GroBa: Growing balloons for calibre measurement on stenotic lumens

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Abstract

This paper describes GroBa, a new lumen calibre measurement technique based on growing balloons. GroBa presents the advantages of cross-sectional based methods, as it is able to cope with irregular, non-tubular vessel structures, such as stenosis or aneurysms, but at the same time it is able to obtain precise calibre measurements even when the estimated centrelines are not accurate. Experimental results using phantoms and real subtracted full-body magnetic resonance angiograms show the potential of this work. GroBa is integrated into a fully automatic system that segments the vasculature, obtains its centrelines, measures the lumen calibre at each detected artery, and presents the calibre information in false colours in the maximum intensity projection exploiting the HSV colour-space; all without any human intervention.

1. Introduction

Magnetic Resonance Angiography (MRA) is a non-invasive imaging modality that has been widely used for stenosis detection and screening [1, 3]. Although MRA provides a 3D volumetric data, stenosis grading is usually performed on 2D Maximum Intensity Projections (MIP), incurring into a loss of anatomical information [5]. However, manual segmentation and width estimation on volumetric data is a very complex and tedious work, even for trained personnel; and it is prone to high inter- and intra-observer variability [7]. An automatic measurement able to exploit the 3D anatomical features of the vascular system with high repeatability and accuracy is therefore desirable.

Severe methods exist in the literature to measure lumen calibre. These can be roughly divided into parametric model fitting and cross-section based methods. Methods based on parametric models try to fit a tube-like model into the lumen, exploiting the fact that vessels are elongated and roughly tubular [9, 8, 5]. These methods work very well on regular and uniform vessels, but a cylindrical model is not well suited to capture subtle variations in lumen calibre, as the consequence of mild stenotic symptoms. Even more, vessels are not always cylindrical and as a consequence irregular shapes, even without cross-sectional area variations, might lead to inaccurate lumen calibre measurements. Another common problem with such methods is that they usually require human intervention to indicate the start and the end of the segment the user is interested on, and to correct the fitting in case it falls in a local minima.

Cross-section based methods rely on finding the longitudinal axis (the centreline) of the lumen and obtain a cross-section perpendicular to such axis. The two-dimensional cross-section can then be used to measure the lumen area and calibre. However, small errors when estimating the vascular longitudinal axis may lead to inaccurate calibre measurements [4].

This paper describes GroBa, a new robust lumen calibre measurement methodology based on growing balloons. GroBa presents the advantages of cross-sectional based methods, as it is able to cope with irregular, non-tubular vessel structures, such as stenosis or aneurysms, but at the same time it is able to obtain precise calibre measurements even when the estimated centrelines are not accurate.

GroBa is completely integrated into a fully-automatic al-
algorithm that given a volumetric full-body subtracted MRA, finds and segments the vessel lumen, extracts their centrelines, and robustly measures the calibre at each centreline point. Finally the lumen calibre information is overlayed into the Maximum Intensity Projections (MIP) using the HSV colour space to facilitate a faster and more reliable stenotic diagnosis by the clinician.

This paper is organised as follows. In Section 2 the methodology is detailed, including lumen segmentation with a penalisation step to promote the extraction of leg vessels; centreline extraction; GroBa, the lumen calibre method; and MIP enhancement. Section 3 provides results for synthetic and real data experiments. Finally, Section 4 concludes with the discussion of the method.

2. Methodology

In this section we describe GroBa, the calibre measurement method, as a part of a fully-automatic system for visual enhancement to facilitate stenosis detection. The overview of the system can be seen in Figure 1.

2.1 Lumen segmentation

Full-body, subtracted MRA is a non-invasive imaging technique that aims to highlight arteries in 3D data. However, the contrast used during the process is not perfectly confined to the arteries and other organs and tissues might appear on the volumetric data. Figure 2a shows a usual clinical situation, where the arteries are shown altogether with the heart, kidneys, bladder, and some muscular/fatty tissue in the abdominal and thoracic regions.

