Color-Coded Depth Information in Medical Volume Rendering

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In this thesis a segmentation to produce a binary 3d-volume is made, followed by a distance transform to approximate the Euclidean distance from the centerline of the vessel to the background. The distance is used to calculate the smallest diameter of the vessel and that value is mapped to a color. This way the color information regarding the diameter would be the same from all the projection angles.

Color-coded MIPs, where the color represents the maximum distance, are also implemented. The MIP will result in images with contradictory information depending on the angle choice. Looking in one angle you would see the actual stenosis and looking in another you would see a color representing the abnormal diameter.

Color-coding, Distance transforms, Volume Rendering, Stenosis
Abstract

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1 Introduction

1.1 The Problem

Volume rendering is one of the new techniques in medical imaging. It gives the possibility to look at large volumes of data in three dimensions, for example data from a Magnetic Resonance Imaging, MRI, or Computed Tomography, CT. In the area of vascular diseases it is desirable to examine the occurrence and rate of stenosis, which is a narrowing of a blood vessel. The problem is that the observer, presumably a radiologist, becomes dependent on the projection angle. To discover these stenosis areas the radiologist needs to twist and turn the volume frequently, which is of course time consuming and dependent on a highly developed interaction system.

In this thesis filtering, segmentation, distance calculations and color-coding of the distance, from the centerline of the vessel to the background, are to be implemented. A clinical evaluation is to be made later on, in another project, to determine whether the detection of stenosis in the volume rendering became easier. In fig 1-1 a synthetic image of a vessel is shown to illustrate the problem. The narrowing is only visible from the left and the right, not from the front or back. Two original volume renderings of a MRI dataset are shown in fig 1-2. The stenosis is in the upper part of the aorta. The stenosis is harder to detect from certain angles.
1.2 Aims

In this thesis a color-coding algorithm is to be implemented that represents distances in the volume. This will be presented both as a volume rendering and as Maximum Intensity Projections. It should be possible to vary the color-coding so it is easy to apply on different vessel sizes. The color-coding should make the narrowing illustrated in fig 1-1 visible from the front and back directions.
Figure 1-2. Two original, volume rendered MRI datasets seen from different angles. The arrow shows the stenosis.
2.1 Magnetic Resonance Imaging

Magnetic Resonance Imaging, MRI, is a technique that uses a strong magnetic field to create images of the inside of the human body. MRI was developed in the beginning of the 1970’s and was built upon nuclear magnetic resonance, NMR, a phenomenon that was discovered in the late 1940’s by F. Bloch and E. Purcell. NMR arises in the nuclei of certain atoms when they are immersed in a static magnetic field (MR-camera) and exposed to a second oscillating magnetic field.

Nuclei with an odd number of protons, and/or an odd number of neutrons have a nonzero nuclear spin. Spin is a property of nature like electrical charge. One of the atoms that have a nonzero nuclear spin is hydrogen, which occurs frequently in our body tissues. The protons become capable of receiving and then transmitting electromagnetic energy. The strength of the transmitted energy is proportional to the number of hydrogen protons in the tissue. This information is used to produce an image [19]. There is a number of different MR imaging techniques to choose from, which bring out some properties and not others. One of them is Magnetic Resonance Angiography, MRA. Fig. 2-1 demonstrates the MRI technique.
1. The scanner’s magnetic field realigns the protons of body’s hydrogen atoms so that the all spin along the same axis.

2. When protons are placed in a magnetic field they behave like a magnet, i.e. they have a north and south pole. The protons are now aligned along the body direction. Particles with opposite sign spin pair up. But there are a few that do not. In MRI it is the unpaired nuclear spins of hydrogen protons that are of importance.

3. The scanner next sends out a radio pulse. The radio pulse makes some of the atoms that are unpaired spin at a particular frequency and direction, depending on the type of tissue they make up. When the pulse shuts down the atoms return to their natural alignment and release energy.

4. The energy gives off a signal that the camera picks up. The computer processes the signal and produces an image of the different types of tissue.

Figure 2-1. The MRI technique

2.2 Magnetic Resonance Angiography

Magnetic Resonance Angiography, MRA, denotes a number of MRI techniques for imaging the flowing blood in the vascular system. It provides stacks of parallel cross-sectional images, which show the blood in the vessels as bright regions, surrounded by dark static tissue.

