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著者	Takemura Akihiro, Kojima Hironori, Ueda Shinichi, Isomura Naoki, Noto Kimiya
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A metric for evaluation of deformable image registration

Akihiro Takemura[*a], Hironori Kojima[b], Shinichi Ueda[b], Naoki Isomura[b], Kimiya Noto[b]

[a] Faculty of Health Sciences, Institution of Medical, Pharmaceutical and Health Sciences, Kanazawa University/ 5-11-80 Kodatsuno, Kanazawa, Ishikawa JAPAN, 920-0942; [b] Department of Radiological Technology, Kanazawa University Hospital/ 13-1 Takaramachi, Kanazawa, Ishikawa JAPAN, 920-8641

ABSTRACT

We propose a new metric, local uncertainty (LU), for the evaluation of deformable image registration (DIR) for dose accumulation in radiotherapy. LU measures the uncertainty of placement of each voxel in an image set after a DIR. The underlying concept of LU is that the distance between a focused voxel and a surrounding voxel on an image feature such as an edge is unchanged locally when the organ that includes these voxels is deformed. A candidate for the focused voxel after DIR can be calculated from three surrounding voxels and their distances. The positions of the candidates of the focused voxel calculated from several groups of any three surrounding voxels would vary. The variation of candidate positions indicates uncertainty of the focused voxel position. Thus, the standard deviation of candidate positions is treated as an LU value. The LU can be calculated in uniform signal regions. Assessment of DIR results in such regions is important for dose accumulation. The LU calculation was applied to a pair of computed tomography (CT) head and neck examinations after DIR. These CT examinations were for the initial radiotherapy planning and re-planning for a treatment course where the tumor underwent shrinkage during treatment. We generated an LU image showing high LU values in the shrinking tumor region and low LU values in undeformable bone. We have proposed the LU as a new metric for DIR.

Keywords: Deformable Image Registration, Validation/Evaluation, Image-Guided Therapy

1. INTRODUCTION

Deformable image registration (DIR) methods are used to assist target/normal tissue contouring for re-planning of radiotherapy treatment plan[1]. Presently, DIR methods have applied four dimensional CT image to contour tissues and a tumor in several breath phases[2]. DIRs could reduce time for contouring and have potential to decrease inter-observer variation of contouring. For contouring DIRs have been validated in several studies[3–5] which evaluated correctness of the placement of organ edges or assessed the similarity of organ volumes. These studies evaluated DIRs performance by Dice similarity coefficient. Dice similarity coefficient assess that the contoured tissue/tumor have correct volume and location.

DIR also have potential to accumulate daily radiation dose in radiotherapy[6–8]. In such dose accumulation, the dose distribution is obtained by summing a deformed dose distribution with a reference dose distribution pixel by pixel. Therefore, dose distribution in contours and/or regions in uniform signal should be deformed correctly. However, The Dice similarity measure cannot assess the correctness of deformation for such uniform signal regions. To obtained correct accumulated dose distribution, DIR should correclty perform not only in volume and shape but also pixel location in the uniform signal regions. Unfortunately, true deformation in regions of uniform signal is hardly obtained because there is no clue for correct deformation inside a uniform signal region. Evaluation of deformation in uniform signal regions is hard. We think that quantifying the uncertainty of deformation in these regions, instead of the correctness of deformation, could improve the reliability of accumulated dose obtained with DIR. We propose a new metric, local uncertainty (LU), for DIR.

*at@mhs.mp.kanazawa-u.ac.jp; phone 81 76 2652538; fax 81 76 2344366

2. METHODOLOGY

2.1. Local uncertainty for deformable image registration

The LU quantifies the uncertainty of a focused voxel position after DIR. Especially, the LU can target regions of uniform signal. The concept of the LU is that the distance between a focused voxel and a surrounding voxel on an image feature such as an organ edge is unchanged in a small region when the organ including these voxels is deformed. A position of a point in three dimensional space can be calculated from three points and three distances. Thus, three surrounding voxels on image features and their distances from the voxel to the focused voxel can calculate the position of the focused voxel in the deformed image. However, another candidate for the focused voxel which was calculated from other three surrounding voxels in the deformed image is not necessarily to correspond to the first candidate. There is a variation of the candidate positions. We defined the variation of the candidate positions as the LU.

