Biomedical Engineering Master

Analysis of structures in Medical Images – Application in Resonance Magnetic Images of Female Pelvic Cavity

Practical Work

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Analysis of structures in Medical Images - Application in Resonance Magnetic Images of Female Pelvic Cavity

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Abstract

The female pelvic cavity is one of the structures of the human body that has more features. The act of procreation is only possible in this cavity.

The use of diagnostic imaging, specifically magnetic resonance is the most effective and least harmful to the diagnosis of the pelvic cavity.

The presence of noise in such image is very common, so one the first steps to be performed in image analysis is the removal of noise and then the enhancement of edges.

The methods discussed here ranging from spatial filters to the basic operations.

This work is the study and implementation of enhancement methods to know which method is most effective.
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I – Introduction
1.1 – Motivation

The pelvic floor is one of the body’s most interesting organs, but unfortunately, also one of the least understood and most neglected. Outside of health-care circles, the pelvic floor is virtually unknown. In contrast, the pelvic floor is vital for the reproductive functions. In women, it becomes even more crucial to the health and happiness of the individual, and is intimately involved in labor and delivery. With its central role in sexuality and childbirth, the female pelvic floor is most intimately concerned with femininity.

Diagnostic imaging technology has transformed the practice of medicine. As a result of these technological changes, medical experts are now able to provide more precise and less invasive care, leading to better prevention of illness and better clinical outcomes. In fact, diagnostic imaging has become the standard of medical care for virtually all major medical conditions and diseases, including cancer, stroke, heart disease, trauma, and abdominal and neurological conditions. Pursuant to this higher standard of care, diagnostic imaging has grown across the world.

A wide variety of imaging methods are nowadays used to visualize the pelvic floor. However, the most widely used method for observation of this area of the body is magnetic resonance.

The medical images contain anatomic and functional important information, which reflects the organ’s shape and the patient condition. For the proper analysis and validation by a technical and also can obtain the potential maximum of the recorded data, the images have to be processed.

The observations of medical images for medical are, in most cases, qualitative and are only based on the experts’ visual sense. To improve the observation and make it more quantitative can be accomplished with an automated processing and a computational vision processing.

Computational vision is the capacity to describe or analyze images by using computational resources. For human, this process is relatively simple, due to power of their vision system. The importance of this system has led a large number of investigators trying to play it computationally, through the development of automatic systems capable of performing some of their functions.

The enhancement of the structures, which are present in medical images, is an important step that can make possible a good segmentation and analyzing.
The segmentation and analyzing of the structures, which are represented in medical images, are one of the areas of Computational Vision that demonstrated a greater development in the last years.

This chapter aims to expose the main objectives of this work, and describe the organization adopted for this report.
1.2 – Objectives

The principal aims of this work are the study about the anatomy of the female pelvic cavity and, based on the analyze of magnetic resonance images, revise and discuss some image enhancement possibilities.

Thus, in this study was realized a comparison between various image enhancement methods and was analyzed the results in order to elect the process that obtains the best enhancement results.

As, in this thesis project, there is the objective to make the segmentation of the female pelvic cavity organs, a fine image enhancement step can lead to suitable segmented images and then to collected robust image features. These characteristics may then give good indications of the organs’ shape and also indicate the presence of pathologies automatically.

Currently in imaging services, these tasks are carried out manually by a medical specialist, which has disadvantages as the long time spent and the fact that the reproducibility is low.

The work presented in this report, had as main objectives the creation of a survey, the study, description and analysis of existing techniques for the enhancement of medical images.
1.3 – Structure

The report is divided in the remaining chapters:

- **Chapter II – Anatomy of the female pelvic cavity**
  In this chapter is made a briefly description on the Anatomy of the female pelvic floor.

- **Chapter III – Medical Imaging**
  In this chapter is described the historical evolution of the medical imaging, and presented the features of a large variety of imaging systems, as well as the detail description of the Magnetic Resonance Imaging technique.