MRA is interesting for many physicians working in vascular diseases because of its ability to non-invasively visualize those diseases. Its potential to replace conventional X-ray angiography that uses iodinated contrast, a substance that contains iodine and appears white on the X-ray, has been recognized for many years. This interest in MRA has been
increased because of the low costs, the evaluations showing satisfied patients and the fact that the invasiveness is minimal.

There are three techniques commonly used in performing MRA. They are:
1. Time-of Flight, TOF, angiography
2. Phase Contrast angiography, PCA.
3. Contrast-enhanced MRA, CE-MRA.  
   The first two methods, TOF and PCA, both rely on the motion of blood to make it visible, and are completely non-invasive. The third, CE-MRA, requires intravenous injection of a paramagnetic substance (commonly gadolinium), which increases the signal intensity of blood upon introduction into the circulatory system [20]. In this thesis, only CE-MRA is considered.

### 2.3 Vascular diseases

Two types of vascular disorders are easily depicted with MRA: *stenosis*, a constriction or narrowing of the artery by the buildup of fat, cholesterol and other substances over time in the vascular wall, and *aneurysms*. An aneurysm is a segmental dilatation or ballooning-out of the wall of an artery, vein or the heart due to weakening of the wall. Aneurysms are often caused by high blood pressure or wall shear stress [1]. In other words, stenosis means decreased diameter and aneurysm means increased diameter of the vessel. The two categories are illustrated in fig 2-2.

![Figure 2-2. The left image shows an aneurysm. The right image shows a stenosis.](image)
2.4 Stenosis detection methods today

2.4.1 Digital Subtraction Angiography

Until now, the most common method for stenosis measurement has been 2D digital subtraction angiography, DSA. In DSA, X-ray images of arterial blood vessels are acquired both before and after intraarterial injection of a radiopaque contrast medium (appears bright in the X-ray image). These images are in digital form and a computer "subtracts" one set of data from another with the resulting image showing only the contrast and, therefore, the shape of the arterial lumen through which the blood is flowing. DSA is widely used due to its high spatial resolution and widespread availability. Unfortunately, this technique suffers from several disadvantages. For example it is invasive i.e. it requires arterial catheterization and injection of contrast agents.

During the X-ray image acquisition the patient is subjected to radiation. Since DSA is a projection technique, vessels may be over-projected, which may make the visualization of the artery of interest more difficult. Since the estimation of the degree of stenosis is based on 2D images, the measurements may not perfectly reflect the dimensions of the 3D vasculature.

2.4.2 Magnetic Resonance Angiography

MRA is gaining popularity as a potential replacement of DSA for diagnostic imaging. It does not involve radiation and the contrast agents used have fewer side effects than those used for DSA. Although it is possible to obtain 3D reconstruction of the vasculature with MRA, it is difficult and the visualization of such images must be made retrospectively by generating projection views. Popular techniques for this are Maximum Intensity Projection, MIP, or Closest-Vessel Projection, CVP visualization technique [2]. In the 3D MRA volume, the voxels representing vessels are expected to have higher intensities than the voxels from static tissues. Therefore, MIP has been used to visualize the vascular structures. Read more about MIP in chapter 2.5.

Usually the radiologist uses the diameter of a normal segment of the artery as a reference. The reference diameter is often calculated as the mean diameter along the artery and proposed by the system to the radiologist that may either validate it or change it. A stenosis is associated with a significant narrowing of the artery and is quantified by such parameters as the diameter reduction expressed as a percentage [3]. Automated methods for this calculation have been introduced.
2.4.3 Doppler Ultrasound

Another method is Doppler ultrasound. Doppler ultrasound is one of the few ways that blood flow information can be obtained non-invasively. It is widely used to measure blood flow and thus useful in the detection of stenosis since a change in the blood flow can be detected in the stenosis area.

The Doppler ultrasound unit has a split transducer in which one half of the transducer sends and the other half receives high frequency sounds. A sound wave is emitted from the left side of the transducer. This wave travels through tissue until it encounters flowing blood in the vessel. Ultrasound is strongly scattered from red blood cells and the Doppler shift of the returning echoes gives information about the velocity of the blood.

The blood flow velocity can then be shown to the physician using various methods e.g. loudspeakers for listening and color-coding. High Doppler frequencies are created by the faster moving red blood cells. The different frequencies are encoded with a color scale that represents the distribution of flow velocities inside the vessel. This information is particularly useful in assessing phenomena like turbulence and blood spurting through constrictions in vessels due to sclerotic plaques in stenosis.

