

Tracking of Non-rigid Targets in 3D US Images: Method and Evaluation on Clinical Data

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Abstract. In ultrasound-guided procedures, such as high-intensity focused ultrasound or radio-frequency ablation, non-rigid clinical targets may undergo displacements due to physiological motions. To cope with that issue, the accurate estimation of the target motion is required in order to adjust the position of medical tools. In this paper, we present a robust approach that allows to track in real-time deformable targets in 3D ultrasound images. Our method combines visual motion estimation with a mechanical model of the target. We demonstrate the good performance of our approach by showing the tracking accuracy results on the MICCAI CLUST'14 challenge database.

1 Introduction

Over the last few years, minimally-invasive procedures, such as high-intensity focused ultrasound (HIFU), or radio-frequency ablation (RFA) have gained significant attention due to the shorter recovery time compared to conventional therapies. However, the quality of these therapies can strongly depend on both the deformations and displacements of the clinical targets since the surgeon needs to continuously adjust the positions of medical tools. Thus, to ensure the target visibility under ultrasound (US) image guidance, several target tracking methods have been developed. The first type of method consists in extracting key features in the US images such as target contours as proposed by Angelova and Mihaylova [1]. Furthermore, the feature extraction can be based on a Bayesian approach for increasing the tracking robustness. However, such features may not be clearly visible in each frame of the US sequence. The second type of methods is based on block-matching techniques, where target motions are computed from small blocks displacements using similarity criterion like Sum of Squared Differences (SSD) [2], or Cross-Correlation (CC) [3]. However, these methods assume that the displacement is the same within a local region block. The last type of approaches are based on the optical flow estimation such as the method proposed by Mikic et al. [4] or Lee et al. [5]. Nevertheless, due to the poor quality of

US images, such approaches may provide undesirable results. To cope with that issue, the motion field can be regularized by using elastic or fluid-like operators as it is proposed by Somphone et al. [6]. In order to ensure physically-plausible motions, we recently presented in [7] a robust approach that combines a mechanical model and visual estimation. The good performance of this method has been showed on data obtained from a deformable soft tissue phantom. In this paper, we demonstrate that this method can achieve high accuracy on real-data by testing our algorithm on the database proposed by the MICCAI CLUST’14 challenge. The rest of the paper is organized as follows. In section 2, we present the method that allows to track deformable target in 3D ultrasound images. In section 3, we describe the performance of our approach on real-data. Finally, section 4 concludes the paper.

2 Method

The objective of our approach is to track the motions of a clinical target in 3D ultrasound sequence. To do so, we first generate a mesh model of the target by using a piece-wise affine warping function. Then, the model is linked to a mechanical component in order to ensure physically-plausible motions. Finally, we estimate the target displacements thanks to an additive-update approach based on intensity variation.

2.1 Piece-wise Affine Model

In 3D US images, a clinical target can be represented by a continuous set of N_v voxels that is delimited by a visible border. Typical examples are shown in Fig. 1. In order to define the target, we first extract its shape at the initial 3D frame of the US sequence by performing a segmentation. To remove sharp edges and discontinuous shapes, a smoothing step is performed on the 3D segmented surface and a corresponding fitted tetrahedral mesh containing N_c vertices is defined. Then, in order to represent the displacements of the voxels, we propose to use a piece-wise affine warp function. Our piece-wise affine warping is parameterized from both the vertice positions and an affine interpolation that uses barycentric coordinates as proposed in [8]. In this way, we can relate all the voxel positions with all the vertices as follows:

$$\mathbf{p}_{im} = \mathbf{M} \cdot \mathbf{q} \quad (1)$$

where \mathbf{M} is a $(3 \cdot N_v) \times (3 \cdot N_c)$ constant matrix defining the set of barycentric coordinates. \mathbf{p}_{im} is a $(3 \cdot N_v) \times 1$ vector defining all the voxels positions, and \mathbf{q} is a $(3 \cdot N_c) \times 1$ vector containing all the vertices positions. Thanks to Eq. (1), we can update the positions of the target when the vertices of the model are displaced. To compensate the lack of smoothness as well as the poor estimation of vertice positions in US images, we combine a mechanical model to the estimation of displacement.

