Location-Aware Encoding for Lesion Detection in ^{68}Ga-DOTATATE Positron Emission Tomography Images

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***Abstract*— *Objective:* Lesion detection with positron emission tomography (PET) imaging is critical for tumor staging, treatment planning, and advancing novel therapies to improve patient outcomes, especially for neuroendocrine tumors (NETs). Current lesion detection methods often require manual cropping of regions/volumes of interest (ROIs/VOIs) a priori, or rely on multi-stage, cascaded models, or use multi-modality imaging to detect lesions in PET images. This leads to significant inefficiency, high variability and/or potential accumulative errors in lesion quantification. To tackle this issue, we propose a novel single-stage lesion detection method using only PET images. *Methods:* We design and incorporate a new, plug-and-play codebook learning module into a U-Net-like neural network and promote lesion location-specific feature learning at multiple scales. We explicitly regularize the codebook learning with direct supervision at the network's multi-level hidden layers and enforce the network to learn multi-scale discriminative features with respect to predicting lesion positions. We also introduce a learnable fusion layer to automatically combine hidden-layer and last-layer output predictions for lesion detection. *Results:* We evaluate the proposed method on a real-world clinical ^{68}Ga-DOTATATE PET image dataset, and our method produces significantly better lesion detection performance than recent state-of-the-art approaches. *Conclusion:* We present a novel deep learning method for single-stage lesion detection in PET imaging data, with no ROI/VOI cropping in advance, no multi-stage modeling and no multi-modality data. *Significance:* This study provides a new perspective for effective and efficient lesion identification in PET, potentially accelerating novel therapeutic regimen development for NETs and ultimately improving patient outcomes including survival.**

***Index Terms*— Lesion detection, PET, neuroendocrine tumors, deep neural networks, location-aware encoding**

I. Introduction

GASTROENTEROPANCREATIC neuroendocrine tumors (GEP-NETs) are rare, difficult-to-detect tumors which commonly present at advanced stages, with the liver as the most common site of metastases [1]. ^{68}Ga- and ^{64}Cu-DOTATATE positron emission tomography-computed tomography (PET/CT) are widely used molecular imaging techniques for NETs [2]–[4] and show very promising results for accurate staging of GEP-NETs [5], [6]. To develop effective treatments, it is critical to quantify the positive disease burden in PET images. Manual NET assessment in ^{68}Ga- and ^{64}Cu-DOTATATE PET images is very labor-intensive, time-consuming and potentially prone to intra-/inter-observer variation in image interpretation. Thus, many automated approaches have been introduced to improve the efficiency, objectivity and reproducibility of lesion or tumor quantification.

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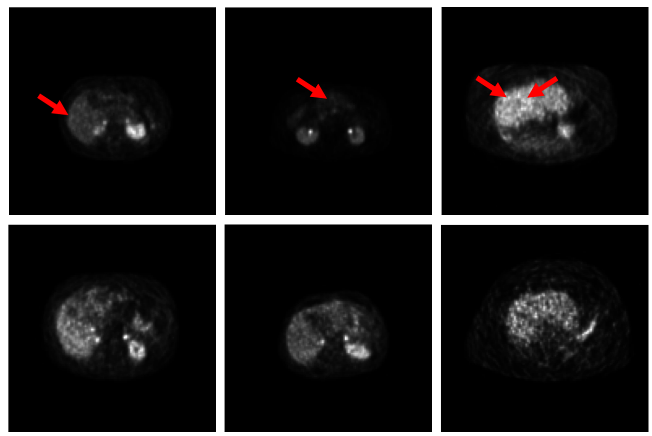


Fig. 1. Some example ^{68}Ga-DOTATATE PET images of livers. Row 1 denotes three different abnormal subjects with each having one or more hepatic lesions (pointed out by red arrows), and row 2 represents three normal subjects without liver lesions.

However, the complex nature of PET images poses significant challenges for automated lesion quantification [7], [8], as shown in Fig. 1. First, PET images usually exhibit low spatial resolution and image contrast such that the boundaries between lesions and surrounding normal regions are not clear. Second, noise is inherently high in PET images compared with anatomical imaging modalities such as CT and magnetic resonance imaging (MRI). In addition, ^{68}Ga-DOTATATE PET imaging typically uses lower administered dose and faster radionuclide decay than the prevailing ^{18}F-fluorodeoxyglucose (FDG)-PET [9] diagnostic tool, so that ^{68}Ga-DOTATATE PET usually has higher image noise and a lower signal-to-noise ratio, thus significantly affecting the lesion detectability. Finally, lesions often show large variability in the shape, size, texture, intensity inhomogeneity and other features, and this further challenges lesion detection and/or segmentation algorithms for PET images.

Thresholding-based methods are commonly used for lesion detection or segmentation in PET images at an early stage, but the assumptions on which these methods rely rarely hold in real-world practice [7], [10]. Afterwards, more advanced imaging processing and statistical or machine learning techniques have been applied to automated lesion quantification, delivering relatively higher accuracy than thresholding [11]–[17]. Recently, deep learning that shows great success in medical imaging [18]–[21] has been applied to PET image analysis, often leading to improved lesion/tumor quantification performance [22], [23]. Many previous approaches take as input a region/volume of interest (ROI/VOI), which is often manually cropped and contains objects of interest only (e.g., lesions or specific organs containing lesions) [7]. These methods use ROIs/VOIs to constrain their outputs and reduce the noise effects outside the ROIs/VOIs, but they require considerable human interaction to isolate the tumors in advance. Some recent deep learning-based lesion quantification methods [23] take the entire PET image as input, but most of them use general-purpose deep neural network architectures without specific algorithm adaptation to PET image data, and there are still much room for improvement. Another large population of approaches use multi-modality images, such as PET/CT or PET/MRI, for lesion quantification [7], [8]. However, these approaches typically require different imaging modalities to be properly aligned or registered, and this may be difficult to achieve in actual practice [7]. In addition, certain tumor boundaries may not be present in CT or MRI images but appear in PET images only, such as liver lesions for GEP-NETs with ^{68}Ga-DOTATATE PET imaging.

