A Method towards Automated Thrombolysis in Myocardial Infarction (TIMI) Frame Counting Using 3D Reconstruction

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Abstract

In coronary angiography the flow can be assessed by tracking the contrast agent in the image sequence. The contrast flow is used to estimate the functional behavior of the coronary arteries using TIMI frame count (TFC). In this paper we propose and evaluate a method towards the automation of TFC. The method creates a two dimensional map of the contrast agent in which the opacification of the vessel centerline is plotted against time. This map is used to find the velocity of the contrast agent and subsequently the TFC. The vessel centerline is obtained using the Fast Marching Method to find the minimum cost path between the catheter point and the manually selected endpoint of the vessel. Results from 22 patients show that the algorithm is able to estimate TFC correctly within the intra observer error range of $\pm$ 5 frames compared to manual TFC if the contrast agent is completely opacifying the coronary arteries within one cardiac cycle.

1. Introduction

Thrombolysis in Myocardial Infarction (TIMI) flow grade is a valuable and widely used qualitative measure in angiographic imaging. Since the flow grades are subjective and intra observer variability can be large, Gibson et al. [1] introduced an associated quantitative method called the TIMI Frame Count (TFC). The TFC is measured by counting the number of frames between the entering of the dye in the artery until the dye reaches the end of the artery (most distal landmark, often a bifurcation). The TFC of *e.g.* the right coronary artery (RCA) is around 20 frames at 30 fps (at other acquisition rates a conversion factor must be applied) for healthy subjects. It appears that the TFC from the RCA is comparable to the TFC from the left circumflex artery (LCx). The TFC of the left anterior descending artery (LAD) is significantly higher, therefore a correction factor of 1.7 must be applied yielding a CTFC (Corrected TFC) for the LAD. Bickel et al. [2] confirmed that the semi-quantitative TFC method is superior to the TIMI flow grading method.

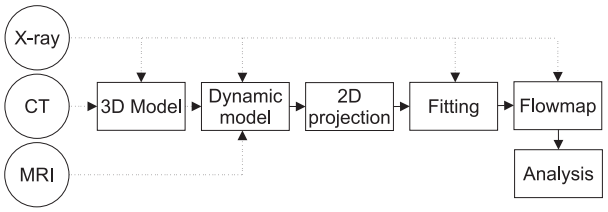

Figure 1: Processings stages and modalities

When hyperemia is introduced pharmacologically using *e.g.* adenosine, we can measure the TFC in basal state and in hyperemic state. Division of the basal TFC (basal flow velocity index) by the hyperemic TFC (hyperemic flow velocity index) results in an estimate of the coronary flow velocity reserve. According to Stoel et al. [3], the Frame Count Reserve (FCR) and the relative FCR (rFCR) can provide a good estimate of Coronary flow Velocity Reserve (CVR) and relative Coronary flow Velocity Reserve (rCVR) and they propose the FCR as a relative simple, fast and inexpensive angiographic method to assess functional significance of coronary stenoses.

Although Computed Tomography Angiography (CTA) is becoming more popular, planar X-ray angiography is still the main modality for coronary intervention and the best modality for measuring contrast flow using TIMI frame counting [4]. In this paper, we propose a method which uses 2D X-ray images to measure the contrast flow velocity based on a similar principle as TIMI frame counting. Figure 1 schematically shows the processing stages and modalities usable for our method. CTA and MRI are only usable partially, while X-ray can be used in all stages of our method.

