosen correctly. Both mathematical and animal models have demonstrated that, when there is reduced cardiac output, intravenous injection will result in a delayed intra-

<table>
<thead>
<tr>
<th>Strategy</th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
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<tbody>
<tr>
<td>Higher iodine concentration</td>
<td>Better opacification</td>
<td>Higher contrast dose</td>
</tr>
<tr>
<td>Higher flow rate</td>
<td>Greater peak enhancement</td>
<td>Larger i.v. access</td>
</tr>
<tr>
<td>Higher injection volume</td>
<td>Greater peak enhancement</td>
<td>Vascular injury</td>
</tr>
<tr>
<td>Iso-osmolar contrast</td>
<td>Less nephrotoxic</td>
<td>Higher iodine load</td>
</tr>
<tr>
<td>Gadolinium</td>
<td>Can be used in urgent situations, for patients with severe allergy</td>
<td>Costlier</td>
</tr>
<tr>
<td>Saline chaser</td>
<td>Less contrast medium</td>
<td>Increased viscosity</td>
</tr>
<tr>
<td></td>
<td>Decreased streak artifact at origin of great vessels</td>
<td>Vascular enhancement not as dense; risk of increased nephrotoxicity with large gadolinium doses</td>
</tr>
<tr>
<td></td>
<td>Greater absolute difference in attenuation</td>
<td></td>
</tr>
<tr>
<td>Fixed delay</td>
<td>Simple, error free, straightforward for multiple technologists in large centers</td>
<td>Must lengthen delay for atrial fibrillation; may be inadequate for timing in very rapid newer (64-slice) scanners</td>
</tr>
<tr>
<td>Bolus tracking</td>
<td>Accounts for different patient physiology</td>
<td>More complex and time consuming for the technologist; scanning may not occur during peak or uniform contrast enhancement</td>
</tr>
<tr>
<td>Test bolus</td>
<td>Accounts for different patient physiology</td>
<td>More complex; requires 10–15 ml of additional contrast, not required for imaging</td>
</tr>
<tr>
<td>Multiphasic injection (Fleischmann)</td>
<td>Uniform plateau of enhancement</td>
<td>Requires test bolus</td>
</tr>
<tr>
<td>Multiphasic, with exponential decay (Bae)</td>
<td>Accounts for varying patient physiology</td>
<td>Requires dual phase CT power injector</td>
</tr>
<tr>
<td></td>
<td>Uniform plateau of enhancement</td>
<td>Requires multiple assumptions regarding patient factors that may not be known or justifiable</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Requires CT power injector capable of exponential decay mode</td>
</tr>
</tbody>
</table>
arterial arrival time, and a greater peak arterial enhancement [62]. In patients with reduced cardiac output, therefore, once again, a 35- to 40-s prep delay is typically employed. Increasing the degree of arterial opacification can be accomplished simply by utilizing either a larger injection volume, or a faster injection rate. With a test bolus, the time density curve generally has a slightly different geometry from that of the main bolus, as shown in Fig. 4.11. This is likely due to injection of a smaller amount of contrast without a saline flush, which has different dilutional effects [55].
As image acquisition takes a finite amount of time, usually longer than the time required for peak arterial enhancement, the resultant CTA image will include portions of the arterial tree that are not at peak enhancement. If the degree of arterial enhancement is not uniform, a thresholding-based reconstructed image may demonstrate a false occlusion [63, 64]. By using bolus-shaping strategies, so that arterial enhancement plateaus instead of peaks, more uniform vascular opacification is achieved and a more representative image of the vascular tree can be displayed. To determine the appropriate injection strategy, mathematical models of vascular enhancement have been developed. Bae et al. [62, 65] used a compartmental approach to model aortic enhancement. From this model, a multiphasic injection was proscribed by solving for the inverse of the model, the input function [66]. They found that an exponentially decelerated injection method yields uniform vascular enhancement [67]. Alternatively, Fleischmann and Hittmair [68] modeled aortic enhancement by treating the entire body as a “black box,” where the input function was defined by the administration of a small contrast test bolus, and the output function was simply the resulting, measured time-attenuation curve. The “patient function” was the “black box,” which substitutes for the multiple compartment model of Bae et al.’s approach. Using a discrete Fourier transform, the injection parameters that will result in a desired degree and duration of contrast enhancement in any given vessel can then be calculated. Unlike Bae et al.’s multi-compartmental model, no estimates or assumptions regarding patient weight, cardiac output, or other individual details needs to be included in the calculation. This tailored bolus can result in more uniform, optimal vascular enhancement, while simultaneously minimizing contrast requirements [69].