In this paper, we propose a novel deep neural network with location-aware feature encoding for single-stage hepatic lesion detection using only PET images (see Fig. 2). Specifically, we design a discriminative codebook learning module and incorporate it into a residual learning-based U-Net-like neural network to enhance feature discriminativeness for lesion detection. We use lesion location labels as auxiliary supervision at hidden layers to directly regularize the training of the codebook, which is thus enforced to encode features that are semantically discriminative with respect to lesion locations. In addition, we introduce a learnable fusion layer to automatically combine the hidden-layer and last-layer output predictions for lesion detection. The entire network is end-to-end trainable and performs lesion identification in a single-pass manner. It requires neither a preprocessing step to crop an ROI/VOI region as model input nor other imaging modalities such as CT or MRI. The proposed method is extensively evaluated on a set of 3D real clinical ^{68}Ga-DOTATATE PET images from 125 subjects and compared with several recent state-of-the-art deep learning approaches. In summary, the contributions are four-fold:

- We build a single deep network for one-stage liver lesion detection using only PET images. The neural network allows both single-stage training and inference, and requires neither image preprocessing (e.g., manual cropping) nor additional deep models to determine ROIs/VOIs a priori. This is significantly different from many current lesion identification methods for PET data.
- We design a novel plug-and-play codebook learning module to assist with discriminative feature learning. We incorporate this module into the network at multi-level hidden layers and highlight lesions in multi-scale feature maps by pruning irrelevant activations. Thus, the network is guided to focus on target lesion regions and alleviate the effects of non-target regions such as image artifacts.
- We apply a learnable fusion layer to automatic information fusion of multi-scale lesion predictions from the intermediate/hidden layers and the last layer. Model training with this fusion layer can boost lesion detection performance compared with the counterpart trained without a fusion layer.
- The neural network does not require additional imaging modalities such as CT or MRI for model training. Using standalone PET image data, the network outperforms recent state-of-the-art deep learning models in hepatic lesion detection.

II. Related Work

Lesion or tumor quantification in PET images is very critical for accurate diagnosis of NETs and assessment of the response to therapy. Early-stage methods are primarily based on fixed, adaptive or iterative thresholding, which usually assumes the image is bi-modal and has uniform intensity inside and outside lesions. However, these thresholding-based methods often produce unsatisfying results due to unrealistic assumptions [7], [10]. Later, more advanced image processing and computer vision techniques are applied to tumor segmentation in PET images, such as active contour/surface models [24]–[27], mean shift [28], difference of Gaussians [29], and other algorithms. Statistical or machine learning has also been adopted to quantify tumors in PET images and has achieved good performance, such as fuzzy C-means clustering [12], [13], Gaussian mixture models [14], [15], fuzzy locally adaptive Bayesian [16], [17], and many others [30]–[36]. However, these methods typically require manual feature engineering for data representation, which is a nontrivial task especially for complex PET images. In addition, recent studies show that traditional machine learning often gives inferior tumor quantification performance in PET images compared with end-to-end deep learning [22], [23], [37].

Deep neural networks, especially convolutional neural networks (CNNs) [38], [39] and their variants [40]–[44], have recently exhibited great power in medical image computing and achieved state-of-the-art performance in various tasks [18]–[21], including outcome prediction of cancer therapy [45] and diagnosis of Alzheimer's disease [46] from PET images. Chen *et al.* [47] have exploited a CNN model to segment initial cervical tumors in ^{18}F-FDG PET images and then applied complex post-processing to tumor refinement. Pfaehler *et al.* [48] have presented a U-Net-based neural network for tumor segmentation in a lung cancer PET image dataset and achieve good lesion quantification accuracy and repeatability. Wehrend *et al.* [49] have adopted a U-Net-like architecture to locate liver lesions in PET images followed

