Volume preserving elastic transformation for local breast-tissue quantification

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Abstract. In this paper, we explore the idea of quantifying local breast-tissue density changes. Breast tissue density has been correlated to breast cancer incidence in numerous studies which have shown a statistical relationship between glandular density and the occurrence of cancer. In particular, postmenopausal women who take HRT run an increased risk of developing cancer due to the "regeneration" of fibroglandular tissue that is often induced by the exogenous hormones. In this paper, we present a method that combines mammogram normalisation and volume-preserving registration, and which can be the starting point for temporal-local breast tissue quantification.

1 Introduction

Hormone Replacement Therapy (HRT) has many beneficial effects for post-menopausal women (e.g. reduction of menopause-related symptoms, and lesser risk of developing osteoporosis). Unfortunately, long-term use has been correlated with an increased risk of breast cancer [1]. Glandular tissue regeneration is often a sign that the woman's breasts are responding to the exogenous hormones and it is crucial to closely monitor the patient in such cases, with the goal of an early diagnosis of a possible HRT-induced cancer. Our objective is to quantify local tissue density changes for women taking HRT using only mammogram data. To achieve this goal, two sub-goals are necessary:

- Computation of the h_{int} mammogram representation [2]. Due to different imaging conditions (e.g. time of exposure) and image degrading factors (e.g. scattering), result in poor intensity correspondence in temporal mammograms. The h_{int} representation estimates the thickness of non-fatty tissue "above" each pixel, by using a physical model of the image formation [2]. This way, we have a *quantitative* representation of the breast which effectively normalises temporal mammogram pairs. Based on this representation (of integrated non-fatty breast tissue), we have calculated measures of tissue density change by exploiting the fact that a global or local change in a specific h_{int} mammogram area reflects the change in the fibroglandular composition of the breast or of that area [3]. Since the sum of h_{int} values (total non-fatty tissue) is expected to be invariant between acquisitions, we are interested in cases where this does not hold (possible pathology, response to HRT).

- Image registration. This is necessary in order to align the two mammograms, for quantification of change in local tissue density. We have developed a method for mammogram registration, based on breast boundary landmarks (detected using curvature) and internal landmarks (using a multi-scale segmentation framework) [4]. However, the elastic registration process involves pixel rearrangement and scaling which can significantly alter the total "volume" of the mammogram image.

The later can be defined as:

$$V_{\text{Image}} = \sum_{i=0}^{N} I(x_i, y_i)$$

Where N is the number of pixels across the image.

This problem is illustrated in Figure 1, where the "volume" of a synthetic image is reduced to 70% of the original, after applying a transformation. In many cases, due mainly to temporal differences in the breast size/compression, the mammogram size and geometry can change significantly between acquisitions. Consequently, aligning temporal data can reduce (or increase) the image volume. This is not important for "un-normalised" mammograms since the intensities of corresponding regions are not necessarily related. However,

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in the h_{int} representation, every pixel value represents the height of “non-fatty” tissue at that point during acquisition. This paper demonstrates that intensity correction based on the registration deformation field can compensate for the volume error introduced by elastic deformation. In the remainder of this paper, we explain our method for volume preserving image alignment (Section 2) before we present some results (Section3) and a brief discussion (Section 4).