4.3.3 Post-processing: Image Reconstruction

4.3.3.1 Image Review

Appropriate image review is as important as optimal scanning technique for the correct interpretation of CTA studies. This applies to both image display parameters and three dimensional reconstruction techniques. With regard to soft copy image display, proper window and level settings are essential for optimizing parenchymal stroke detection. During image review of both NCCT and CTA images, windows and level settings should be chosen so as to exaggerate gray matter, white matter contrast, as discussed in Chapter 3 [70].

The CTA dataset is commonly reformatted into image slices twice that of the acquired scan width, however, with some degree of overlap. Even without overlap, this method may result in a prohibitive number of images for visual review. We therefore typically have two sets of axial reformatted images – one set with thick slices, for minimal quantum mottle, a minimal number of slices, and therefore optimal visual image review, and a second set with thin slices and maximal overlap, for use in 2D and 3D reconstruction. This second set minimizes certain reconstruction artifacts, such as the “zipper” effect.

By reconstructing the reformatted images into a composite view, with both thicker axial slices and optimally reconstructed images – including MIP, multi-planar volume reformat (MPR), curved reformat, shaded surface display (SSD), or VR – it is possible to
evaluate the vessel in its entirety. This becomes important for lesions at branch points, such as aneurysms. While occlusions of large vessels may be obvious from sequential axial images, cutoff of a tortuous vessel, such as a distal M2 segment of the MCA, may not be entirely clear without a detailed and lengthy review of the source images.

Review of reconstructed 2D and 3D images assists review of the 2D axial source images by providing a composite image of the vessels. Once this review is complete, referral to the axial source images can be made for further evaluation, such as the degree of stenosis. Confirming the presence of a small aneurysm or venous sinus thrombosis may be better accomplished on the thin axial source images as well. Finally, vascular dissection can be directly visualized on the axial source images.

Both 2D and 3D reconstructions are time intensive. At our institution, there are dedicated 3D technologists who use stand-alone workstations to output the reconstructed images. Of note, however, many scanners have easy-to-use reconstruction software that can be applied directly at the CT console by the technologist, with only minimal interruption of patient throughput. Specifically, our technologists routinely reconstruct axial, coronal, and sagittal MIP views through the brain on all or our head CTAs. These are done using 3-cm-thick slabs with 0.5 cm of overlap, take only approximately 30 s per imaging plane to reconstruct, and contribute greatly to a quick overview of the anterior cerebral artery (ACA, sagittal plane), MCA (axial and coronal plane), posterior cerebral artery (PCA, axial plane), and vertebral-basilar (coronal plane) anatomy, as shown in Figs. 4.1 and 4.2.

One potential pitfall of all of these rendering techniques is the obscuration of internal blood vessel lumens by immediately adjacent calcified structures. This is most problematic for the characterization of cavernous sinus aneurysms, and for the measurement of residual lumen diameters in the presence of heavy circumferential calcifications, especially in cases of severe carotid bifurcation occlusive disease [28, 71]. A pitfall unique to SSD, which makes it unsuitable for CT venographic reconstructions, is its elimination of image pixels by thresholding. This feature makes it possible to falsely create the appearance of a vascular stenosis if bone adjacent to a vessel is over-thresholded. In what follows, the various reconstruction methods are discussed in more detail:

### 4.3.3.2 Maximum Intensity Projection

As with MRA, a projected 2D image can be constructed by displaying only pixels with the maximum, or highest, CT attenuation along a given ray. One of the earliest descriptions of creating MIP images for CTA was by Napel [72]. As noted above, MIP images can be quickly and easily reconstructed at the scanner console, and are thereby readily available and convenient for review. These images can be constructed with a user-defined thickness. Furthermore, they are less sensitive than axial source images and SSD images to varying window and level settings. At Stanford, they have suggested that MIP is superior to SSD for renal artery stenosis evaluation [73].

![MIP image of normal carotid bifurcation; atherosclerosis may obscure the underlying enhancing lumen. (BIF Bifurcation, RCCA right common carotid artery)](image-url)
groups have suggested similar results for carotid artery stenosis.

As this technique relies on detecting the highest pixel on a given ray, it is sensitive to overlap from adjacent bony and opacified venous structures, as shown in Figs. 4.12 and 4.13. Currently attempts are being made to subtract the underlying bony structures, but to date these have been limited by multiple technical factors, not least of which is subtle patient motion resulting in inaccurate representation of the composite image.