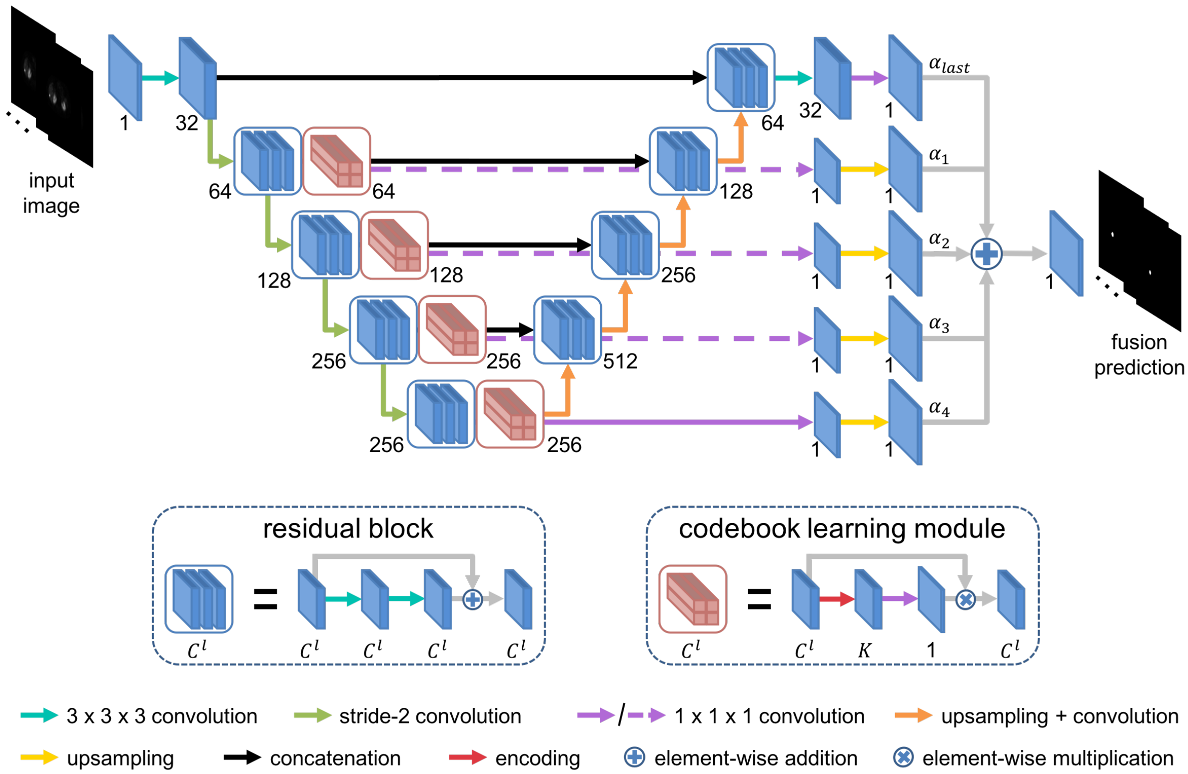


Fig. 2. The proposed neural network with location-aware encoding for single-stage hepatic lesion detection in PET images. The blue boxes represent the feature maps with the number of channels presenting below. The arrows with different colors denote distinct operations. To avoid cross connections, we use dashed lines to represent the $1 \times 1 \times 1$ convolutions on some codebook learning modules for side-output predictions. The encoding layer (red arrow) in the codebook learning module takes as input a C^l-channel feature map and produces a K-channel output feature map, where K is the number of codewords in the codebook.

by using annotated liver masks to refine network predictions. These methods require a pre-defined ROI/VOI to isolate lesion regions from the background, but these ROIs/VOIs are often manually determined and thus need additional human effort.

Some deep models recently take as input the entire PET image for tumor quantification and do not require ROI/VOI cropping in advance. Leung *et al.* [50] have used a U-Net-like architecture [42] to quantify lung tumors with a multi-stage training pipeline, Lu *et al.* [51] have modified the U-Net with a dropblock technique to segment tumors in lung cancer FDG PET images, and Liu *et al.* [52] have applied a Bayesian encoder-decoder neural network to oncological PET segmentation. By considering the information between adjacent slices, Blanc-Durand *et al.* [53] have built a 3D U-Net model for gliomas segmentation in ^{18}F-fluoro-ethyl-L-tyrosine (^{18}F-FET) PET images. Iantsen *et al.* [54] have incorporated residual learning blocks [55] and squeeze-excitation blocks [56] into a 3D U-Net [43] for automated tumor uptake segmentation in cervical cancer PET images and have obtained better performance than conventional thresholding and standard U-Net models. Deep models based on 3D U-Nets have also been exploited to segment head and neck tumors [57] as well as lymphomas [58] in FDG PET images. Despite promising performance of these methods, there is still large room for improvement, especially for those multi-stage models that introduce additional variability in tumor quantification. In addition, none of these approaches are designed and evaluated on DOTATATE PET image data, which typically has higher image noise and lower lesion detectability than FDG PET [9], [49].

A popular line of research for lesion quantification in PET images is to use multimodal imaging data, such as PET/CT or PET/MRI [7], [59]–[62]. Xue *et al.* [63] have introduced a multimodal neural network for lesion segmentation, which uses shared down-sampling blocks between the PET and CT encoding branches for feature co-learning. Jin *et al.* [64] have built a two-streamed neural network to quantify gross tumor volume for esophageal cancer and conducted both early and late feature fusion for PET and CT images. Kumar *et al.* [65] have used a co-learning CNN model to quantify the importance of each modality's features and fuse complementary information from multimodal image data for tumor detection and segmentation. Zhu *et al.* [66] have applied early fusion of PET, CT and tumor distance maps to a multi-branch U-Net-like network for lymph node gross tumor volume estimation, and a similar fusion strategy is used in some other methods that leverage multimodal images for lesion detection or segmentation [67]–[71]. Guo *et al.* [72] have employed different fusion networks to segment tumors using a mixture of PET, CT and MRI images and demonstrated that information fusion at a feature level produce better performance than fusion at the model output level. Learning with multimodal images for tumor quantification assumes an appropriate registration between different modalities, but this assumption might not always hold in reality [7]. Meanwhile, for some diseases, lesions may not be present in the anatomical modality and thus there is no correspondence between tumor boundaries in PET and CT (or MRI) images, such as GEP-NETs with ^{68}Ga-DOTATATE PET imaging. In this paper, we will thus focus on lesion detection using only PET images.

III. Single-Stage Lesion Detection in DOTATATE PET

Our end-to-end lesion detection neural network is built on a 3D encoder-decoder architecture with long-range skip connections, as shown in Fig. 2. The network learns an inherent codebook on-the-fly, which consists of a set of visual codewords to model input data distribution, with multi-scale auxiliary supervision for lesion location-aware feature encoding. The auxiliary supervision is directly linked to multi-level hidden layers of the neural network and enhances the discriminativeness of learned features to facilitate lesion identification. In addition, the network adopts a learnable fusion layer to combine the hidden-layer and last-layer outputs for model training, which allows the network to automatically adjust the contribution of each output prediction and optimize a weighted fusion for lesion detection.

Fig. 2 shows our U-Net-like network architecture, which mainly consists of one contracting path (encoder) and one expanding path (decoder), with each containing four residual learning blocks [73]. The contracting path uses stride-2 convolutional layers to stack the residual blocks, while the expanding path links up its residual blocks with interpolation-based upsampling followed by convolutional operations. One codebook learning module is added on top of each residual block in the contracting path, and the output of each codebook learning module is copied and concatenated with feature maps of the corresponding residual block in the expanding path via a long-range skip connection. A learnable fusion layer is applied to information aggregation of the outputs from the hidden layers and the last layer. All convolutional layers in the residual blocks use a 3D kernel of $3 \times 3 \times 3$, and each is followed by an instance normalization layer [74] and an exponential linear unit [75].

