International Journal of Scientific Engineering and Applied Science (IJSEAS) – Volume-2, Issue-1, January 2016
ISSN: 2395-3470
www.ijseas.com

A Survey on Segmentation –Based Correction of Partial Volume Artifacts for the Accurate Volumetric Measurement of Solid Lung Lesions on CT Images

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Abstract

In medical application, estimating the accurate volume of a tumor needs measurement of the change in its size. This is an important issue for monitoring the cancer therapeutics and this requires tumor's description in three dimension (3-D). For achieving this, segmentation should be performed. Segmentation is used to find the tumor area on the captured lung image from the Computed Tomography (CT) scan report. Segmentation can be performed using different methods. But there is a challenging feature present on the CT medical images which is called partial volume artifacts. The partial volume effect (PVE) is a significant limitation in medical imaging that results in blurring when the boundary between two features of interest falls within a voxel. The removal of these artifacts is also a major concern for volumetric measurements. This paper is a survey on segmentation-based partial volume correction which is useful for volume measurement of solid lung lesions on CT images. Survey involves different 3-D segmentation methods and partial volume correction procedures for accurate volumetric measurements.

Keywords: *Computed Tomography, 3-D segmentation, Partial volume artifacts, Partial Volume Correction.*

1. Introduction

Oncology is a subordinate of medicine that deals with tumors. Oncology related therapy -the diagnosis of any tumor and monitoring of tumor patients- decides whether the treatment is successful or not [13]. Some of the techniques which are commonly used to diagnose the tumor are X-RAY, magnetic resonance imaging (MRI) or positron emission tomography (PET). In these techniques systemic scan is used which takes several x-ray images from different angles to form 3-D.It provides more detailed view than others and also determines how much the tumor has affected the nearby tissue. If the tumor size is shrinking and also there is no new arise, therapy is successful. That is the size of tumor in CT images is an important parameter for assessing the treatment success.

Since tumors grow and shrink irregularly, diameter measurement of tumor is not an accurate parameter for assessing its size. Another standard for size calculation based on segmentation is voxel-counting. A voxel counting defines the volume which is given by the number of all voxels within the output of segmentation and then multiplied by their volume. But partial volume artifacts will pose challenges in the accuracy and reproducibility of volumetric measurements by voxel counting method [1]. Partial volume correction is an important task before measuring the volume of lesion. There are different 3-D segmentation procedures and partial volume correction methods. This survey involves collection of those methods for the purpose of finalizing the accurate and the best method for volumetric measurement of lesions.

A robust and automated algorithm to segment lung nodules in three dimensional CT volume dataset is one of the earliest methods. Here the nodule is segmented out in slice–per–slice basis. This approach can be consistently and robustly used to segment not only the solitary nodules but also the nodules attached to lung walls and vessels. However, since any completely automated algorithm may fail in a complicated case, they have required user interaction for the corrections of segmentation results [2]. Another novel approach is the automatic segmentation of lung nodules in a given volume of interest (VOI) from high resolution multi-slice CT images by dynamically initializing and adjusting a 3D template and analyzing its cross correlation with the structure of interest [3]. However, since this method is developed for small and ellipsoidal shape nodules, the nodules attached to the chest wall, an ellipsoid shape is usually not a good approximation [4]. A three-dimensional method for the segmentation, analysis, and characterization of small pulmonary nodules imaged using CT uses a semi-automatic classification of the target nodule into one of four nodule models -solitary, vascularized, Pleural tail, and juxtapleural nodules. But formulating mathematical models of each class and developing separate segmentation schemes accordingly made this method a difficult one. A robust statistical estimation and verification framework for characterizing the ellipsoidal geometrical structure of lung nodules in the multi-slice X-ray CT images is another approach for segmenting the lesion. But due to irregular nodule growth the change in shape of lesion is a potential drawback of this ellipsoid approximation approach [14]. An automated method for 3D image analysis for segmenting lung nodules in HRCT has been proposed but it is harder to integrate as a plug-in to existing workstations or CAD systems if there is any dependence on a preprocessing step [6].

