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Unsupervised Echocardiography Registration through Patch-based MLPs and Transformers

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Abstract. Image registration is an essential but challenging task in medical image computing, especially for echocardiography, where the anatomical structures are relatively noisy compared to other imaging modalities. Traditional (non-learning) registration approaches rely on the iterative optimization of a similarity metric which is usually costly in time complexity. In recent years, convolutional neural network (CNN) based image registration methods have shown good effectiveness. In the meantime, recent studies show that the attention-based model (e.g., Transformer) can bring superior performance in pattern recognition tasks. In contrast, whether the superior performance of the Transformer comes from the long-winded architecture or is attributed to the use of patches for dividing the inputs is unclear yet. This work introduces three patch-based frameworks for image registration using MLPs and transformers. We provide experiments on 2D-echocardiography registration to answer the former question partially and provide a benchmark solution. Our results on a large public 2D-echocardiography dataset show that the patch-based MLP/Transformer model can be effectively used for unsupervised echocardiography registration. They demonstrate comparable and even better registration performance than a popular CNN registration model. In particular, patch-based models better preserve volume changes in terms of Jacobian determinants, thus generating robust registration fields with less unrealistic deformation. Our results demonstrate that patch-based learning methods, whether with attention or not, can perform high-performance unsupervised registration tasks with adequate time and space complexity.

Keywords: Unsupervised Registration · MLP · Transformer · Echocardiography.

1 Introduction

Image registration is essential for clinical usage; for example, the registration of cardiac images between end-diastole and end-systole is meaningful in myocardium deformation analysis. Non-rigid echocardiography image registration is one of the most challenging image registration tasks, as finding the deformation field between noisy images is a highly nonlinear problem in the absence of

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ground truth deformation. Specifically, various image registration problems require the mapping between moving and fixed images to be folding free [2, 7, 32]. Traditional non-learning approaches rely on the optimizing similarity metrics to measure the matching quality between image pairs [10, 26, 31]. With the rapid promotion of deep learning, various frameworks of convolutional neural networks (CNN) have been introduced in image registration and have shown impressive performance in many research works.

We consider a 2D non-rigid machine learning-based image registration task in this work. With two given images: $I_{fix}^N, N \in \mathbb{R}$ and $I_{mov}^N, N \in \mathbb{R}$, we want to learn a model $\mathcal{T}_\omega(I_{mov}, I_{fix}) \rightarrow \phi(\theta)$ that generates a constrained transformation $\phi(\theta)$ based on a similarity measurement \mathcal{M} to warp the moving image by minimising the loss function:

$$L = \arg \min_{\theta} \mathcal{M}(I_{fix}, I_{mov} \circ \phi(\theta)) + \lambda \mathcal{C}(\phi(\theta)) \quad (1)$$

where the transformation ϕ is parameterised by the parameter θ and constrained by a regularisation term $\mathcal{C}(\phi(\theta))$ to ensure ϕ to be a spatially smooth transformation. However, iterative optimization of Eq. 1 is very time-consuming, whereas a well-trained CNN does not need any iterative minimization of the loss function at test time. This advantage drives researchers' attention to learning-based registration. Learning-based registration methods can be categorised into supervised and unsupervised registration approaches.

Supervised Registration The supervised learning registration methods [6, 27, 33] are primarily trained on a ground-truth training set for which simulated or registered displacement fields are available. The training dataset is usually generated with traditional registration frameworks or by generating artificial deformation fields as ground truth for warping the moving images to get the fixed images. [28, 34]. One of the limitations of the supervised registration approaches is the registration quality, which is highly influenced by the nature of the training set of deformation map, although the requirement in terms of training set can be partially alleviated by using weakly-supervised learning [4, 5, 12, 14, 15].

Unsupervised Registration In unsupervised registration [4, 9, 13, 17, 24], we rely on a similarity measure and regularisation to optimize the neural network for learning the transformations between the fixed and moving images. Usually, a CNN is used directly for warping the moving images, which is then compared to the fixed-image with the similarity loss. The displacement field can also be obtained from a generative adversarial neural network, which introduces a discriminator neural network for assessing the generated deformation field quality. [11, 23, 29, 36].