A. Lesion Location-Aware Encoding

Incorporating codebook learning into deep neural networks can enhance expressive power of the networks' feature representations and has produced improved performance in different computer vision tasks, compared with the counterpart without codebook learning [76]–[78]. Inspired by [76], we construct a novel lesion location-aware codebook learning module to encode multi-scale spatial information within input images for lesion localization. Specifically, we tailor the codebook learning technique in [76] and make the following significant improvements: 1) we extend the module to learn rich hierarchical features from 3D volumes for object detection instead of 2D image classification, 2) we use auxiliary, side-output supervision to directly regularize the module learning such that the lesion location-relevant information can be encoded in feature learning, and 3) we insert this module into multiple hidden layers of the neural network to extract multi-scale location-aware features, instead of placing it on only the penultimate layer that learns much coarse-scale features, which may contain limited local details of lesions. Our codebook learning module is also different from [77], [78], which aggregate codebook-encoded features across the entire image such that spatial information is lost; instead, our module learns to capture spatial context and highlight salient regions with auxiliary supervision.

Our 3D codebook learning module mainly consists of an encoding layer, a $1 \times 1 \times 1$ convolutional layer followed by an instance normalization layer [74], and a sigmoid activation function. Specifically, we modify the encoding layer in [76] by changing 2D to 3D operators, removing the aggregation operation for encoded features to avoid losing of image spatial information, and outputting the coding coefficients to directly highlight target regions. Formally, let $\boldsymbol{Z} \in \mathbb{R}^{C^l \times D^l \times H^l \times W^l}$ denote the input feature map of our improved encoding layer, where C^l, D^l, H^l and W^l represent the channel, depth, height and width of the feature map, respectively. The goal of the encoding layer is to learn a visual codebook and use it to encode discriminative features for lesion detection. Specifically, the encoding layer first interprets the feature map $\boldsymbol{Z}$ as a set of C^l-dimensional, voxel-level visual descriptors $\{\boldsymbol{z}_i \in \mathbb{R}^{C^l}\}_{i=1}^{M^l}$, where $M^l = D^l \times H^l \times W^l$. Then, it simultaneously learns an inherent codebook $\boldsymbol{B}$ composed of K codewords, $\boldsymbol{B} = \{\boldsymbol{b}_k \in \mathbb{R}^{C^l}\}_{k=1}^{K}$, and produces an output feature map $\boldsymbol{U} \in \mathbb{R}^{K \times D^l \times H^l \times W^l}$, which contains a group of K-dimensional coding coefficient vectors $\{\boldsymbol{u}_i \in \mathbb{R}^K\}_{i=1}^{M^l}$, one for each input visual descriptor. Instead of relying on hard-assignment coding that is widely used in the traditional bag-of-visual-words (BoVW) model [79], [80], we adopt a soft-assignment coding strategy [81], [82] to address codeword ambiguity and make the codebook learning module differentiable, so that the entire neural network can be trained with standard backpropagation in an end-to-end manner. Specifically, the j-th component of the i-th coding coefficient $\boldsymbol{u}_i$ is

$$u_{ij} = \frac{e^{-s_j ||\boldsymbol{z}_i - \boldsymbol{b}_j||_2^2}}{\sum_{k=1}^{K} e^{-s_k ||\boldsymbol{z}_i - \boldsymbol{b}_k||_2^2}}, \tag{1}$$

where $\{s_k\}_{k=1}^{K}$ are scalar-valued smoothing factors for the assignment, one for each codeword. These factors are automatically learned during model training so as to allow for a finer modeling of the distribution of input descriptors $\{\boldsymbol{z}_i\}_{i=1}^{M^l}$. The $||\cdot||_2$ is an l_2 norm to measure the distance between each pair of input descriptor and codeword. The u_{ij} denotes the degree of membership of descriptor $\boldsymbol{z}_i$ to codeword $\boldsymbol{b}_j$, i.e., soft assignment, and a higher value of u_{ij} means that $\boldsymbol{z}_i$ is closer to $\boldsymbol{b}_j$.

Given the coding coefficients $\{\boldsymbol{u}_i\}_{i=1}^{M^l}$ for the voxel-level input descriptors $\{\boldsymbol{z}_i\}_{i=1}^{M^l}$, it is common to perform an aggregation operation, e.g., $\sum_{i=1}^{M^l} u_{ij}$ summing over all the voxels, to obtain an image-level representation for different visual tasks [76]–[78], [82]. However, this aggregation operation removes the spatial information about the locations of target objects in the input feature map and thus may pose challenges for object localization, such as lesion detection in PET images. Thus, instead of conducting an aggregation operation, we propose to use the voxel-wise coding coefficients $\{\boldsymbol{u}_i\}_{i=1}^{M^l}$ to directly highlight the target lesion regions and suppress irrelevant activations within the feature map. To this end, we add a $1\times1\times1$ 3D convolutional layer followed by instance normalization on top of the encoding layer, and use a sigmoid function as the activation to produce a voxel-wise scaling feature map $\boldsymbol{V}\in\mathbb{R}^{1\times D^l\times H^l\times W^l}$, which contains M^l scaling factors $\{v_i\in\mathbb{R}\}_{i=1}^{M^l}$ (see Fig. 2). Then, we apply a voxel-wise multiplication to $\boldsymbol{V}$ and $\boldsymbol{Z}$, and output a scaled feature map $\boldsymbol{Z}'\in\mathbb{R}^{C^l\times D^l\times H^l\times W^l}$ to emphasize the lesion locations and prune irrelevant responses in other regions. Formally, this computation is formulated as

$$\boldsymbol{V}=\sigma(g(\boldsymbol{U})), \tag{2}$$
$$\boldsymbol{Z}'=\boldsymbol{V}\otimes\boldsymbol{Z}, \tag{3}$$

where $g(\cdot)$ represents the convolutional operation followed by instance normalization, $\sigma(\cdot)$ denotes the sigmoid activation function, and $\otimes$ means the voxel-wise multiplication.