- **Chapter IV – Image Analysis**
  Afterwards, in chapter IV is considered the principal theme of this work. Hence, the chapter contains the description of digital image and refers the processes that are involved in creating a digital image, and enumerates its characteristics. Then is presented the fundamentals of the computational vision, and is study further the image enhancement and some of the existing methods. Furthermore, in this chapter is analyzed some algorithms of image enhancement applied in magnetic resonance images.

- **Chapter V – Final Considerations**
  Finally, this chapter presents the final conclusions and perspectives of future work.
II – Anatomy of Pelvic Cavity
2.1 – Introduction

In this Chapter it will be presented the Anatomy of pelvic cavity. The study has a particular interest to be analyzed because pelvic cavity represents a group of organs or elements that will be digitally analyzed from medical images.

Furthermore, the study of anatomy should allow describes the boundaries of the greater and lesser parts of this region.

Therefore, it is important to know their macroscopic form in order to facilitate its recognition and know which characteristics that can be used to the digital processing later.
2.2 – Anatomy

The pelvic floor is characterized by a complex morphology because different functional systems join here. These regions are subdivided according to functional and clinical requirements. The actual clinical subdivisions discern an anterior, middle and a posterior compartment. [Crocco, 1988; Hamm, 2007]

A clear understanding of the pelvic anatomy, Figure 2.1.a, is crucial for the diagnosis of female pelvic diseases.

The anterior and posterior compartment may also be found in male.

2.2.1 - Posterior Compartment

The skeletal elements of the sacrum and the coccyx dorsally are the borders of the posterior compartment. These are completed by the anococcygeal body, a stratified non-ligamentous structure in which fleshy muscle attachments underlie a tendon, Figure 2.a, dorsocaudally and by the components of the levator ani muscle, Figure 2.b, laterally and caudally. All components of the levator ani muscle are found in the posterior pelvic compartment, the pubococcygeus muscles and the
iliococcygeus muscles constitute an irregular plate and insert into the coccyx, Figure 2.1.b. The inferior compartment, the puborectalis muscles, does not insert into any skeletal structure, but is continuous with the external anal sphincter caudally, although it is not a totally circular muscle.

The rectovaginal fascia, plate of dense connective tissue, smooth muscle, cells and nerves, locally arranged between rectum and vagina, constitutes an incomplete border ventrocranially, Figure 2.2.c. Ventrocaudal border is composed of the serine al body, fibromuscular rather than tendinous and quite unlike the centrum tendineum of the diaphragma.

The only organ is the anorectum, which is constituted by the rectum and the anal canal, Figure 2.2.d. The major part of the posterior pelvic compartment is filled by the anorectum and the perirectal subcompartment, an identical situation happen with the rectal adventitial tissue, which is mainly consisted by adipose tissue subdivided by several connective tissue septa. The nerves of the inferior hypogastric plexus, Figure 2.2.e, are attached to the uterosacral ligament, Figure 2.2.f. The uterosacral ligament is a dense connective tissue running from the edges of the cervix uteri to the region of the sacrospinous ligament., that is directly situated between the rectal fascia and the inferior hypogastric plexus

The ventral border of the perirectal compartment, which filled by the rectal adventitia including nerves, vessels, and lymph nodes, represents the border between the posterior and middle compartments.

The macroscopic distinction between both muscles is provided by the anococcygeal body. [Crocco, 1988; Hamm, 2007]
The important vessels, nerves and lymphatics of the posterior compartment are: superior rectal artery, rectal nerves, rectal lymph nodes, inferior hypogastric plexus, superior hypogastric plexus, common iliac artery and internal iliac artery. [Crocco, 1988; Hamm, 2007]

Figure 2.2: In this figure is possible see the anococcygeal body (a), the levator ani muscle (b), the rectovaginal fascia (c), the anorectum (d), the hypogastric plexus (e), the uterosacral ligament (f), the tendinous arch (h), and the perineal membrane (g) (images from Hamm, 2007).