To remove the non-arterial information from the volume we first apply a vessel enhancement technique [2] that promotes the extraction of tube-like structures, producing the *vesselness* (likelihood of vessel) volume in Figure 2b, where each voxel contains a scalar level of vesselness at that position.

To be able to segment the clinically relevant vessels in the legs and neck, while minimising the false positives generated by the contrast present in the organs of the abdominal area, we apply a boost function, Eq. *(1)*, that increases the vesselness values of voxels the further they are from the abdominal area.

\[
p_i = \frac{v_i (w - e^{-\frac{(x_i - c)^2}{2\sigma^2}} (w - 1))}{p_i}
\]

where \(v_i\) is the vesselness value, \(p_i\) is the vesselness value after boost, \(x\) is the normalised coordinate point in the top-bottom axis where \(x = 0\) corresponds to the top voxel and \(x = 1\) to the bottom one, \(w\) controls how much penalisation is applied to the abdominal area, \(c\) denotes where is the centre of the region we want to penalise, and \(\sigma\) the size of that region. In our experiments we have used \(w = 3\), \(c = 0.35\), and \(\sigma = \frac{b}{8}\). The weighting function can be seen in Figure 2e. The penalised volume can be seen in Figure 2c.

Usually full-body MRA are in fact four different scans obtained at slightly different times stitched together by the acquisition machine. As a consequence differences in contrast and brightness might occur between these four regions and a threshold to segment the arteries from the rest of the volume uniform for all regions might produce suboptimal results. Nevertheless, in the data we have examined the bottom region is always the darkest one and with very uniform brightness in the vessels, so we use that region to find the automatically the threshold.

The threshold \(t\) is then obtained as \(t = \frac{b}{2}\), where \(b\) is the maximum vesselness level from the bottom fourth of the volume, roughly corresponding to the knee down part of the body. As it is better to have false positives than false negatives, \(t\) is set to a deliberately conservatively low level; as a consequence with \(t = \frac{b}{2}\) and \(t = \frac{b}{4}\) the results are very similar, once \(t > \frac{b}{2}\) the false negatives grow rapidly. Each voxel with a vesselness level under \(t\) is considered non-vessel and set to 0. Figure 2d shows the vesselness volume with the non-vessel voxels set to 0.

2.2 Centreline extraction

Using the binary segmented volume, a skeletonisation methodology based on fast-marching techniques [6] is applied in order to obtain the centreline of all the main arteries with sub-pixel accuracy. The lumen calibre will be now measured at each centreline point.

2.3 Calibre estimation using GroBa

GroBa exploits the fact that vessels present a tube-like structure, however it is able to capture small asymmetric variations in shape and cross-sectional area as we show in the experiments section. The methodology is as follows.

Give any arbitrary point \(p = [x, y, z]^T\), a balloon of 1 voxel volume is initialised centred at \(p\). The balloon is iteratively grown using binary dilation inside the segmented area.
lumen (only voxels that have been previously segmented as lumen are considered) until the balloon height is twice as large as its calibre or no new neighbours are found. At each iteration, the diameter and the height of the balloon is computed exploiting the fact that the lumen has an elongated structure. First the two most distant voxels in the balloon are obtained by finding the furthest voxel to the initial balloon centre and then finding the most distant voxel to it, the euclidian distance between these points is set as \( d \) and it corresponds to the three-dimensional diagonal of the cylindrical balloon.

Assuming that the balloon has a roughly cylindrical shape, it is possible to compute its radius and height given the volume \( v \) of the balloon (the sum of the voxels) and its diagonal \( d \), given the formula of the volume of the cylinder in Eq. (2), the Pythagoras rule in Eq. (3), and the symbols explained in Figure 3,

\[
\begin{align*}
  v &= \pi r^2 2h \\
  r^2 &= s^2 - h^2
\end{align*}
\]

where \( r \) is the unknown radius of the cylinder, \( h \) is the unknown half-the-height of the cylinder and \( s \) is the known half-the-diagonal of the cylinder, being \( s = \frac{d}{2} \), as seen in Figure 3. Using substitution we obtain Eq. (4).

\[
v = \pi 2h(s^2 - h^2) \\
v = 2\pi s^2 h - 2\pi h^3
\]

where the only unknown is \( h \), so we can obtain \( h \) as the roots of Eq. (5).

\[
-2\pi h^3 + 2\pi s^2 h - v = 0
\]

Once we have obtained \( h \), the cylinder height \( c \) is obtained as \( c = 2h \), the radius \( r \) as \( r = \sqrt{\frac{2}{\pi}} h \), and the lumen calibre \( l \) at the last iteration as \( l = 2r \).

