Chapter 6

Final Results

In this chapter, the algorithm's performance will be analyzed, based on the lesion's location, volume and maximum voxel value on the compensated image as compared to the gated and uncorrected ones. Some registered images are displayed. The way in which the choice of the reference phase affects the method's performance will also be investigated.

6.1 Materials and methods

6.1.1 Materials

The gated PET/CT data set provided by [28] was utilized to evaluate the algorithms. Ten different PET/CT reconstructions corresponding to different respiratory phases are available. The uncorrected PET image can be estimated as an average of the ten available frames. This is the best approximation that can be made, as no further information about how much time is spent in average in each phase is available (which would make possible to weight the frames).

6.1.2 Methods

The proposed correction algorithm is based on transforming all of the PET frames (referring to different breathing phases) to a reference phase and then calculating the average. The transformations are found out from the corresponding CT images. It is then possible to compare this corrected image to the not-compensated average and to the "sharp" reference frame.

The first step was to crop the image to a 90x90x40 voxel volume around the lesion in order to reduce the computational cost of the operations. This box is big enough to easily make the lesion fit into it in every frame. The next step was setting to zero all the voxels that have a value of less than one third of the image's maximum vale. This proportion (one third) to threshold the lesion was arbitrarily defined. In the end, the lesion was isolated in the image (every non-zero voxel belongs to the tumor).

The following quality measures are employed to evaluate the performance:

1. The tumor's center of gravity: if the position in cartesian coordinates is regarded as a three-dimensional random variable (considering physical coordinates, not pixel ones), and the image's value in every point as the value of the probability density function, it is possible to calculate the distribution's mean. This mean represents the tumor center of gravity. The closer the center of gravity of a registered image to the reference one's, the better

2. The tumor's volume: this measure can be estimated by counting the number of thresholded voxels and multiplying it by the volume of one voxel. The closer the volume of the registered image to the reference one's, the better. As the effect of the respiratory motion on the images is to make them blurry, the tumor's volume appears to be larger in the uncorrected image. That is why it can be said that, the smaller the volume, the better.

3. The maximum voxel value: another important feature of a PET image that can be distorted due to respiratory motion blurring is the maximum voxel value (which is proportional to the maximum SUV, standard uptake value). The SUV is used for differentiation of malign and benign tumors. Checking if the algorithm can recover the maximum SUV (the blurring due to motion is supposed to decrease it) is also a good quality measure for the compensation algorithm.

The experiments were repeated using all the respiratory phases as reference in order to investigate if the algorithm's performance was independent of the choice for the standard phase.

6.2 Results

The quality measures for both the original and improved demons algorithm, configured as in the previous chapter (14-10-6 iterations, $\sigma = 0.75$ mm for the original algorithm, 7-5-3 with $\sigma = 1.75$ mm for the improved one), are shown in tables 6.1 and 6.2. Phases 0 and 9 are chosen as reference, as these frames represent the minimum and maximum in the breathing cycle. Phase 0 is in fact the closest one to full inhale.

	COG Deviation (mm)	max. value	Volume(cm^3)
Gated frame 9	0	4491	18.9
Uncorrected PET	6.05	3986	24.1
Comp. orig. demons	2.28	4244	22.6
Comp. impr. demons	2.16	4391	23.3

Table 6.1: Center of gravity deviation respect to frame 9, maximum voxel value and volume for the frame 9 itself, the uncorrected PET image and the "compensated" ones.

	COG Deviation (mm)	max. value	Volume(cm^3)
Gated frame 0	0	4502	18.0
Uncorrected PET	9.39	3986	24.1
Comp. orig. demons	2.75	4268	21.2
Comp. impr. demons	3.22	4370	22

Table 6.2: Center of gravity deviation respect to frame 0, maximum voxel value and volume for the frame 0 itself, the uncorrected PET image and the "compensated" ones.

The tables show how the center of gravity displacement is reduced between a 60% and a 70%. The volume increment is reduced (roughly, and in average) from 5.5 cm^3 to 3.75 cm^3, or a 33%. The tables also show that the original demons algorithm performs slightly better than the improved one.

The voxel values are very important because, as already mentioned, they help the doctors to tell apart malign and benign tumors. The same tables and the image histograms (figure 6.1) show that the uncorrected image has the bins packed to the left. In fact, there are no bins beyond 4000, when the sharp frames would reach approximately 4500. One can appreciate how the compensated versions tend to stretch the histogram again (although they do not reach 4500, but intermediate values around 4350-4400). This time, it is the improved algorithms that performs slightly better than the original one.