2.5 Visualization methods

2.5.1 Maximum Intensity Projection

MIP displays at each pixel in the image the maximum of the intensities of the voxels that project on that particular pixel. The method of calculating a projected pixel value is simple: The maximum gray value that is detected along the projection ray is transmitted into the observer image plane, as shown in fig 2-3.
MIP processing makes the anatomy easy to understand but it has a non-physical intensity behavior and does not have the relationship to density that is familiar to the reader of radiographic films. Furthermore, MIP tends to underestimate vessel width and overestimate the extent of stenosis. Vessel width is reduced because of the lower signal intensity in marginal pixels due to laminar flow i.e. it flows in straight lines at a constant velocity. MIP also has the drawback of giving no information about the relative position of vessels along the projection axis [4]. Fig 2-4 gives two examples of MIPs.

Considering the simplicity of the method, the resulting images are of remarkably good quality, at least at a first glance. Nevertheless a number of deficiencies result from the algorithm's tendency to ignore relevant pieces of information, which are present in the original image data:

- Loss of small, low-contrast vessels.
- Reduction of vessel diameters.
- Lacking depth-cues.
- No reference to fixed tissue.

MIP reduces the original contrast between vasculature and background. The reason for this is that rays, which do not hit a vessel, show the maximum value encountered, on the viewer plane. Thus for "background rays" the maximum noise is considered which results in increased noise. Especially smaller vessels with low contrast are painted over by noise from adjacent or distant sections in the volume. Another unwanted effect is the reduction of vessel diameters. Physical phenomena, such as turbulence, slow flow and small vessels in relation to pixel size, can also cause the edges of vessels to be less intense than the centers. These regions are likely to be obscured by background
noise. This is a problem of particular relevance for the estimation of vessel diameters, e.g. in the diagnosis of stenosis.

For the rendering of 3D scenes depth-cues are very important. Since the MIP algorithm ignores available depth information, the three-dimensional spatial relationships of neighboring vessels cannot be correctly analyzed with a single projection image. To a certain degree, depth recognition may be provided by motion studies, in which image series with slightly different viewing angles are displayed. But in many situations analysis of anatomic details requires a stable scene.

Another argument put forward particularly by neurosurgeons is the lack of reference to stationary tissue in MIP images. MIPs focus is solely on vascular structures, therefore suppressing soft tissue information. In this way MIP images are similar to Digital Subtraction Angiography projections [5].

![Figure 2-4. Two MIPs. To the left, calves. To the right, heart and aorta.](image)

### 2.5.2 Volume Rendering

3D data can be hard to visualize. One technique is to use a volume renderer to create images from a 3D data set. One way to understand the idea of volume rendering is to compare it to another technique for visualizing 3D data, surface rendering.

With surface rendering, the idea is that a threshold value is used to decide where an isosurface (3D contour) should be drawn. For example if there is a volume of data consisting of values from -10.0 to 10.0, a decision can be made that anything less than 0.0 is inside the volume of interest, and anything greater than 0.0 is outside. Calculations of a
surface at this threshold level can be made and then rendering of this surface is made using polygon rendering techniques [19]. This has the advantage that it is fast and easy to do, once the calculations have been made the first time. However, it also throws away tremendous quantities of data that probably is of interest.

With volume rendering the data is preserved because every voxel within the data set is considered. For example, with a data set consisting of intensity values, rendering of volumes with transparencies assigned to each voxel can be made, which would lead to a much better picture of the considered data.

With surface rendering, it is hard to manipulate the rendered volume in useful ways. It is often helpful to be able to take slices from the rendered volume at arbitrary angles, cut parts out of it and so forth. If just the surface of the volume is rendered this becomes difficult and may not necessarily produce desirable results. Conversely, when every point in the volume is considered as in volume rendering, actions such as clipping, taking slices from and parts out of the volume are much simpler [6].

There is a number of different techniques currently in use for volume rendering. For example ray casting [22], shear warp methods [21] and 3D texture mapping techniques. Texture mapping was originally developed to provide the appearance of high surface complexity when rendering geometric surfaces. As they found their way into standard graphics hardware, researchers began exploiting these new capabilities to perform volume rendering.