Most of the work for correcting partial volume artifacts focus on techniques like MRI, PET as well as single-photon emission computed tomography (SPECT) [8]. An algorithm for identifying the distribution of different material types in volumetric datasets uses a probabilistic Bayesian approach is introduced by D. H Laidlaw [7]. Due to the limited resolution of this modality one approach for correcting PET images for the PVE is based on iterative deconvolution with the use of a 3D maximum likelihood expectation-maximization (MLEM) algorithm [9]. This method for unsupervised estimation, in single-channel image data, of partial volume fractions simultaneously estimates partial volume fractions, by means of the different tissue classes, as well as the locations of tissue boundaries within the image [10].

Another method introduced to restore the ideal boundary by splitting a voxel into sub-voxels and distributing the signal into the sub-voxels. Each voxel is divided into four or more sub-voxels by nearest neighbor interpolation. The gray level of each sub voxel is treated as materials which is able to move between sub voxels but it is not same as in the case of movement between voxels [11]. To create interpolated 3D images corrected for partial volume without enhancement of noise there is another anisotropic diffusion method.

The remaining part of the paper is organized as follows. In Section II survey of all methods will be described in detail. The paper concludes with a brief summary in section III.

2. Literature Survey

In medical applications of image processing, the change of tumor's size is an important issue for monitoring the cancer therapeutics and to decide whether the treatment is in right way. For this, oncologist needs the correct delineation of tumor in 3-D to measure its size in terms of volume. To achieve this requirement, proper segmentation has to be done on the CT image. In 3-D image, because of limited spatial resolution, some voxels will loss at the border of the segmentation output due to partial volume artifacts. To overcome this problem, partial volume correction is used. This survey includes the details of 3-D segmentation and partial volume correction methods which are useful for accurate volumetric measurement of lesions.

N. Xu et al. [2] described a method for segmentation of nodule. To segment lung nodules in 3-D CT volume dataset, they introduced a robust and automated algorithm. The nodule is segmented out in slice–per–slice basis. For this, first process each CT slice individually to extract two dimensional (2D) contours of the nodule. Thereafter these can then be stacked together to get the whole 3D surface. Due to the usage of dynamic programming based optimization algorithm, the extracted 2D contours are optimal. To extract each 2D contour, a shape based constraint is utilized. This approach can be used to segment the solitary nodules as well as the nodules attached to lung walls and vessels in a consistent and robust manner. Here, the approach uses dynamic programming to find the optimal lung nodule boundary. However, many times the initialization of the shape based approach fails due to the calcification of the pulmonary nodules. This problem can be resolved by pre–processing the CT volume dataset, locally around the nodule selected by the physician, using Expectation Maximization (EM) algorithm. This detects and removes the calcification from the nodules. This process can also feasibly give information about the pattern of calcification of the nodules. Method is completely automated. Only the initial point on the surface of the nodule requires manual assistance. However, since any completely automated algorithm may fail in a complicated case, they have provided user interaction for corrections of the segmentation results [4].

L. Fan et al. [3] explained another innovative approach to the automatic segmentation of lung nodules. It works by initializing and adjusting a 3D template dynamically and analyzing its cross correlation with the structure of interest. Firstly, thresholding techniques are used to separate the background voxels. The structure of interest consisting of a candidate nodule and possible attached vessels and it is then extracted by excluding any part of the chest wall present inside the VOI. From the proposed segmentation method finds the core of the structure of interest, which corresponds to the nodule can be found out. According to this result, algorithm analyzes its orientation and size, and initializes a 3D template. Then, the template gradually expands and its cross correlation is being computed at each step. Based on the analysis of the cross correlation curve, the template is optimized. An AND operation is performed between the extracted structure and the optimal template in order to perform the segmentation of the nodule. Then it is refined by a spatial reasoning method. The method is suitable for small, ellipsoid nodules but requires interactive correction in the case of irregularly shaped nodules. However, for nodules attached to the chest wall, an ellipsoid shape is usually not a good approximation.