With voxel-level lesion labels (e.g., 3D binary images) on the last layer of a neural network and an appropriate loss function for lesion detection, we can train the network including the codebook learning module using the standard backpropagation algorithm [83]. However, the supervision from only the last layer may not provide sufficient support to the codebook for learning discriminative features in (early) hidden layers [84], potentially leading to performance degradation of lesion detection. Therefore, we propose to directly add auxiliary supervision, i.e., lesion labels, on top of the codebook learning module and enforce it to understand spatial information about lesion locations for enhancement of feature discriminativeness. Specifically, we place another $1\times1\times1$ 3D convolutional layer on the module for a side-output prediction of lesion locations and introduce an auxiliary lesion detection loss to explicitly regularize the module training (see Fig. 2). In this way, the codebook learning module can directly receive gradients from this side-output loss, in addition to the supervision backpropagated from the last layer. Thus, the codebook significantly improves the discriminativeness of encoded features with respect to predicting lesion locations and is specifically optimized for the lesion detection task. Note that our codebook learning module simultaneously builds the codebook $\boldsymbol{B}$ and encodes the features $\boldsymbol{Z}'$ in an end-to-end, supervised manner, by taking advantage of the readily available lesion annotations from training data. This is different from the traditional BoVW model [79], [80], which conducts codebook learning and feature encoding in a separate, unsupervised mode and thus may not be appropriate for supervised-learning downstream tasks.

B. Learnable Fusion of Multi-Scale Predictions

NETs typically exhibit significant scale variation in PET images with volumes ranging from a few to hundreds of voxels. This may pose a great challenge for deep neural networks to learn effective feature representations for different-sized lesions. In particular, the high layers in neural networks extract coarse-scale features and ignore local details, and thus may have difficulty in capturing information for small lesions after conducting several downsampling operations. Inspired by [85], [86], we introduce multiple side-output predictions to multi-level layers of the neural network (see Fig. 2), enforcing it to learn multi-scale feature representations for lesion detection. Specifically, we insert multiple codebook learning modules to the network, with each linking to a residual block in the contracting path, so that each module is responsible for encoding discriminative feature maps at a certain scale. With lesion location-aware feature encoding, only task-relevant activations are progressively merged via skip connections at each scale and upsampled back to the high-resolution space for lesion detection.

In order to directly take advantage of side-output predictions from the codebook learning modules, we incorporate an additional learnable fusion layer into the network so that the hidden-layer, side-output predictions and the last-layer output prediction are fused via a weighted sum to produce a fused prediction. Because the fusion weights are automatically learned during training, the network can dynamically adjust the relative importance of each prediction for lesion detection. Formally, let $\boldsymbol{A}^l\in\mathbb{R}^{D^l\times H^l\times W^l}$ be the output prediction map, before applying a sigmoid activation function, of the l-th codebook learning module, where $l=1,2,...,L$. Similarly, denote $\boldsymbol{A}^{last}\in\mathbb{R}^{D\times H\times W}$ as the output prediction map of the network's last layer, before using the sigmoid function. The prediction $\hat{\boldsymbol{Y}}^{fuse}\in[0,1]^{D\times H\times W}$ of our fusion layer is

$$\hat{\boldsymbol{Y}}^{fuse}=\sigma(\sum_{l=1}^{L}\alpha_l\cdot f(\boldsymbol{A}^l)+\alpha_{last}\boldsymbol{A}^{last}), \tag{4}$$

where $\{\alpha_l\}_{l=1}^{L}$ and α_{last} are the learnable fusion weights for side-output and last-layer predictions, respectively. The $f(\cdot)$ denotes an interpolation-based upsampling operation to resize the side-output predictions to the original scale. In our modeling, we incorporate $L=4$ codebook learning modules into the neural network (see Fig. 2). We do not place a codebook learning module to the first convolutional layer, i.e., on the first skip connection, because it does not capture sufficient semantic context for lesion localization.

C. Loss Function

We formulate lesion detection as a binary voxel-wise classification problem, i.e., lesion voxels versus non-lesion voxels, and optimize the neural network using a weighted binary cross-entropy loss. Because lesions account for a lower proportion of each PET image than non-lesion regions, we assign a higher weight value to the lesion voxels in the loss function for addressing the data imbalance. Note that we add this lesion detection loss to the network's last layer, each codebook learning module, and the fusion layer. Let $\{(\boldsymbol{X}_i,\boldsymbol{Y}_i)\}_{i=1}^{N}$ denote the training data set of N 3D PET images, where $\boldsymbol{X}_i\in\mathbb{R}^{C\times D\times H\times W}$ and $\boldsymbol{Y}_i\in\{0,1\}^{D\times H\times W}$ respectively represent the i-th training image and its associated gold-standard label, which is a binary 3D image with 1's for lesion voxels and 0's for the others. Denote y_{ij}, $\hat{y}_{ij}^{last}$, $\hat{y}_{ij}^{l}$, and $\hat{y}_{ij}^{fuse}$ the j-th voxel value of $\boldsymbol{Y}_i$, $\hat{\boldsymbol{Y}}_i^{last}$, $\hat{\boldsymbol{Y}}_i^{l}$, and $\hat{\boldsymbol{Y}}_i^{fuse}$, respectively. Note that the side-output predictions from the codebook learning modules are upsampled to the original scale $\hat{\boldsymbol{Y}}_i^{l} = \sigma(f(\boldsymbol{A}_i^l)) \in [0,1]^{D\times H\times W}$ for $l = 1, 2, ..., L$, and $\hat{\boldsymbol{Y}}_i^{last} = \sigma(\boldsymbol{A}_i^{last}) \in [0,1]^{D\times H\times W}$. The full loss function $\mathcal{L}$ of lesion detection for the i-th training image $\boldsymbol{X}_i$ is

$$\mathcal{L} = \mathcal{L}^{last} + \sum_{l=1}^{L} \mathcal{L}^{l} + \mathcal{L}^{fuse}, \quad (5)$$

$$\mathcal{L}^{last} = \frac{-1}{|\boldsymbol{Y}_i|} \sum_{j=1}^{|\boldsymbol{Y}_i|} (\beta y_{ij} \log \hat{y}_{ij}^{last} + (1 - y_{ij}) \log(1 - \hat{y}_{ij}^{last})), \quad (6)$$