2. Method

A planar X-ray image sequence, $I(n)$, containing a conspicuous projection of the vessel during and after the injection of the contrast agent is required, with $n = [1..N]$, n the frame in the image sequence and N the number of frames in the sequence. The recorded ECG signal is used to estimate how many frames are recorded in one cardiac cycle. All images are contrast enhanced using log-subtraction of the background. The background, I_b, is created by morphologically dilation with a disc structuring element of size $[15 \times 15]$ pixels. The cardiac cycle is subdivided into phases. From each phase we calculate a three dimensional model using annotation of the main branches in two projection. Each segment of the tree is traced in the image using a minimum cost algorithm using the Fast Marching Method (FMM) [5]. Each trace is converted to a euclidean distance map. The two distance maps are used as cost function for the 3D spline algorithm. In our algorithm, the two traces are reconstructed in 3D space to a 3D spline, then the 3D spline is projected onto the 2D plane. Next, the 3D spline is moved using a minimization criterion based on the two distance maps. The control points of the 3D spline are redistributed so that they are equispaced in 3D space. A maximum of 10 iterations is used to reduce computation time. In this paper from all images the catheter endpoint and the vessel endpoint are manually selected because we want to make sure that the startpoint and endpoint of the vessel are correct. A contrast flow map, $\Gamma(s, n)$, is created from the selected vessel [6]. The flow map consist of N rows. The spatial position s is the normalized length $s = [0..1]$ of the vessel from the start $s = 0$ to the endpoint $s = 1$. A typical flowmap is show in figure 2, the temporal data $I(n)$ is plotted along the vertical axis and the spatial s is plotted along the horizontal axis. Each row represents the opacity at the centerline of the vessel in which the length of the vessel is parametrised by s. The flowmap is created using the centerline of the vessel by tracking it throughout the image sequence.

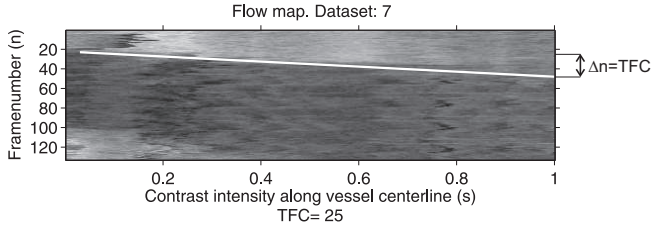

Figure 2: Flowmap for dataset 7.

The vessel centerline is found using the FMM. The FMM is used to build a map containing the travel time between the startpoint and all other points using a speed function. The minimum cost path from the starting point to an end point can be found and is defined by $x(s)$ and $y(s)$ in which s runs from the start of the vessel to the endpoint. $x(s)$ and $y(s)$ are the locations of the vessel in the image at position s. The FMM solves the Eikonal equation:

$$\|\nabla T(x, y)\| F(x, y) = 1 \tag{1}$$

In which $T(x, y)$ is the arrival time of the front, $\|\cdot\|$ is the euclidean norm and $F(x, y)$ is the speed function.

The speed function selected as input for the Fast Marching method is based on the contrast enhanced image $I_c(x, y)$:

$$F(x, y) \quad = \quad (1 - (I_c \otimes G_\sigma))^\gamma \tag{2}$$

where the operator $\otimes\, G_\sigma$ is defined as gaussian filtering with standard deviation σ. The image data is inverted so that low intensity means higher speed. Then the discrimination between low and high intensity is increased using a power of γ. In this research we have used values of $\sigma = 1.5$ and $\gamma = 10$. The fastest route from the startpoint to the endpoint is used as vessel centerline. A typical centerline found by this algorithm is shown in figure 3.

The contrast enhanced image I_c is used to find the pixel values on the vessel centerline. The pre-contrast agent frames do not have a clear vessel centerline. For these cases we use vessel centerlines found in the frame $n + N_p$, in which the constant N_p is defined as the number of frames between two R tops of the ECG. This centerline is written as $\mu(I_c(n), s)$. The data of the rows in the flow map are defined as:

$$\Gamma(n, s) = \mu(I_c(n), s) \tag{3}$$

Pixel values lower than the pixel values measured in the previous N_p frame are used in the contrast flow map $\Gamma(s, n)$. The fastest path through a vessel in one frame is not always of the same length as in another frame. Therefore the whole line of pixel data is scaled using linear interpolation to the length of the contrast flow map.

