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ASSESSMENT OF THE QUALITY OF RADIOTHERAPY WITH THE USE OF PORTAL AND SIMULATION IMAGES – THE METHOD AND THE SOFTWARE[1]

Quality assessment in external beam radiotherapy necessitates for an efficient and robust tool for comparing the planned and realised geometry of the treatment. Such a tool using the modified Hausdorff distance measure has been developed and successfully introduced to clinical practice. The majority of steps of the method are automatic. The user specifies the share of edge data extracted from the images to be used in calculations, not *before*, but *after* these calculations are actually performed. Thus, the results of the choice can be seen immediately. This mechanism makes the method extremely robust against partially erroneous or missing data.

1 MOTIVATION

Introduction of quality standards in radiotherapy of cancer makes it necessary to assess the quality of each session of the radiation therapy. In the case of radiotherapy with external beams, this assessment comprises the measurement of precision with which the geometry of the treatment conforms to the planned geometry of the system formed by the relevant part of the patient's body, the beam of the therapeutical radiation and the additional equipment, like the shields or bolus. Irradiation accuracy is of critical importance for the result of the treatment [1].

The actual geometry in a specified therapeutical session is recorded in the *portal image*. The planned geometry is recorded in the *simulation image*, normally made before the therapy begins. The simulation image should be registered with each of the portal images, made during each of the therapeutical sessions. The simulation image is an X-ray of high quality. 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. The registration of the simulation and the portal image is therefore a non-tivial task.

Designing a general-purpose image processing software to assist the radiotherapy physicists in the process of quality assessment of the radiation therapy of cancer is still a challenge. One of the major problems is to overcome the unavoidable presence of errors in the data derived from the registered

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images. These errors consist in that some features in one image have no counterparts in the other image, and hence, should not be registered. Before the registration is done, it is not obvious which are these features.

Full automation of registering the simulation and portal images, with edges used as the matched features, has been attempted, as for example in [8, 9, 10, 11]. In some methods the special features of the edges of the irradiation field are used to ease the process, as in [8], where the fact that the irradiation field has rectilinear edges is utilised. Satisfactory results are easier to receive in the case of pelvis [9, 11], where edges of thick bones are easy to detect. Algorithms in which landmarks to be matched should be shown manually were also recently used in [5]. The literature on the general methodology on image registration is extremely broad. The surveys can be found in [4, 14, 17].

In this paper we shall describe the method and the software for measuring the geometrical accuracy of the treatment. The image registration method used is based on the concept of the Hausdorff distance measure, according to [13, 16]. This measure has been modified in that its single parameter is evolving in a way to make the method more stable and user-friendly. Other applications of the distance measures based on the Hausdorff distance, including those closely related to image registration, can be found for example in [6, 15].

2 THE MEASUREMENT OF IRRADIATION ACCURACY

The aim of the registration is to find out if and how the irradiation field moved with respect to the anatomical structures of the patient, if the simulation image is taken as the reference state, and the portal image is treated as the current state. To this end, the initial registration is performed by overlaying the edges of the irradiation field in both images. This state is considered as the initial state for registration of the edges of the anatomical structures of the patient. The parameters of the movement of the portal image from this initial state, to the state in which the edges of the anatomical structures are registered, are the result of the measurement. Obviously, if the actual geometry of the treatment, represented in the portal image, is an accurate replica of the geometry planned in the simulation image, than after registration of the irradiation field edges, also the edges of the anatomical structures should be registered accurately. Hence, any inaccuracy of the registration of the anatomical structures of the patient, after the field edges are registered, accounts for an error in the geometry of the therapy.

The registration of the field edges in both considered images is the compensation of the differences in the geometries of the imaging system of the simulator and the portal. Thanks to this compensation, the registration of the anatomical features is always starting from a *reasonable initial state*. This state is easy to find and can be fully automatised, as the irradiation fields have simple shapes. The same method is used for registering both used sets of features.

2.1 THE REGISTRATION METHOD

The proposed registration method, presented previously in [12] in its less advanced forms, will be briefly desribed here. The description will be based on the classification used in [4].