$$\mathcal{L}^{l} = \frac{-1}{|\boldsymbol{Y}_i|} \sum_{j=1}^{|\boldsymbol{Y}_i|} (\beta y_{ij} \log \hat{y}_{ij}^{l} + (1 - y_{ij}) \log(1 - \hat{y}_{ij}^{l})), \quad (7)$$

$$\mathcal{L}^{fuse} = \frac{-1}{|\boldsymbol{Y}_i|} \sum_{j=1}^{|\boldsymbol{Y}_i|} (\beta y_{ij} \log \hat{y}_{ij}^{fuse} + (1 - y_{ij}) \log(1 - \hat{y}_{ij}^{fuse})), \quad (8)$$

where $\mathcal{L}^{last}$, $\mathcal{L}^{l}$, and $\mathcal{L}^{fuse}$ are the losses for the network's last layer, the l-th codebook learning module, and the fusion layer, respectively. The $|\boldsymbol{Y}_i|$ represents the cardinality of $\boldsymbol{Y}_i$, and β denotes the weighting parameter to control the relative importance between lesions and non-lesion regions. Applying the loss in Eq. (5) to all the training images, we train the entire neural network including the codebook learning module using both standard supervision from the network's last layer and auxiliary supervision from side-output layers and the fusion layer.

During the testing stage, for each new input image $\boldsymbol{X}$, we have multiple 3D prediction maps from the last year ($\hat{\boldsymbol{Y}}^{last}$), the codebook learning modules ($\hat{\boldsymbol{Y}}^{l}$, $l = 1, 2, ..., L$) and the fusion layer ($\hat{\boldsymbol{Y}}^{fuse}$). Considering that the computation in the last and fusion layers takes into account multi-scale feature representations, we apply an average aggregation operation to the predictions from these two layers and obtain a final prediction map for lesion detection as follows

$$\hat{\boldsymbol{Y}}^{final} = \frac{1}{2}(\hat{\boldsymbol{Y}}^{last} + \hat{\boldsymbol{Y}}^{fuse}). \quad (9)$$

To reduce the effects of noisy predictions, we remove the responses with low values in the final prediction map, i.e., those voxels with values not greater than a threshold τ are suppressed, and then apply connected component analysis to individual lesion identification.

IV. Experiments and Discussion

A. Experimental Setup

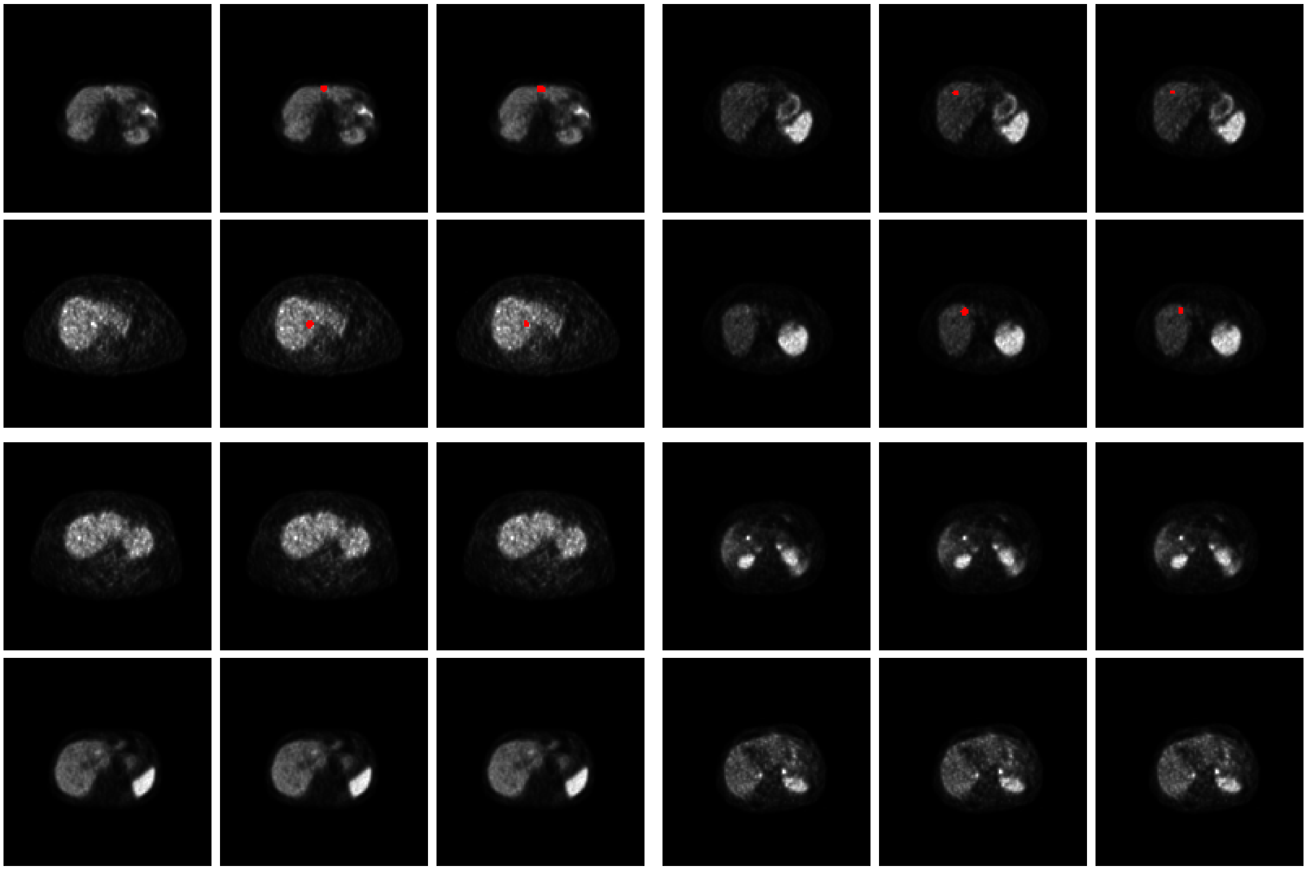
1) Dataset: We acquire a real clinical ^{68}Ga-DOTATATE PET liver image dataset using a photomultiplier tube-based PET scanner. The dataset has 125 subjects with 58 abnormal (i.e., patients with hepatic lesions) and 67 normal cases. Each subject has one 3D PET volume consisting of a certain number of 128×128 transaxial slices, and the number of slices in the liver volume varies from 23 to 71 for different subjects. Each abnormal PET volume has one or more hepatic lesions. Following [49], we randomly split the dataset into training, validation and test sets with a ratio of 6:2:2. This study is determined to be exempt from IRB review by the Colorado Multiple Institutional Review Board at University of Colorado Anschutz Medical Campus.