4.3.3.3 Multiplanar Volume Reformat

Instead of taking the highest attenuation in a pixel along a given ray, an MPR image is formed by the mean CT attenuation. It not routinely used by our group for image review, as the vascular images are more commonly confounded by overlap from adjacent bone or opacified venous structures. The slab can be reconstructed at an arbitrary slab thickness, as in an MIP; here the vertebrobasilar system is shown in Fig. 4.14. Unlike SSD and VR techniques, 2D MPR reconstructions do not obscure partially occlusive thrombus.

Figure 4.13

a Coronal and sagittal MIP images demonstrate an apparent aneurysm of the callosomarginal artery (arrows). b Axial contrast-enhanced source image confirms the presence of a density in the location shown on the MIP images (arrow). c Comparison with noncontrast image at the same location demonstrates calcification of the falx, there is no aneurysm (arrow)
4.3.3.4 Curved Reformat

To quickly delineate and review the entire course of a long, tortuous vessel, such as the carotid or vertebral arteries, a curved reformatted image is helpful. Here, the vessel is traced along its course, with the user selecting the pixels to display on consecutive axial images. The resultant reconstructed image is displayed in a 2D format. The process of creating curved reformatted (CR) images is the most time intensive of any of the rendering techniques discussed here. It is also subject to interpretative error, analogous to that of a conventional angiogram; hence, two orthogonal views are required to accurately screen for vascular stenosis (which may be elliptical, rather than spherical in nature), as shown in Fig. 4.15. CR is especially useful for quickly screening for dissection, of both the cervical internal carotid artery and the vertebral artery at the skull base, provided these reconstructions are included with the reconstructed image set, demonstrated in Fig. 4.16.

4.3.3.5 Shaded Surface Display

This is a thresholding technique to display all pixels with attenuation values greater than a user-specified Hounsfield threshold. Bone, opacified vessels, and calcification will be captured with a threshold of 80–100 HU; however, most parenchymal structures will be excluded [74]. SSD is not as useful in acute stroke evaluation, as only pixels on the surface are displayed. This limits evaluation for partially occlusive thrombus. It has been most successfully used for surgical planning, such as for paraclinoid aneurysms.

4.3.3.6 Volume Rendering

With VR, unlike with SSD, nonsurface pixels are included in the dataset [75]. This is advantageous, in that, using thresholding, layers of the vessel can be “peeled” away or made transparent, so as to demonstrate underlying structures, as shown in Fig. 4.17. To date, however, there have been no convincing studies demonstrating the efficacy of VR in evaluation for acute stroke. This is likely due to the intensive user interaction required to set user-defined parameters of opacity, window, and level to create diagnostically appropriate VR projections.
Figure 4.15
Occlusion of left internal carotid artery (LICA). Early and delayed axial images confirm the occlusion. Curved reformatted images of the LICA display the occlusion. Included are AP and lateral CR images of a normal LICA for comparison.
Figure 4.16
Left vertebral artery (LVA) dissection on both axial images and as shown on a curved reformatted image.
Figure 4.17

a Axial noncontrast and CTA source images with a subarachnoid hemorrhage (SAH) and the suggestion of an underlying top of carotid aneurysm. b Curved reformat, axial MIP, and volume rendered image confirm aneurysm at the top of the right carotid artery.
4.4 CTA Protocol for Acute Stroke

Multislice scanners allow flexibility in designing a rapid, efficient CTA protocol for acute stroke. The intracranial and extracranial vasculature can be imaged in less than 60 s. Even with the rapid scanning protocols designed for acute stroke, it remains necessary to tailor the acquisition, so that the images provide optimal visualization not only of an occlusive circle-of-Willis thrombus, but also of the possible origin or source of this thrombus. Because detection of an acute intracranial embolus is our first priority, the initial phase of our protocol is focused at the circle of Willis. By acquiring the intracranial CTA source images (also known as: “whole brain perfused blood volume images” – see Chapter 5) first, we simultaneously acquire the best possible CTA images of the circle of Willis. These images have the least venous contamination, as well as the most uniform arterial opacification, which is especially critical in the region of the cavernous sinus, Fig. 4.18.