2.2 Mechanical Component

Our approach combines a mass-spring-damper system to the mesh model previously described. Thus, the vertice displacements are constrained by linking each connected vertice pair with a spring ensuring physically-plausible and coherent displacements of the vertices. Furthermore, the mass-spring-damper system can be specifically characterized by setting a mass value to each vertex, together with elastic and damping coefficients on each spring depending on the soft-tissues homogeneities. These values can be accurately estimated from elastography images. From this model, we can compute the force $\mathbf{f}_{ij} = [f_{x_{ij}} \ f_{y_{ij}} \ f_{z_{ij}}]^T$ applied on a vertice \mathbf{q}_i from a neighbor vertice \mathbf{q}_j . This force can be expressed as follows:

$$\mathbf{f}_{ij} = K_{ij}(d_{ij} - d_{ij}^{init})(\mathbf{q}_i - \mathbf{q}_j) + D_{ij}(\dot{\mathbf{q}}_i - \dot{\mathbf{q}}_j) \circ (\mathbf{q}_i - \mathbf{q}_j) \quad (2)$$

where d_{ij} and d_{ij}^{init} respectively represent the distance between the vertices \mathbf{q}_i and \mathbf{q}_j at their current positions and at their initial positions. The \circ operator expresses the Hadamard product, K_{ij} is a scalar value denoting the stiffness of the spring that links the two vertices while D_{ij} is the damping coefficient value. By combining the previous equation for all the vertices, we can express the total amount of forces \mathbf{f}_i exerted on each vertice \mathbf{q}_i of the mesh model as follows:

$$\mathbf{f}_i = \sum_{n=0}^{N_i} \mathbf{f}_{in} + G_i \dot{\mathbf{q}}_i \quad (3)$$

N_i denotes the number of neighbors vertices connected to the vertice \mathbf{q}_i . G_i represents the velocity damping coefficient associated to the vertice \mathbf{q}_i . In order to obtain the displacements $\Delta \mathbf{d}$ associated to the mass-spring-damper system, we integrate the forces expressed in Eq. (3) with a semi-implicit Euler integration scheme. Such mechanical constraint can ensure the smoothness warping function of the deformation and limits the noise sensitivity of the intensity-based approach.

2.3 Additive Update Tracking

Let us recall that the main objective of our approach is to estimate the new positions of the target in 3D US sequence. To do so, we use an intensity-based method that consists in evaluating the vertice displacements by minimizing a dissimilarity function E . Therefore, we can express the cost function which minimizes image dissimilarity from the relationship described in Eq. (1) such that:

$$C(\Delta \mathbf{q}) = E(I_t(\mathbf{p}_{im}(t)), I_{t_0}(\mathbf{p}_{im}(t_0))) = E(I_t(\mathbf{M}\mathbf{q}(t)), I_{t_0}(\mathbf{M}\mathbf{q}(t_0))) \quad (4)$$

where I_{t_i} is a vector representing the US intensity of the volume acquired at time index t_i . $\Delta \mathbf{q}$ denotes the vertices displacements. $\mathbf{p}_{im}(t_i)$ and $\mathbf{q}(t_i)$ represent respectively the voxel positions and the vertice positions at time index t_i . In order to determine the dissimilarity function E , we assume that the intensity values of soft tissues are consistent over the time. Consequently, we propose to

use the Sum of Squared Differences (SSD) in order to measure the image error. The cost function can now be expressed as:

$$C(\Delta\mathbf{q}) = (I_t(\mathbf{M}(\mathbf{q}(t)) - I_{t_0}(\mathbf{M}(\mathbf{q}(t_0))))^2 \quad (5)$$