Finally, in order to evaluate the algorithm's robustness against noise, the experiments where repeated with all the phases as reference. The quality measures depending on the chosen phase are shown in figures 6.2, 6.3 and 6.4.

The center of gravity deviation is quite similar for both the original and the improved demons algorithm, and tends to grow towards the phases that correspond to the full inhale and full exhale positions. This was expected, as these positions lead to larger average distances between the lesion in the images and thus to more difficult registration problems. In the phases where

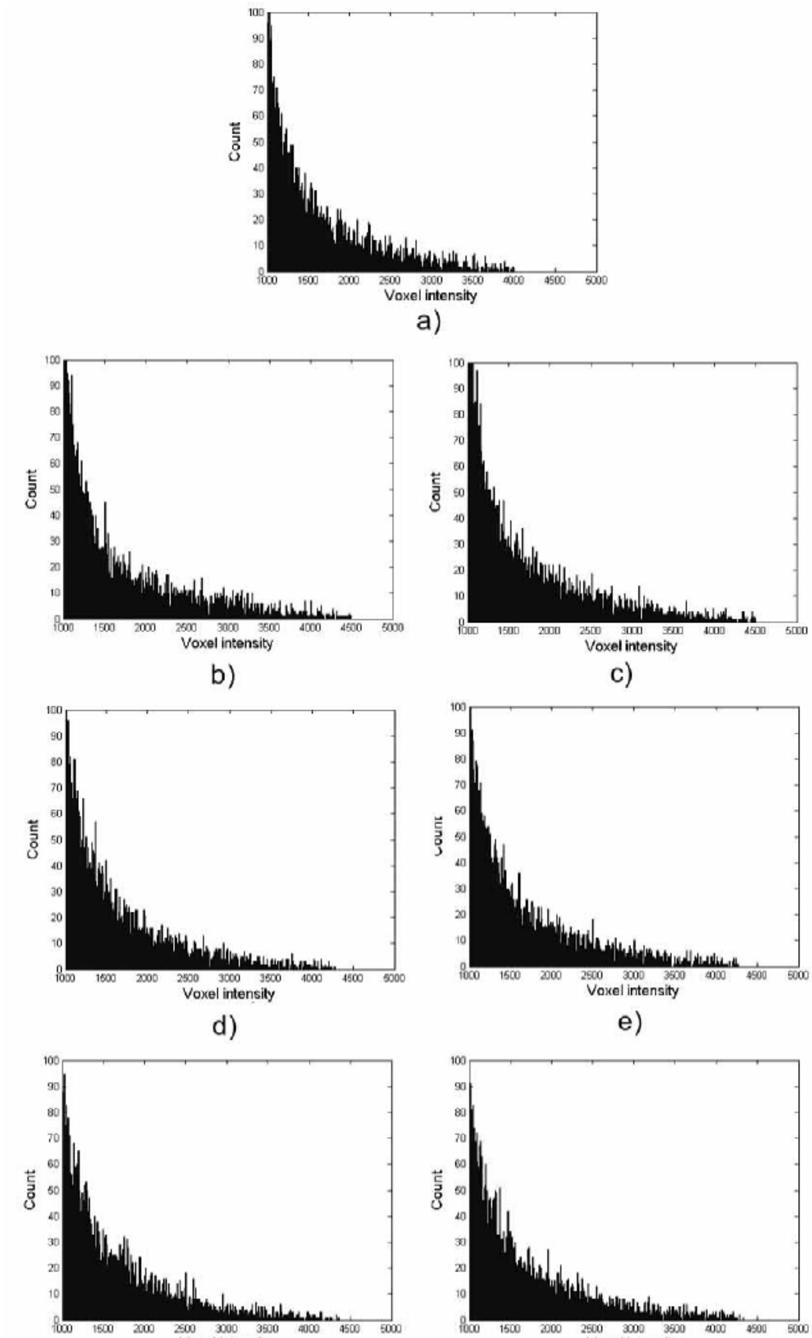

Figure 6.1: Histograms for the voxel values in the lesion: a) Blurry PET b) Frame 9 c) Frame 0 d) Original demons to frame 9 e) Improved demons to frame 9 f) Original demons to frame 0 g) Improved demons to frame 0.

