INTRODUCTION

In the evaluation of patients with suspected coronary artery disease (CAD), the role of non-invasive imaging has increased exponentially over the past decades, particularly in patients with an intermediate likelihood of CAD. Non-invasive imaging plays an important role in risk stratification and selection of further treatment strategies ***(Underwood et al., 2004)***.

Traditionally, the detection of CAD by non-invasive imaging was based on assessment of the hemodynamic significance of the stenoses through visualization of inducible ischemia. For this purpose, myocardial perfusion imaging (MPI) with gated single-photon emission computed tomography (SPECT) has been used extensively ***(Underwood et al., 2004).***

More recently, multi-slice computed tomography (MSCT) has been proposed as an alternative imaging modality for evaluation of patients with suspected CAD ***(Leber et al., 2005; Leschka et al., 2005).***

With the recently introduced 64-slice MSCT, high sensitivity and specificity for the detection of significant (>50% luminal narrowing) stenoses have been reported ***(Mollet et al., 2005; Raff et al., 2005).***

However, because MSCT visualizes coronary artery stenoses directly, rather than the hemodynamic significance of the lesions, it is important to recognize that, unlike MPI, the technique identifies atherosclerosis rather than ischemia ***(Pugliese et al., 2006; Ropers et al., 2006).***

Recent advances in multi-detector-row computed tomography (MDCT) technology have continuously improved the quality of non-invasive coronary artery imaging. As a result, various studies have demonstrated a high accuracy of coronary angiography with 64-slice CT for the diagnosis of CAD ***(Ehara, 2006; Nikolaou, 2006).***

In particular, the high negative predictive value with an overall sensitivity of 96.4% and specificity of 97.5% for the detection of significant coronary stenoses has made non-invasive coronary angiography using 64-slice CT a modality that allows significant coronary stenoses to be reliably excluded ***(Fox, 2006)***.

Consequently, the Task Force on the Management of Stable Angina Pectoris of the European Society of Cardiology has recently recommended in their guidelines that CT coronary angiography be performed in patients with stable angina who have a low pre-test probability of CAD, and an inconclusive exercise electrocardiogram (ECG) or stress imaging test ***(Fox, 2006).***

Conventional invasive coronary angiography (CAG) has been the "gold standard" for the diagnosis of coronary artery disease. However, CAG shows only luminal stenosis and the extent of coronary atherosclerosis. It does not provide information on plaque composition. Non-invasive coronary MDCT can visualize the coronary artery lumen, artery wall, and atherosclerotic plaque; even the lipid pool can be visualized, which is fibrous, calcified, and heavily laden with cholesterol ***(Leber et al., 2004; Schroeder et al., 2007).***

In 2 studies, it was confirmed that contrast-enhanced MDCT permits accurate identification of coronary plaques, and that CT density values within plaques reflect echogenicity and plaque composition ***(Leber et al., 2004; Schroeder et al., 2007).***

Little data exists on accuracy of MSCT to assess the lesion severity (% area stenosis & lesion length) ***(Wehrschuetz et al., 2010).***

The value of MSCT in guiding Interventional Cardiologists about the Stent Length & Diameter is still unclear with no available data ***(Wehrschuetz et al., 2010).***

AIM OF THE WORK

To evaluate the role of MSCT in assessing lesion severity, diameter & length, compared to conventional coronary angiography QCA, & hence its role in defining reference value for stent diameter & length before PCI.

Chapter 1

NATURAL HISTORY OF MSCT

CT overview:

Computed tomography (CT) was introduced in the early 1970s and has revolutionized not only diagnostic radiology, but also the entire practice of medicine. In 1979, ***G. Hounsfield and A.M. Cormack*** received the Nobel Prize for their significant contributions to the development of computed axial tomography. Using computer reconstruction techniques, Hounsfield demonstrated that the internal structures of an object could be reconstructed based on the attenuation pattern of an X-ray beam that had passed through the object at different angles. In 1971, Hounsfield had constructed the first CT scanner that could image the brain ***(Ohnesorge, 2007)***.

The basic principle of CT is that a fan-shaped, thin X-ray beam passes through the body at many angles to allow for cross-sectional images. The corresponding X-ray transmission measurements are collected by a detector array. Information entering the detector array are used to produce thin sections. The data recorded by the detectors are digitized into picture elements (pixels) with known dimensions. The gray-scale information contained in each individual pixel is reconstructed according to the attenuation of the X-ray beam along its path using a standardized technique termed "filtered back projection". Gray-scale values for pixels within the reconstructed tomogram are defined with reference to the value for water and are called "Hounsfield units", or simply "CT numbers" ***(Ohnesorge, 2007)***.