2.4 Image enhancement

Once the lumen calibre is obtained for each centreline point of the segmented lumen, the MIPs are enhanced in order to show the calibre information to provide an easier way to diagnose and screen stenosis-related pathologies, exploiting the HSV (Hue, Saturation, Value) colour-space.

The goal is to generate 3 MIPs (frontal, lateral and sagittal) that will show the original volume with calibre information coded as colour. To do that we use the HSV space, so the H and S channels will contain the calibre information, and the V channel will be the MIPs of the original subtracted MRA volume.
Figure 4: Sagittal, lateral, and frontal projections with calibre information added as colour. Notice how the 2-pixel calibre variation in the femoral artery is easily spotted thanks to the enhanced visualisation.

To extrapolate the calibre information into each voxel previously segmented as lumen, we use morphological dilation, so each lumen voxel would be associated with a scalar (normalised between 0 and 1) that indicates the local calibre. The MIPs of this volume correspond to the hue channel of the enhanced MIPs, so each different calibre will show in a different colour).

The saturation channel of the enhanced MIPs is set to 1 for pixels that show any segmented lumen (so they will show a fully saturated colour) and 0 for pixels that do not show any (so they will be gray).

Finally, the value channel is set to the MIPs of the original MRA volume averaged with the vesselness volume. This is done so the vessel pixels show brighter colours.

The final result can be seen in Figure 4. The colour range is limited to $[0, 20]$, so any lumen with a calibre over 20 is shown in bright red. Notice how this visualisation allows to clearly see the 2-pixel calibre variations in the femoral artery.

Figure 5: The green profile shows the synthetic lumen cross section as a result of subtracting the stenotic profile (in dashed blue) with $s = 1$ and $\sigma = \frac{3}{4}$, from the original uniform lumen (in red) of $r = 4$.

3 Experiments

Synthetic and real data has been used to validate the lumen calibre measurement technique.

3.1 Synthetic data

First, a synthetic lumen has been created using a polar coordinate system in order to generate data showing asymmetric stenosis and aneurysms. The contour of such lumen is defined using Eq. (6).

$$\rho = r(1 - se^{-\frac{(\theta - \theta_0)^2}{2\sigma^2}})$$  \hfill (6)

where $r$ is the lumen radius, $s$ is the amount of asymmetric stenosis (being 0 for non stenotic lumen, and negative for aneurysms), $\theta = [0, 2\pi]$, and $\sigma$ defines the shape of the stenosis denting in the shape. Figure 5 shows the contours of cross-section of a synthetic lumen using this technique. Figure 6c shows an entire luminal volume created using this technique.

A 200 pixel-long synthetic lumen has been generated using this method. We used a variable $r$ and $s$ along it. This phantom allowed us to test the GroBa technique compared with the ground truth and other lumen calibre measurement techniques. The ground truth calibre is obtained by averaging the radius of the parametric vessel, integrating over the polar coordinates from 0 to $2\pi$ and then multiplying it by 2. The centreline has been obtained as the centroid of each perpendicular cross-section. To test the robustness of GroBa to noisy centrelines, Gaussian noise with a standard deviation of 1 pixel is added to the centreline coordinates. Trying to reproduce usual techniques used in real world applications, the noisy centreline is then smoothed with a uniform filter with a length of the mean lumen calibre ($\sim 10$ pixels in our experiment) in order to generate a more realistic centreline.
Table 1: Experimental results: Sum of absolute error.