The method for texture mapping is to store the volume as a solid texture in the graphics hardware, then to sample the texture using planes parallel to the image plane and finally composite them into the frame buffer using the blending hardware [13]. These steps are made in the following order:

1. Load the volume data into 3D texture memory.
2. Generate a set of polygons parallel to the viewing plane.
3. Render the polygons with 3D texture coordinates in front-to-back order.

When using texture mapping for rendering volume data, no gradient estimation is supported in the hardware. To circumvent this limitation, one can store the pre-calculated gradient together with the volume data. The gradient is used to calculate shading.
3D texture mapping hardware has been recognized as a very efficient acceleration technique for volume rendering, right after the first SGI RealityEngine, which is a powerful computer, was introduced [23]. The shipment of the first SGI RealityEngine made 3D texture mapping hardware an available interactive feature. With respect to volume rendering, polygons parallel to the viewing plane are placed in the volume. When using perspective projection, 3D texture mapping becomes more complicated since one needs to account for the correct blending. Thus parallel projection is applied in most cases. In fig 2-5 the principles of 3D texture mapping are demonstrated. One of the problems involved with 3D texture mapping is its limited availability. It is currently supported in hardware on most mid- and high-end SGI platforms, on HP fx class machines and on the ATI Radeon graphics card. A texture can be stored as a pure density volume interpreting the interpolated density values as indices into a large lookup table with at least 256 entries. This OpenGL extension is only available on mid- and high-end SGI platforms [14].

![Figure 2-5. 3D texture mapping technique.](image)

### 2.6 Color as an Information Illustrator

The human eye is extremely sensible to color variations. A trained person can separate over one million different color shades in a test with comparisons of colors in pairs [8]. An untrained person can discern approximately 20,000 color shades. In management information systems research studies have been made regarding the usage of color in graphical presentations. Conclusions that have been made are:
- Color improves the performance in recall tasks.
- Color improves the performance in search-and-locate tasks.
- Color improves the performance in memory tasks.
- Color improves the ability to understand instructions.
- Color improves the performance in decision judgment tasks.
However it should always be kept in mind that color is a subtle variable that can give tremendous positive effects if it is used the right way. An uncritical use however does not automatically give a good result.

### 2.6.1 Color Mapping

Color mapping is a common scalar visualization technique that maps scalar data to colors, and displays the colors on the computer system. The scalar mapping is implemented by indexing into a color lookup table as shown in fig 2-6. Scalar values serve as indices into the lookup table.

![Color mapping using a lookup table.](image)

**Figure 2-6.** Color mapping using a lookup table.

In fig 2-7 a more general form of the lookup table is presented, transfer functions. A transfer function is any expression that maps scalar values into a color specification. The key to color mapping for scalar visualization is to choose the transfer function carefully [7].

![Scalar values are mapped into separate intensity values for the red, green and blue component.](image)

**Figure 2-7.** Scalar values are mapped into separate intensity values for the red, green and blue component.
3 Implementation

3.1 Implementation process

The volume renderer used in this thesis is created by Andreas Sigfridsson [12] and uses a 3D texture mapping technique. The software that has been used is Matlab 6.5. The volume renderer runs on a SGI Onyx machine. The process that has been implemented is shown in fig 3-1.

3.2 The Magnetic Resonance Angiography Data

The angiography data that are processed in this thesis were acquired in a 1.5 T scanner (GE Signa) using a 3D SPGR sequence after intravenous injection of gadolinium. The images were transferred in digital form.
from the Picture Archiving and Communication System (PACS) of Linköping University Hospital. The data were in DICOM-format [20]. DICOM is the standard for Digital Imaging and Communications in Medicine. The standard is general and comprises image formats not only for radiology but also for all of medicine, with additional specifications for messaging and communication between imaging machines, with PACS, and with hospital information systems.

The volume has dimensions in voxels for example 256x256x124 voxels. Each voxel has spatial dimensions for example 0.8mm x 0.9mm x 2mm. The data is normally stored as a list of raw data values with a separate header.

