Edge-based Robust Image Registration for Incomplete and Partly Erroneous Data*

Piotr Gut[1], Leszek Chmielewski[1], Paweł Kukołowicz[2], and Andrzej Dąbrowski[2]

[1] Institute of Fundamental Technological Research, PAS,
Świętokrzyska 21, PL 00-049 Warsaw
pgut@ippt.gov.pl; lchmiel@ippt.gov.pl
[2] Holycross Cancer Centre,
Artwińskiego 3, PL 25-734 Kielce
PawelKu@onkol.kielce.pl; AndrzejDa@onkol.kielce.pl

Abstract In image registration it is vital to perform matching of those points in a pair of images which actually match each other, and to postpone those which do not match. It is not always known in advance, however, which points have their counterparts, and where are they located. To overcome this, we propose to use the Hausdorff distance function modified by using a voting scheme as a fitting quality function. This known function performs very well in guiding the matching process and supports stable matches even for low quality data. It also makes it possible to speed up the algorithms in various ways. An application to accuracy assessment of oncological radiotherapy is presented. Low contrast of images used to perform this task makes this application a challenging test.

Keywords: image registration, Hausdorff distance, robust statistics

1 Introduction

The availability of numerous imaging modalities makes it possible to visualise in a single image the phenomena available in different modalities, if the images coming from them are precisely overlaid on one another, or *registered*. The choice of landmarks in the images to be used as nodes for registration seems to be the most fundamental task. Contents of the two registered images can differ in such a way that it is difficult to state which particular pixel in one image should be matched with which one in the second image.

Quality assessment of the teleradiotherapy of cancer is an important application which has all the features of a very challenging image registration task. In an ideal treatment, the positions of the irradiation field and the shields with respect to the anatomical structures of the patient conform to the planned ones and are constant in all the radiotherapy sessions. A change of the applied dose by one per cent can lead to a change in time to local recurrence of the disease by as much as several percents [1]. Imprecise realisation of the geometry can imply the loss of chance for permanent remission of the disease or to severe post-irradiation effects.

* W. Skarbek (Ed.): CAIP 2001. LNCS 2124, pp. 309-316, 2001.
© Springer-Verlag Berlin Heidelberg 2001

To assess the accuracy of the treatment of cancer it is necessary to compare the planned geometry with the actual location of the irradiation field and anatomical structures. The actual geometry in a specified therapeutical session can be recorded in the *portal image*. The planned geometry is recorded in the *simulation image*, routinely made during therapy planning. The *simulation image* should be registered with each of the *portal images* made. The simulation image is an X-ray of high quality (see Fig. 1a for an example). The portal image is produced by the therapeutical beam of the ionising radiation, and is inherently of low contrast due to that different tissues, like bones and muscles, attenuate the radiation very similarly (Fig. 1b).

The literature on image registration in the context of radiotherapy refers to the portal images made with beams generated in accelerators rather than with the cobalt apparata. In Poland more than a half of patients are treated with cobalt, which produces the portal images of even worse quality. In any case, the features to be compared are difficult to find in the portal images, and the problem of missing and spurious features arises. In consequence, the comparison of geometries in these images is difficult and time-consuming.

Although full automation of the method has been attempted, as for example in [9,10,11,12], the majority of algorithms lack generality. In [9] the rectilinearity of edges of the irradiation field is used. In [12] the mutual rotation of images is not taken into account. Satisfactory results were received only in the case of pelvis [10], where edges of thick bones are clearly visible. In other localisations the registration of the simulation and portal images is generally a difficult task, even for humans, due to lack of clear natural landmarks. Algorithms in which landmarks to be matched should be shown manually are also in use [5,6]. Apart from the case of the portal films, the image registration literature is extremely broad. The surveys can be found in [4,13,19].

In the literature, the main effort seems to go to the proper choice of landmarks. The approach proposed here resolves this by applying the partial Hausdorff distance [14,18] as the registration accuracy measure. This measure removes the necessity of finding the corresponding landmarks. All that is needed is to specify the sets of pixels to be matched in both images. Moreover, it is allowed for these sets to contain incomplete and partly erroneous data. Such an approach makes it easier to build an image registration system with more adaptivity than systems using strict correspondence. In the method proposed here, it is not necessary for the software to exhibit a deep understanding of the contents of the images, like the ability to recognise any specific anatomical details. Hence, the complexity of the system is not excessively large.