The objective is to find iteratively the vertice displacements by minimizing the cost function C . To do so, we perform a Taylor expansion of the previous equation that leads to:

$$C(\Delta\mathbf{q}) \approx (\mathbf{J}\Delta\mathbf{q} + I_t(\mathbf{M}(\mathbf{q}^{k-1}(t))) - I_{t_0}(\mathbf{M}(\mathbf{q}(t_0))))^2 \quad (6)$$

where $\mathbf{q}^{k-1}(t)$ represents the estimation of the parameters at time t at iteration $k-1$ of the optimization algorithm. \mathbf{J} denotes the Jacobian matrix associated to the cost function. This matrix relates the variation of the parameters $\Delta\mathbf{q}$ with the intensity variation of I . It can be computed as follows:

$$\mathbf{J} = \nabla I \cdot \mathbf{M} \quad (7)$$

where ∇I denotes the gradient of the current 3D US frame. In order to obtain the optimal displacements of the vertices, we chose to use a forward-additive steepest gradient strategy as it is computationally efficient since it does not require the calculation of pseudo-inverse of large Jacobian matrix. We therefore obtain:

$$\Delta\mathbf{q} = -\alpha \mathbf{J}^t [I_t(\mathbf{M}(\mathbf{q}^{k-1}(t))) - I_{t_0}(\mathbf{M}(\mathbf{q}(t_0)))] \quad (8)$$

where $\alpha > 0$ denotes the step size of the minimization strategy. \mathbf{J}^t represents the transpose matrix of the Jacobian \mathbf{J} . As stated previously, in order to prevent inaccurate results, we propose to combine this motion estimation with the internal displacements of the mass-spring-damper system. This can be performed by iteratively estimate the optimal displacement as follows:

$$\mathbf{q}_k(t) = \mathbf{q}_{k-1}(t) + \Delta\mathbf{q} + \Delta\mathbf{d} \quad (9)$$

where $\Delta\mathbf{d}$ is the internal displacements obtained from the integration of forces expressed in Eq. (3). $\Delta\mathbf{q}$ represents the external displacements from the steepest gradient strategy in Eq. (8). $\mathbf{q}_{k-1}(t)$ denotes the estimation of vertice position at iteration $k-1$ and at time index t . In order to balance the influence of the mass-spring-damper system regarding to the motion estimation between $\Delta\mathbf{q}$ and $\Delta\mathbf{d}$, we can tune the α coefficient that represents the step size of the minimization strategy in Eq.(8).

3 Results

3.1 Description of our Evaluation Environment

Our approach has been tested on real-data and has been implemented with C++/GPU code by using Cuda and VTK libraries. The segmentation step in the first volume is performed with the ITK-SNAP [9] software and can be executed in less than 3 minutes. The mesh is generated thanks to the tetGen [10] software. The resulted computation time of the online tracking is 350 ms allowing thus real-time capabilities. The code was executed on a Windows 7 machine with an Intel core i7-3840qm(2.80GHz).

3.2 Validation Results on Real-data

In order to evaluate our method, we used the database provided by the workshop MICCAI CLUST’14 challenge. The main goal of this challenge is to compare different state-of-the-art methods for tracking anatomical landmarks in US sequences. For this purpose, a database containing 2D/3D ultrasound sequences of volunteers under free breathing is provided. Furthermore, in order to generate ground truth data, the positions of the target landmarks are identified from expert annotations for each frame. Thus, a comparison can be performed between the ground truth landmark positions and the warped point positions (estimated from our model) over each frame. It is worth mentioning that both the ultrasound sequences and the annotations are provided by several research institutes. Thus, the approach has been tested by tracking 32 different anatomical features acquired from 16 3D US sequences. In these experiments, we set empirically the elastic and damping parameters such that $K_{ij} = 3.0$ and $D_{ij} = 0.1$ for all the springs, along with $G_i = 2.7$ for all vertices. The step size of the steepest gradient method has been set to $\alpha = 2 \times 10^{-6}$. In future work, we plan to automatically estimate these parameters by using elastography images. The final results are reported in table 1 and presented in the workshop website⁵. In the Fig. 1, the