A schematic explanation of the candidate calculation for LU in a two-dimensional image was shown in Fig. 1. First, the distances (r1, r2, r3) of each of three surrounding voxels (p1, p2, p3) to the focused voxel p0 before DIR is found. Then these three distances and three positions of the moved surrounding voxel (p1', p2', p3') after the DIR are used for calculation of a candidate for the focused voxel position. The p1', p2', p3' and their distance can depict circles. An intersection point c0' of the p1' centered circle with radius r1, the p2' centered circle with radius r2, and the p3' centered circle with radius r3 is a candidate for the moved focused voxel. The candidate c0' to cn' are obtained from groups of any three surrounding voxels. To calculate the LU, n should be at least 3. The LU value is calculated from the variance of each coordinate of the candidates as;

$$\mathrm{LU} = \sqrt{\sigma_x^2 + \sigma_y^2 + \sigma_z^2} \tag{1}$$

$$\sigma_x = \sqrt{\frac{\sum_{i=0}^{n}(x_{ci'} - \bar{x})}{n-1}} \tag{2}$$

$$\sigma_y = \sqrt{\frac{\sum_{i=0}^{n}(y_{ci'} - \bar{y})}{n-1}} \tag{3}$$

$$\sigma_z = \sqrt{\frac{\sum_{i=0}^{n}(z_{ci'} - \bar{z})}{n-1}} \tag{4}$$

where, $\sigma_x, \sigma_y, \sigma_z$ are the standard deviation of x, y, z coordinates, respectively. $x_{ci'}$, $y_{ci'}, z_{ci'}$ are x, y, z coordinates of the i th candidate ci. n is a number of the candidates.

We employed tissue edge in a CT image as the image feature. The voxels which had the difference of 40 or larger in Hunsfield Unit from the voxel value of the focused voxel were searched as the surrounding voxels. We have assumed that the LU calculation should be applied to the DIR result of CT image to CT/cone-beam CT image because dose accumulation have been targeted. Thus, we employed 40 Hunsfield Unit.

Large number of groups of any three surrounding voxels can be used for the calculation of the LU. However, Too many groups takes much calculation cost. Therefore surrounding voxels along 18 neighbor voxel directions (Fig. 2) regarding to each focused voxel were searched and 12 groups of three of the surrounding voxels were used for a demonstration of the LU calculation.

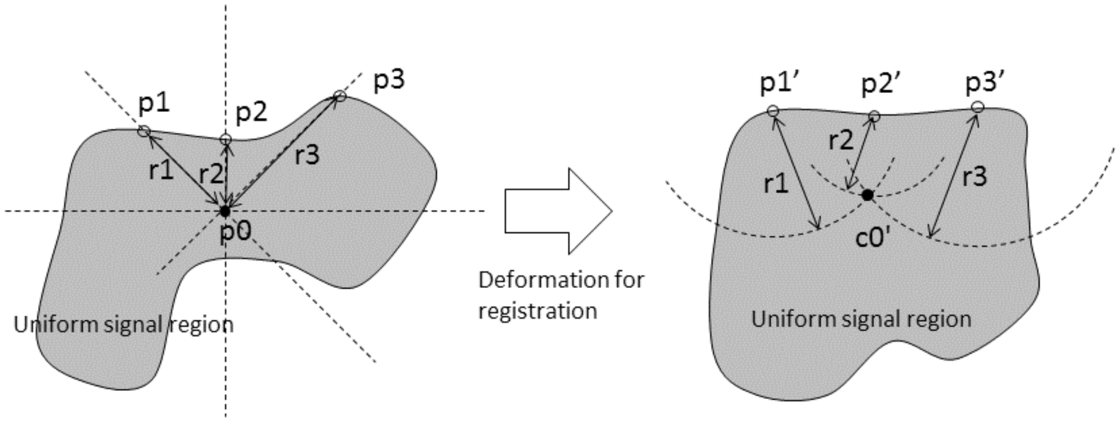


Figure 1 Schematic explanation of the candidate calculation for LU in a two-dimensional image. The p0 is a focused voxel, the p1 to p3 are surrounding voxels before DIR. After DIR, the p1 moves to p1′, the p2 to p2′, and the p3 to p3′. An intersection of three circles, whose centers are p1′, p2′, and p3′ and radii are r1, r2, and r3, is a candidate for the focused voxel c0′.

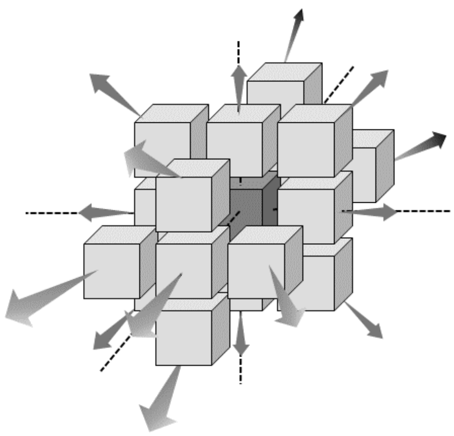


Figure 2 Directions for searching a pixel on image features. A dark gray box means a focused voxel and light gray boxes mean 18 neighbor voxels. Voxels on image features was searched along the direction from the focused voxel to each neighbor voxel.

