The advent of CT colonography (CTC) (or, virtual colonoscopy) in 1994 was made possible by the development of spiral CT technology, which provides a volumetric coverage of a cleansed and air-distended colon within a single breath-hold (Vining et al. 1994). The introduction of multidetector row computed tomographic (MDCT) scanners in late 1998 opened a new era for CT in general and CTC in particular (Berland and Smith 1998). The use of multiple detector arrays along the z-axis offers substantial benefits related to anatomic coverage, scanning time and longitudinal spatial resolution compared with single-slice spiral CT (SSCT) (Beaulieu et al. 1998; Fenlon et al. 1999; Hara et al. 2001).

In principle, using similar parameters on both SSCT and MDCT results in wider anatomic coverage and faster scanning time with MDCT. On the other hand, MDCT provides sub-millimeter collimation, improves z-axis resolution and generates isotropic voxels, thus resulting in better image quality of reformatted planes as well as three-dimensional reconstructions. The drawback is represented by data explosion, with generation of over 1000 images per scan per patient, making data viewing and analysis (“data workflow”) a key issue to be solved within the next period of time (Sherbondy et al. 2005).

Another major issue is represented by dose delivery, usually higher with CTC compared with a standard abdominal CT study due to routine use of prone and supine scans (Chen et al. 1999). Radiation exposure has also substantially increased over the past few years due to the widespread use of thinner collimations and the consequent increase of tube current setting in order to reduce image noise. Low or ultra-low dose MDCT protocols together with new automatic dose modulation software may help in solving this problem representing a crucial issue for proposing VC as a screening method for colonic polyp in healthy subjects (Iannaccone et al. 2003a).

A further variable in CTC scanning parameters is due to technological differences among CT vendors and MDCT generations. It does exist quite a large experience on investigations using 4-slice MDCT scanners, but, on the other hand, there are only few manuscripts on 16-slice MDCT and almost no experience, at the time of this paper, on new 64-slice scanners. As technology continues to advance, there will be a continuing need to reassess the relative tradeoffs between scan width, image noise, patient dose, image artefacts, breath-hold times, and the number of reconstructed images to be viewed and archived.

As a consequence of what is discussed above, the “panorama” of technical approaches is expanding, offering a wide spectrum of different possibilities. For these reasons, the answer to the question “which are the right parameters to be used for CTC?” may be puzzling.
6.2 Collimation

Collimation is the parameter that – more than the others – has dramatically changed since the development of CTC in parallel with continuous evolution of scanner technology.

“Thin” collimation is a mandatory pre-requisite for a CTC study. The question is how thin is a thin collimation? The answer may be either political or technical. In fact, collimation is strictly related to the size of the target lesion. Since on CT you are not able to detect a lesion smaller than the effective slice thickness due to partial volume effect, the size of the ideal target lesion should be defined. This should avoid the risk of searching for the thinnest possible collimation as soon as a new equipment becomes clinically available.

On a technical point of view, a consensus was achieved (Barish et al. 2005). If considering SSCT, the maximum accepted collimation is 5 mm with overlapped image reconstruction (usually set at 3 mm) (Hara et al. 1997a). In the first CTC studies, due to limited tube capacity, two or three consecutive breath-holds were needed to cover the entire colon. With the progress in tube technology a single breath-hold of 40–50 s was made possible on SSCT scanners (Taylor et al. 2003).

One of the major benefits of MDCT was represented by the possibility of further reducing the collimation compared with SSCT. Although some authors, at the beginning of the era of MDCT, still proposed 5 mm as an optimal collimation due to acquisition time, breathing artefacts as well as image workload (Hara et al. 2001), it is now widely accepted that a collimation no thicker than 3 mm is mandatory (Barish et al. 2005).

A major drawback of the use of thin collimation is the increased current tube setting, necessary in order to reduce image noise and to maintain an acceptable image quality. This is the reason why several researchers have performed investigations on the potential benefits of thin collimation protocols (Whiting et al. 2000; Fletcher et al. 2000; Rogalla and Meiri 2001; Gillams and Lees 2002; Macari et al. 2002; Taylor et al. 2003; Wessling et al. 2003) looking for the clinical impact of thin collimation protocols in terms of polyp detection and characterization.

Several in-vitro evaluations were reported. Although phantom models have inherent limitations represented by the ideal conditions of the study design (a true colon is a “moving” organ, with peristaltic motion as well as motion related to the anatomic location, partly intra-peritoneal and partly retroperitoneal) they may provide useful data to be tested on patients.

In a personal experience (Laghi et al. 2003), we built a phantom model with 12 lesions ranging in size between 3.2 mm and 12 mm; lesion morphology simulated sessile polyps and flat and depressed lesions. Different scanning protocols were compared based on 1.0-mm, 2.5-mm and 5-mm collimations with effective slice thicknesses of respectively 1.25 mm, 3 mm and 5 mm. Results showed the best performance when using 1.0-mm collimation protocol (no lesions missed), although a statistically significant difference was observed only among 1.0/2.5-mm protocols in comparison with 5.0-mm protocol. If only lesions larger than 10 mm were considered no differences were observed among the three different protocols (Fig. 6.1).

In another study (Wessling et al. 2003), a phantom with simulated polypoid lesions ranging in size between 2 mm and 12 mm was built. Different protocols were compared with collimation ranging between 1.25 mm and 5 mm. Results showed no significant differences for detection of lesions larger than 10 mm, but for lesions smaller than 6 mm a clear benefit was observed when a thin collimation protocol (1.25 mm) had been used.

Finally, in another in-vitro experience (Taylor et al. 2003), where a human colonic specimen of a patient with familiar adenomatous polyp syndrome after total colectomy was used as a phantom, collimation had a significant effect on the polyp detection rate which, when unadjusted for size, was 50% higher at 1.25 mm collimation than at 2.5 mm. This effect was most marked for polyps less than 5 mm, for which a small but significant improvement in detection also occurred with increased tube current. For polyps 5 mm or larger the benefit of decreased collimation was significant resulting in 7% improved detection rate. The use of thin collimation is also associated with a possible increase of specificity, especially of small lesions, since the detection of tiny air bubbles might be easier.

Although thin collimation protocols may provide benefits in terms of detection of small polypoid lesions, the application to patients is limited by technical restrictions of four-slice MDCT. In fact, with four-slice MDCT a compromise between scanning time and collimation is necessary since the use of 1 mm is associated with scanning time over 30 s (Fig. 6.2). On the other hand, the increased radiation exposure as compared with thicker collimation