The data sets that have been used in this thesis contain header information. This information can be revealed in MatLab 6.5 using the `dicominfo` function. From this header information concerning for example slice spacing, slice location, pixel bandwidth etc. can be obtained.

```
FileSize: 534284
Format: 'DICOM'
FormatVersion: 3
Width: 512
Height: 512
PixelBandwidth: 122.0703
FlipAngle: 30
SliceLocation: -0.7000
```

Usually some sort of overlapping between the image slices in the MRI scanner takes place during the MRA acquisition. This is a problem when distance computation should be performed. It is necessary that the spatial dimensions are uniform in order to perform the distance calculations correctly.

### 3.3 Noise reduction

Noise is any undesirable signal. Noise gets introduced into the data via electrical systems used for storage, transmission and processing. A filter can suppress some of the noise. Filtering is a neighborhood operation, in which the value of any given pixel in the output image is determined by applying some algorithm to the values of the pixels in the neighborhood of the corresponding input pixel. A pixel's neighborhood is some set of pixels, defined by their locations relative to that pixel. Numerous filter techniques are available.

#### 3.3.1 Linear filtering

Linear filtering is filtering in which the value of an output pixel is a linear combination of the values of the pixels in the input pixel's
neighborhood. Linear filtering of an image is accomplished through an operation called convolution. In convolution, the value of an output pixel is computed as a weighted sum of neighboring pixels. The matrix of weights is called the convolution kernel, also known as the filter.

Averaging or Gaussian filters can be used in linear filtering. For example, an averaging filter is useful for removing grain noise from a photograph. Because each pixel is set to the average of the pixels in its neighborhood, local variations caused by grain are reduced. Linear filtering is a very general and powerful tool of processing and may be used for various purposes.

### 3.3.2 Median filtering

Median filtering is a simple and very effective noise removal filtering process. Median filtering is similar to using an averaging filter, when each output pixel is set to an "average" of the pixel values in the neighborhood of the corresponding input pixel. However, with median filtering, the value of an output pixel is determined by the median of the neighborhood pixels, rather than the mean. The median is much less sensitive than the mean to extreme values. Median filtering is therefore better for removing these extreme values without reducing the sharpness of the image [15]. The median filtering has been implemented in 3D, because of the fact that the data is a volume. A 3x3x3 neighborhood has been used. The result can be seen in the fig 3-2 below. Note that the large structures appear smoother while the smallest vessels become less visible.
3.3.3 Adaptive filtering

One so-called adaptive filtering method is to apply a Wiener filter (a type of linear filter) to an image adaptively, tailoring itself to the local image variance. Where the variance is large, the Wiener filter performs little smoothing. Where the variance is small, the Wiener filter performs more smoothing [15]. This approach often produces better results than non-adaptive filtering. The adaptive filter is more selective than a comparable linear filter, preserving edges and other high frequency parts of an image. This filtering technique requires more computation time.

3.4 Segmentation

Segmentation is a process that separates objects in an image or volume. Segmentation is a partition of the image/volume in a number of regions, each region related to a spatial pattern of the data.

A segmentation of the 3D MRA data is necessary but not trivial. It is a complicated process due to many reasons. One of them is that the segmentation method should be able to detect vessels across a range of scales since the width of vessels can vary significantly.
A variety of methods have been developed for detecting vessels within MRA. One type of methods segments or classifies voxels within the image into either vascular or nonvascular regions. The simplest segmentation method is intensity thresholding whereby points are classified as either greater or less than a given intensity. However, non-uniform distribution of the contrast agent (gadolinium) can lead to significant intensity variation along vessels defeating such methods [9].

Frangi et al. [16] have created an alternative method for enhancing vessel structures with the eventual goal of vessel segmentation. Their technique would probably have given a nice segmentation result but has not been implemented due to time limitations. Frangi et al. present a vessel enhancement filtering technique based on the eigenvalues of the Hessian matrix to determine the likelihood that a vessel is present locally. The Hessian matrix $H$ describes the second-order structure of local intensity variation around each point of a volume. The filter is applied with different kernel widths or scales $S$ to enhance both small and large vessels.