Multi-layer Perceptron and Transformers MLP is one of the most classical neural networks and consists of a stack of linear layers along with non-linear activation [30]. For several years, CNN has been widely used due to its performance on vision tasks and its computation efficiency [19]. Recently, several alternatives to

the CNN have been proposed such as Vision Transformer (ViT) [1] or MLP-Mixer [16] which demonstrated comparable or even better performance than CNN on classification or detection tasks. There are currently intense discussions in the community of whether patching, attention or simple MLP play the most important role in such good performance.

In this paper, we propose three MLP/Transformer based models for echocardiography image registration and compare them with one representative CNN model in unsupervised echocardiography image registration. There are already works using transformers to register medical images, such as TransMorph [8] and Dual Transformer [35], but they are mostly restricted to high signal-to-noise medical images, such as MRI images and CT images. While ultrasound images (2D) are actually the most popular imaging modality in the real world. Our inspiration not only comes from the trending debate over Transformer and MLP, but also stems from the intuition that patch-based learning methods share similar logic to traditional block matching method for cardiac tracking.

Our contributions are two-folded. First, we show the effectiveness of patch-based MLP/Transformer models in medical image registration compared with a CNN-based registration model. Second, we conduct a thorough ablation study of the influence of different structures (MLP, MLP-Mixer, Transformer) and different scales (single scale or multiple scales). Our results provide empirical support to the observation that the attention mechanism may not be the only key factor in the SOTA performances. [21,25], at least in the field of unsupervised image registration.

2 Methodology

2.1 Diffeomorphic Registration

We estimate a diffeomorphic transform between images, which preserves topology and is folding-free. Our model generates stationary velocity field $v(\theta)$ [3] instead of generating displacement maps, thanks to an integration layer applied to the velocity field leading to diffeomorphism $\phi(\theta)$. Formally, the diffeomorphic transformation ϕ is solution to a differential equation related to the predicted (stationary) velocity field V [9]: $\frac{\partial \phi_t}{\partial t} = v(\phi_t)$; $\phi_{t=0} = Id$. In a stationary velocity field setting, the transformation ϕ is defined as the exponential of the velocity field $\phi = \exp(v)$ [2]. The integration (exponential) layer applies the scaling and squaring method to approximate the diffeomorphic transform [18]. The obtained transformation ϕ is then used by a spatial transform layer to deform the image.

2.2 Proposed frameworks

Given two images, I_{fix} and I_{mov} , we would to estimate the transformation $\phi(\theta)$ that transforms the moving image to the fixed image so that $I_{fix} \approx I_{mov} \circ \phi(\theta)$. We approximate the ideal $\phi(\theta)$ by the following proposed frameworks. The following three propositions are all based on patch-wise manipulations and

share a similar general architecture. As shown in Fig.1, I_{fix} and I_{mov} are both processed by an identical feature extractor (green block) separately. The two feature maps are then passed through the cross feature block (blue block). After two linear layers, we obtain their corresponding velocity field. The velocity field passes through an integration layer and we obtain the final displacement field by upsampling it to the original image size.

Pure MLP registration framework The same MLP block (Block I in Fig.1) are used for feature extractor and the cross feature block in this model. The outputs from two separate feature extractor (shared weights) are added together before feed into the cross feature block. We note this model **PureMLP** for abbreviation in the following paper.

MLP-Mixer registration framework The MLP-Mixer registration framework is very similar to the former Pure MLP framework. The only difference is that the three MLP blocks used for separate feature extraction and cross feature processing are replaced by MLP-Mixer blocks [16] (Block II in Fig.1). The MLP-Mixer block has an identical structure as the MLP block, but with feature map transpose to obtain channel-wise feature fusion (the red cell of Block II in Fig.1). We note this model **MLPMixer** for abbreviation in the following paper.