Subsequently, CTA acquisition of the neck is performed, to evaluate for carotid stenosis as a possible source of emboli. In this “neck” phase of our protocol, it is not detrimental if venous opacification is present. Importantly, however, delayed scanning through the arch typically permits contrast that has pooled in the subclavian or brachiocephalic veins to be cleared prior to imaging, decreasing the resultant streak artifact from highly concentrated contrast in these vessels; subsequently, the origin of the great vessels can be more clearly evaluated. Figure 4.19 shows how streak at the origin by early scanning can result in misinterpretation.

After acquiring the CTA of the neck, single slab quantitative cine CT perfusion can additionally be performed, depending on the clinical need for these data in acute stroke triage. This is discussed at length in Chapter 5, along with specific scanning parameters. A sample stroke CTA protocol, optimized for the GE 16-slice scanner, is, however, provided for reference in Table 4.5. Using the aforementioned techniques for contrast reduction, including saline bolus chaser and bolus shaping, it is increasingly possible to obtain maximal uniform arterial enhancement—and hence diagnostic CTA images—without compromising the contrast dosage required for the addition of CT cine perfusion.
Figure 4.19
Dense contrast results in streak artifact in the right innominate vein and accounts for apparent nonopacification of the right common carotid artery (RCCA). On an axial image superior to the apparent obstruction, there is uniform enhancement.
4.5 Accuracy and Clinical Utility of CTA in Acute Stroke

4.5.1 Optimal Image Review

As noted above, review of both “thick” reformatted axial source images and 2D/3D reconstructions is required for appropriate stroke CTA interpretation. Indeed, given the enormous number of thin slice axial sections required for 3D reconstruction, a systematic, practical approach to image review is required.

Initially, the axial 5-mm-thick noncontrast CT images are reviewed, primarily to exclude hemorrhage – an absolute contraindication to thrombolysis – but also to assess for parenchymal hypodensity (a strong relative contraindication to thrombolysis, if more than one-third of a vascular territory is involved). Care is taken to perform image review using narrow “window width” and “center level” display settings, so as to maximize the subtle difference between gray matter and white matter attenuation – thereby maximizing the detection of subtle, edematous, hypodense ischemic regions, as discussed at length in Chapter 3 [70].

Subsequently, the 2.5- to 3-mm-“thick” axial CTA source images are reviewed, typically at the scanner console. Large, proximal circle-of-Willis occlusions, such as top-of-carotid T-lesions, are easily recognizable, even from the axial source images, and the “stroke team” can be immediately activated. Simultaneously, at most modern scanner consoles, the CTA dataset can be reformatted into “thin” (1.25 mm or less) slices, and quickly reconstructed into axial, coronal, and sagittal MIP images, facilitating more sensitive detection of secondary and tertiary branch occlusions. Even more detailed reconstructions, including curved reformat of the entire neurovascular system, from arch to vertex, can be created offline on a 3D workstation by a dedicated technologist.

4.5.2 Role of CTA in Acute Stroke

A major clinical role of CTA in acute stroke management remains the exclusion of unnecessary i.a. thrombolytic therapy in patients presenting with acute embolic stroke, but who do not have large vessel occlusions amenable to thrombolysis. Such occlusive stroke mimics include, but are not limited to, small vessel strokes (e.g., lacunar infarcts), TIAs, migraine headaches, seizures, and hypoglycemic events. CTA offers a convenient solution to the problem of efficiently diagnosing primary, secondary, and – increasingly – tertiary branch levels of intracranial vascular occlusion, prior to the initiation of thrombolytic therapy.

### Table 4.5: Sample acute CTA/CTP protocol for GE 16-slice scanner. (CTA CT angiography, CTP CT perfusion, FOV field of view, ICA internal carotid artery, NCCT noncontrast CT)