In the present clinical practice the assessment of the accuracy of radiotherapy is done manually by an experienced observer. As a rule, this tedious procedure is not performed routinely. There is a need to support and objectify this process. The methodology presented below is general and requires little user intervention. However, full control of the physician over the process will be maintained, according to the requirement of a human decision in the therapeutical process.

At the time of preparation of the paper the studies were at the stage of stabilisation of the algorithms and the first series of experiments. The methodology will be thoroughly tested, first in laboratory tests, with phantoms used in radiological practice, then in clinical tests in the Holycross Cancer Centre in Kielce.

2 The method

The following description of the proposed method will be structured according to the classification used in [4]. The practical information, like the hints on the values of parameters, will be given in Chapt. 4.

2.1 Features to be matched

Edges can be considered as the most natural and easily detectable features to be used as landmarks for matching. Edges, classically understood as loci of largest brightness gradient, can be very efficiently found with the *zero-second-derivative* filter. The location is the pixel where the second derivative along the gradient direction changes its sign, and the intensity is the gradient modulus. This typical filter was modified by scaling up its masks and convolving them with a circular-symmetrical Gaussian, with the deviation chosen such that the 3σ circles are tangent to each other. The scale should be chosen suitably to the level of noise in the images and the scale of the edges sought.

The resulting edges are one-pixel wide. The thresholding was complemented with an edge-following algorithm which lets the thresholded edges prolong to their neighbouring nonzero pixels from the non-thresholded image, with the ability to jump over gaps up to two pixels wide, along the direction close to that of the preceding edge fragment. The threshold is selected manually. There is a possibility of manually selecting the most relevant parts of edges of the chosen structures – anatomical structures or irradiation field. In the portal image, the edges of the anatomical structures are more easily found if the image is enhanced with a typical histogram equalisation procedure (see Fig. 1b, d).

2.2 Transformation space

Affine transformation is used. This resolves to five parameters. According to the day-to-day radiological experience, this mathematically simple transform is enough for the application. An iterative algorithm described below is used to find the optimum transformation. Throughout the iteration process the current transformation is always calculated starting from the original image, which reduces the geometrical distortions of the images to a negligible level.

2.3 Measure of the registration accuracy

For the sake of robustness of the whole registration process, the partial Hausdorff distance measure was used. This measure was proposed in [14,18], and then used in various applications, including those closely related to image registration (*e.g.* [7,16]). This unsymmetrical measure of similarity of two sets of points has an intrinsic property of neglecting the *outliers* without biasing the result. Let B be the base set and O the overlaid set, and let $d(o, b)$ be the Euclidean distance between points $o \in O$ and $b \in B$. The partial Hausdorff distance is

$$H^r(O, B) = Q^r_{o \in O} \min_{b \in B} d(o, b) \ , \tag{1}$$

where $Q^r_{x \in X} g(x)$ is the *quantile* of rank r of $g(x)$ over the set X, $r \in \langle 0, 1 \rangle$. In the considered application, B and O are the discrete sets of edge pixels in the simulation and the portal image. Let us explain the notion of the quantile for this discrete case in th following way. Note that for each pixel of the overlaid set $o \in O$ we have one measurement of distance; let us call it $m(o) = \min_{b \in B} d(o, b)$. We can sort these measurements for all the set O in an ascending order. Now, the quantile of rank r can be defined as the s-th smallest element $m(o), o \in O$, where $s = r/|O|$, and $|O|$ is the power of the set O – the number of measurements.

For example, setting r to 0.60 in Eq. (1) can be understood as follows: "see what is the maximum distance for which 60% of the points in the overlaid set vote". Hence, the partial Hausdorff distance can be considered as a kind of a *voting scheme*, in which the *outliers* are rejected and the *inliers* are taken into account in a natural way.