Swin-Transformer registration framework This model uses the MLP block (Block I in Fig.1) to first extract patch based features for both I_{fix} and I_{mov} . For cross feature block, we adapt the recent Swin Transformer [22] to do the cross-patch attention locally (Swin Block in Fig.1). Our Swin block accepts feature input from both images (I_{fix} and I_{mov}), where key K and value V are normalised I_{fix} features while query Q comes from normalised I_{mov} features. Swin block calculates the cross-attention within a pre-defined window region. We perform normal window partition for I_{fix} features and one normal partition, one shifted partition for I_{mov} features. The cross-attention under the two types of window partition configurations are summed together before feeding to the final linear layers. Due to the page limit, we invite interested readers to refer to [22] for detailed description of Swin transformer mechanism. We note this model **SwinTrans** for abbreviation in the following paper.

2.3 Multi-scale features

In order to enforce different reception fields for patch-based models, we decide to combine multi-scale models together. This is accomplished by adopting models of different patch sizes together. It is quite similar to how CNN achieves this goal, by applying larger kernel or adding pooling layers. In particular, our multi-scale model consists of several parallel independent single-scale child models. The output of each child model is upsampled and then combined together to form the final estimation of velocity field $v(\theta)$

$$v(\theta) = \sum_{C=1}^N \omega_C Out_C \quad (2)$$

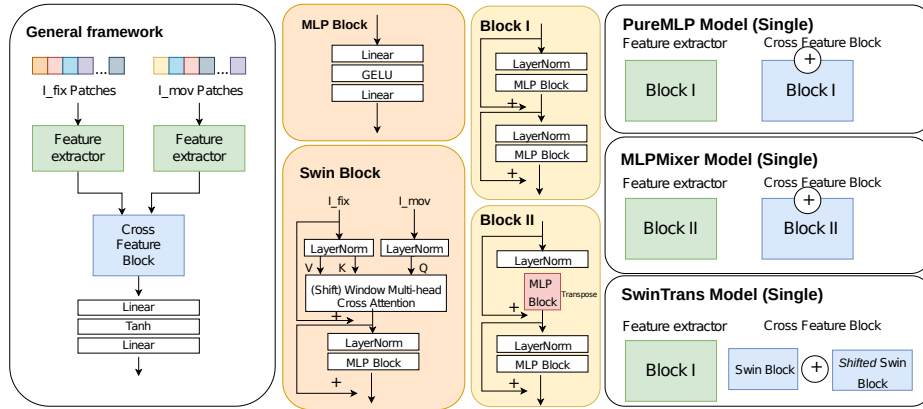


Fig. 1. The detailed composition of proposed three frameworks. Here we only show single-scale models. Please read Section.2.2 for more description.

where Out_C is the output of child model C . The final velocity field v is then passed to calculate the final transform ϕ as depicted in former subsections.

3 Experiments and Results

3.1 Dataset

To evaluate the effectiveness of our unsupervised registration models, we use a publicly accessible 2D echocardiography dataset CAMUS¹. This dataset consists of 500 patients, each has 2D apical 4-chamber (A4C) and 2-chamber (A2C) view sequences. Manual annotation of cardiac structures (left endocardium, left epicardium and left atrium) were acquired by expert cardiologists for each patient in each view, at end-diastole (ED) and end-systole (ES) [20]. The structure annotations of 450 patients are public available while that of the other 50 patients unreleased. In total we have 1000 pairs of ED/ES images and we randomly split (still considering age and image quality distribution) the 900 pairs (with annotations) into training (630), validation(90) and test data (180). The 100 pairs (without annotations) are included into the training set (730).

3.2 Implementation

We compare our proposed three models with a very popular CNN registration model VoxelMorph [4]. To be consistent with our setting, we make use of the diffeomorphic version of VoxelMorph model (we use the abbreviation **Vxm** in the following paper). We train all the models with input images resized to 128x128 pixels and use an Adam optimiser (learning rate = 0.0001). We set training epoch to be 500 and training is early stopped when there

¹ <https://www.creatis.insa-lyon.fr/Challenge/camus/>

is no improvement on validation set over 30 epochs. Our codes are available (https://gitlab.inria.fr/epione/mlp_transformer_registration).