Participants	Mean (mm)	SD (mm)	95th (mm)
Our method	1.62	2.19	4.81
Somphone O., et al. [6]	2.55	2.46	7.98
Rothlbbers S., et al. [11]	2.80	2.96	7.94
Lubke D., and Grozea C. [12]	4.63	4.03	12.44

Table 1. Results of 3D point-landmark tracking. The first column of the table details the reference to each candidate method. The subsequent columns represent respectively the mean error, the standard deviation, and the 95th percentile expressed in millimeters for each approach.

performance of our approach is illustrated by showing the tracking results at different frames on four landmarks representing hepatic vein bifurcations. The Fig. 2 and Fig. 3 detail results for each target tracking task. From these figures, we can notice that the mean tracking error is under 2 mm for most of the ultrasound sequences. However, we can observe that we obtained some unsatisfactory results when the 3D mesh model goes out of the field of view (SMT-04_01), or when the target follows high deformation regarding the provided elastic parameters (EMC-03_01). Furthermore, the error can be also higher due to the low resolution of the ultrasound volume (e.g. ICR-02_01). However, the overall results are very encouraging since our approach performs more accurate tracking of landmark points than other candidate methods and achieves real-time capabilities. In addition, we can notice that our method remains robust with empirical parameters.

⁵ <http://clust14.ethz.ch/results.html>

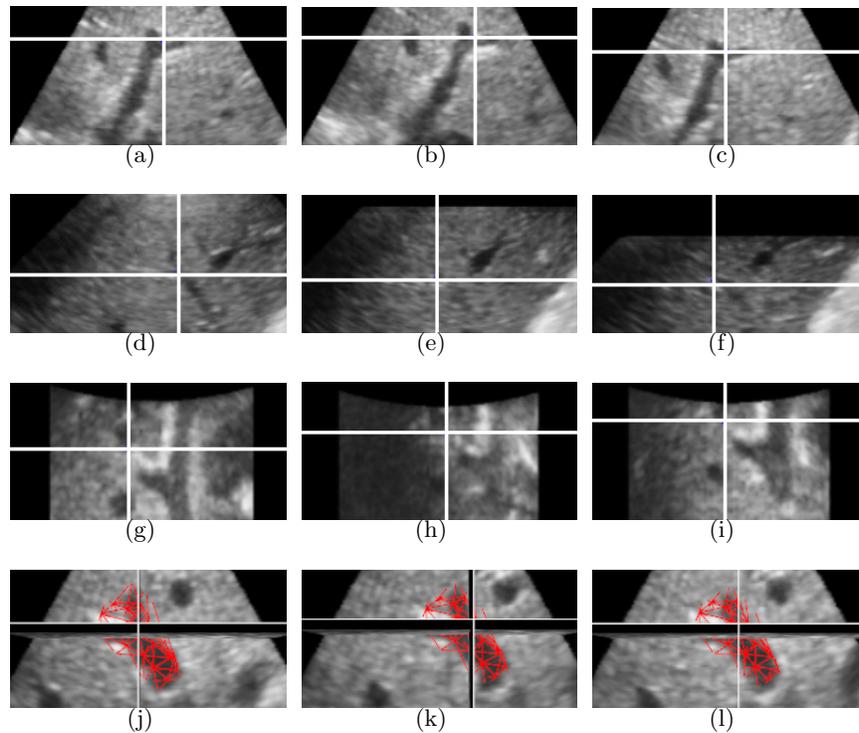


Fig. 1. Example of the tracking task on several sequences. (a-b-c) Tracking of landmark representing hepatic vein bifurcation. The white cross represents the point position in Y-slice at frame index 00 (a), 08 (b), 12 (c). (d-e-f) Tracking of landmark representing vein bifurcation in another 3D US sequence at frame index 00 (d), 05 (e), 12 (f). (g-h-i) Tracking of landmark representing portal vein bifurcation at frame index 00 (g), 23 (h), 59 (i). (j-k-l) Tracking of landmark representing first degree bifurcation of hepatic bile duct. The red model represents the associated 3D tetrahedral mesh model at frame index 00 (j), 28 (k), 78 (l).