2) Implementation Details: We empirically set $\beta = 10$ in Eqs. (6) $\sim$ (8) and $K = 16$ for codebook learning. We train our neural network using stochastic gradient descent with Nesterov momentum [87] and set the parameter values as: momentum = 0.99, learning rate = 10^{-3}, weight decay = 10^{-6}, batch size = 2, and maximum number of iterations = 10^5. For each data batch during training, we load 64 slices for each subject and use zero-valued slice padding for subjects with less than 64 slices, i.e., $C = 1$, $D = 64$, $H = 128$ and $W = 128$. We apply data augmentation to model training, including random rotation within $(-10^o, 10^o)$, random horizontal and vertical translation with a displacement in $(-0.125W, 0.125W)$ and $(-0.125H, 0.125H)$ respectively, and random scaling with a factor in $[0.8, 1.2]$. We stop the training process if the performance on the validation set does not improve for successive 2×10^4 iterations. During testing, we use $\tau = 0.1$ to suppress low-valued predictions for lesion detection.

3) Evaluation Metrics: We follow [49], [67] to use precision, recall and F_1 score as the evaluation metrics for lesion detection. We associate automatedly detected lesions with the corresponding gold-standard 3D lesion annotations using the Hungarian algorithm [88]. One detected lesion can correspond to at most one gold-standard annotation, and vice versa. An automatic detection is defined as true positive (TP) if the intersection over union (IoU) between this detection and its associated gold-standard lesion is greater than 5%, otherwise false positive (FP). In addition, automated detections are considered false positive (FP) if they are not matched with any gold-standard lesions, and gold-standard annotations are viewed as false negative (FN) if they do not have associated automated detections. Based on these definitions, we can quantify the metric values as: $precision = TP/(TP + FP)$, $recall = TP/(TP + FN)$, and $F_1\ score = 2 \cdot precision \cdot recall/(precision + recall)$.

B. Model Evaluation

1) Comparison with State of The Art: We compare the proposed method with several recent state-of-the-art deep models, including 3D U-Net [43], V-Net [44], residual squeeze-excitation U-Net (resU-Net) [54], attention U-Net (attU-Net) [89], 3D U-Net with concurrent spatial and channel squeeze-excitation learning (scU-Net) [90], and project-excite FCN (peFCN) [91], as shown in Table I. Each model is run 5 times with different random seeds, and the mean and standard deviation of each metric are reported. As we can see, our method outperforms other competitors by a large margin in all the three metrics, with a range of $3.10\% \sim 52.58\%$ for precision, $12.31\% \sim 28.21\%$ for recall, and $13.77\% \sim$



(a) Original images (b) Predictions (c) Gold standard (d) Original images (e) Predictions (f) Gold standard

Fig. 3. Qualitative lesion detection results using our method. Rows $\mathbf{1} \sim \mathbf{2}$ represent model predictions on multiple abnormal subjects (lesions marked with red color), and rows $\mathbf{3} \sim \mathbf{4}$ denote predictions on several normal subjects without hepatic lesions. Columns (a)/(d), (b)/(e) and (c)/(f) represents the original images, model predictions and gold standard annotations, respectively.

34.27% for F_1 score. In particular, our method provides significantly better performance than the others with p-value < 0.05 in Student's t-test in terms of F_1 score. Although V-Net gives a high precision, it mis-detects a number of lesions and produces a very low recall, thus leading to a 65.15% F_1 score. The attU-Net adopts an attention mechanism to suppress irrelevant responses in feature maps and thus facilitate lesion localization, delivering a relatively higher F_1 score than V-Net as well as other competition approaches. However, all these approaches are significantly outperformed by our method that produces an average F_1 score of 83.24% (with a maximum F_1 of 86.96%), demonstrating the effectiveness of our method on lesion detection in PET images. Fig. 3 shows qualitative lesion detection results using our method on several example PET images.

TABLE I

COMPARISON WITH STATE-OF-THE-ART METHODS IN LESION DETECTION IN TERMS OF THE MEAN AND STANDARD DEVIATION (STD) OF EACH METRIC: $\boldsymbol{mean \pm std}$. THE HIGHEST VALUE OF EACH METRIC IS HIGHLIGHTED WITH BOLD, AND THE $*$ INDICATES THERE IS A STATISTICALLY SIGNIFICANT DIFFERENCE ($\boldsymbol{p}$-VALUE $<$ 0.05) BETWEEN OUR METHOD AND OTHERS IN TERMS OF $\mathbf{F_1}$ SCORE.

	Precision ($mean \pm std$)	Recall ($mean \pm std$)	F_1 score ($mean \pm std$)
3D U-Net [43]	43.46 ± 1.72	56.41 ± 6.49	$48.97 \pm 3.05^*$
V-Net [44]	92.44 ± 5.75	50.77 ± 5.71	$65.15 \pm 4.24^*$
resU-Net [54]	79.70 ± 6.03	45.64 ± 6.36	$57.79 \pm 5.75^*$
attU-Net [89]	84.61 ± 2.44	58.97 ± 2.29	$69.47 \pm 1.92^*$
scU-Net [90]	42.96 ± 4.97	61.54 ± 2.29	$50.32 \pm 2.65^*$
peFCN [91]	57.47 ± 11.92	57.95 ± 4.47	$56.72 \pm 3.73^*$
Ours	$\mathbf{95.54} \pm 3.79$	$\mathbf{73.85} \pm 2.51$	$\mathbf{83.24} \pm 1.93$

2) Ablation Study: In order to evaluate the effectiveness of each component of our method, we conduct an ablation study to report the lesion detection performance of the following model variants: 1) $Baseline$: train a lesion detection model with neither codebook learning nor prediction fusion, i.e., using the loss $\mathcal{L}^{last}$ only; 2) CL : train a model with codebook learning but without prediction fusion, i.e., using the loss $\mathcal{L}^{last} + \sum_{l=1}^{L} \mathcal{L}^l$; 3) PF : train a model with prediction fusion but without codebook learning , i.e., using the loss $\mathcal{L}^{last} + \mathcal{L}^{fuse}$; 4) $Ours$: the proposed method that trains a model with both codebook learning and prediction fusion, i.e., using the loss $\mathcal{L}^{last} + \sum_{l=1}^{L} \mathcal{L}^l + \mathcal{L}^{fuse}$. We run each model 5 times with different random seeds, and report the mean and standard deviation for each metric.