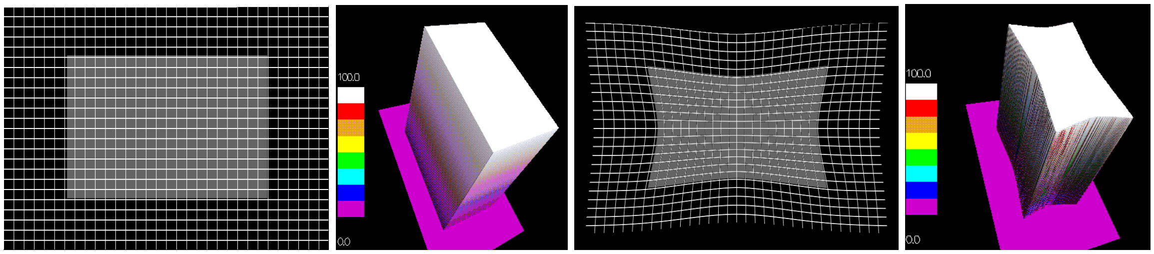
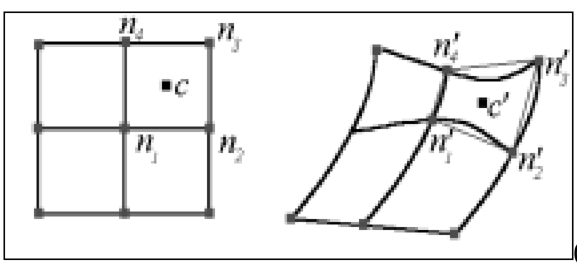


Figure 1. (a) The original rectangle undeformed, (b) the original rectangle plotted as a 3-D surface, (c) Image (a) is deformed using thin-plate spline interpolation, (d) The intensity remains constant and as a result the total image volume is reduced by 70%.

2. Designing a “volume-preserving” transformation method

A 2-d elastic transformation (e.g. thin-plate spline interpolation) usually changes the image “volume”. This produces an undesirable effect in h_{int} images where the “height” of each pixel, consequently the image volume, are meaningful quantitative measures of the breast anatomy. Our method for intensity correction is summarised in the next section: if we take an orthogonal cell in the image to be transformed and calculate a “deformation measure” for that cell, then we can “adjust” the intensity (height) of the deformed cell so that the “volume” is preserved. In Figure 2 (a), we illustrate this concept. The orthogonal cell represents the image before registration. The central node c, and the 4 surrounding nodes n_i, (i= 1, 2, 3, 4) are transformed to c' and n_i' according to the image based calculated transform.



(a)

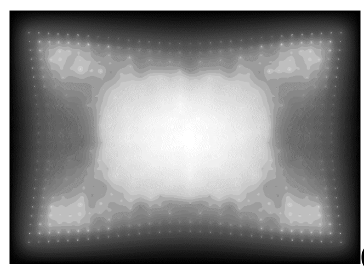

(b)

Figure 2. (a): A local affine approximation is used to describe the deformation of an orthogonal image cell (defined by the nodes n_1, n_2, n_3, n_4), (b): The calculated intensity-correction field based in all the nodes of Figure 1(a) and 1 (c).

Based on those points, we can calculate the area of the cell before and after the transformation (we use a local affine approximation of the transformed region). The ratio of the areas is then used to assign a modification value to the transformed node c':

$$M(c') = \frac{A}{T(A)}$$

Where A is the area of the orthogonal cell, T is the applied transform. After we calculate the modification value for all the nodes of the image, we calculate an intensity displacement field using finite differences (Laplacian interpolation). This is shown in Figure 2 (b). In order to compensate for the image “volume” changes, we multiply the image intensities by the intensity correction field. The remaining volume (error volume) can be equally distributed along the image pixels so a 100% volume preservation can be achieved. This requires the calculation of an adjustment (“plateau”) value according to the formula:

$$p = \frac{V_{original} - \sum_{i=1}^{N} I(x_i, y_i) \cdot C(x_i, y_i)}{N}$$

Where p is the plateau value, $V_{original}$ is the volume before the transformation, I is the image and C the intensity-correction field. In the next section we illustrate this concept with a synthetic example followed by a real patient case.

3. Results

In Figure 3(a), the same rectangle as Figure 1 (b) (of constant intensity) is deformed using thin-plate spline interpolation. However, the intensity of the square does not change, resulting in loss of volume as it is shown in Figure 3(b). Figure 3(c) is the corrected image using the method described in the previous section. The maximum intensity rises from 100 to 173 (on an 8-bit grey-scale) and the intensities are distributed according to the intensity-correction field shown in Figure 2 (b). The error volume is distributed equally in all pixels and the resulting volume is the same as in the unreformed image.

