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REVIEWS

STATE-OF-THE-ART PAPERS

New Applications of Cardiac Computed Tomography

Dual-Energy, Spectral, and Molecular CT Imaging

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ABSTRACT

Computed tomography (CT) has evolved into a powerful diagnostic tool, and it is impossible to imagine current clinical practice without CT imaging. Because of its widespread availability, ease of clinical application, superb sensitivity for the detection of coronary artery disease, and noninvasive nature, CT has become a valuable tool within the armamentarium of cardiologists. In the past few years, numerous technological advances in CT have occurred, including dual-energy CT, spectral CT, and CT-based molecular imaging. By harnessing the advances in technology, cardiac CT has advanced beyond the mere evaluation of coronary stenosis to an imaging tool that permits accurate plaque characterization, assessment of myocardial perfusion, and even probing of molecular processes that are involved in coronary atherosclerosis. Novel innovations in CT contrast agents and pre-clinical spectral CT devices have paved the way for CT-based molecular imaging. (J Am Coll Cardiol Img 2015;8:710–23) © 2015 by the American College of Cardiology Foundation.

Since the advent of 64-detector row computed tomography (CT) in 2005, coronary computed tomographic angiography (CTA) has been demonstrated as a promising noninvasive technique for the evaluation of coronary artery stenosis (1,2). In the past few years, numerous technological advances in CT have occurred, including dual-energy computed tomography (DECT), spectral CT, and CT-based molecular imaging. Early studies of these methods have been largely promising and shown improved cardiac and coronary evaluation. An understanding of these advanced CT principles is required to fully appreciate the promise of the applicability of these technologies for the evaluation of patients with cardiac and coronary disease.

THE PRINCIPLES OF DECT

Improvements in CT, although rapid in recent years, are nevertheless constrained by the physical principles underlying this technology, which are a function of x-ray attenuation detected from multiple orientations around an imaged object. In a basic sense, these principles are generally 2-fold and include the photoelectric and Compton effects when considering x-ray photons within the diagnostic energy range

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(Figure 1). The former is highly dependent on the photon energy level and is related to the atomic number and photon energy level, whereas the latter is independent of the photon energy level but rather related to material density. For a proper grasp of the advances in DECT and spectral CT, a basic understanding of these principles is required.

The photoelectric effect is the ejection of an electron from the innermost shell of an atom (the K shell) by a photon with a greater energy than the binding energy of the K shell. As a result, the total energy of any incoming photon is absorbed (Figure 1). The binding energy of electrons in the K shell is material specific and is proportional to the atomic number (Z). For a given photon energy, the photoelectric effect scales on a magnitude order of Z^3. However, this does not imply that an element with $Z = 100$ yields an 8 times greater attenuation than a material with $Z = 50$. This can be explained by the spike in attenuation seen related to a maximal photoelectric effect (the K edge), which occurs when the photon energy level is just greater than the electron binding energy of the K shell of an atom. The K-edge value varies for each material and is higher with increasing atomic numbers. Paradoxically, the photoelectric effect is maximal at the K edge of the absorber and is reduced with increasing photon energy levels, inversely proportional to the photon energy cubed ($1/E^3$). Therefore, the probability of the photoelectric effect is dependent on both the atomic number and the photon energy level according to Z^3/E^3.

The Compton effect is the collision of photons with valence electrons of the outermost shell of an atom. In contrast to the photoelectric effect, the energy of the incoming photon is not totally absorbed, giving rise to photon scattering (Figure 1). Compton scattering is dependent on the density of electrons, and because all elements have approximately the same amount of electrons per unit mass, the atomic number is of less relevance for the occurrence of Compton scattering.

The principles of DECT are based largely on the photoelectric effect and can be achieved by exploiting the energy-dependent attenuation of materials when exposed to 2 different photon energy levels. These physical principles can be exploited for in vivo human imaging, because DECT is based on dissimilar tissue characteristics with respect to their energy-dependent x-ray attenuation. Subsequently, DECT enables the distinct differentiation between 2 basis materials (Figure 2). These materials can be chosen arbitrarily, as long as their K edges are sufficiently different (i.e., attenuation profiles), such as water and iodine. Any other material with an attenuation spectrum different than that of the chosen basis materials will be reflected as a combination of the 2 basis materials (Figure 3). As such, by exploiting differences in energy-related attenuation of tissues, DECT provides information about tissue composition that is unobtainable with conventional single-energy computed tomography (SECT).

The advantage of using different energy x-ray levels for decomposition of tissues has been known for a long time and was even mentioned by Hounsfield (3) in his original paper on CT 4 decades ago: "Two pictures are taken of the same slice, one at 100 kV and the other at 140 kV so that areas of high atomic numbers can be enhanced. Tests carried out to date have shown that iodine (Z = 53) can be readily distinguished from calcium (Z = 20)". However, this approach at that time was subject to technological limitations and was therefore abandoned.

ABBREVIATIONS AND ACRONYMS

CAC = coronary artery calcium

CTA = computed tomographic angiography

CT = computed tomography

DECT = dual-energy computed tomography

MPI = myocardial perfusion imaging

SECT = single-energy computed tomography

SPECT = single-photon emission computed tomography

VUE = virtual unenhanced

DECT METHODS

Although SECT is typically performed with polychromatic energy levels of photons set to 120 or 140 kVp, energy levels of photons with DECT are typically 80 and 140 kVp for the acquisition of low- and high-energy-dependent tissue attenuation profiles, respectively. The exploitation of 2 polychromatic energy spectra by DECT can be achieved by at least 3 different methods (Figure 4): 1) 2 x-ray source and detector pairs, with each source operating at a different tube voltage; 2) a single source-detector pair with an x-ray tube capable of rapidly switching between low and high tube potential or by switching tube potential between gantry positions; and 3) an x-ray source operating at constant tube voltage with a double-layer detector capable of differentiating between low- and high-energy photons.

CLINICAL APPLICATIONS OF DECT

MYOCARDIAL PERFUSION IMAGING. Compared with SECT, DECT may allow better tissue characterization and therefore enhanced visualization of myocardial perfusion defects, thus encouraging its use for ischemia assessment. Given the unique ability of DECT to allow differentiation of iodine attenuation characteristics when it is exposed to different photon energy levels, DECT allows the mapping of iodine distribution in the myocardium as a quantitative, albeit surrogate, marker for perfusion and blood volume (4) (Figure 5). There is an early body of evidence showing the clinical feasibility of a DECT myocardial