2.2. Application to the CT to CT deformable image registration

To demonstrate the LU, the LU calculation was applied to a deformable image registration result of two computed tomography (CT) examinations of a patient with a head and neck cancer. A plug-in software for ImageJ (NIH) was developed to demonstrate the LU calculation and it was used for this demonstration. The CT images were obtained from the CT system, Toshiba Aquilion LB (Toshiba Medical Systems, USA). The Scan was performed at 120 kV with 100mA, 3 sec/rotation and a 480 mm field of view. The CT images had the matrix of 512 x 512(pixel size: 0.94 x 0.94 mm^2) and the slice thickness of 2 mm.

One of the CT examinations was performed for a radiotherapy treatment plan before the start of a radiation therapy course and the other was performed one month later during the radiation therapy course. The tumor in a lymph node was presented on the right neck in the image of the first scan (Fig. 3a). The second CT examination was performed for re-planning and the tumor had decreased in size (Fig. 3b).

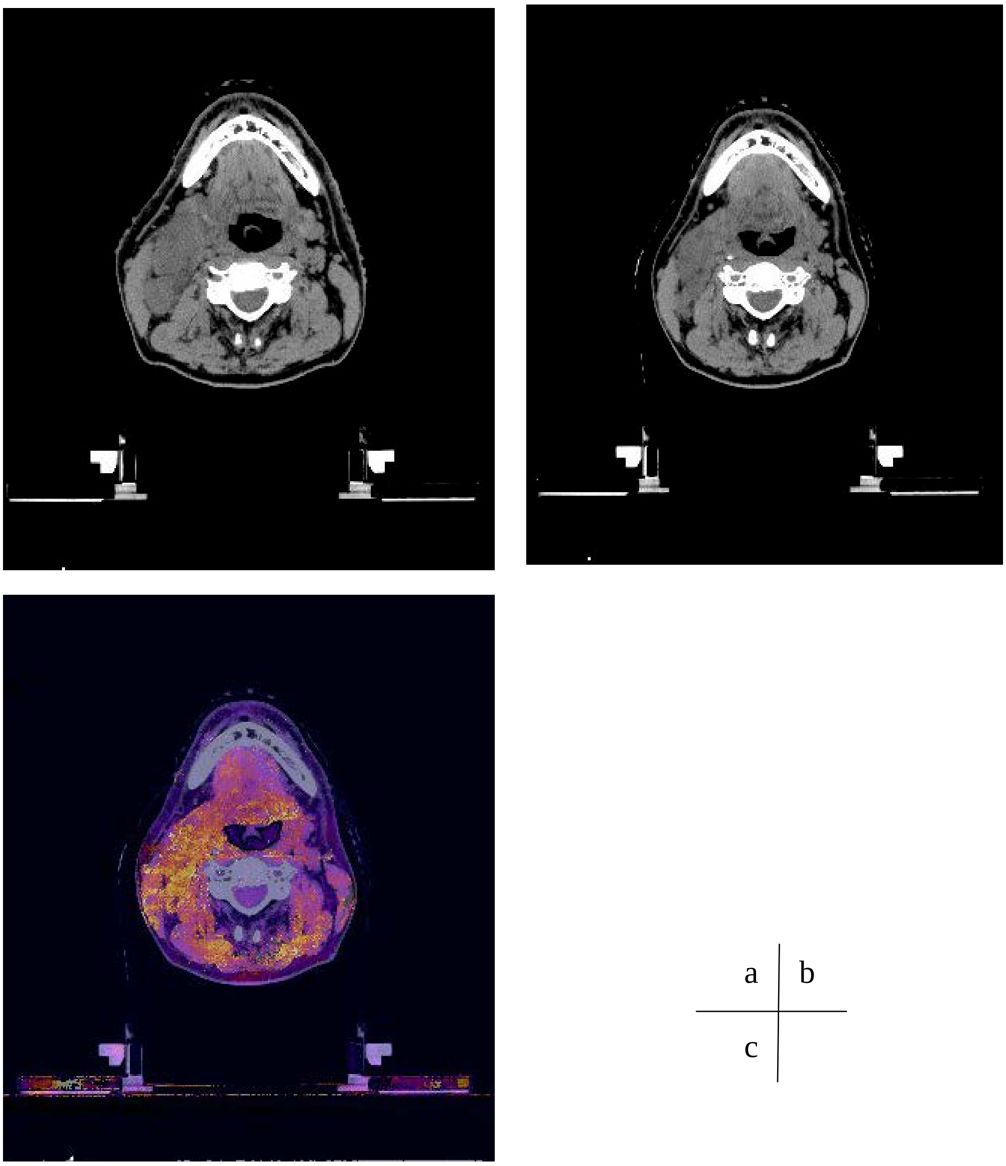


Figure 3 Result of the LU calculation. Image (a) is a representative slice from the pretreatment CT examination, image (b) is a representative slice from the CT examination acquired during treatment. The images illustrate the shrinkage of the tumor during treatment. Image (c) is the LU image. High LU is depicted in red and low LU is depicted in blue. The region in the shrinking tumor shows high LU and uninvolved bony structures show low LU.

