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Preliminary research on body composition measurement using X-ray phase contrast imaging

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ABSTRACT

Body composition measurement is of cardinal significance for medical and clinical applications. Currently, the dual-energy X-ray absorptiometry (DEXA) technique is widely applied for this measurement. In this study, we present a novel measurement method using the absorption and phase information obtained simultaneously from the X-ray grating-based interferometer (XGI). Rather than requiring two projection data sets with different X-ray energy spectra, with the proposed method, both the areal densities of the bone and the surrounding soft tissue can be acquired utilizing one projection data set. By using a human body phantom constructed to validate the proposed method, experimental results have shown that the compositions can be calculated with an improved accuracy comparing to the dual energy method, especially for the soft tissue measurement. Since the proposed method can be easily implemented on current XGI setup, it will greatly extend the applications of the XGI, and meanwhile has the potential to be an alternative to DEXA for human body composition measurement.

1. Introduction

Body composition measurement has great significance in medical and clinical applications and has been studied extensively since the middle of last century [1–4]. Dual-energy X-ray absorptiometry (DEXA), which is based on the differential X-ray attenuation by bone and soft tissue measured at two different photon energies, was primarily developed for the bone mineral density (BMD) measurement and soon extended to body composition measurement. Currently, by using instrument specific algorithms, three major components of human body (i.e. the fat tissue mass, lean tissue mass and the bone mineral mass) can be acquired with a single DEXA scan [1,2]. Furthermore, merits including low radiation dose (less than 1.5×10^{-2} mSv per scan), high precision (measurement precision, 1–2% for BMD and 2–6% for body-composition scans) and short scanning time (within 5 min for whole body scan) make the technique widely applicable [1,4].

DEXA requires two different X-ray energy spectra, which can be achieved either by switching the tube voltage between two potentials (Hologic models) or by employing a cerium filter to absorb the X-ray photons preferentially above the cerium absorption edge (GE Lunar and Stratec models) [4]. Depending on the X-ray generation models, different types of detectors should be used for data acquisition. For the GE Lunar and Stratec models, an energy resolving detector is required to obtain two images at different X-ray photon energies within one exposure. The Hologic models work with conventional detectors, but the body under observation should be kept static during two exposures to avoid the motion artifacts. Briefly speaking, the DEXA equipment requires some technical modifications comparing with the conventional radiographic device. A further deficiency of DEXA is that the measurement precision of the soft tissue density is generally less accurate (around 2–6%) than that of BMD (around 1–2%) [1]. The poorer reproducibility and accuracy in the estimate of the soft tissue density is due to the weaker absorption and smaller differences between mass attenuation coefficients at two different X-ray photon energies, which is also a problem in the conventional radiography.

Starting in the 1990s, the X-ray phase-contrast imaging (XPCI) emerged as a novel and powerful tool due to the high imaging contrast for weakly absorbing materials [5–8]. Various phase-sensitive methods have been developed to exploit the phase-contrast information in the X-ray regime [6–13]. Among those XPCI techniques, the differential XPCI

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based on X-ray grating interferometer (XGI), such as Talbot-Lau interferometer, is one of the most promising approaches for medical and industrial applications [14]. Furthermore, XGI provides access to three complementary image modalities simultaneously: absorption, differential phase and small-angle scattering ("dark-field") contrasts [15]. The absorption contrast is generated by the attenuation of the X-ray photons when they pass through a material, and quantitative information can be retrieved from radiographic imaging with knowing auxiliary information about the samples. Specifically, the attenuation coefficient of a certain material varies according to the X-ray photon energy, which is the physical fundamental of the DEXA [3]. Additionally, the differential phase-contrast (DPC) imaging and dark-field imaging can provide valuable and supplementary information to the conventional absorption imaging, and quantitative measurement by combining the different information has been reported [16–18]. Wang et al. proposed a method for material discrimination by defining an R value acquired by taking ratio of the absorption and small-angle scattering signals [16]. However, because the small-angle scattering information depends not only on the material composition but also strongly on the sub-pixel structures (that is, unresolved fine scatterers) in the sample [19,20], this method is applicable in limited occasions or questionable [21]. On the other hand, similarly to the absorption information, the X-ray phase information, which in theory can be retrieved simply by one-dimensional integration of the DPC imaging, depends only on the composition and density of the sample [22]. Qi et al. proposed a method by using the absorption and phase computerized tomography (CT) images obtained by XGI to calculate the effective atomic number and electron density [18].

Recently, we proposed using the absorption and phase-contrast information simultaneously acquired by a XGI setup for the estimation of body composition in radiographic observation (not CT). Simulation work under illumination of parallel monochromatic X-ray was performed to demonstrate the feasibility of the proposed method [23]. While in this work, the proposed method is extended to a polychromatic Talbot-Lau system, a non-linear regularized integration (NRI) algorithm [24] is applied for phase retrieval from the DPC information and a quadric surface fitting algorithm [25] is employed for the beam-hardening correction. Experimental results of a body phantom have demonstrated the feasibility of the proposed method with an improved measurement accuracy for the soft tissue areal density compared with DEXA.