<table>
<thead>
<tr>
<th></th>
<th>Max</th>
<th>Mean</th>
<th>Std. Dev.</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Noisy centreline</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>GroBa</td>
<td>0.710</td>
<td>0.212</td>
<td>0.159</td>
</tr>
<tr>
<td>MIPS</td>
<td>1.241</td>
<td>0.453</td>
<td>0.314</td>
</tr>
<tr>
<td>Cross-section</td>
<td>36.357</td>
<td>6.875</td>
<td>8.315</td>
</tr>
<tr>
<td><strong>Smoothed centreline</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>GroBa</td>
<td>0.651</td>
<td>0.207</td>
<td>0.166</td>
</tr>
<tr>
<td>MIPS</td>
<td>1.094</td>
<td>0.447</td>
<td>0.302</td>
</tr>
<tr>
<td>Cross-section</td>
<td>1.547</td>
<td>0.236</td>
<td>0.263</td>
</tr>
</tbody>
</table>

GroBa is tested against two different calibre measurement techniques, a cross-section based method, and a second method that uses the MIPs to obtain the luminal calibre. The cross-sectional method cuts the lumen volume at each centreline point $p_i$ with a plane oriented according to the normal vector $(p_{i+1} - p_i)$. The area of such cross-section is obtained and the calibre is estimated as $2\sqrt{\pi a}$. The MIPS method measures the lumen calibre at the frontal and lateral MIPs and an average value is obtained.

Table 1 shows the maximum, mean and standard deviation of the calibre error measured as the sum of the absolute differences for GroBa and the comparing methods. GroBa achieves the best results with 0.21 pixels of mean error (measured as the absolute difference in calibre) in both cases: using a very noisy centreline and its smoothed version, showing its robustness to inaccurate centreline positions. The cross-sectional method obtains very similar results when the centreline has been smoothed, but it can be seen in Figure 6a that a noisy centreline leads to large calibre overestimation due to the fact that the cross-sections are not perpendicular to the lumen. Calibre measurement obtained on the 2D MIPs are less accurate as anatomical data is lost in the projection step.

Figure 6b shows how GroBa is able to cope with small variations in lumen calibre due to a narrowing (stenosis) and widening (aneurysms). The 3D representation of the synthetic vessel used in this experiment can be seen in Figure 6c.

3.2 Real data

The full automatic algorithm, including lumen segmentation, centreline extraction, lumen calibre estimation and MIPs enhancing was run in data from 8 different patients with a resolution of 80,000,000 voxels, and a voxel size of 1mm. The 8 MRA volumes were screened by a clinician and any stenotic region was annotated. The algorithm showed its ability highlight stenotic regions allowing for an easier and quicker visual inspection; Figure 4 shows that a narrowing of only 2 pixels in the femoral arteries can be easily spotted on the enhanced MIPs. A detail of a different patient also with stenosis in the femoral artery is shown in Figure 7.

As the method relies on an initial lumen segmentation in order to measure accurately the lumen calibre, errors in that step might result in poor calibre measurements. GroBa has also the limitation of being inaccurate close to junctions, nevertheless it gives reliable measurements at a distance of only 1 lumen calibre from the junction (less than 10 pixels in some cases).

4 Discussion

This paper has presented GroBa, an algorithm based on growing balloons that is able to measure with sub-pixel accuracy the calibre of irregularly shaped lumens. GroBa is part of an automatic system that is able to find and segment the lumen in subtracted MRA volumes, find its centrelines and measure the calibre for the main arteries without any
human intervention. Finally, the HSV colour space is exploited to visually enhance the MIPs, adding the calibre information as colour.

The method has been tested on synthetic data, where it has shown a mean error of 0.21 pixels, even when the centreline included a large amount of noise; and in real data, where the method has shown its potential for an easier and quicker screening of stenosis.

As future work, an automatic stenosis detector is intended to be developed. This new method will use the lumen calibre obtained by the current system and detect regions with stenosis so they can be highlighted. This will also allow to compare quantitatively the output of the system with clinical annotations in real data.

References