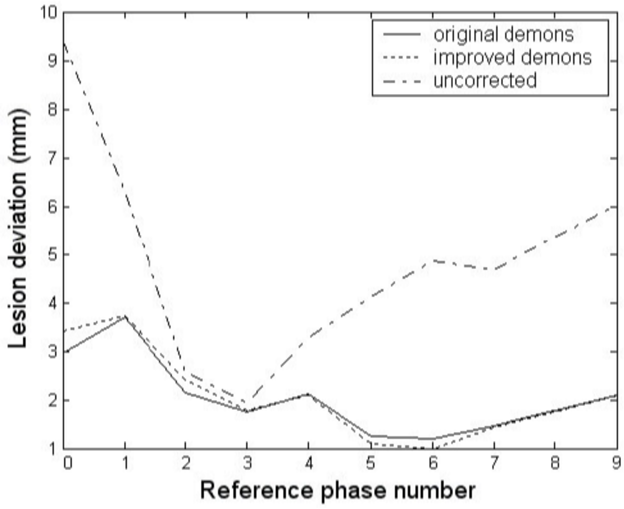

Figure 6.2: Distance between the centers of gravity of the reference frame and the compensated image, for the original and the improved demons algorithm, as well as for the uncorrected blurry PET.

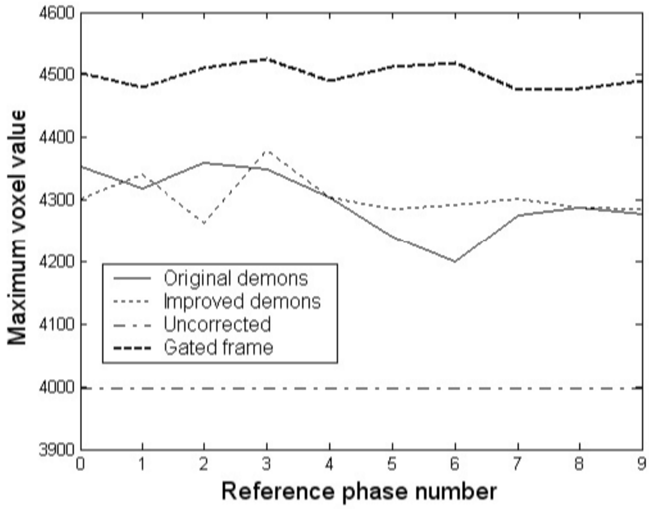

Figure 6.3: Maximum voxel value value for the compensated image depending on the chosen reference frame, for the original and the improved demons algorithm, as well as for the uncorrected blurry PET and gated frames.

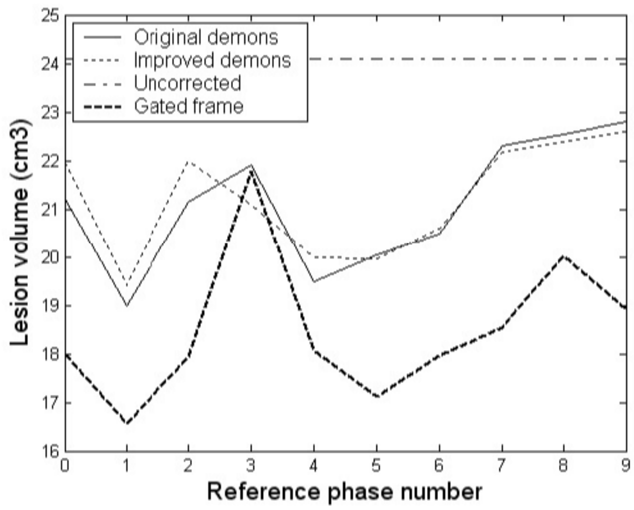

Figure 6.4: Lesion volume for the compensated image depending on the chosen reference frame, for the original and the improved demons algorithm, as well as for the uncorrected blurry PET and gated frames.

the uncorrected image's center of gravity is closer to the gated image's, the improvement with the algorithm is almost non-existent.

The maximum voxel value has a quite irregular behavior with the reference phase (probably due to the noisy nature of PET images), fluctuating around values close to 4250 (halfway between the uncorrected and sharp PET images). The improved demons algorithm outperforms the original one in most of the cases.

The lesion volume is also very similar for both algorithms. It is far from being constant with the chosen reference phase. This is because of the tumor's volume being different in every gated PET image, when it should not. The noisy nature of PET is again to be blamed.