2. Material and methods

2.1. Physical principle

The absorbance f of monochromatic X-rays by a material of uniform composition and density follows the Beer-Lambert law:

$$f = -\ln\left(\frac{\mathrm{I}}{\mathrm{I}_0}\right) = \mu\ \mathrm{t} = \mu_\rho M, \tag{1}$$

where I and I_0 are the intensities of the emergent and incident X-rays, respectively. μ is the linear attenuation coefficient and t is the thickness of the material along the path of the X-ray beam. For the BMD measurement, a projection areal density $M = \rho t$ (g/cm^2), where ρ is the material density, is usually measured. And $\mu_\rho = \mu/\rho$ is the mass attenuation coefficient. Polychromatic beam generated by X-ray tube is always applied for medical imaging. Because the attenuation coefficient depends upon the X-ray photon energy E, the attenuation of the X-ray intensity is expressed as the integration over the X-ray spectrum $S(E)$ additionally weighted by the spectral efficiency of the detector $D(E)$, leading to Ref. [26]:

$$f = -\ln\frac{\int S(E)D(E)\exp(-\mu(\mathrm{E})\mathrm{t})\mathrm{dE}}{\int S(E)D(E)\mathrm{dE}} \tag{2}$$

On the other hand, apart from the attenuation, the phase of an X-ray wave front traversing a homogenous object is shifted by g that is written as

$$g = \gamma t = \gamma_\rho M. \tag{3}$$

Here, $\gamma = 2\pi\delta/\lambda$ is the linear phase-shifting coefficient, where λ is the wavelength and δ is the refraction index decrement from unity of the material. So g is referred to as the phase shift. Similarly, we define the mass phase-shifting coefficient as $\gamma_\rho = \gamma/\rho$. Thus, relationship between phase information and sample thickness is derived similarly to Eq. (1). However, the linear phase-shifting coefficient γ has a dependence on the photon energy ($\gamma \sim E^{-1}$) [26] differently from the linear attenuation coefficient μ ($\mu \sim E^{-3\sim-4}$). For a more general case of polychromatic beam, the total phase shift becomes the incoherent sum of spectrum-weighted phase shifts at different photon energies:

$$g = \frac{\int S(E)\gamma(E)t\mathrm{dE}}{\int S(E)\mathrm{dE}} \tag{4}$$

.

Over the diagnostic range, any biological sample can be decomposed into two basis materials [25]. For example, when using DEXA for body composition estimation, Plexiglas (Polymethyl methacrylate, PMMA) and aluminum (Al) are always chosen for calibration [27] due to their similarities of attenuation behavior to soft tissue and bone, respectively. Therefore, the following derivations are based on this two-material body composition model.

Supposing that an object combined by PMMA of thickness p and Al of thickness q is traversed by monochromatic X-rays of a photon energy E_0, $f(p, q)$ can be reduced to a linear expression based on Eq. (1):

$$f = \mu_{PMMA}(E_0)p + \mu_{Al}(E_0)q. \tag{5}$$

Both the attenuation coefficients $\mu_{PMMA}(E)$ and $\mu_{Al}(E)$ are smooth and monotonically decreasing functions of the X-ray photon energy E within the range of diagnostic X-rays. When polychromatic X-rays are used, an average attenuation coefficient can be defined by the first partial derivatives of $f(p, q)$, written as $\overline{\mu}_{PMMA} = \partial f/\partial p$ and $\overline{\mu}_{Al} = \partial f/\partial q$. Moreover, for the photon energy $E \leqslant E_P$, where E_p is the maximum energy in the spectrum, the average attenuation coefficients decrease smoothly, monotonically, and asymptotically to $\mu_{PMMA}(E_P)$ and $\mu_{Al}(E_P)$ with the increase of the X-ray traversing length.

Similarly, for the same body model and monochromatic beam, the phase shift g can be rewritten as

$$g = \gamma_{PMMA}(E_0)p + \gamma_{Al}(E_0)q. \tag{6}$$

A smooth and monotonically decreasing function of energy is also available to describe the linear phase shifting coefficient γ, and we define average phase-shifting coefficients, $\overline{\gamma}_{PMMA} = \partial g/\partial p$ and $\overline{\gamma}_{Al} = \partial g/\partial q$ in polychromatic X-ray case.

Therefore, symbolically, the body composition measurement problem can be written as

$$0 = S_p(f, g) - p, \tag{7a}$$

$$0 = S_q(f, g) - q, \tag{7b}$$

where S_p and S_q are the functions to describe the bone thickness and soft tissue thickness, respectively. For monochromatic X-rays, S_p and S_q can be solved from Eqs. (5) and (6), assuming simple forms of linear polynomials. However, for polychromatic X-rays, due to the beam hardening effect, the problem is searching for a good approximation of S_p and S_q that are denoted by $P(f, g)$ and $Q(f, g)$, respectively. It has been shown for DEXA that second order functions describing smooth, ripple-free, and monotonic surfaces can be used for the approximation [25]. The surfaces exhibit correct linear asymptotic behavior for both large and small sample thicknesses, allowing well approximation beyond calibration regions. Due to the similarities between the absorption and phase information mentioned above, this method can be easily