<table>
<thead>
<tr>
<th>Procedure</th>
<th>Contrast Range</th>
<th>Slice thickness (mm)</th>
<th>Image spacing (mm)</th>
<th>Table feed (mm/s)</th>
<th>Pitch</th>
<th>Tube voltage (kV)</th>
<th>Tube current (mA)</th>
<th>Rotation time (s)</th>
<th>Scan FOV</th>
<th>Display FOV</th>
</tr>
</thead>
<tbody>
<tr>
<td>NCCT</td>
<td>None</td>
<td>C1 to vertex</td>
<td>2.5</td>
<td>2.5</td>
<td>5.62</td>
<td>0.562</td>
<td>140</td>
<td>220</td>
<td>0.5</td>
<td>Head 22</td>
</tr>
<tr>
<td>CTA Head</td>
<td>4 ml/s for 40 ml, 25 s delay</td>
<td>C1 to vertex</td>
<td>2.5</td>
<td>2.5</td>
<td>5.62</td>
<td>0.562</td>
<td>140</td>
<td>200</td>
<td>0.5</td>
<td>Head 22</td>
</tr>
<tr>
<td>CTA Neck</td>
<td>0.8 ml/s for 30 ml</td>
<td>Arch to C1</td>
<td>2.5</td>
<td>2.5</td>
<td>5.62</td>
<td>0.562</td>
<td>140</td>
<td>250</td>
<td>0.5</td>
<td>Large 22</td>
</tr>
<tr>
<td>Cine CTP</td>
<td>7 ml/s for 40 ml, 5 s delay</td>
<td>Top of ICA</td>
<td>5</td>
<td>None, cine</td>
<td>None</td>
<td>N/A</td>
<td>80</td>
<td>200</td>
<td>1</td>
<td>Head 22</td>
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When considering the possible need for thrombolysis, assessment of Circle-of-Willis anatomy, as well as of collateral flow distal to an occlusion, was demonstrated on CTA as early as in 1998 [25]. Indeed, Kucinski et al. [80] recently showed that collateral circulation is an independent predictor for outcome in acute ischemic stroke. Associated lesions in the setting of an acute intracranial vessel occlusion, such as aneurysms, which would preclude thrombolytic therapy, although rare, can also be demonstrated by CTA [21], as shown in Figs. 4.20 and 4.21.

CTA has been shown to be highly accurate in delineating the presence and extent of intracranial thrombus, Fig. 4.22. In a study of 44 consecutive intra-arterial thrombolysis candidates who underwent both CTA and the “gold standard” of catheter arteriography, CTA demonstrated a sensitivity and specificity of 98.4% and 98.1%, respectively, for the detection of proximal large vessel thrombus [76].

The degree and level of occlusion have been shown to be important factors in planning acute stroke therapy. In early studies, intravenous rt-PA for clot lysis was more likely to be effective in secondary and tertiary MCA branch occlusions than in larger, more proximal occlusions [6, 81]. In another study using CTA, there was little benefit from intravenous rt-PA when there was poor collateralization, autolyzed thrombi, or proximal “top of carotid” saddle emboli [82]. In still other studies, “top of ICA” carotid terminus occlusion was demonstrated to be a better predictor of fatal outcome than an admission une-nhanced CT showing greater than one-third MCA territory hypodensity [83, 84].

Moreover, the source images from the CTA dataset, assuming an approximate steady-state level of contrast enhancement during scan acquisition, are intrinsically blood volume weighted. Like DWI, these images can be used to determine tissue with a high likelihood of infarction in the absence of early, complete recanalization [77]. This topic is discussed more fully in Chapter 5.

Unlike unenhanced (or perhaps even enhanced) MRA, CTA is an anatomic imaging technique, and is therefore not highly likely to yield false-positive results for occlusion due to slow flow or artifact. Carotid sonography is also known to be inaccurate in distinguishing a true carotid occlusion from a hairline lumen. Indeed, accuracy for distinguishing hairline residual lumen from total occlusion of the internal carotid artery using single-slice helical CTA was found to be excellent, compared to a gold standard of catheter arteriography [78]. Accuracy rates were 95%
Stroke CT Angiography (CTA)

CTA has also been used to perform serial monitoring of patients with proven internal carotid artery occlusions. Surprisingly, spontaneous recanalization of an occluded internal carotid artery has been demonstrated [79]. In such cases, serial catheter angiography would have been prohibitive in terms of time, cost, and risk of complications.

Figure 4.21
Acute stroke in a patient with endocarditis. The unenhanced and corresponding diffusion-weighted image (DWI) demonstrates acute infarct in the left parietal/temporal lobe. Evaluation with CTA MIP images clearly demonstrates a bilobed aneurysm (arrow), which is confirmed on the catheter cerebral angiogram.

and 80% for two independent raters, with no statistically significant difference in accuracy between the two readers.

CTA has also been used to perform serial monitoring of patients with proven internal carotid artery occlusions. Surprisingly, spontaneous recanalization of an occluded internal carotid artery has been demonstrated [79]. In such cases, serial catheter angiography would have been prohibitive in terms of time, cost, and risk of complications.