W. J. Kostis [5] described a three-dimensional methods for the segmentation, analysis, and characterization of small lung nodules on CT images. It uses a semi-automatic classification of the target nodule into one of four nodule models -solitary, vascularized, pleural tail, and juxtapleural nodules. Separation from adjacent high density structures is performed by morphological methods after an initial segmentation using a fixed threshold. However, since it was specifically designed for small nodules, several assumptions were made concerning especially the removal of attached vasculature that are not necessarily transferable to lesions of arbitrary size and morphology. Instead of making general assumptions about the connectivity and treating every nodule the same, the novel adaptive approach described in the following finds the theoretically optimal strength of morphological opening for each specific nodule such that adjacent vasculature is removed while the nodule shape is preserved and its exact boundaries can be restored.

Since the varied local geometry is not amenable to a single method, the techniques for the segmentation of each of these nodule classes differ from one another. Therefore, formulate mathematical models of each class and develop separate segmentation schemes accordingly. This is a notable difficulty of this method. K.

Okada et al. [14] explained a powerful statistical assessment and verification scheme for characterizing the geometrical structure of ellipsoidal lung nodules in the multi-slice CT images. This solution estimates the target's center location, ellipsoidal boundary approximation, volume, maximum or average diameters, and isotropy by robustly and efficiently fitting an anisotropic Gaussian intensity model. The purpose of the design is to enhance the robustness against data with large deviations from the chosen model, margin-truncation induced by neighboring structures, and marker location variability. The system is generic and can be applied for the analysis of blob-like structures in various other applications. This automated method to approximate ground glass opacities as well as solid nodules by ellipsoids using anisotropic Gaussian fitting, the volume of the nodule is calculated by the volume of the ellipsoid. The approach is intriguing due to its applicability to nonsolid nodules in case of shape changes due to irregular growth of nodule. This is a potential drawback of this ellipsoid approximation approach.

A fully-automated 3D image analysis method [6] is proposed to segment lung nodules in HRCT. They proposed a 3-D approach for providing automatic segmentation of nodules with sizes ranging from 2 to 20 mm diameter irrespective to their spatial location. This approach relies on a specific morphological operator, the selective marking and depth constrained (SMDC) connection cost. It features specific properties in terms of selectivity -ensures size-independent nodule detection- and topographical connectivity preservation -makes possible to discriminate the nodules from other dense structures and to differentiate the three classes of nodules. This method is designed for small lung nodules. For the purpose of doing segmentation, here they used some morphological methods which follows initial thresholding. But during the detection procedure which processes the complete lung, this method can take advantage of global information acquisition. For accurate local assessment, the global information has a great role. To analyze a complete nodule implies the analysis of more than 300 CT image slices. It is not suitable for fast, interactive one-click methods except a preprocessing step is performed earlier. It is harder to integrate a method as a component to existing workstations or CAD systems if there is any dependence on preprocessing step.

An algorithm that works on volumetric datasets produced with MRI or CT for identifying the dispersion of different material is explained by D. H. Laidlaw et al. [7]. Two aspects for this method are: 1) voxels can contain more than one material, due to partial-volume effects, or blurring, so compute the relative proportion of each material in the voxel and 2) by reconstructing a continuous function from the samples, incorporate information from neighboring voxels into the classification process and also looking at the distribution of values that the function takes on within the region of a voxel. A histogram taken over the region of the voxel is used to represent the distribution of those values. The mixture of materials that are identified within the voxel using a probabilistic Bayesian approach. This matches the histogram by acquiring the mixture of materials within each voxel most likely to have created the histogram.

The improvements of this method as compared with other methods arises due to: 1) the reconstruction of continuous function from the samples, 2) use of histograms taken over voxel-sized regions to represent the contents of the voxels, 3) model the sub-voxel partial-volume effects caused by the band-limiting nature of the acquisition process, and 4) the use of Bayesian classification approach.

A. S. Kirov et al. [9] explained one approach for correcting PVE which arises due to limited resolution of PET images. Here authors used a post-reconstruction PVE correction based on iterative deconvolution using a 3D MLEM algorithm. A one-step late (OSL) regularization procedure based on the assumption of local monotonic behavior of the PET signal is used for achieving the convergence. To selectively control variance depending on the local topology of the PET image, this technique is further modified by authors. There is a promising approach for partial volume effect corrections in PET which is a regularized iterative deconvolution with variance control based on not only the properties of the PET image but also on estimated image noise.