Loss function In order to enforce the diffeomorphic property of our registration model, we apply a symmetric loss function for all the unsupervised models:

$$\arg \min L = L_{mse}(\hat{\phi}(I_{move}), I_{fix}) + L_{mse}(\hat{\phi}^{-1}(I_{fix}), I_{move}) + \lambda * L_{diff}(\hat{\phi}) \quad (3)$$

where $\hat{\phi}^{-1}$ is the inverse of $\hat{\phi}$ and L_{diff} is a diffusion regularizer for smoothness $L_{diff} = \int \|\nabla_x \phi + \nabla_y \phi\|^2$ and set $\lambda = 0.01$ according to [4].

Data augmentation In order to improve model generalisation and avoid overfitting, we apply the same random data augmentation tricks for each image pair during training phase. The following augmentation techniques: rotation, cropping and resizing, brightness adjustment, contrast change, sharpening, blurring and speckle noise addition are conducted with a probability of 0.5 separately. No augmentation is applied during validation nor test phase.

3.3 Experiments

Multi-scale models (abbreviation: model name + _M) we apply three child models for PureMLP, MLP Mixer and SwinTrans (with patch size of 4x4, 8x8 and 16x16 respectively). For child model of size 4x4, 8x8 and 16x16 in SwinTrans, we set the number of window size to be 8, 4 and 2, the number of heads to be 32, 16 and 8 respectively. The dimension of patch-embedding is set to be 128 for all patch-based methods. ω_C in Equation.2 is set to be 0.5, 0.3, 0.2 for child model with patch size of 4x4, 8x8, 16x16 separately.

Single-scale models (abbreviation: model name + _S) we run single-scale models for PureMLP, MLP Mixer and SwinTrans three proposed frameworks (using patch size of 4x4 pixels). The same configuration is set as for child model with patch-size 4x4 in multi-scale models.

3.4 Results

Since our SwinTrans model relies mostly on features of I_{fix} (with skip-connection of I_{fix} features), we report only the metrics related to the transformation $\phi(\theta)$ that $I_{fix} \approx I_{move} \circ \phi(\theta)$. For CAMUS test dataset, we report the Dice score, Hausdorff distance (HD) and mean surface distance (MSD) between ground truth ED mask and transformed ES mask and the Jacobian determinant in the area of myocardium region.