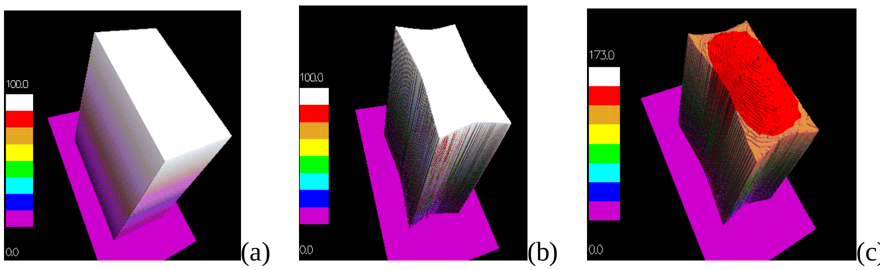


Figure 3. The intensity correction algorithm for the deformed square in Figure 1. (a) The original square, (b) After transformation the volume is reduced by 30%, (c) Using the intensity correction field (Figure 2 (b)) and the “plateau” value p, we correct the intensities so that the image volume is preserved (notice that the colour scale range has changed from 100 to 173).

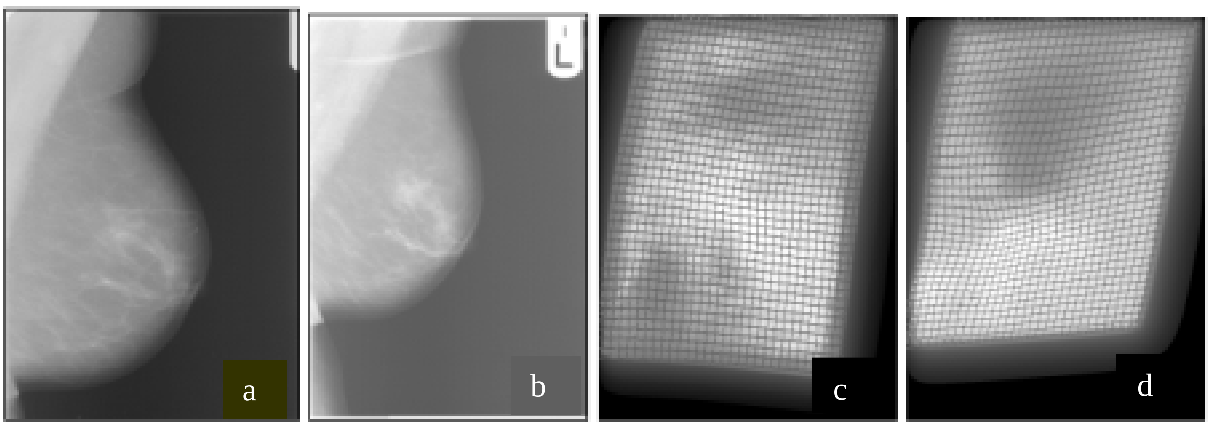


Figure 4. (a), (b): A temporal HRT pair, where the breast compression is significantly different, (c), (d): The intensity correction fields for boundary-based, and internal landmark registration respectively