The matched features

As said before, edges are considered as the most natural and easily detectable features to be used as landmarks for matching. The information on correspondence of landmarks in the registered images is not necessary.

Transformation space

Affine transformation is used. According to the day-to-day radiological experience, five parameters are enough to express the deformation of images necessary for registration: two movements, two scalings and one rotation.

Measure of the registration accuracy

The modified Hausdorff distance measure is used. This measure, proposed in [13, 16], is an unsymmetrical measure of similarity of two sets of points. The measure has a very desirable property: it postpones the data which can be treated as the *outliers* in the data set, while the *inliers* are taken into account. This is done by the method itself, without any user intervention. The robustness of the measure towards the partially missing or erroneous data and its relatively simple structure makes it an efficient image registration tool, also beyond the specific but challenging application considered here.

Let B be the base set and O the overlaid set of points, 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, $r \in (0,1\rangle$, of the variable $g(x)$ over the set X. In the considered application, B and O are the discrete sets of edge pixels in the simulation and the portal image, respectively.

Let us explain the notion of the quantile for this discrete case in the 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 $m(o)$, $o \in O$. Namely, $s/|O| < r < (s+1)/|O|$, and $|O|$ is the power of the set O, that is, the number of measurements. The values $m(o)$ can be very quickly found if the distance transform of the set B is precalculated, as described in [2]. If $r = 1$, the well known chamfer matching technique is obtained.

The formal definition of the quantile for the discrete case is as follows ([3], p. 785). Let v be a discrete random variable with the probability $P(v)$, and the probability measure of a set $\tilde{P}\{v : \text{cond}(v)\}$, where $\text{cond}(v)$ is an arbitrary condition on v. Then

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

Here, $v = m(o)$. In the calculations, the probability $P(v)$ is replaced by the frequency $F(v) = F[m(o)]$, $o \in O$, which is represented by a histogram. Then, the probability measure of a set $\tilde{F}\{m : \text{cond}(m)\}$ is the sum of the histogram cells for the set of measurements for which $\text{cond}(m)$ holds. Finding the histogram is more effective than sorting.

The smaller the function $H^r(O,B)$, eq. (1), the better the registration of the sets of edge pixels O and B. This function is actually the *dissimilarity* function, and should be minimised. Note that it has a single parameter – the quantile rank r. The choice of this parameter influences the result, as r is actually the *share* of pixels in the overlaid set O – the edges in the portal image – which are taken into account in the calculations as *inliers*, while the remaining pixels are treated as *outliers*. This share is not known before the registration is done, however. In the literature there are examples of an *a priori* choice of r (see for example [6, 15]), equivalent to the requirement that the *majority* of the results vote: slightly more than 50%, or 60%. As described in [12], we resign of this arbitrariness. The method for this is described in the next paragraph.

Strategy of search in the space of transformations

At present the quickest descent optimisation is used to find the minimum of the dissimilarity function.

To find the right value of the quantile rank, or the share of inliers r, the optimisations should be performed for various values of $r \in (0,1)$, and the best result should be chosen. To avoid this, in the beginning of the optimisation r is set to the "reasonable initial value" of $\alpha_1 \in (0.5,1\rangle$ and the minimum is found for this value of r. If the dissimilarity measure is zero (which is very unlikely), the algorithm stops. Otherwise, r is reduced by a small value, say, 0.01, and a new minimum is found. This is iterated until either the dissimilarity or r is zero, which is the final stop condition. In this way we obtain a range of results for various values of r. The best one is chosen manually by the operator. This is in conformity with the requirement of letting the most important decision be made by the human, not the software.