6.3 Conclusions and sample images

The results corroborate the expectations from the CT co-registration chapter: the resulting transforms can be applied for compensating correlated PET data. The tumor's volume in the corrected images decreases compared to the uncorrected one, and the uptake value increases. Moreover, the tumor position is brought much closer to the real position than in the uncorrected image. As the effects of the blurring due to respiratory motion are to increase the tumor's volume, decrease the uptake value, and change the tumor position, it can be stated that they are partially compensated with the algorithm proposed in this master's thesis.

The described partial represents an important benefit when radiotherapy is applied, as the region to radiate will become smaller, decreasing the amount of unnecessarily radiated healthy tissue. The physician can moreover choose the reference phase that he wishes, as the results are satisfactory for every respiratory phase (although they are better in the phases halfway between full inhale and full exhale).

Another promising feature of the results is that they were not obtained with images from an hybrid PET/CT scanner. It can hence be expected that they will improve when such a machine is used for the acquisition.

The most problematic aspect of the results is that they are unfortunately based on just one data set. Further experiments with more image samples would have to be performed in order to validate the algorithm.

To finish this chapter, some image samples are shown in figures 6.5 and 6.6, and in their zoomed versions 6.7 and 6.8. They show the sharp, uncorrected and compensated versions for frames 0 and 9.