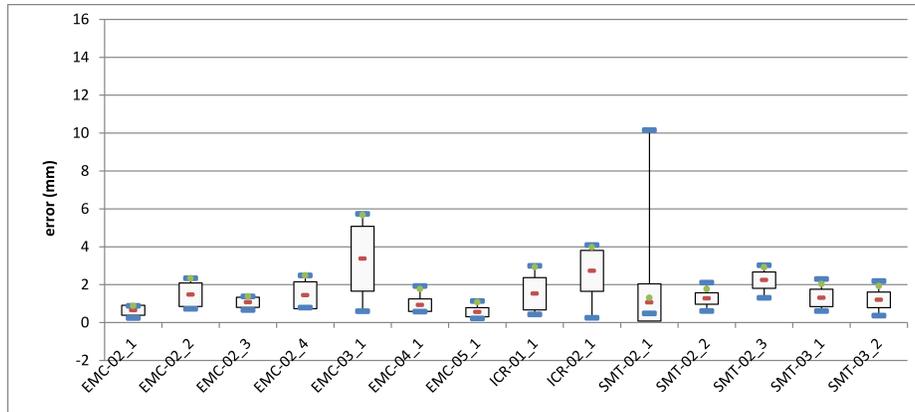


Fig. 2. Tracking error results for each sequence. The x-axis and y-axis represent respectively each ultrasound sequence and the associated tracking error expressed in millimeters. The name of the sequence (EMC-02_1) represent both the acronym of the institute and the sequence index. (Red) Mean tracking error estimated from euclidean distance. (Black box) Mean error \pm standard deviation. (Whiskers) Minimum and maximum errors. (Green dot) 95th percentile of error.

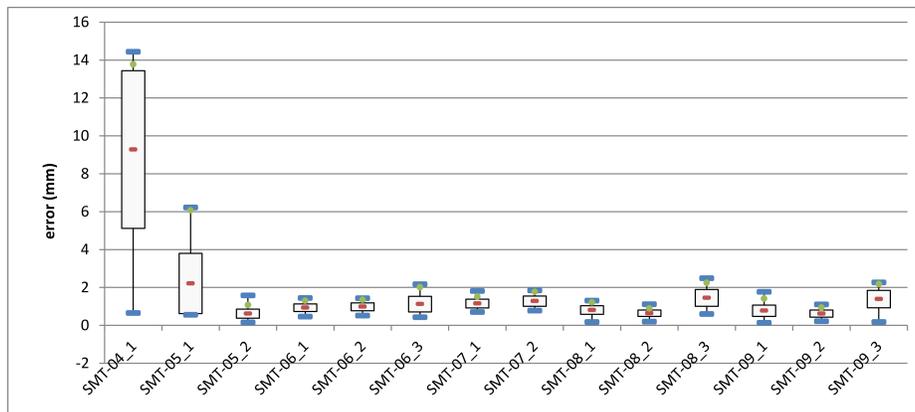


Fig. 3. Tracking error results for each tracking task. The x-axis and y-axis represent respectively each ultrasound sequence and the associated tracking error expressed in millimeters. The name of the sequence (e.g. SMT-04_1) represent both the acronym of the institute and the sequence index. (Red) Mean tracking error estimated from euclidean distance. (Black box) Mean error \pm standard deviation. (Whiskers) Minimum and maximum errors. (Green dot) 95th percentile of error.

4 Conclusion

In this paper, we presented a method for tracking and automatically compensating the displacements of a deformable target in 3D ultrasound images. The robustness of our tracking method is ensured by combining a mechanical model to the displacement estimation. We evaluated the good performance of our approach through CLUST'14 challenge database. In future work, we plan to automatically estimate the elastic parameters by using elastography images.

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