A formal definition of the quantile for the discrete case is as follows. Let v be a discrete random variable with the probability density $P(v)$. Then,

$$Q^r v = q : P(v \leq q) \geq r \ \wedge \ P(v \geq q) \geq 1 - r \ . \tag{2}$$

Here, $v = m(o)$. In the calculations, the density $P(v)$ is replaced by the frequency $F(v) = F[m(o)], o \in O$, which is represented by a histogram. Finding the histogram is more effective than sorting. Moreover, the minimum distance of a given pixel $o \in O$ can be directly read out from the pre-calculated distance transform of the edge image B ([2,8,17], see *e.g.* [7,15] for similar applications).

2.4 Strategy of search in the space of transformations

At present we apply a maximum gradient optimisation, starting from a reasonable initial state, provided manually by the user by "dropping" the overlaid image onto the base one. In spite of the obvious drawback of possibly falling into local minima, this simple method performs sufficiently well. The algorithm starts with the quantile rank $r = 1.0$. Together with the application of the distance transform, this resembles the chamfer matching technique [3], applied in [7], but the similarity ends here. After a minimum is reached, the rank is reduced by a small value (say, 0.01), and the algorithm goes on, until the rank reaches zero or the distance becomes zero. This procedure exhibits some analogy to simulated annealing, with the quantile rank related to temperature.

As the quantile rank r is reduced in the optimisation algorithm, the registering transformation stabilises and the result tends to an optimum. However, when the rank approaches zero, the number of edge pixels considered in the matching is going down, which can degrade the result. The approach we chose to find the optimum is as follows. During iterations, a series of registration versions is received, each of them easy to store in a form of the transformation matrix. The set of results for the last iterations of the optimisation process is presented to the user, and this is the user who chooses the best one. This eliminates the unreasonable results and also has a very important *virtue*: the final choice is made by the user. This is in line with the requirement of letting the most important decision be made by the clinician, not the software.

3 Example

Let us now present a practical example of calculations performed for a pair of images of the pelvis, shown in Fig. 1, both 500×500 pixels large.

The simulation image of Fig. 1a is a relevant fragment of the therapy simulation image, and the portal image b shows the irradiation field and some of the anatomical details inside it. Figs. c and d show the respective edge images, both received with the scale 4. This corresponds to $3\sigma = 4$ pixels, and the filter mask of 9×9 pixels. The edge images were thresholded. For registration, the edges of the anatomical structures (bones) considered as relevant were selected by the user. In Fig. f there is the result of registration which started from the initial state of Fig. e, with the reference image taken as the base one, and the portal image as the overlaid one (see Eq. (1)). Note that both images contain the edge fragments having no counterpart in the other image, which did not influence the quality of the result.

The result of the 174th iteration of 177 was chosen as the best one, with the distance measure 2.00 and the rank $r = 0.54$. The final 177th iteration gave zero distance at $r = 0.20$ after about 50 s, on an 800 MHz Pentium III processor.