### 3.4.1 Thresholding

In general, thresholding techniques are difficult to apply to MRI data due to large-scale intensity inhomogeneities and are less suitable for MRAs and Computed Tomography Angographies, CTAs, due to flow artifacts (e.g., contrast dissipation and laminar flow effects that produce irregular cross-sectional intensities) [10]. Despite these problems, thresholding was used for segmentation in this thesis. Thus the focus was mainly on the color-coding. There is a large number of segmentation methods available that are much more accurate and sophisticated than intensity thresholding. Throughout this thesis constraints had to be made due to the time limitation. Segmentation methods such as area thresholding and labeling have been tested with mixed results. Labeling is when connected objects are labeled. The voxels labeled 0 are the background. The voxels labeled 1 make up one object, the voxels labeled 2 make up a second object, and so on. An intensity threshold is shown in fig 3-3.
3.5 Distance Calculation Algorithm

3.5.1 The Euclidean Distance
The distance between two points of a set is defined as the Euclidean distance. The Euclidean distance for two points, \( P \) and \( Q \) in a 3D space is:

\[
P = [p_x, p_y, p_z] \quad Q = [q_x, q_y, q_z]
\]

\[
ED = \sqrt{(p_x - q_x)^2 + (p_y - q_y)^2 + (p_z - q_z)^2}
\]

The advantage of the Euclidean distance is the fact that it is intuitively obvious. The disadvantages are costly calculations due to the square root, and its non-integer value. Because of the heavy calculations when using the exact Euclidean distance a distance transform can instead be used to approximate the Euclidean distance.

3.5.2 Distance Transform
Distance transforms, DT, can be used to make an approximation of the Euclidean distance. A distance transform converts a binary digital image, consisting of feature and non-feature pixels, into an image where all non-feature pixels have a value corresponding to the distance to the nearest feature pixel.

Computing the distance from a pixel to a set of feature pixels is essentially a global operation. Unless the digital image is very small, all global operations are costly to compute. Therefore algorithms that consider only a small neighborhood at a time but still give a reasonably good result are necessary. Propagating local distances, i.e. distances...
between neighboring pixels, approximate the global distances in the image.

A DT mask is introduced. The constants in the mask are the local distances that are propagated over the image. The size of the neighborhood can vary. The computation of the DT is either parallel or sequential. In the parallel case the center of the mask is placed over each pixel in the image. The local distance in each mask-pixel the constant is added to the value of the image pixel below it. The new value of the image pixel is the minimum value of all the sums. The process is repeated until no pixel value changes. In the sequential case the mask is split into two masks and the masks are passed over the image one time each [17].

![Parallel DT computation](image)

Figur 3-4. Parallel DT computation. The original image has one feature pixel in the center.

In this thesis DT are considered in 3D. This makes it more complicated. The conditions are a little bit different. In 3D cubic space each voxel has three types of neighbors: six area neighbors, 12 edge neighbors and eight point neighbors. A path between two voxels in the 3D volume can thus include steps in 26 directions. A decision can be made concerning the voxel connectivity. The voxels could be 6- 18- or 26-connected. Therefore there are three different path-generated distances \(D^6\), \(D^{18}\) and \(D^{26}\), where the distance is defined as the number of steps in the minimal path between voxels with 6- 18- or 26-connectedness. In \(D^6\) only steps to the six face neighbors are allowed. In the \(D^{18}\) steps can also be taken to edge neighbors and in \(D^{26}\) point neighbors are also considered. This is illustrated in fig 3-5. To compute the DT with 6-connectedness
propagation is made as in the 2D case. Each voxel is assigned the minimum value of the current voxel, added by one, and the values of the already visited area neighbors [11].

The distance label of any voxel in the DT can be interpreted as the radius of a digital ball centered at that voxel. The balls look different for $D^6$, $D^{18}$ and $D^{26}$. In $D^6$ the ball is an octahedron. In $D^{18}$ it is a dymaxion (or cuboctahedron) and in $D^{26}$ the ball is a cube [18]. The distance transform implies that each ball is as large as possible still included in the object, in this thesis the vessel.

The spatial resolution of the voxels is important here because of the fact that the distance is measured in the number of voxels across the vessel. The ideal voxel is cubic so that no interpolation is necessary.

![Figure 3-5. Connections used in the distance transform.](image)
3.6 Mappings

3.6.1 Opacity choice
The opacity varies from 0-255, 255 being entirely opaque and 0 being completely transparent. In the visualization of these color-coded blood vessels it is important with a relatively low opacity so that the colors inside the vessels become visible. The user is able to control the opacity when using the application.

3.6.2 Color choice
The volume renderer in this thesis can display volumes that only have an opacity value. However it can also handle colors. The color mode is based on four channels. One channel for R (red), one for G (green), one for B (blue) and one for the alpha value (opacity) i.e. the input is a four dimensional volume.