Since CT uses X-ray absorption to create images, the differences in the image brightness at any point will depend on physical density and the presence of atoms with different atomic numbers. The absorption of the X-ray beam by different atoms will cause differences in CT brightness on the resulting image. Blood and soft tissue (in the absence of vascular contrast enhancement) have similar density as they consist of similar proportions of the same atoms (hydrogen, oxygen, carbon). Bone has an abundance of calcium. Fat has an abundance of hydrogen. Lung contains air which is of extremely low physical density. The higher the density, the brighter the structure on CT ***(Ohnesorge, 2007)***.

Calcium is bright white, air is black, and muscle or blood is gray. Computed tomography, therefore, can distinguish blood from air, fat and bone but not readily from muscle or other soft tissue. The densities of blood, myocardium, thrombus, and fibrous tissues are so similar in their CT number that non-enhanced CT cannot distinguish these structures ***(Budoff, 2006)***.

The distinction of blood and soft tissue (such as the left ventricle, where there is no air or fat to act as a natural contrast agent) requires injection of contrast with CT. Similarly, distinguishing the lumen and wall of the coronary artery also requires contrast enhancement. The accentuated absorption of X-rays by elements of high atomic number like calcium and iodine allows excellent visualization of small amounts of coronary calcium as well as the contrast-enhanced lumina of medium-size coronary arteries ***(Budoff, 2006)***.

Evolution of Spiral CT:

In the early 1990s, the introduction of spiral CT constituted a further evolutionary step in the development and ongoing refinement of CT imaging techniques ***(Kalender et al., 1990)***.

Until then, the examination volume had to be covered by subsequent axial scans in a "step-and-shot" mode, the so-called "sequence scan" technique. Consequently, axial scanning required long examination times because of the inter-scan delays necessary to move the table incrementally from on scan position to the next, and it was prone to misregisteration of anatomical details due to the potential movement of the relevant anatomical structures between two scans, e.g. by patient motion, breathing or swallowing ***(Rubin et al., 1995)***.

With spiral CT, the patient table is continuously translated while scan data are acquired, the X ray forms a helix across the longitudinal plane of the patient (hence the name helical or spiral CT). The prerequisite for the success of spiral scanning was the introduction of slip-ring gantries, which eliminated the need to rewind the gantry after each rotation and enabled continuous data acquisition during multiple rotations. For the first time, volume data could be acquired without the danger of misregistration or double-registration of anatomical details ***(Rubin et al., 1995)***.

Images could be reconstructed at any position along the patient axis (longitudinal axis), and overlapping image reconstruction could be used to improve longitudinal resolution. Volume data became the very basis for applications such as CT angiography, which has revolutionized non-invasive assessment of vascular disease ***(Rubin et al., 1995)***.

The ability to acquire volume data also paved the way for the development of 3D image-processing techniques, such as multi-planar reformations (MPR), maximum intensity projections (MIP), and volume-rendering techniques (VRT). These have become vital components of CT imaging today ***(Kalender, 1995)***.