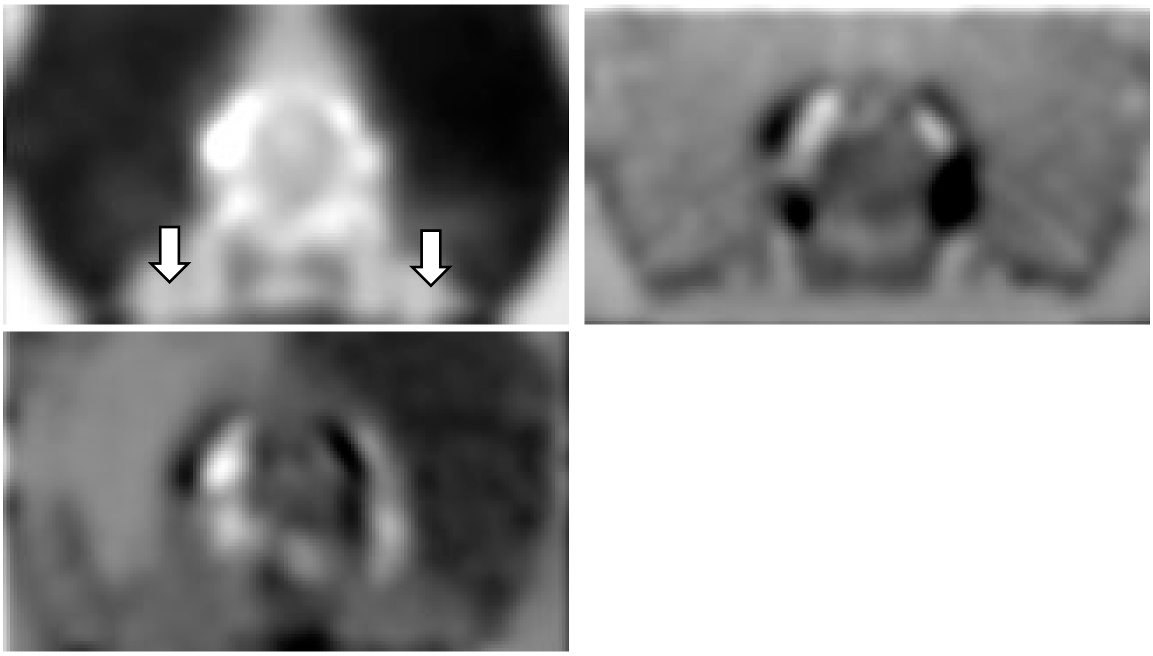
The second CT image set was deformed to match the first CT image set with an image registration software, MIM maestro 5.6.3 (MIM software Inc. Cleveland, OH, USA). After the deformation, the deformed image set was checked and these CT image sets and displacement vector field data were exported in DICOM-RT format. The displacement vector field had the voxel spacing of 3x3x3 mm^3.

3. RESULTS

Figure 3 shows representative slices from the pretreatment and during-treatment examinations, and the resulting image after applying the LU calculation. The LU image shows high values in the shrinking tumor and low values in undeformable bone structures, illustrating the high uncertainty of the DIR result in the tumor region. There were high value regions on the shell holder in the LU image. The shell holder is made from plastic and they are basically undeformable. Displacement vector on these regions represented the deformation on X axis (Figure 4). This is wrong deformation in uniform signal regions. The LU calculation could show these deformation error which usually could not depicted in deformed images.

Precision of the LU calculation is depend upon the resolution of the displacement vector field data and CT images. The resolution of displacement vector field data is also depend upon the DIR software. The resolution of 3x3x3 mm^3 for the displacement vector field data was used for this demonstration. This resolution was default for exporting the deformation result on the DIR software we used. Dose distribution data usually have 2 to 5 mm resolution, so that 3 mm resolution might be sufficient. However, the best resolution of displacement vector field data should be found in the future.

Sequential CT image data of one case was used for the demonstration of the LU calculation. More cases should be applied for checking the adaptability and consistency of the LU calculation. Then, the LU calculation should be applied for several CT image sets of a patient for dose accumulation.



a b

c

Figure 4 Displacement vector field data. The displacement vector fields along X, Y and Z axes are in image a, b and c, respectively. The Slice position is the same as the images in Figure 3. The location of the shell holder had varied intensity in the image a (white arrows) and that means these regions were deformed.

4. CONCLUSIONS

This new metric for evaluation of DIR, which quantifies the uncertainty of each voxel after DIR, even in regions with homogeneous signal, can potentially result in more accurate dose accumulation of delivered dose during a radiotherapy course.

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