MOTION CORRECTION IN POSITRON EMISSION TOMOGRAPHY CONSIDERING PARTIAL VOLUME EFFECTS IN OPTICAL FLOW ESTIMATION

Daniel Tenbrinck\textsuperscript{1,2}, Mohammad Dawood\textsuperscript{2}, Fabian Gigengack\textsuperscript{1,2}, Michael Fieseler\textsuperscript{1,2}, Xiaoyi Jiang\textsuperscript{1,2}, Klaus Schäfers\textsuperscript{2}

\textsuperscript{1}Dept. of Mathematics and Computer Science University of Münster, Germany \textsuperscript{2}European Institute of Molecular Imaging University of Münster, Germany

ABSTRACT

Motion correction in Positron Emission Tomography (PET) using optical flow estimation can lead to image artifacts due to Partial Volume Effects (PVE). These artifacts appear especially in cardiac gated PET images and cause blurred edges in the averaged gates. In this paper we propose a new method to motion correct PET images considering the PVE during optical flow estimation. For this purpose we introduce a local intensity correction algorithm and combine it with the optical flow computation in an iterative scheme. The results of our approach show a qualitative and quantitative improvement of the motion corrected PET gates in examinations of both human patients and laboratory mice for pre-clinical research.

Index Terms— Motion Correction, Optical Flow, Partial Volume Effect, PET, Gating

1. INTRODUCTION

Reconstructing images from PET data is not trivial from a mathematical point of view and one has to face many problems to gain reliable information for clinical diagnosis and therapy. One problem in PET imaging is the relative long duration of data acquisition (from minutes to an hour depending on the tracer and the aim of the study). Due to cardiac and respiratory motion of the patient during the examination the resulting images become blurred in regions with large motion.

1.1. Gating

One basic approach to avoid motion artifacts in PET data is called \textit{gating}. The idea of gating is to divide the process of motion into different phases and group the measured data according to these phases. Each resulting image of its corresponding phase is called a \textit{gate}. Every gate contains only the information from the same phase of motion, thus the motion within one gate is reduced significantly [1]. Using information from only one gate decreases the signal-to-noise ratio (SNR), leading to difficulties in quantification. To get images from PET data without blurring but good SNR one has to register all gates to a reference gate and average the registered gates.

1.2. Motion correction using optical flow

Optical Flow (OF) has proven its usefulness for PET gate registration [2]. The advantage of OF estimation algorithms lies in the mathematical foundation of these algorithms. The basic assumption of optical flow computation is called \textit{Brightness Constancy Constraint} (BCC):

\[ I(x, y, z, t) = I(x + u, y + v, z + w, t + 1) \]

with \( I(x, y, z, t) \) representing the intensity of a voxel \((x, y, z)\) in an image \( I \) at time \( t \) and \((u, v, w)\) the motion of this voxel. This constraint requires that the intensity values of corresponding voxels are constant in presence of motion.

However, this assumption is violated in some real world applications. Among others, the partial volume effect can lead to a violation of this constraint. In the following the gate that has to be motion corrected is called \textit{floating gate} \( I_F \), as the target of the correction is called \textit{reference gate} \( I_R \).

1.3. Partial volume effect

Due to the limited spatial resolution of PET scanners edges in the reconstructed images often appear blurred and washed out. One reason for this is the so called \textit{Partial Volume Effect} (PVE). Information of structures smaller than the resolution of the PET scanner is spread out in the local neighborhood. Thus, without considering PVE it is difficult to quantify the uptake of a tracer in certain Regions of Interest (ROI). It has been shown that especially for small ROIs the partial volume effect leads to large errors in quantification [3]. The spatial resolution of the PET scanner is therefore decisive for the magnitude of PVE. The partial volume effect not only makes quantification of the PET data difficult, but also influences the results of motion correction using optical flow estimation.
1.4. Effect of PVE on OF estimation

On the one hand the BCC is a fundamental equation for robust optical flow estimation. On the other hand the requirement of intensity consistency under motion may be too strict for images gained from real data. Especially in positron emission tomography the PVE leads to a violation of the BCC and thus to errors in OF estimation. A one-dimensional toy example in Fig. 1 explains the violation of the BCC due to PVE in a simplified manner. On the left it shows two one-dimensional point-source signals, which have been blurred by the PVE. As the peaks of the signals move towards each other on the right, they merge due to the PVE and are not separable any more. Pixels, that lie between the two peaks, do not have corresponding pixels with the same intensity and so the BCC gets violated at this location.