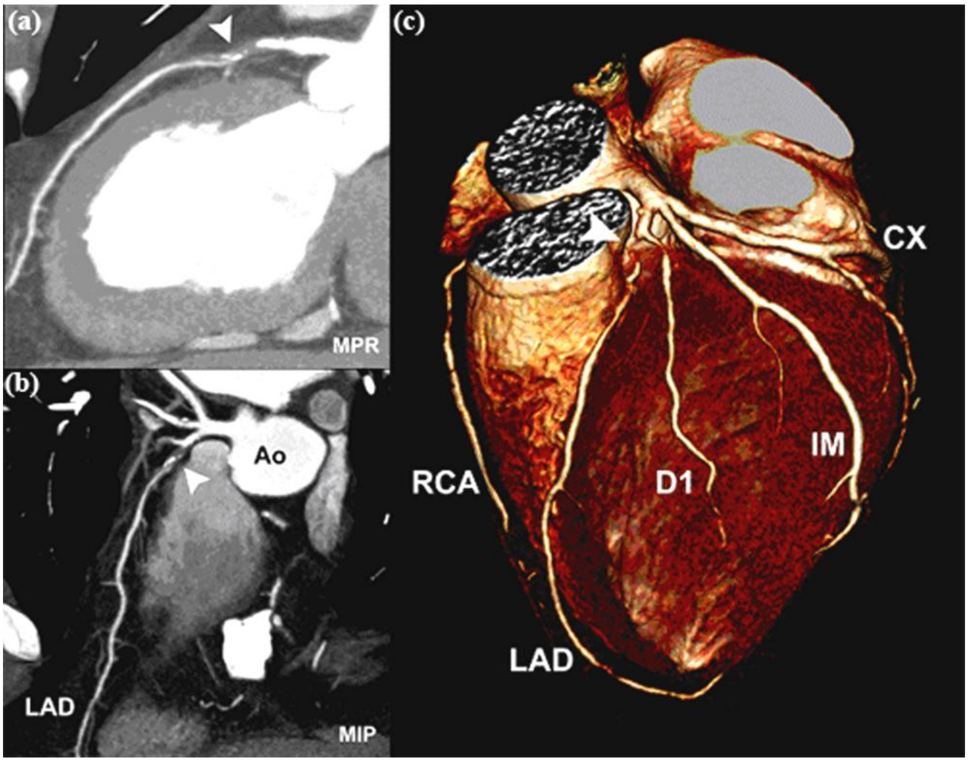

Figure (1): MSCT imaging techniques: (a) The curved multiplanar projection (MPR), (b) the maximum intensity projection (MIP) and (c) the colored-volume-rendered image (Ao=aorta, LAD=left anterior descending artery, CX=left circumflex artery, D1=first diagonal, IM= intermediate branch, RCA=right coronary artery) ***(Annick and Pim 2009).***

Ideally, volume data are of high spatial resolution and isotropic in nature, i.e., the data element (voxel) of each image is of equal dimensions in all three spatial axes, and forms the basis for image display in arbitrarily oriented imaging planes ***(Kalender, 1995)***.

For most clinical scenarios, however, single-slice spiral CT with 1-s gantry rotation time is unable to fulfill these prerequisites. To avoid motion artifacts and to optimally use the contrast bolus, spiral CT body examinations need to be completed within a certain time frame of, ordinarily, one patient breath-hold (25-30 s). If a large scan range, such as the entire thorax or abdomen (30 cm), has to be covered with single-slice spiral CT within a single breath-hold, a thick collimation (slice thickness) of 5-8 mm must be used. While the in-plane resolution of a CT image depends on the system geometry and on the reconstruction kernel selected by the user, the longitudinal (z-plane) resolution is determined by the collimated slice width and the spiral interpolation algorithm. A thick collimation of 5-8 mm results in a considerable mismatch between the longitudinal resolution and the in-plane resolution, which is usually 0.5-0.7 mm depending on the reconstruction kernel. Thus, with single-slice spiral CT, the ideal of isotropic resolution can only be achieved for very limited scan ranges ***(Kalender, 1995)***.

Development of Multi-Slice CT:

Strategies to achieve more substantial volume coverage with improved longitudinal resolution have necessitated the simultaneous acquisition of more than one slice at a time and a reduction of the gantry rotation time. Interestingly, the very first medical CT scanners were 2-slice systems, such as the EMI head scanner, introduced in 1972, or the Siemens SIRETOM, introduced in 1974. With the advent of whole body fan-beam CT systems for general radiology, 2-slice acquisition was no longer used ***(Flohr and Ohnesorge, 2007)***.

The simultaneous acquisition of N number of slices results in an N-fold increase in speed of the scan if all other parameters, such as slice thickness and gantry rotation speed, are unchanged. This increased performance of multi-slice CT compared to single-slice CT allowed for the optimization of a variety of clinical protocols. The examination time for standard protocols could be significantly reduced, which proved to be of immediate clinical benefit for the quick and comprehensive assessment of trauma victims and non-cooperative patients ***(Rubin et al., 2001)***.

Alternatively, the scan range that could be covered within a certain scan time was extended by a factor of N, which is relevant for oncological staging or for CT angiography with extended coverage, for example, of the lower extremities ***(Rubin et al., 2001).***

In 1998, all major CT manufacturers introduced MSCT systems, which typically offered simultaneous acquisition of 4 slices at a rotation time of down to 0.5 s. This was a considerable improvement in scan speed and longitudinal resolution and offered better utilization of the available X-ray power ***(Klingenbeck et al., 1999)***.

The most important clinical benefit, however, proved to be the ability to scan a given anatomic volume within a given scan time with substantially reduced slice width, at 4 times increased longitudinal resolution. This way, for many clinical applications, the goal of isotropic resolution was within reach with 4-slice CT systems. Examinations of the entire thorax or abdomen could now routinely be performed with a slice width of 1 mm or 1.25 mm. Multi-slice CT also expanded into areas previously considered beyond the scope of conventional spiral CT scanners, such as routine vascular diagnosis, high resolution low dose CT of the lung, virtual CT colonography ***(Flohr and Ohnesorge, 2007)***.