The dissimilarity function is not convex and the described algorithm can fall into local minima. To make this less likely, the following modification is introduced. As soon as the rank r attains some specified value $\alpha_2 < \alpha_1$, it is set to a larger value $\alpha_3 \in (\alpha_2, \alpha_1\rangle$, and the iterations continue. Hence, r is decreasing from $r = \alpha_3$ to α_2 for the next time. When r attains α_2, then a condition is checked, whether this time an improvement of similarity has been reached. If not, then the iterations are performed with r decreasing further, until the final stop condition is reached. If yes, then r is set back to α_3 and the process goes on. The currently used values of the parameters are: $\alpha_1 = \alpha_3 = 2/3, \alpha_2 = 1/3$. It must be stressed that finally the user can choose *any* value of $r \in (0, \alpha_1\rangle$, irrespective of the other parameters, having the possibility to directly see the result calculated for this particular value.

The rationale for such a procedure can be informally described as follows. If the current minimum of the dissimilarity function is the global one, then a larger number of matched pixels attain their counterparts in the other image than in the case of a local minimum. Increasing the rank will throw the process out from the "worse" minima more forcefully than from the "good" ones. A correspondence between this simple strategy and the simulated annealing can be noticed.

2.2 DETECTION OF THE EDGES

In general, the relevant edges in the analysed images can be found with the *zero-second-derivative* filter with the anisotropic, adaptive scale, controlled by the mechanism described in [18]. The resulting edges are almost one-pixel wide. There is a possibility of manually selecting the most relevant parts of edges of the chosen structures – anatomical structures or irradiation field, and editing the images of edges, including the drawing and deleting the edges manually.

In the simulation image the field edges are not the maxima of brightness gradient, but they are represented as the middle lines of the images of wires, with which the field is marked during therapy planning. The wires are not always visible in their full length, as can be seen in Fig. 1a. A method using the dedicated detector of the roof edges and a version of the hierarchical Hough transform is used to find the central lines and edges of the irradiation field. The central lines represent the coordinate system in which the registration process and the measurements are performed.

The method used will be outlined here; the details can be found in [7]. First the roof edges are found with a fast roof edge detector designed for finding the centres of bright lines, located generally far from each other (Fig. 1b). The assumption of symmetry of the brightness pattern around the line is used. Then the Hough transform is used to find the dominating direction of the lines in the whole image. The transform space is folded by superimposing the ranges of the angle corresponding to angle quadrants, so that any two perpendicular directions are treated as one. Next, the second Hough transform of the image is found, but now only around the angle previously found, and with
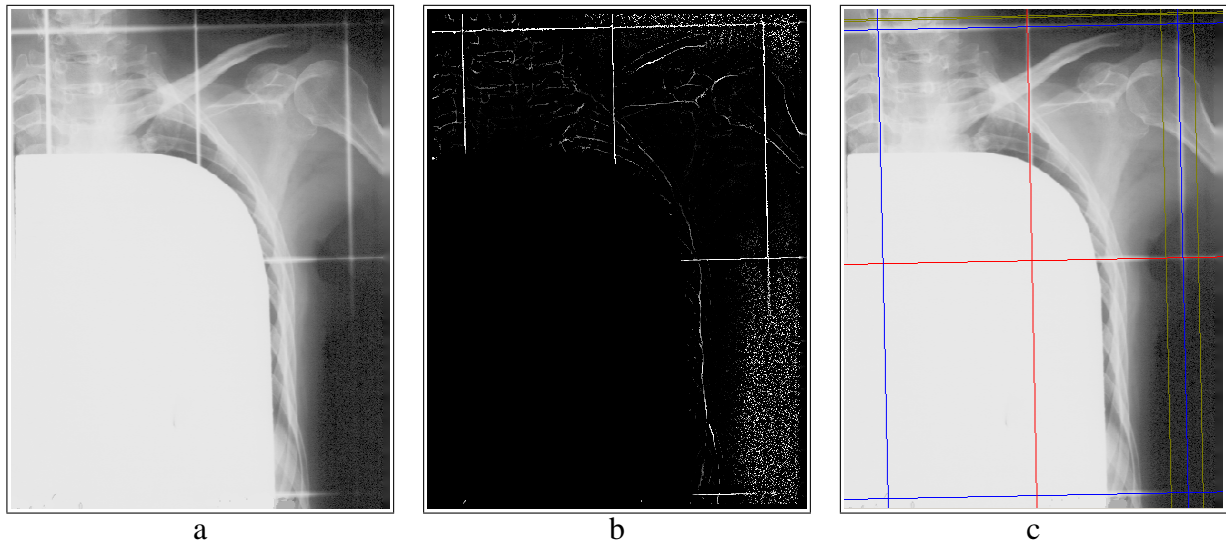
Fig. 1: Detection of edges of the irradiation field in the simulation image. a: original image with wires visible; b: result of the roof edge detector; c: from two perpendicular families of lines found, the hypothesis of central (red) and lateral (blue) lines is formed, and less prominent lines (yellow) are rejected.