In PET imaging situations similar to the example in Fig. 1 can occur. Examinations of the human heart with a high uptake of radionuclides in the region of the myocardium can lead to a heavy influence of the PVE on optical flow estimation. In cardiac gating it is aimed to correct the motion between different gates of the contracting myocardium. As the heart is a relatively small structure with an relatively high uptake of radioactive tracer in these studies, it is easy to predict problems in the computation of OF. For a gate representing the end of diastole the inner region of the myocardium appears widely opened and only low intensity can be found for these voxels. When the heart contracts the myocardial wall thickens and the inner region gets smaller as these walls move towards each other. Especially at the end of systole almost no space is left inside the myocardium and for this reason a similar situation to the examples above arises. Because of the heavy violation of the BCC in cardiac gated PET images the results of motion correction often show image artifacts. These artifacts get even more disturbing for small animal PET data in pre-clinical research as compared to human patient data. The reason for this is the relatively small ROI, e.g. mouse myocardium, together with the low resolution of a small animal PET scanner.

Fig. 4c) shows the results of motion correction on cardiac gated mouse data using OF estimation from [2]. In a) one can see the gate that has to be motion corrected (end phase of systole). In b) is the reference gate for the computation of OF (end phase of diastole). It can clearly be seen that the PVE causes image artifacts in the inner myocardium during motion correction. These artifacts lead to an immense loss of anatomical information in this case.

To deal with these challenges it is necessary to modify the estimation of optical flow and adjust this technique to the special circumstances of PET imaging.

2. METHODS

Our aim is to present a new method for motion correction of cardiac gated PET data, taking into account the influence of the partial volume effect on OF estimation. The idea is to localize erroneous regions in the corrected gate and enhance this gate by redistributing the surplus of intensity which led to image artifacts. With this approach we want to minimize image artifacts after motion correction and thus improve the image quality for better quantification.

2.1. Data acquisition

To validate our method we chose the data of ten different PET studies containing five examinations of human patients and five pre-clinical data sets of laboratory mice. The tracer used in these studies is a radioactive marked glucose analog called $^{18}$F-Fluorodeoxyglucose (FDG). This type of experiment was chosen because of the high uptake of FDG in the myocardium. Each patient data set is split into ten gates of $175 \times 175 \times 47$ voxels. To acquire PET data of human patients we used a Siemens Biograph Sensation 16 PET/CT Scanner with a spatial resolution of approximately 6 mm. The PET data sets of mice were acquired using a quadHIDAC scanner which has a spatial resolution of approximately 1 mm. The mouse data is divided into eight gates of $80 \times 80 \times 80$ voxels.

2.2. Gating and optical flow estimation

We used amplitude-based gating techniques because they are the most precise methods from all evaluated techniques, like e.g. time-based gating methods [1, 4]. For amplitude-based gating the blurring of anatomical structures, induced by motion, is at least significant. This leads to better results in motion correction and to clearer edges. For the human patient data sets we chose gate 10 as the reference gate for OF estimation and analogously gate 8 for the mouse data. Those gates represent the end phase of myocardial diastole, at which anatomical structures can be quantified best.

The optical flow estimation was performed by an extension [2] of one of the most precise methods from the literature [5] to work with PET data. To perform the motion correction using the calculated flow fields we use a tricubic interpolation method that warps the gate with help of the computed flow vectors. This approach gives excellent results for respiratory gated PET data and is relatively stable to the impact of noise. Yet, it may also produce image artifacts for cardiac gated PET data because it is also based on the BCC.
2.3. Local intensity correction

To deal with the impact of the PVE on OF estimation one has two different options. The first option is to correct all gates a priori with an Partial Volume Correction (PVC) algorithm, e.g. in [6]. It is a hard task to determine the source image without decreasing the SNR of the given data. For this reason we decided to choose another way to deal with this problem. The second option, which is the one of our approach, is to take into account the influence of the PVE during the process of OF estimation. In the following we are going to introduce a method called Local Intensity Correction (LIC), which we used to challenge the described problems.

Let \( I_R \) and \( I_F \) be two gates. \( I_R \) is defined as the reference gate for OF estimation whereas \( I_F \) is the floating gate. The first goal of our approach is to identify regions in which the BCC has been violated severely. Because we want to avoid image artifacts in the corrected gates we concentrate on these specific regions. One can identify voxels as part of image artifacts by looking at the difference between the reference gate \( I_R \) and the result of motion correction \( I_{MC} \) using the given OF algorithm. We use a threshold \( T > 0 \) for determining if a voxel is part of an image artifact or not. For large differences between \( I_R \) and \( I_{MC} \) we assume that motion correction failed for these voxels. We restrict this identification to the case of positive differences, because they lead to the unwanted image artifacts. With this threshold we can classify each voxel \( \bar{x} \) by the following indicator function:

\[
f(\bar{x}) := \begin{cases} 1, & \text{for } I_R(\bar{x}) - I_{MC}(\bar{x}) > T \\ 0, & \text{for } I_R(\bar{x}) - I_{MC}(\bar{x}) \leq T \end{cases}
\]

(2)

We have to choose \( T \) with respect to the SNR of the given gates. Otherwise differences induced by noise could lead to a wrong classification of these regions.