The mean component is removed from all spatial components by subtracting the mean value in the temporal direction:

$$\Gamma_d(n, s) \quad = \Gamma(n, s) - \tfrac{1}{N} \textstyle\sum_{n=1..N} \Gamma(n, s) \tag{4}$$

Next, we calculate a threshold using a 256 bins histogram from Γ_d . In general, the maximum peak in the histogram belongs to the contrast agent (dark) values. We assume that we can seperate the opacified arteries from the transparent arteries using the value of the maximum peak position in the histogram + 1/8 of the number of bins as threshold. Γ_d is smoothed in temporal direction using a Gaussian with $\sigma = 5$. After this, the threshold is applied. A canny edge detector finds the lines in the contrast flow

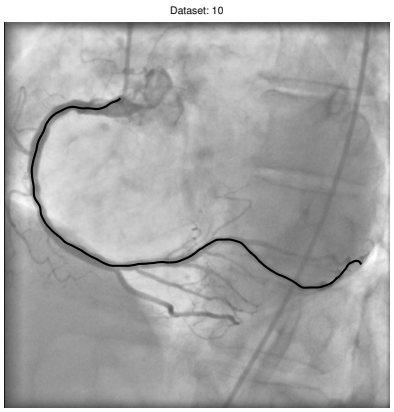

Figure 3: A typical vessel centerline found using the Fast Marching method. In this case only the start and end point is manually selected.

map. One line (or curve) in the contrast flow map indicates where the contrast agent enters parts of the vessel. The assumption that the contrast agent propagates with a constant velocity through the arteries allows us to use a line fit. The lines are found using the hough transform [7]. All lines that do not have a negative slope are discarded as well as lines with a starting point in the second half of the total aquisition time (we assume that the injection and full opacification of the coronary arteries happens in the first half of the recording). Evaluation of the algorithm is done by comparing the results of the TFC measurement algorithm with the manually counted TFC.

3. Experiments and results

The method is evaluated on the coronary arteries of 22 patients from which 6 patients are imaged using a Philips Integris H (P) and 16 patients using a Siemens Axiom-Artis (S) C-arm. All the images are $[512 \times 512]$ pixels. The catheter location and the endpoint of one coronary artery are manually annotated throughout the whole image sequence by an expert. After this, the algorithm is applied. The ground truth (TFCM) was obtained by manual selection of the start and end frames by an expert. The automatic TFC (TFCA) is the output of our algorithm. The measurement is qualified as correct if the absolute difference between TFCA and TFCM is ≤ 5, since 5 equals the intra observer difference [8]. We are working on three-dimensional modelling of the coronary arteries to assist temporal artery tracking and length quantification. Preliminary results are plotted in figure 5.

4. Discussion

The proposed method can estimate the contrast flow using TIMI frame counting in 13 of 22 cases within the expected range of $\pm$ 5 frames. Note that the 22 cases are randomly selected from our patient database without quality pre-selection. In figure 2, a good quality flow map is drawn. It clearly shows an edge between the contrast and non-contrast region, which is also detected by the algorithm. Figure 4b shows a flow map of the data outside the error range. There are a few reasons why the algorithm fails: Inaccurate vessel traces may occur when the selected artery is not completely in view throughout the sequence. In this case, it may happen that the boundary of the image is false identified as part of the artery. Another false tracking may occur when the catheter is clearly visible, the catheter may be tracked as part of the artery. The diaphragm can also be identified as a vessel. Since we assume that the contrast agent propagates linear throughout the vessel, the algorithm fails when we the contrast agent is not linear propagating. Most of the time this occurs when the distal part of the vessel is opacified one cardiac cycle later. This can also be seen in the flowmap from dataset 12. Furthermore, we have to take into account that we are dealing with a two-dimensional projection of a three dimensional tubular structure. Therefore, vessel structure perpendicular to the projection plane result in a higher opacity. This is clearly seen in the flow map of dataset 12.