a larger angular resolution – for example $0.2°$. As the result, two perpendicular families of lines are found, each having a specified number of lines, for example, five. From these families, a number of hypotheses corresponding to the possible locations of central and lateral lines representing the coordinate system and the edges of the irradiation field are formed. The first hypothesis, formed from the six most prominent lines seen in the image of Fig. 1a, is seen in Fig. 1c, shown in red and blue. If the user is unable to choose an acceptable hypothesis among those presented by the software, than it is possible to manually point to the lines representing the irradiation field, or in the case of the absence of any of these lines, it is possible to show the pixels through which they should pass.

3   EXAMPLE OF CALCULATIONS – THE SOFTWARE SYSTEM AutoPort

The cycle of operations which the user should perform to measure the accuracy of irradiation recorded in one pair of images – one simulation and one portal image – will be described here. To analyse another portal image representing another irradiation session of the same therapeutical cycle, planned in the same simulation image, it is necessary to repeat only the actions relevant to the new portal image.

The analysis has been performed with the software system AutoPort developed by the authors.

The source images are shown in Fig. 2. The contrast of the portal image can be enhanced by linear transformation and by local adaptive histogram equalisation (Fig. 2c).

The edges of the anatomical structures and the irradiation field, and the central lines of the irradiation field in the simulation image, which determine the coordinate system for the analysis, are found as described in section 2.2. For images of Fig. 2 the edges are shown in Figs. 3 and 4. As already said, the registration of the edges of the anatomical structures starts from the state found by registering the edges of the irradiation field. The result of registration, which determines the sought measurement of the irradiation accuracy, is shown in Fig. 5, where the pixels finally estimated as *inliers* and *outliers* in the measurement are marked with colours. In this case, the result was found in 22 iterations (7s, Pentium III 1000MHz). The user has chosen the 17th iteration as the best, with the dissimilarity measure equal 2.80 pixels and quantile rank r – share of inliers – equal 0.52. The choice
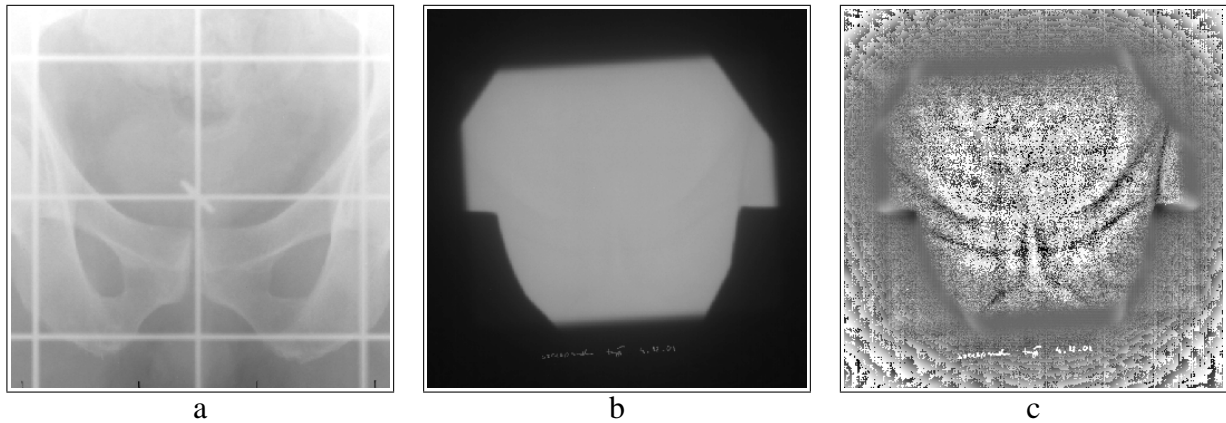
Fig. 2: Source images for image registration. a: simulation image; b: portal image, raw; c: portal image, contrast enhanced with local adaptive histogram equalisation.
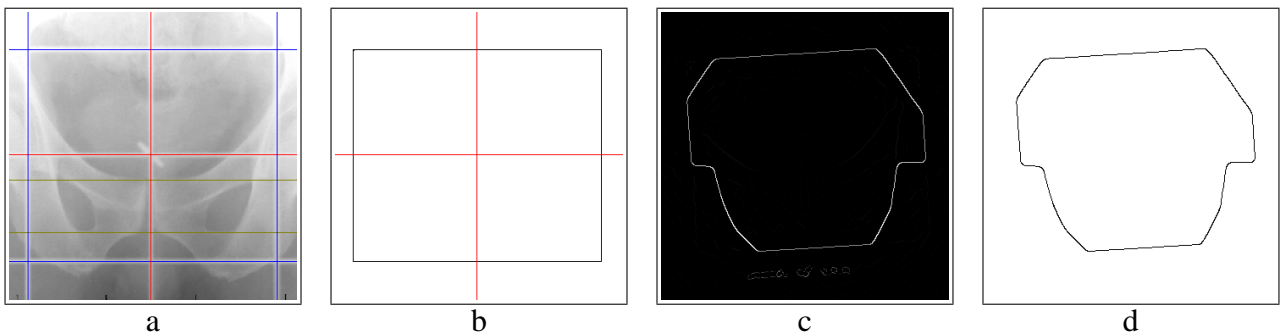