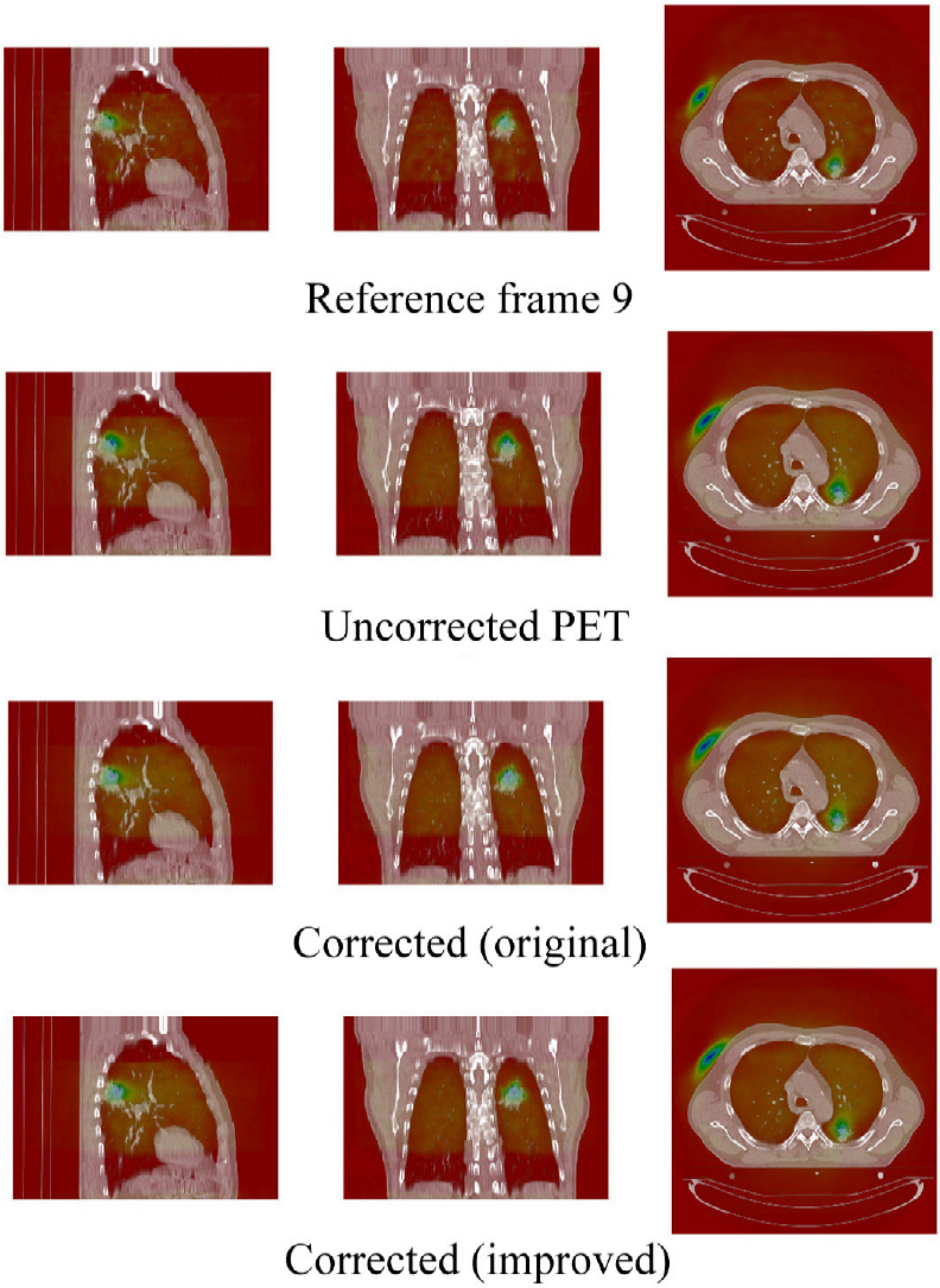

Figure 6.5: Sharp CT at phase no. 9 fused with the gated PET at the same phase, the uncorrected one, and the compensated ones.

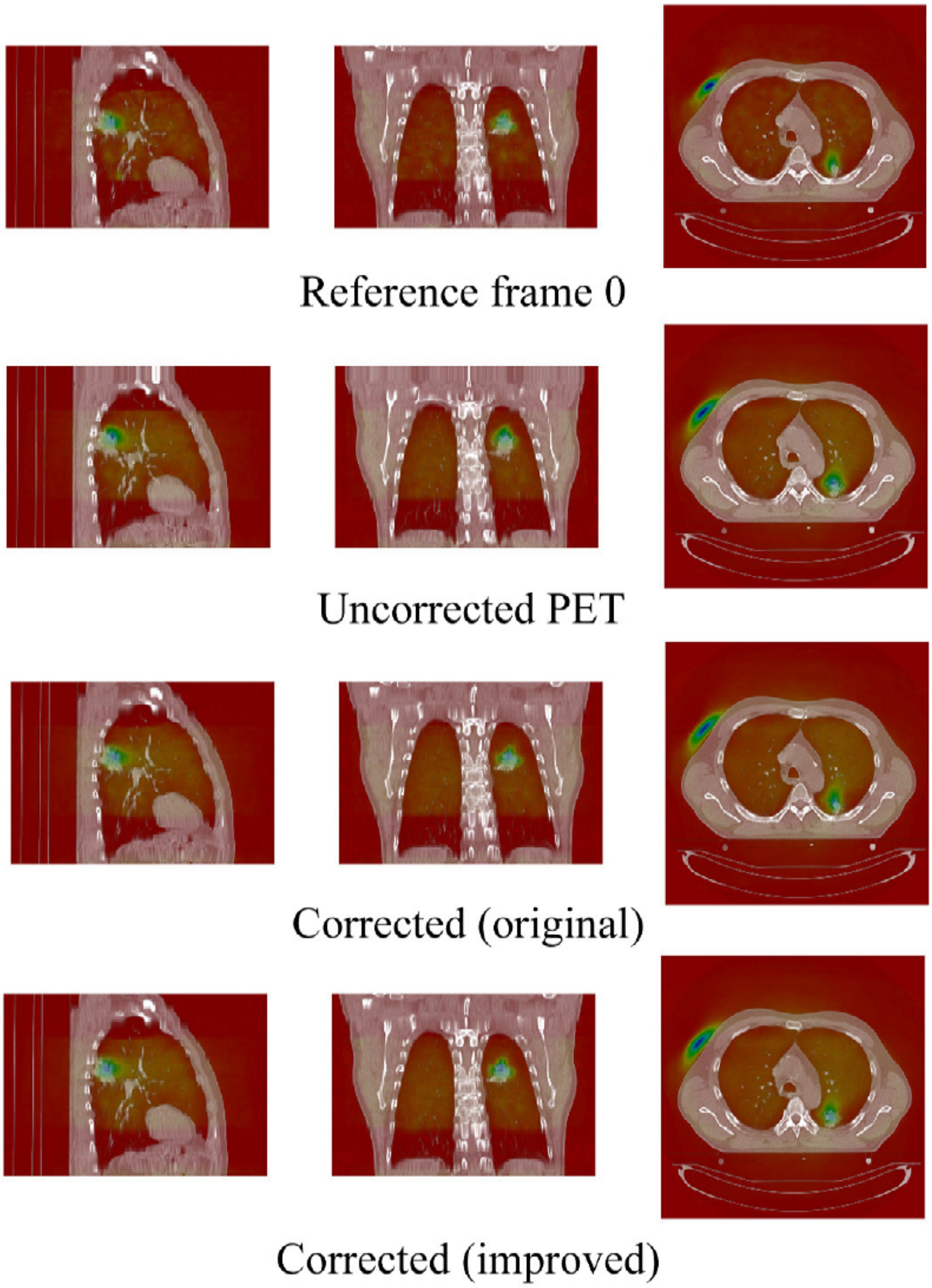

Figure 6.6: Sharp CT at phase no. 0 fused with the gated PET at the same phase, the uncorrected one, and the compensated ones.