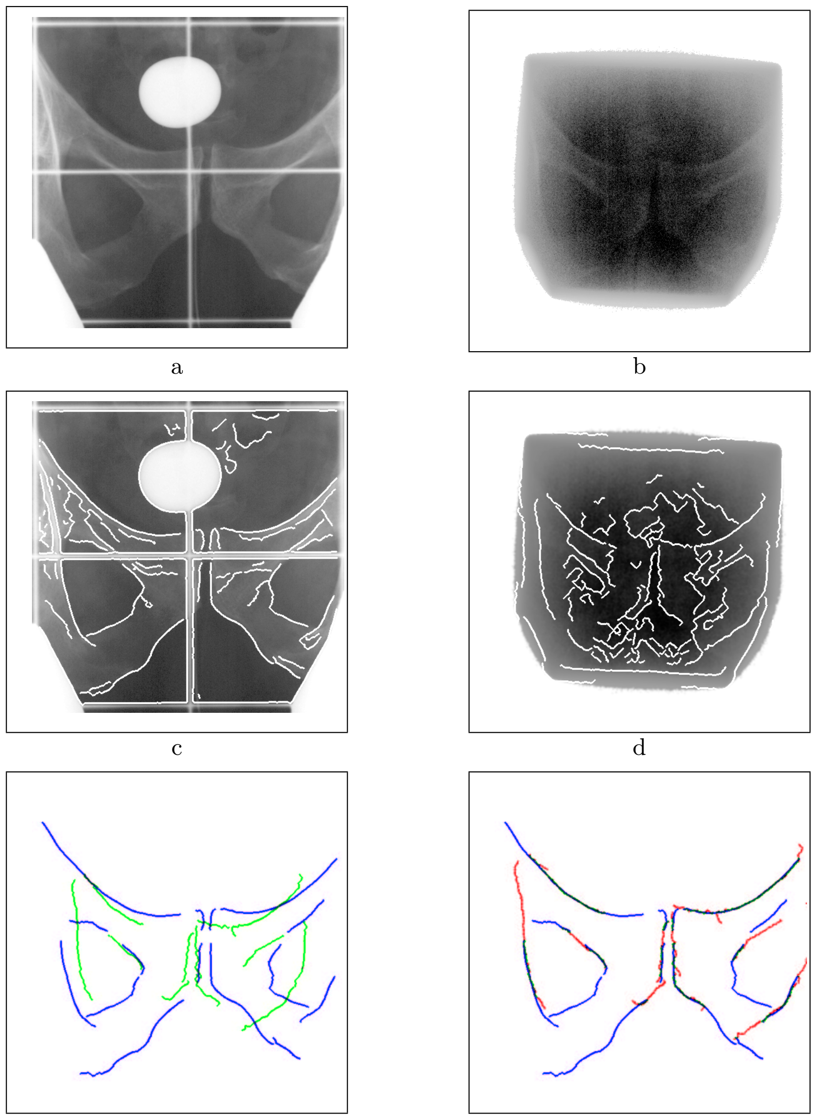
4 Practical hints

The following practical hints can be formulated on the basis of the experience with the described method gathered until now.

Roles of images In general, not always the more accurate simulation image should be used as the base one, in the meaning of Eq. (1). As the base image the one having a larger number of edge pixels *presumably* having the counterparts, in reference to all its edge pixels, should be taken.

Scale and thresholds of the edge detector The scale should be chosen suitably to the scale of the edges sought. For the structures in the simulation image and the shields in the portal image the scales can be small (2-3). For the edges of bones in the portal image, scale should be larger (*cf.* [9]). Obviously, larger scale means less noise and fewer details.

The thresholds should be chosen to receive a maximum subjective ratio of visibility of important edges with respect to irrelevant ones, to reduce further



a b c d e f

Figure 1. Example of results for the case of pelvis. a, b: source images – simulation and portal, respectively (portal image enhanced by histogram equalisation). c, d: edges after thresholding, overlaid on source images (portal image Gaussian filtered). e, f: input and result of registration of the relevant fragments of edges of anatomical structures manually selected from those of c and d. Base, portal image: blue; overlaid, simulation image: green – matched pixels (*inliers*), red – pixels rejected from matching (*outliers*). Contrast enhanced and edges thickened for better visualisation.

manual editing of the edge images. High thresholds are used to find the edges of the irradiation field, which is very clear in the portal image. In the simulation image the edges of the irradiation field is represented by wires, also clearly seen (see Fig. 1a). For anatomical structures lower thresholds are used.

For a given anatomical localisation and constant imaging parameters, the scales and thresholds found will be valid for large series of images.

5 Further developments

Some of the above hints and the hitherto experiments with the method lead to the concepts which will be considered for use in the further work.

Directional information on edges This will be utilised in the distance function to improve the reliability of matches. It seems that this enhancement will be necessary in the case of more complex images like those for the head, neck or breast, but only if the portal images are of sufficiently good quality.

Pre-calculated distance maps for virtual elementary transformations Assuming the centre of rotation and scaling is fixed with respect to the base image, from such maps the maximum distances for pixels of transformed images can be read, without performing the transformations. This is profitable if a number of steps made is enough to compensate for the difference in time of calculating a transformed image, which is quick, and pre-calculating a distance map, which is slower. The described speed-up was already implemented and tested, but the above mentioned assumption is not always true. Further investigation of the applicability of this concept is necessary.

Hierarchical algorithm This speed-up and the pre-calculation of the distance maps for virtual transformations, proposed above, will be used only if calculation time with a processor available when the final version of the software is delivered is excessively long.