Fig. 3: Edges of the irradiation field. a, b: simulation image; a: centre lines of wires; b: extracted field edges (black) and coordinate system (red); c, d: portal image; c: result of a standard edge detector; d: field edges – image c thresholded.

is performed by moving a slider through the values of r, while the result of registration with the view of inliers and outliers, together with the present dissimilarity measure, changes on-line in the screen. In the Fig. 6 an example of a snaphot of the screen presented by `AutoPort` at the end of the analysis is shown. The user can require a report to be produced, which can be stored in the patient's files.

The program `AutoPort` has been tested at the Holycross Cancer Centre in Kielce for over six months and has become the main tool for measuring the accuracy of external beam radiotherapy in
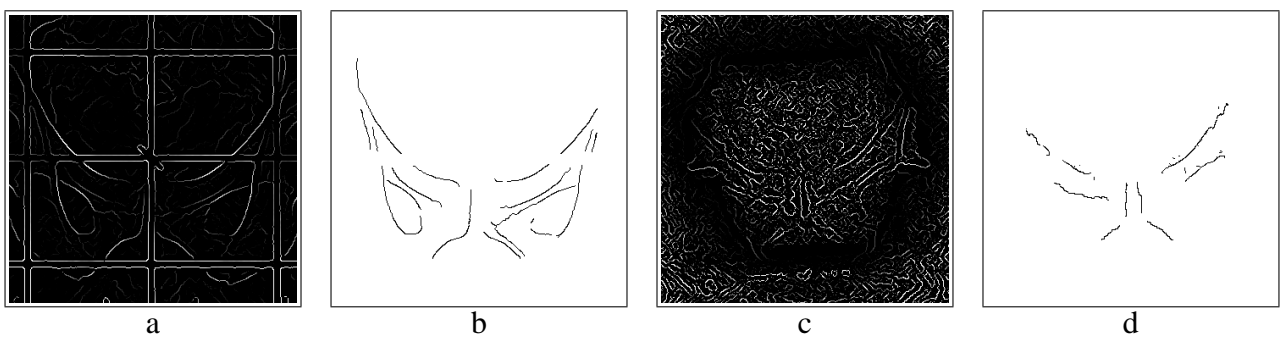