Data is mapped to a color by converting the raw data value into a color table index using a linear transfer function. Providing a minimum and maximum value specifies the transfer function. These values correspond to the beginning and the end of the color table. Data values which lie below the minimum transfer function value will be colored with the first value in the color table (index #0) while values that lie above the maximum transfer
function value will be colored with the last value in the color table. Data values which lie between the minimum and maximum transfer function values will be converted to an index based upon a linear interpolation, the resolution of which is the number of entries in the color table.

Several color maps were chosen in the final version. Various maps have been evaluated, both continuous and non-continuous. Fig 3-7 shows an example of color maps that can be used. The bottom right is an example of a non-continuous color map i.e. there are only a few colors represented and the transmission between them are abrupt. The upper right example map only one color, red, the rest are gray values. When using this the decision of the distance max can be made easier. Read more about distance max in chapter 3.8. The top left color map is continuous and goes from black to red, red being the greatest distance.

![Color maps](image)

**Figure 3-7. Different color maps (MIPs).**

### 3.7 Visualization

#### 3.7.1 Visualization pipeline

The visualization pipeline is shown in fig 3-8. The input is a MRI volume. The volume is filtered and segmented using thresholding. DT is then performed on the binary volume and the result is color-coded. The color channels and an opacity channel are sent to the volume renderer.
The opacity is based on the original gray value \( b \) or a constant value between 0-255 \( a \).

\[ \text{MRI volume} \]
\[ \text{Median/Averaging/Wiener filtering} \]
\[ \text{Intensity thresholding} \]
\[ \text{Distance Transform} \]
\[ \text{Color-coding in R,G,B-channels} \]
\[ \text{Opacity} \]
\[ \text{MIP} \]
\[ \text{3D texture mapping volume rendering} \]

**Figure 3-8.** The visualization pipeline.

### 3.7.2 Ray Casting

In ray casting, a ray is followed from the eye position through the point on the view plane corresponding to the pixel. For each surface in the scene, the distance the ray must travel before it intersects with that surface is calculated. The surface with the shortest distance is the closest; therefore it blocks the others and is the visible surface. Ray casting is therefore a hidden surface removal method.

Ray casting has other applications as well. Volumes of nonstandard objects may be estimated with ray casting methods. Ray casting may also be applied in rendering volume data.

Ray casting is used to make a MIP of a volume. In fig 3-9 MIP-images of a 3D synthetic data set with byte information, representing the DT, in every voxel is demonstrated. It illustrates how colors can be used for detection of narrowing when the projection angle is not optimal. Looking from the front the narrowing at the top of the cylinder is visible as an actual narrowing while looking from the right the same narrowing (of vessels) is represented as a color code. The contrary holds for the bottom of the cylinder.
Ray casting was not used for the volume rendering in this thesis, although it was considered. The reasons for this were several, first of all it is a slow method, the interaction would have been very limited. The choice of volume renderer fell on a renderer based on 3D texture and therefore ray casting could not be easily implemented. A significant 3D “feeling” was also of high priority and because it is slow some of the feeling is lost.

3.8 Graphical User Interface

A graphical user interface, GUI, was implemented to make the manipulation of parameters more manageable. The GUI is demonstrated in fig 3-11. The GUI consists of a load button, to load a new data set, a section of parameter settings, a section of render settings, a render button and a MIP display. The rendered volume is displayed in a separate window, where it can be interactively rotated and scaled. The parameters that can be changed are the filter, the color map, the intensity threshold, the distance threshold and the opacity. The evaluations of the color-coding in the future evaluation project are to be made from this GUI.

- **Load volume:** A new volume is loaded. The maximum size is 256x256x256 voxels.
- **Select noise removal filter:** A suitable filter is chosen. There are four filters to choose from. An averaging filter in 2D, a median filter in 2D, a median filter in 3D and a Wiener filter in 2D. Generally the 3D median filter gives the best result but is also the most time consuming.
- **Select uniform opacity max value/Keep original opacity:** It is possible to keep the original opacity in the dataset i.e. the opacity is set to correspond to the gray value in the voxel. Otherwise the opacity is set to a value from 0-255.
- **Select intensity threshold:** The intensity threshold should be very carefully chosen so that no important information is lost.

- **Go:** When all the parameters are set the “Go” button can be pressed and the computer calculates the distance transform of the filtered, thresholded volume.