After we have identified the regions in which the BCC has been violated, we correct the surplus of intensity by redistributing it in the local neighborhood. This local intensity correction is done by adding the surplus of intensity to voxels within the range of the PVE featured by the probability that they are the source of this surplus. We use the image gradients of \( I_R \) to get these voxels. The higher the gradient in an adjacent voxel the higher the probability that this voxel spilled over intensity due to the PVE. This redistribution of intensity in a local neighborhood \( \Omega(\bar{x}) \) can be formulated by the following rule:

\[
I_{F,LIC}(\bar{y}) = I_F(\bar{y}) + \frac{G(\bar{y})}{S_G(\bar{x})} \cdot I_D(\bar{x}), \quad \text{for } \bar{y} \in \Omega(\bar{x})
\]

(3)

\( G(\bar{x}) := |\nabla I_R(\bar{x})| \) is the magnitude of the image gradient of \( I_R \) in a voxel \( \bar{x} \) and \( S_G(\bar{x}) := \sum_{\bar{y} \in \Omega(\bar{x})} G(\bar{y}) \) is the sum of gradients in the local neighborhood \( \Omega(\bar{x}) \). This formula is used for every voxel \( I_D(\bar{x}) := I_{MC}(\bar{x}) - I_R(\bar{x}) \) with \( f(\bar{x}) = 1 \). This gradient-based distribution of intensity assures that the sum of intensity keeps constant in a gate and that LIC is only performed within the range of the PVE for each voxel. The only necessary condition for the local intensity correction is an initial step of motion correction which results in a sufficiently corrected gate. The results of LIC can be used to correct the first step of OF estimation in an iterative scheme.

2.4. Iterative computation scheme

In order to reduce image artifacts in motion corrected gates we combine the chosen algorithm for OF estimation with the previously presented LIC method in an iterative computation scheme. The idea is to calculate the correct optical flow iteratively, while using the resulting gates of LIC as reference gates. After computing a first step of motion correction, we identify the erroneous regions in the result gate \( I^1_F \) with the indicator function \( f \). After that a local intensity correction is performed on \( I^1_F \) to reduce image artifacts. We define the result of this LIC as \( I^1_{F,LIC} \) and use it as new reference gate. In the next step we compute the OF between \( I^1_F \) and \( I^1_{F,LIC} \) and gain \( I^2_F \) from this. Fig. 2 demonstrates this iterative computation scheme.

It can be repeated for several iterations but leads to a loss of overall intensity in each step due to the usage of interpolation methods for motion correction. For this reason we combine the calculated flow fields to a global flow field afterwards that performs the same correction in one step instead of going through all iterations. Doing this limits the loss of intensity, because only one interpolation step is performed. Due to space limitations we do not go into details any further. Furthermore the number of performed iteration steps is limited with respect to the range of the PVE. Otherwise it was not be guaranteed that the overspill keeps in the region of its origin. Summed up, this method reduces the appearance of image artifacts in the motion corrected gates because in every step of the iteration scheme the distance between the result and the reference gate is reduced.

3. RESULTS

We validated our method by testing it on the ten cardiac gated PET data sets mentioned above. We used Cross-Correlation (CC) as similarity measure for the distance between the motion corrected gate and the reference gate to make our results
4. DISCUSSION

In this work we presented an iterative method that performs motion correction with the help of OF estimation while considering the influence of the PVE on the violation of the brightness constancy constraint. The robustness of our approach can be well demonstrated in the case of mice. Although the use of regularisers is a more common approach we were not yet able to formulate a regulariser that tackles the problems of PVE, which led us to the proposed heuristic method. In future we will deal with the problem of different amounts of intensity in the given gates, which is encountered in PET imaging. Additionally, we will try to incorporate the effect of the PVE more explicitly into our redistribution formula, so that the choice of threshold $T$ and the neighborhood size can be calculated. In conclusion we can say that the quality of PET images has improved by using the local intensity correction method, which led to better quantification results.

5. ACKNOWLEDGEMENTS

This work was supported by grants from the Deutsche Forschungsgemeinschaft (DFG), Sonderforschungsbereich 656 Münster, projects B2, B3.

6. REFERENCES