Fig. 4: Edges of the anatomical structures. a, b: simulation image; c, d: portal image; a, c: results of edge detection; b, d: thresholded and edited images a and c, respectively.
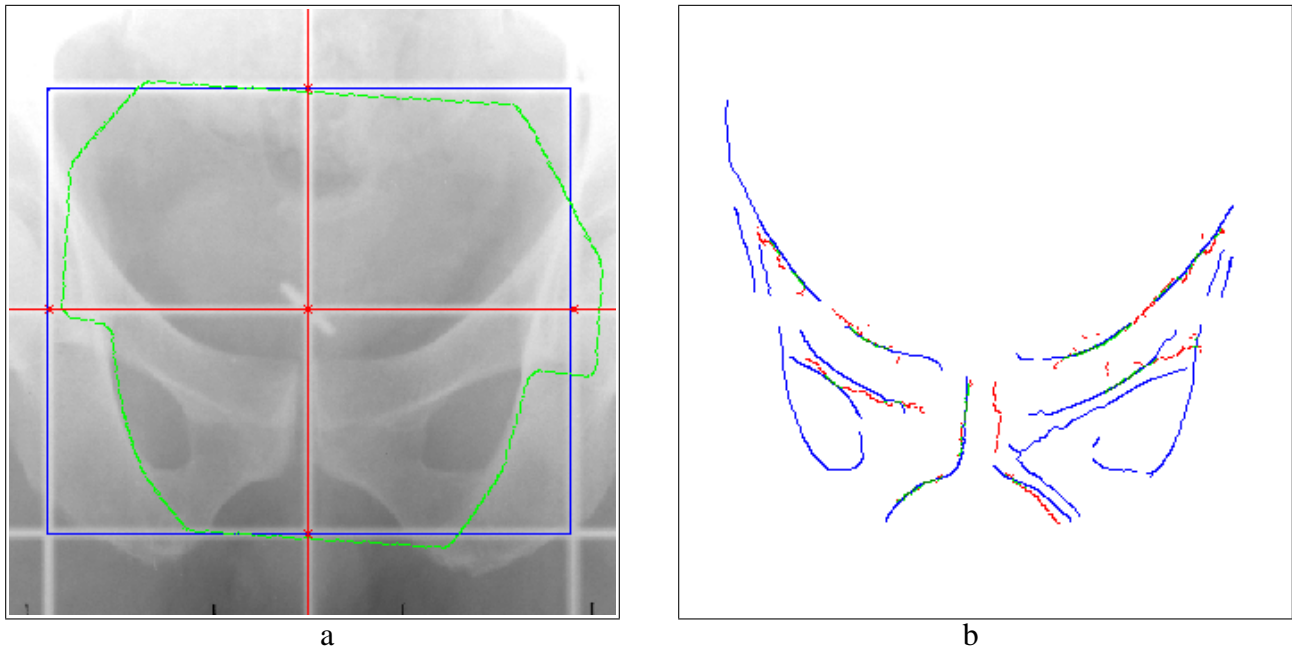
Fig. 5: Results of registration and measurement. a: displacement of field edges: 4.4 mm to the right, 1.0 mm down and rotation 3.9° clockwise; b: registration of edges of the anatomical structures: edges in simulation image – blue, in portal image used in analysis – green (*inliers*), and rejected – red (*outliers*).

normal clinical practice. It has been used to register and measure the therapy accuracy in various localisations, including pelvis, breast, head and neck.

4 CONCLUSION

The problem of quality assessment in the external beam radiotherapy has been solved by using the image registration technique based on the robust measure of dissimilarity – the Hausdorff distance measure, with modifications. The method makes it possible to efficiently register the simulation and the portal image to find out what is the difference between the planned and the realised geometry of the therapy.

The decision on the choice of the final result from a set of results proposed by the method is made by the user, not the software, which is believed to be the important virtue of the proposed method. The choice corresponds to specifying the share of data used in the calculations – the *inliers*, thus rejecting the *outliers*, which can always be present in the data originating from such "difficult" source as the portal images.