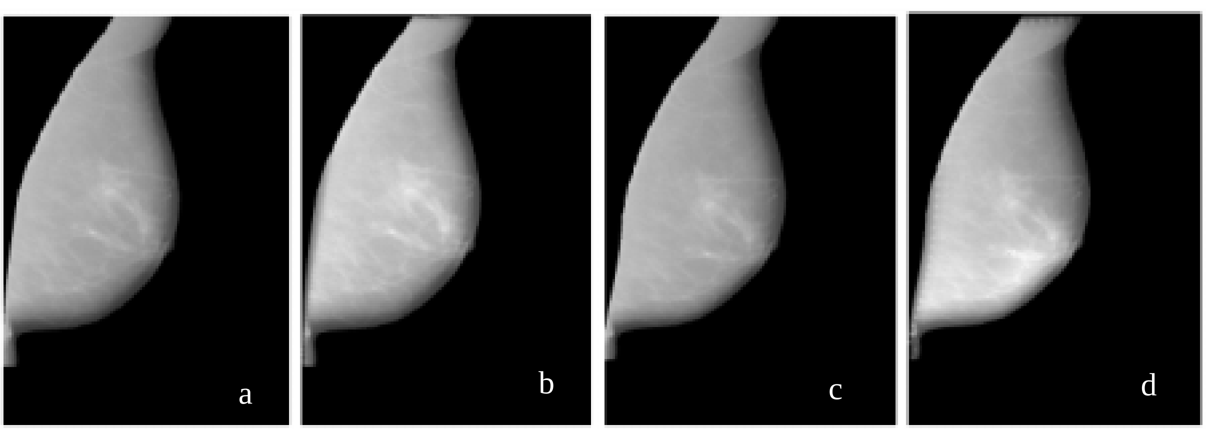


Figure 5. (a): The mammogram in (Figure 4 (a)) registered to the one in Figure 4 (b) using only boundary points, (b): Using that transformation we calculate the intensity corrected image, (c): The same as (a) but using internal landmarks to better approximate the deformation, (d): Again based on the calculated transformation we calculate the intensity corrected image.

In Figure 4 (a) and (b), we show a temporal HRT pair. The difference in breast compression (and size) between the two acquisitions is significant. When the images are registered, the image “volume” is reduced by 33 %. By using our mammogram registration method [4] we align the images and calculate the intensity correction fields, first using only boundary points, and then using internal landmarks to better approximate the deformation (Figures 4 (c) and (d)). The corrected images for these two registration scenarios are shown in Figure 5 (b) and (d) respectively.

4. Discussion

For women that decide, or are advised, to take HRT, it is important to monitor the response to therapy so that the increased risk for developing cancer is assessed. This risk is dependent on whether or not the exogenous hormones stimulate glandular tissue regeneration in certain locations in the breast. Our previous work on HRT temporal quantification was mainly focused on “globally” characterising the changes in the breast based on the h_{int} representation [3]. Although this could find a very interesting clinical application in assessing the response to the therapy, the ultimate goal is to be able to *locally* describe the induced changes in the breast. This could highlight regions that exhibit strong glandular regeneration and would require closer clinical examination in order to early diagnose a potential abnormality.

In this paper, we have presented a method that aims to preserve the h_{int}-based anatomical information content after registration, so that local quantitative comparison is possible. As shown in Figure 5, the result of the intensity correction depends on the complexity of the calculated registration transformation. For this reason, in Figure 5 (d) the glandular tissue is more pronounced as the internal glandular areas are taken into consideration in the registration process. In contrast, Figure 5 (b) calculates a smoother transform (and intensity correction) based only on the boundary alignment. The accuracy of the corrected mammograms depends on the accuracy in the calculation of the geometrical transformation that relates the temporal mammogram pair. The corrected image is calculated mainly for accurate comparison of tissue density in temporal HRT pairs. However, it still needs to be clinically meaningful and in our future work we aim to validate that assumption. In an experiment to validate the intensity preserving alignment method presented here, we used differential compression data [2] (the patient is kept in the same position, and the breast is imaged at the same time, in two different compression levels). This could be considered as a ground truth, since (unlike temporal mammograms where the breast changes between acquisitions) the resulting h_{int} images (after registration and intensity correction) should be exactly the same. Table 1, summarises the results for the whole mammogram and for a segmented region of interest (cancer):

Volume units in Pixels x (8-bit greyscale)	Target Volume	Volume before correction	Volume after correction
Whole mammogram	29439392	24856420	29251438
Segmented cancer	716235	581858	685431

Table 1: Volume preservation results in a differential compression mammogram pair.

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