5. Conclusion

We have presented a method towards the automation of TIMI frame counting. Our algorithm follows the procedure used in clinical practice, the exception is that we not only measure the opacification of the vessel at the begin and endpoint of the vessel, but along the whole vessel centerline. As far as we know, the method presented is the only computer assisted method available for TIMI frame counting. It has the potential to reduce the intra observer variability and increase the reproducability of TIMI frame counting. Our method is able to give a TFC within the error range of $\pm$ 5 frames from 13 out of 22 randomly chosen patients. Furthermore, quantification of the coronary flow can give use more specific information about the functional impact of an apparent lesion. To do this, we have to accurately find the vessels in the images. A 2D projection of a three dimensional object like the coronaries of the heart, likely results in overlapping and foreshortening of vessels. 3D model creation from these 2D images can help to separate these vessels. Messenger et al. [9] indicate that 3D reconstruction is useful for obtaining accurate measurements such as vessel length. Our preliminary results show good reconstruction results and at this time of writing measurements with the reconstructions are in progress.

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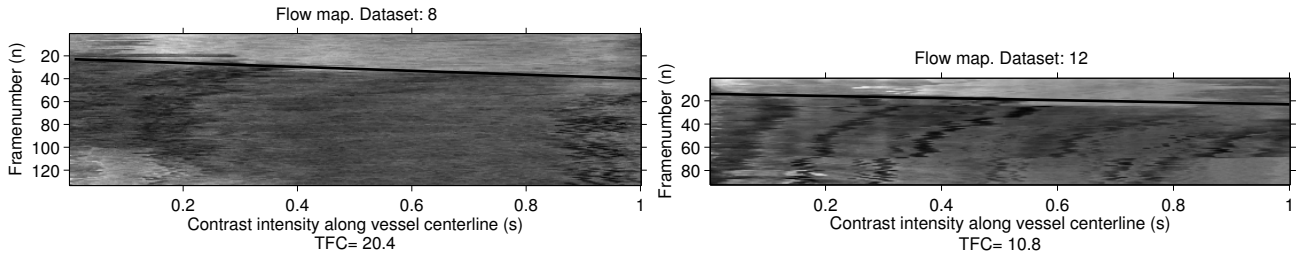

Figure 4: Example flowmaps. Overlayed on the image is the estimated slope which correlates with the TFC.

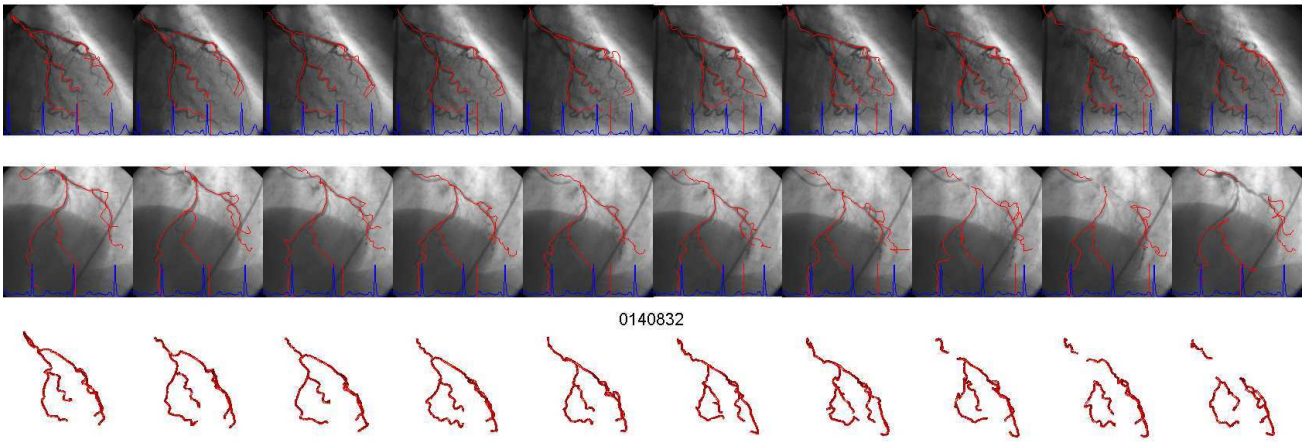

Figure 5: Complete cardiac cycle. The 3D reconstruction in the bottom row is created from pertinent image pair.

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