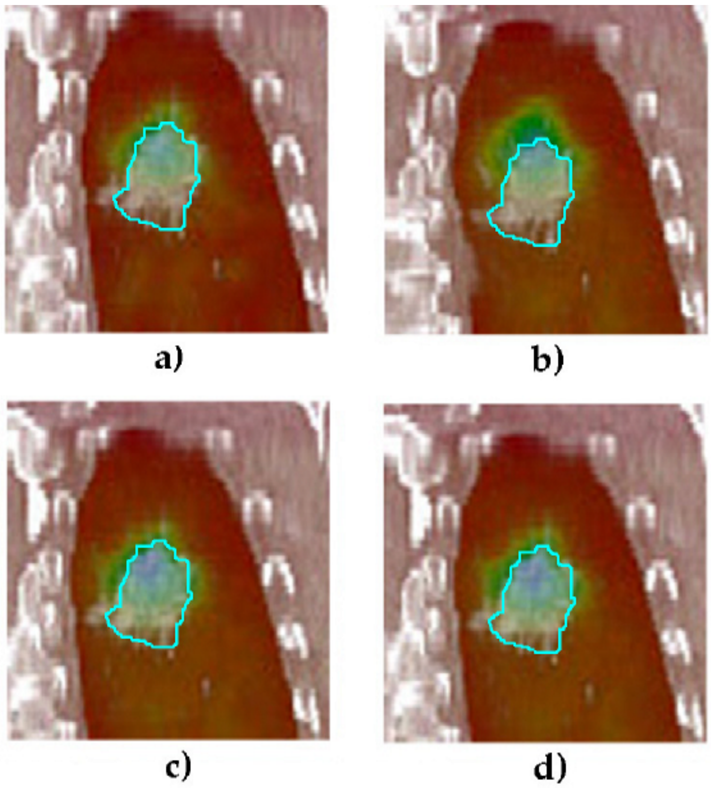

Figure 6.7: Zoomed lesion from the coronal views of figure 6.5, with the tumor delineated in blue: a) Sharp frame b) Uncorrected PET c) Demons (original) d) Demons (improved)

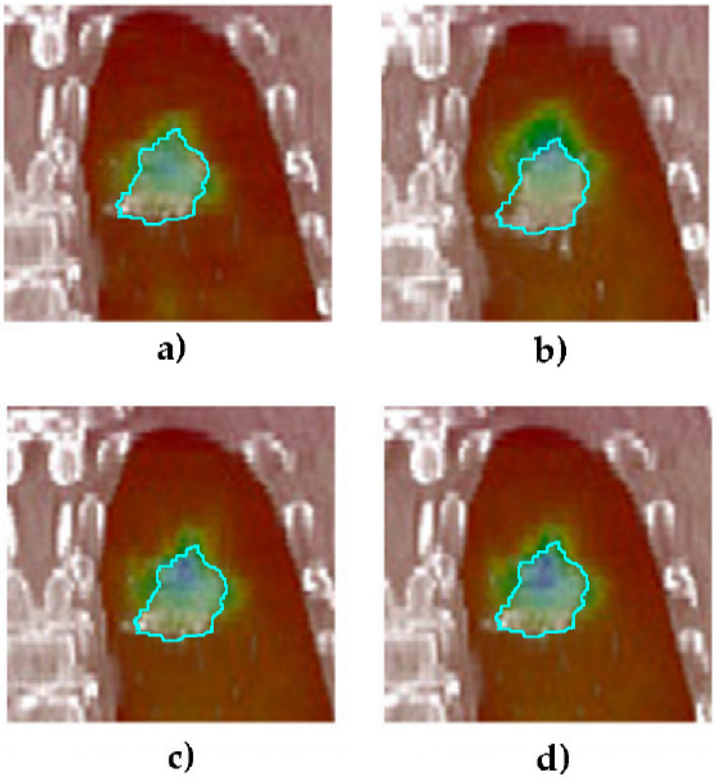

Figure 6.8: Zoomed lesion from the coronal views of figure 6.6, with the tumor delineated in blue: a) Sharp frame b) Uncorrected PET c) Demons (original) d) Demons (improved)