The software tool `AutoPort` in which the described method has been implemented proved its usefulness in the requiring day-to-day clinical practice of the oncological centre.

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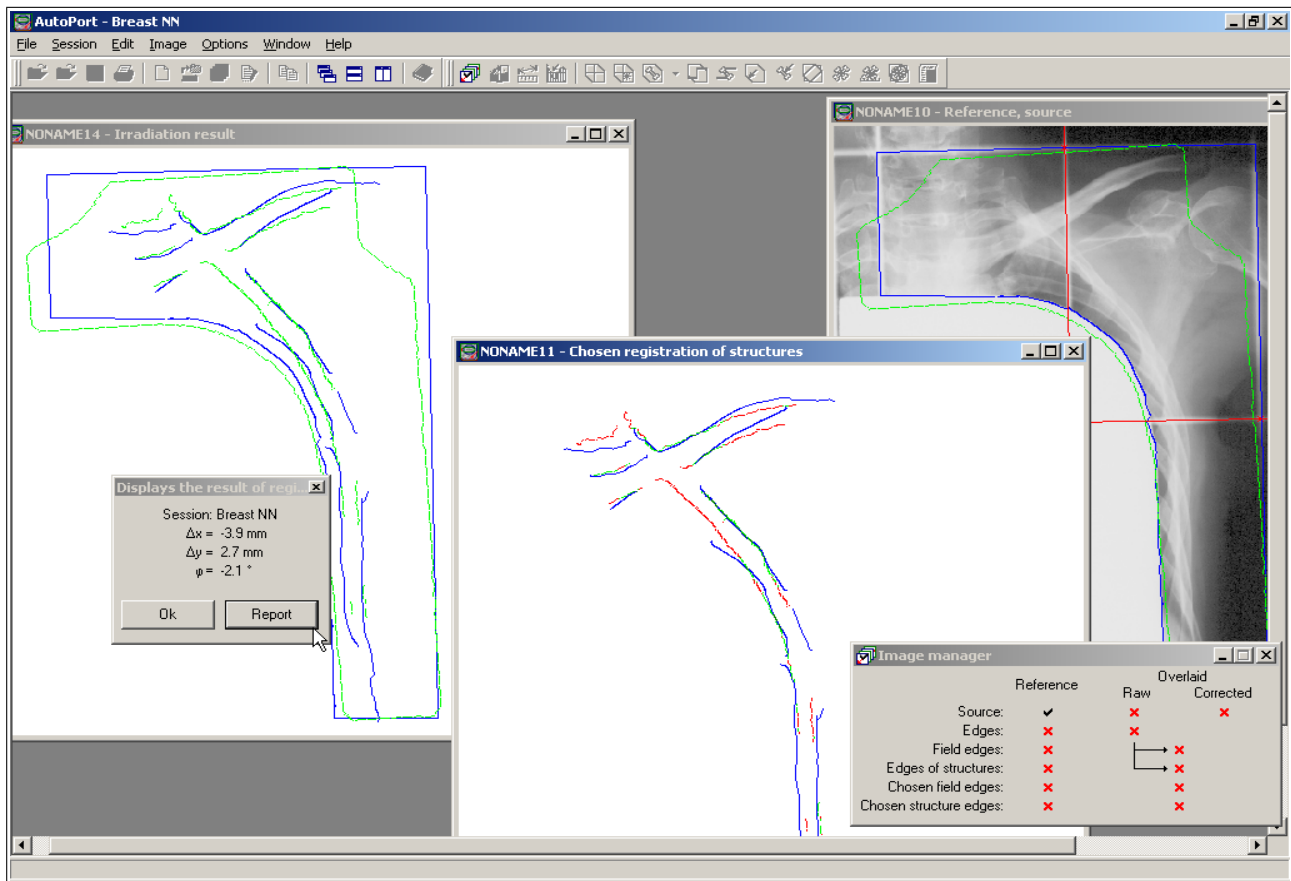
Fig. 6: `AutoPort` – snapshot of the screen at the end of analysis.

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