Due to the presence of partial volume artifacts in medical images that blur the boundaries between different regions, making accurate description of anatomical structures difficult. A method for unsupervised estimation of partial volume fractions in single-channel image data is proposed by D. L. Pham and P. L. Bazin [10]. The proposed algorithm simultaneously measures partial volume fractions by the means of the different tissue classes and also estimates the locations of boundaries of tissue within the image. The second methods allows the partial volume fractions to be forced to represent nearly pure or pure tissue except along tissue boundaries. They demonstrated the application of this algorithm on simulated as well as real MRIs.

They incorporated boundary information into the estimation procedure is introduced. Tissue boundaries within the image play a key role in partial volume effects because they define where partial volume artifacts will occur. To represent pure tissue away from boundaries and also to constrain the partial volume fractions to be smooth, knowledge about the boundary locations is required. For estimating partial volume fractions and boundaries, the overall framework is based on the Mumford-Shah functional approximation. This leads to an iterative algorithm for the estimation of mean intensities of the tissue classes, the boundaries and the partial volume fractions. To solve the discretized partial differential equation in the boundary estimation, there is a multigrid algorithm is used.

O. Salvado et al. [11] proposed a method to restore the ideal boundary by splitting a voxel into sub-voxels and reapportioning the signal into the sub-voxels. This method is designed to correct MRI 2D slice images. The partial volume will be a considerable limitation of this method. Each voxel is divided into sub-voxels by nearest neighbor interpolation. The gray level of each sub voxel is treated as materials which is able to move between sub voxels but it is not same as in the case of movement between voxels. For creating a reverse diffusion process, a partial differential equation is written to allow the material to flow towards the highest gradient direction. This flow is subject to constraints that tend to create step edges. Material is conserved in the process thereby conserving MR signal. The method repeats until the flow decreases to a low value.

This method does not depend on the classification and histogram modeling. They modelled the voxel indication as the integral of the signal over the surface bounded by the voxel boundary limits. Sub-voxels are created using nearest neighbor interpolation. The material able to move between sub-voxels is the grey-level of each sub-voxel. A highly unstable equation will arise due to the change of direction of the time arrow. These methods do not include restrictions to address the presence of partial volume artifacts, and it would require characteristic regularization functional to dispatch this issue. They presented some of the simple constraints which control the reverse diffusion in the context of image processing. Flow is based on constraints that favor to create step edges between regions of different intensities on captured image. Material is conserved in the process so conservation of the MR signal ca also be achieved. Since it uses more information, and resolving the drawbacks of the pattern recognition techniques, this method should lead to better results than pure interpolation methods.

O.Salvado et al. [12] introduced a new anisotropic diffusion method that allows to create interpolated 3-D images corrected for partial volume artifacts, without any enhancement of noise present on the captured image. They applied a modified version of the anisotropic diffusion approach after a zero-order interpolation. For high gradient values, the diffusion coefficient turns into negative values. Therefore, the new scheme restores edges between regions that have been blurred by partial volume artifacts. But this acts as normal anisotropic diffusion in flat regions, where it reduces noise. They add constraints to maintain the method and model partial volume present on the CT image. That is, the sum of neighboring voxels must equal to the signal in a voxel. This voxel should be kept within its neighbor's limits and the signal in the original low resolution voxel. The method performed properly on a different set of synthetic images and MRI scans. There is no notable artifact that persuaded by interpolation with partial volume correction, and also reduces noise in homogeneous regions.

3. Conclusion

This survey has been performed for collecting the details of 3-D segmentation and partial volume correction methods which are useful for accurate volumetric measurement of lesions. As tumor grow and shrink irregularly, diameter measurement of tumors and voxel- counting are not accurate parameters for assessing its size. Partial volume artifacts must be taken care of, for the accurate volume measurements. This survey helps in identifying all possible segmentation methods and partial volume correction methods.

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