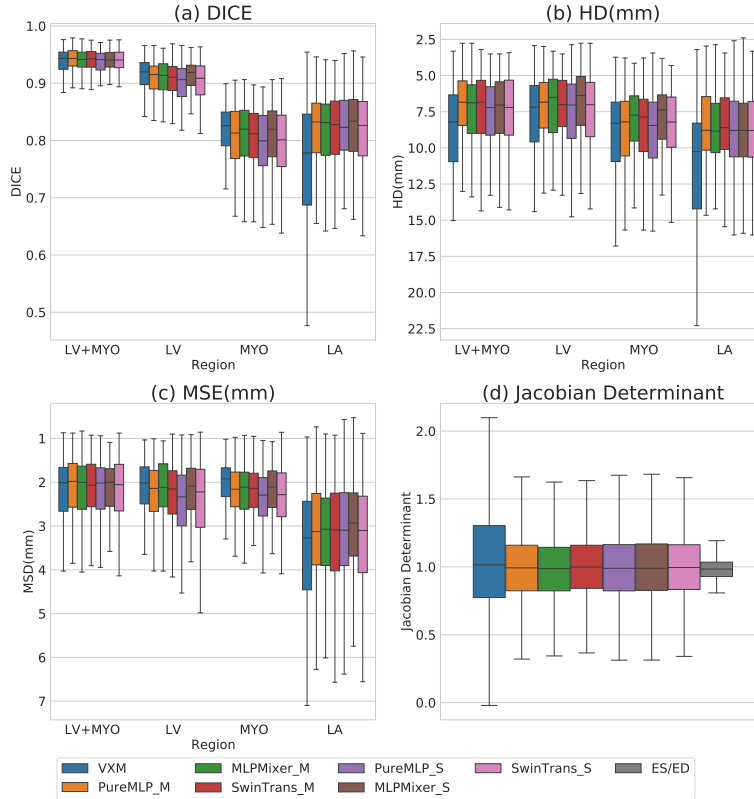


Fig. 2. Comparison of evaluation metrics (Dice score, Hausdorff distance (HD), mean surface distance (MSD) and Jacobin determinant) on test dataset of CAMUS. The Jacobin determinant is only computed in the myocardium region. Except Jacobin determinant figure, the higher the boxplot is in the figure, the better performance it will be.

Evaluation on CAMUS dataset From Fig.2 we can observe that on CAMUS test dataset, almost all the proposed models, no matter it is multi-scale or single-scale, no matter what kind of sub-block it contains (MLP or Transformer or MLP-Mixer), have achieved comparable performance than CNN model (Vxm), in particular for the whole left ventricle and left atrium registration. In addition, the distribution of Jacobin determinant shows that our patch based methods tend to generate more plausible transform, i.e. closer to real ES/ED myocardium area change. This is consistent with the Hausdorff distance results, which indicates that while preserving comparable registration performance, patch-based methods are more resistant to estimation of false large deformation (see the atrium

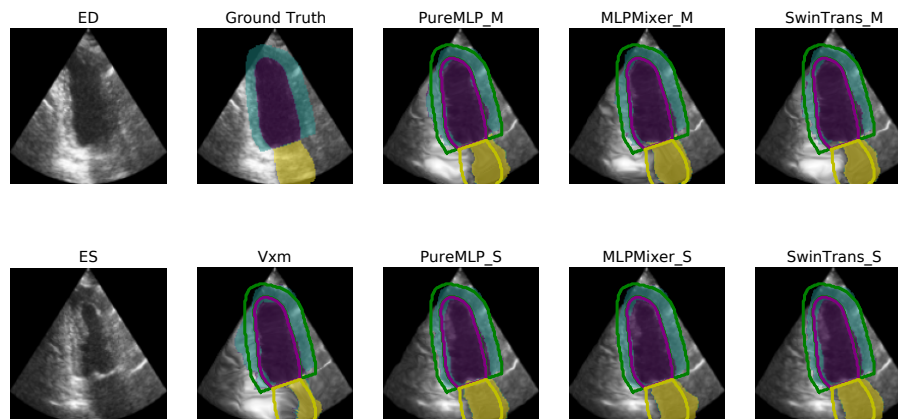


Fig. 3. The same registration example on CAMUS test data with transformed ES masks. Colourful patches are corresponding estimations while bold contours are the ground truth (Yellow: left atrium, Purple: left ventricle, Green: myocardium).

and myocardium region of example in Fig.3). What’s more, single-scale models and multi-scale models have similar performance. With single-sized patches, we are already capable to let feature information flow through the whole image area and estimate registration transform efficiently (see time and space complexity in Table.1).

Table 1. Time and space complexity between different models (evaluated on a GTX 2080Ti)

Model	GPU Memory	Train time (s/pair)	Test time (s/pair)
Vxm	1365 MiB	0.020	0.0047
PureMLP_S	1411 MiB	0.020	0.0038
MLPMixer_S	1447 MiB	0.020	0.0038
SwinTrans_S	1479 MiB	0.029	0.0047

4 Conclusion

In summary, we propose three novel patch-based registration architectures using only MLPs and Transformers. We show that our single and multi-scale models perform similarly and even better to CNN-based registration frameworks on a large echocardiography dataset. The three proposed models demonstrate similar performance among themselves. Our experiments show that patch-based models using MLP/Transformer can perform 2D medical image registration. We shared a similar conclusion with previous works [21, 25] that the success of Transformer

in vision tasks cannot be simply attributed to the attention mechanism, at least in image registration task. Future works will concentrate on the application of MLP/Transformer in time-series motion tracking.

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