Neck CTA can often elucidate the etiology of an intracranial occlusion. In older patients, carotid plaque can be accurately assessed by CTA. In a study of 82 patients, CTA was shown to be comparable to, and at times more accurate than, carotid sonography in determining the degree of stenosis [89]. Agreement in determination of vessel abnormality, plaque morphology, and ulceration between the two methodologies was reported to be 82%, 89%, and 96%, respectively. Moreover, 11 tandem lesions reported by
CTA were not detected by sonography. When compared to a surgical gold standard, using North American Symptomatic Carotid Endarterectomy Trial (NASCET) criteria, degree of stenosis was correctly shown on CTA 89% of the time, as opposed to 83% with sonography. CTA did not overestimate stenosis.

In younger patients, dissection of the carotid and/or vertebral arteries must be included in the differential diagnosis of acute stroke [85–87]. In 1999, Oelerich [88] reported satisfactory results with MRA.

**Figure 4.22**
CT angiography (CTA) maximum intensity projection (MIPs) images with four different orientations demonstrate occlusion of the M1 segment of the right middle cerebral artery.
in demonstrating craniocervical artery dissection. Although MR can demonstrate intramural hematoma, this is typically time-dependent, and, as noted previously, CTA is less prone to error from slow flow or other artifacts, and can provide a true anatomical image of the intravascular lumen. CTA can also provide information regarding the false lumen, flap, and extra-luminal structures – information not necessarily provided even by catheter arteriography.

4.6 Future Directions

As MDCT scanner technology advances, both acquisition speed and extent of z-direction coverage continue to improve. These advances should make possible the use of lower total contrast volumes and more rapid scanning, without loss of image quality or significantly increased radiation dose.

Increased coverage will not only be of benefit for cine CT perfusion studies, which are limited to the detector size of the given MDCT scanner (see Chapter 5), but it may also facilitate more complete neurovascular coverage than is currently obtainable, namely from the aortic arch to the vertex. Specifically, it may be possible to extend the neck portion of the CTA acquisition inferiorly to include the left atrium and left ventricle, so as to assess for endoluminal thrombus in stroke patients who present with atrial fibrillation as a possible embolic source. If successful, this technique has the potential to obviate the need for more invasive trans-esophageal echocardiography (TEE), which carries the rare but life-threatening risk of esophageal rupture [90]. CTA could therefore prove to be of greater value than TEE in acute stroke patients with atrial fibrillation [91].

Additionally, faster gantry rotation speeds may make possible “cine CTA fluoroscopy,” with which difficult-to-detect lesions, such as cavernous sinus aneurysms or arterial venous malformations, could be visualized with greater temporal resolution during arterial, capillary, and venous phases of contrast enhancement. Dedicated studies could display a temporally parsed image, which would mirror the temporal resolution of the standard cerebral catheter arteriography.

4.7 Conclusion

CTA is a highly accurate tool for defining the level of intracranial vascular occlusion in patients presenting within 6 h of embolic stroke onset. CTA data will likely prove valuable in the rapid triage of such patients to appropriate therapy, including i.v. and i.a. thrombolytic treatment. For the evaluation of acute stroke patients, CTA is fast, simple, accurate, and convenient. Clinically relevant information regarding both vascular patency and parenchymal perfusion can be obtained during the first pass of a single contrast bolus in CTA.

Compared with single detector row CTA, there are fewer restrictions on designing a scanning protocol with the newer multislice CT scanners, especially those employing 16 channels or more – facilitating increased longitudinal coverage within a shorter scanning time. Complete imaging of the arterial tree – from arch to vertex – to evaluate for both acute occlusions and chronic stenoses – can be accomplished during peak arterial opacification. Contrast dose can be minimized with the use of a saline bolus and multiphasic injection. CTA not only compares favorably with catheter angiography for detection of acute large vessel occlusion, but can also provide information regarding collateral flow and lesion extent. By including neck CTA for acute stroke evaluation, embolic source and plaque burden at the origins of the internal carotid arteries can additionally be determined. Other stroke etiologies, such as dissection, can also be displayed. Finally, CTA of the neck affords an opportunity to plan catheter angiography, decreasing the time and contrast load required for i.a. thrombolysis, should it be required.

References


S. Sheikh · R.G. González · M.H. Lev
### Stroke CT Angiography (CTA)

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<td>34.</td>
<td>NRC Committee on the Biological Effects of Ionizing Radiation BV (1990) In: Health effects of exposure to low levels of ionizing radiation. National Academy Press, Washington DC</td>
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</tbody>
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Chapter 4