- **Select color map:** The color maps to choose from vary in both color and continuity.

- **Select distance max:** The distance max is chosen so that the color map can be controlled depending on how deep the vessels of interest are.

- **Render:** When the render settings are chosen the information is sent to the volume renderer. The volume render result is shown in fig 3-12.

- **Show MIP, 0°/45°/90°/135°:** A 2D MIP can still be helpful for the radiologist. The MIP of the volume is performed from four different angles 0°, 45°, 90° and 135°, shown in fig 3-10.

![Figure 3-10. MIPs of the volume are made from four different angles: 0°, 45°, 90°, and 135°.](image)

- **Notes:** Notes can be taken during the observation and they are stored in a log file.

- **Save and load settings:** When saving, both the volume and the current parameters are saved. The parameters are saved in the log file together with the notes. The volume is saved in a separate file with the same name. When loading old calculations the volume is sent to the volume renderer and an image of the GUI as it appeared during saving is shown.
Figure 3-11. The Graphical User Interface.

Figure 3-12. The color-coded volume in the volume renderer. The volumes are scaleable and rotatable.
A color-coding algorithm has been implemented that represents the distances in the volume. It has been presented both as a volume rendering and as MIPs. It is possible to vary the color-coding, which makes it possible to apply it on different vessel sizes. A GUI has been implemented to simplify the setting of the parameters. The method is ready for a clinical evaluation.

Examples of color-coded volume renderings and MIPs have already been given in fig 3-6, 3-7 and 3-9 (synthetic images) and in fig 3-12 (clinical dataset).

In fig 4-1 the distance max parameter is illustrated. All values over a certain threshold are mapped to the red color. Fig 4-2 shows a rendering and MIPs from two different angles. The color map is continuous and goes from black to red.
Figur 4-1. Selection of different distance max values represented by the red color. The top images are MIPs. The bottom images are volume renderings.
**Figure 4-2.** Top: MIPs from two different angles. Bottom: Volume rendering with a continuous color map that goes from black to red.
The method of presenting color-coded distances that has been presented in this thesis has not yet been evaluated clinically. This is to be arranged in the near future. The work has resulted in a small, easily controlled GUI with attached log-file. The aim was not to develop an application for a radiologist to use in a clinical context, but to explore the method itself.

An important quality that the volume must possess is cubical voxels. This is absolutely necessary to give a correct result. So far only volumes taken from an MRI scanner have been experienced. It is possible that volumes generated from a CT scanner would cause obstacles not yet discovered.

There are some shortcomings in the application. Defects in the segmentation occur to some extent depending on the quality of the input volumes. There is also a size limitation that occasionally can cause problems. The size of the volume cannot exceed 256x256x256 voxels due to restrictions in MatLab. The calculations are also fairly slow depending on the size of the volume.

If time was not an issue there are of course a number of things that could have required more work. The major thing is probably the segmentation method. It would also have been interesting to make an improved filtering and to analyze volumes from a CT scanner as well.

A clinical evaluation will be realized and will hopefully show if the application is practical. Since the volume renderer is based on 3D texture, some artifacts occur when very small vessels are rendered. The artifacts are that the parallel planes that the renderer contains become visible.
Future work may include a more developed MIP that obtains the angle from where the viewer looks at the volume, from the volume renderer and makes a MIP from that particular angle. It would also be fascinating to explore the possibilities of a tool to measure the distance in millimeter in the volume renderer, for example an interactive, scalable ellipse or sphere.

As the work proceeds with its ups and downs you realize, independent of success, that everything takes longer than the original plan. Much work is done in vain and has to be thrown away, but this is a part of the process and a very important experience.

I have throughout this master thesis learned a lot about difficulties in medical visualization. An image is not just an image, it is a piece of information, where some of it is vitally important and some is insignificant. It is significant to know the difference between those two and therefore extensive collaboration between physicians and technicians is necessary.

I have carried out this work on my own and there is a lot to learn from the experience to have full responsibility and control over a comprehensive project. There are both pros and cons connected with such work. Sometimes it has been frustrating to not be able to discuss the everyday work with someone with similar experiences. I have evidently discussed the major issues in the project with my supervisors but the decisions have been up to me. It feels good to have had the full responsibility of an extensive project. It the future though, I hope to be involved in more cooperation within projects.
References