Table II lists the experimental results of the ablation study. Both CL and PF outperform the $Baseline$ model, indicating that incorporating either codebook learning or prediction fusion into model learning is beneficial to lesion detection.

TABLE II

ABLATION STUDY OF LESION DETECTION IN TERMS OF THE MEAN AND STANDARD DEVIATION (STD) OF EACH METRIC: $mean \pm std$. THE HIGHEST VALUE OF EACH METRIC IS HIGHLIGHTED WITH BOLD, AND THE $*$ INDICATES THERE IS A STATISTICALLY SIGNIFICANT DIFFERENCE (p-VALUE < 0.05) BETWEEN OUR METHOD AND OTHERS IN TERMS OF F_1 SCORE.

	Precision ($mean \pm std$)	Recall ($mean \pm std$)	F_1 score ($mean \pm std$)
Baseline	69.82 ± 7.45	64.62 ± 4.1	$66.97 \pm 5.04^*$
CL	92.08 ± 3.45	65.13 ± 2.05	$76.28 \pm 2.37^*$
PF	85.44 ± 3.86	$\mathbf{73.85} \pm 2.51$	$79.13 \pm 1.75^*$
Ours	$\mathbf{95.54} \pm 3.79$	$\mathbf{73.85} \pm 2.51$	$\mathbf{83.24} \pm 1.93$

We note that the CL model increases the F_1 score from 66.97% to 76.28% compared with the $Baseline$, suggesting that codebook learning can encourage the neural network to learn discriminative feature representations for lesion identification. Combining codebook learning and prediction fusion, our method further significantly improves the F_1 score to 83.24%, and this confirms the effectiveness of our method.

3) Effects of Model Parameters: Our method has an important hyperparameter, the number of codewords (K) in each codebook, which controls the expressive power of the codebook. The left panel of Fig. 4 shows the lesion detection performance of our method using different K values. We see that the F_1 score is relatively low when $K = 4$ or 8, compared with $K >= 16$. This suggests that a small K value may not be sufficient for the codebook to capture the input data distribution and thus leads to poor lesion detection. When $K > 16$, the performance improvement gets saturated. This demonstrates that our codebook learning technique is very effective for feature representation encoding such that we do not need a large codebook to model data distribution.

The β in Eqs. (6) $\sim$ (8) is another critical parameter that is used to highlight lesions during model training. The right panel of Fig. 4 shows the experimental results of our method using different β values. As we can see, a small β probably misses many true lesions and gives a very low recall (and thus a low F_1 score), especially for the case $\beta < 1$ which de-emphasizes the lesions for model training. A higher value such as $\beta = 10$ produces much better lesion detection performance with an F_1 score of over 83%. However, a too large β, e.g., 100, may lead to more false positives and thus a lower precision and F_1 score.

V. CONCLUSION

In this paper, we propose a novel U-Net-like neural network for single-stage lesion detection in PET images. It introduces a newly designed codebook learning module into the encoder for multi-scale discriminative feature encoding, and then applies a learnable fusion layer to multi-scale prediction aggregation for lesion identification. The proposed neural network supports single-stage model training and inference, and does not require manual cropping or cascaded models to select ROIs/VOIs as model inputs, which are required by many existing lesion detection methods with PET imaging. In addition, our model is trained with only PET images and does not need other imaging modalities such as CT or MRI, thus eliminating the non-trivial image registration between different modalities. More importantly, this property makes it well suitable for lesion identification in diseases like GEP-NETs, which typically do not have lesion boundaries present in other modalities but PET imaging.

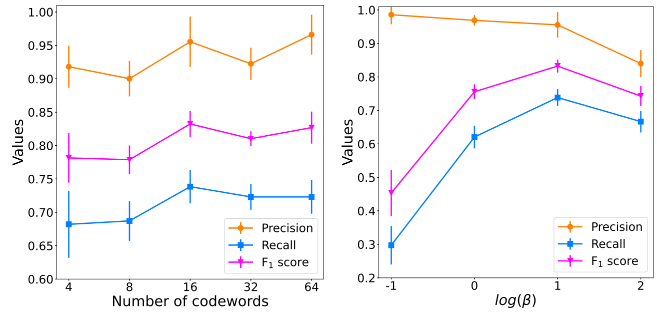


Fig. 4. Lesion detection performance of our method using different numbers of codewords for codebook learning (left) and different β values in Eqs. (6) $\sim$ (8) (right). Each curve represents the mean value of 5 runs with different random seeds, and the vertical lines in each curve denote the standard deviation. Note that the x-axis in the right plot is $\log(\beta)$.

The experimental results demonstrate that the proposed method significantly outperforms recent state-of-the-art deep learning models in lesion detection, with a p-value < 0.05 in statistical tests. We note that the codebook learning module can effectively boost the performance with a small number of codewords (e.g., 16), compared with the baseline model, and this indicates the great ability of discriminative feature encoding. The multi-scale prediction fusion can also improve the baseline model, demonstrating its importance of addressing scale variation of lesions. The experiments also show that it is necessary to tackle the data imbalance issue, i.e., lesions occupy only a very small proportion of each PET image, and appropriately highlight the lesions for model training.

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