6 Conclusion

The image registration algorithm using edges as landmarks and the modified Hausdorff distance as the measure of the registration accuracy performs reasonably well in the application to the quality assessment of the teleradiotherapy of cancer. Required user intervention is reduced to a necessary minimum. The present experience and the concepts worked out in the hitherto experiments make it possible to develop a software tool to support and objectify the otherwise difficult and tedious process of precisely comparing the location of anatomical structures and irradiation field during the radiotherapy sessions with the planned geometry. The software tool emerging as the method matures will be tested first with images received from human-like phantoms, and later in clinical practice. Detailed information on the accuracy of the method will be available then.

Acknowledgement The work is supported by the Committee for Scientific Research, Poland, under the grant no. KBN 4 P05B 064 18.

References

1. P. Aaltonen-Brahme, A. Brahme, et al. Specification of dose delivery in radiation therapy. *Acta Oncologica*, 36(Supplementum 10), 1977.
2. G. Borgefors. Distance transformations in digital images. *Comp. Vision, Graph., and Image Proc.*, 34(3):344–371, 1986.
3. G. Borgefors. Hierarchical chamfer matching: A parametric edge matching algorithm. *IEEE Trans. PAMI*, 10(6):849–865, 1988.
4. L. Gottesfeld Brown. A survey of image registration techniques. *ACM Computing Surveys*, 24(4):325–376, 1992.
5. J. Cai, J. C. H. Chu, A. Saxena, and L. H. Lanzl. A simple algorithm for planar image registration in radiation therapy. *Med. Phys.*, 25(6):824–829, 1998.
6. J. Cai, S-Q. Zhou, J. Hopkinson, A. V. Saxena, and J. Chu. Alignment of multi-segmented anatomical features from radiation therapy images by using least squares fitting. *Med. Phys.*, 23(13):2069–2075, 1996.
7. D. Chetverikov and Y. Khenokh. Matching for shape defect detection. In *Proc. Conf. Computer Analysis of Images and Patterns (CAIP'99)*, volume 1689 of *Lecture Notes on Computer Science*, pages 367–374, Ljubljana, Slovenia, Sept 1999. Springer Verlag.
8. P. E. Danielsson. Euclidean distance mapping. *Comp. Graph. and Image Proc.*, 14:227–248, 1980.
9. K. Eilertsen, A. Skretting, and T. L. Tennvassas. Methods for fully automated verification of patient set-up in external beam radiotherapy with polygon shaped fields. *Phys. Med. Biol.*, 39:993–1012, 1994.
10. K. G. A. Gilhuijs, A. A. H. El-Gayed, M. van Herk, and R. E. Vijlbrief. An algorithm for automatic analysis of portal images: clinical evaluation for prostate treatments. *Radiotherapy and Oncology*, 29:261–268, 1993.
11. K. G. A. Gilhuijs and M. van Herk. Automatic on-line inspection of patient setup in radiation therapy using digital portal images. *Med. Phys.*, 20(3):667–677, 1993.
12. L. M. Giraud, J. Pouliot, X. Maldague, and A. Zaccarin. Automatic setup deviation measurements with electronic portal images for pelvic fields. *Med. Phys.*, 25(7):1180–1185, 1998.
13. S. R. Arrige H. Lester. A survey of hierarchical non-linear medical image registration. *Pattern Recognition*, 32:129–149, 1999.
14. D. P. Huttenlocher and W. J. Rucklidge. A multi-resolution technique for comparing images using the Hausdorff distance. In *Proc. IEEE Conf. on Computer Vision and Pattern Recognition*, pages 705–706, New York, Jun 1993.
15. D. Kozińska, O. J. Tretiak, J. Nissanov, and C. Oztruk. Multidimensional alignment using the Euclidean distance transform. *Graphical Models and Image Proc.*, 59(6):373–387, 1997.
16. D. M. Mount, N. S. Netanyahu, and J. Le Moigne. Efficient algorithms for robust feature matching. *Pattern Recognition*, 32:17–38, 1999.
17. A. Rosenfeld and J. Pfaltz. Distance functions on digital pictures. *Pattern Recognition*, 1:33–61, 1968.
18. W. J. Rucklidge. Efficiently locating objects using the Hausdorff distance. *Int. J. Comput. Vision*, 24(3):251–270, 1997.
19. P. A. van Elsen, E. J. D. Pol, and M. A. Viergever. Medical image matching – a review with classification. *ACM Computing Surveys*, 24(4):325–376, 1992.