Despite of all of these advances, these 4-slice CT systems were considered defective in several aspects; true isotropic resolution could not be achieved because the longitudinal resolution was 1 mm which didn't match the 0.5-0.7 in plane resolution (in axial slices), scanning large volumes such as CT angiography of the lower limbs necessitated even thicker slice thickness to perform the scan in a reasonable time which further more deranged the isotropic resolution ***(Rubin et al., 2001)***.

As a next step, the introduction of an 8-slice CT system in 2000 enabled shorter scan times, but did not yet provide improved longitudinal resolution with a slice thickness of 1.25 mm ***(Flohr and Ohnesorge, 2007)***.

With the introduction of the 16-slice CT systems it became possible to routinely acquire true isotropic submillimeter volumes. The improved spatial resolution goes hand in hand with the considerably reduced scan times that enable high-quality examinations. Clinical practice with these scanners suggested the potential of 16-slice CT angiography to replace invasive carotid angiography. Entire thorax examination with sub-millimeter collimation required a scan time of approximately 11 seconds. This necessitated significantly shorter breath hold time, compared to previous scanners, with the ability to reliably and accurately diagnose pulmonary embolism ***(Schoepf et al., 2003)***.

The race for more slices is on-going. In 2004, all major CT manufacturers introduced the next generation of multi-slice CT systems, with 32, 40, and even 64 simultaneously acquired slices, which brought about a further leap in volume coverage speed. Whereas most of the scanners increase the number of acquired slices by increasing the number of the detector rows, some of the new scanners use additional refined z-sampling techniques with a periodic motion of the focal spot in the zdirection (z-flying focal spot). This so-called "double z" sampling technique can further enhance longitudinal resolution and image quality in clinical routine ***(Flohr and Ohnesorge, 2007)***.

These developments were quickly recognized as revolutionary improvements that would eventually enable users to do real isotropic 3D imaging. Consequently, all vendors pushed towards more and more slices, turning the number of slices into the most important performance characteristic of a CT scanner. Interestingly, analogous to "Moore’s Law" in the computer industry, the increase in the number of slices has been exponential, approximately doubling every 18 months ***(Flohr and Ohnesorge, 2007)***.

In 2008, ten years after the introduction of the 4-slice scanner, Toshiba has declared its most recent multi-slice CT scanner with the ability to acquire 320 slices with one gantry rotation.

Challenges for performing cardiac CT scanning:

The cardiac CT scanning represents a major challenge due to several imaging and physiological factors. The heart is a continuously moving organ (heart beats), in a continuously moving chamber (respiratory motions) ***(Ohnesorge et al., 2000).***

Secondary to the introduction of scanners acquiring more slices per rotation, cardiac imaging necessitated also faster gantry rotation speeds to achieve adequate temporal resolution for visualization of the cardiac anatomy. The introduction of 4-slice CT with a gantry rotation time of 0.5 s and dedicated image reconstruction approaches represented a breakthrough for mechanical CT in cardiac imaging. The temporal resolution for the acquisition of an image was improved to 250 ms and less ***(Ohnesorge et al., 2000)***, sufficient for motion-free imaging of the heart in the mid- to end-diastolic phase at slow to moderate heart rates ***(Hong et al., 2001)***. With four simultaneously acquired slices, coverage of the entire heart volume with thin slices and ECG gating within a single breath-hold became feasible, enabling non-invasive visualization of the cardiac morphology and coronary arteries ***(Achenbach et al., 2000)***.

For ECG-gated coronary CT angiography, the impaired isotropic resolution, characteristic of those 4-slice scanners, constituted a challenge in assessing stents and segments with severe calcification due to partial volume effect due to insufficient longitudinal resolution ***(Nieman et al., 2001)***.

Additionally, in patients with relatively higher heart rate, the relatively insufficient temporal resolution of these scanners necessitated careful selection of separate reconstruction intervals for different coronary arteries ***(Kopp et al., 2001)***.

Lastly, to perform an adequate scan to cover the whole cardiac volume (approximately 12 cm), the breath hold time needed to be about 40 seconds with this 4-slice CT scanners which was almost impossible for many patients with manifest heart disease ***(Flohr and Ohnesorge, 2007)***.

As regards cardiac applications, the development of scanners with increased number of simultaneously acquired slices, was paralleled with appreciable increases in gantry rotation speed. These systems delivered an increased temporal resolution, in addition to the increased spatial resolution, due to increased gantry rotation speed down to 0.375 second ***(Nieman et al., 2002)***.