Figure 2.3: Mid-Sagittal MR image of an adult female: The perasacral space is indicated by the arrows and the rectum region by r (image from Hamm, 2007).
2.2.2 - Anterior Compartment

The anterior compartment is constituted by the bladder, Figure 2.1.a, urethra, Figure 2.1.a, pubovesical ligament, Figure 2.4, or cord, which is an ascending course from the pubic bone, Figure 2.1.a, to the neck of the bladder, and the tendinous arch of the pelvic fascia. The bladder is covered by adipose tissue. The fat pad sometimes may be crossed by variable branches from the obturator vessels.

Within the anterior compartment two structures are found that are composed of dense connective tissue: the tendinous arch, Figure 2.2.h, of the pelvic fascia that originates from the pubic bone, Figure 2.4, and that is connected to the pelvic parietal fascia covering the elevator ani muscle on its visceral side and the semicircular fibrous sheath that covers the ventral and lateral wall of the bladder and the urethra.

The fibrous structures of the anterior compartment build up a hammock like construction for bladder and urethra.

An additional fibrous structure can be found to close the hiatus ventrally: a plate of dense connective tissue fills the space between pubic bone and urethral sphincter, thus constituting the perineal membrane, Figure 2.2.g. [Crocco, 1988; Hamm, 2007]

![Figure 2.4: Anterior compartment. Axial section (400µm) of a 24-week-old female fetus with the semicircular urethral sheath (indicated by arrows) (x12) (a); Sagittal section (500µm) of a 13 to 14-week-old female fetus with pubovesical ligament (indicated by white spots); Pubic Bone, pbo; Urethra, u; Levator ani muscle, lam (images from Hamm, 2007).](image)
The striated muscles of the anterior compartment are the ventral parts of the levator ani muscle, the pubococcygeus and the puborectalis muscle, Figure 2.5. As they are covered by the superior fascia of the pelvic diaphragm, they are clearly separated by the adjacent organs and the external urethral sphincter.

Important vessels, nerves and lymphatics of the anterior compartment are: inferior vesical artery, branches to the ureter, superior vesical artery, vesical lymph nodes, internal iliac lymph nodes, internal iliac artery, inferior hypogastric plexus and paravesical fat pad. [Crocco, 1988; Hamm, 2007]

![Figure 2.5: Axial section (400µm) in a female newborn specimen (x4); Urethra - u; Vagina - v; Rectum – r; Puborectalis muscle (arrows) (image from Hamm, 2007).]

### 2.2.3 - Middle Compartment

Within the middle compartment the adipose tissue surrounding uterus and vagina. This adipose tissue has regular connective septa tissue and it is continuous with the broad ligaments laterally. The broad ligaments themselves are part of the recto-uterine pouch and consist of colagenous fibers. This tissue has developed to the uterosacral ligaments. Subperitoneally, the middle compartment and its organs abut the anterior compartment ventrally. This area is predominated by the dense connective tissue bridge intimately connecting the ventral vaginal wall with the dorsal urethral wall. The border between the middle compartment and the posterior compartment (dorsomedially) is demarcated by the rectovaginal...
fascia that is composed of dense connective tissue, elastic fibers and smooth muscle cells. The uterine tubes lie is attached on mesosalpinx, Figure 2.6.a. [Crocco, 1988; Hamm, 2007]

The ovaries lie in the ovarian fossa, which are extra peritoneal structures, specially the ureter and the internal iliac vessels as well as the origin of the uterine artery.

The mesometrium, Figure 2.6.b, may be considered the largest part of the broad ligament extending from the pelvic floor to the uterine body enclosing the uterine artery or the connective tissue lying directly beneath the peritoneal covering of the uterus.

The important vessels, nerves and lymphatic of middle compartment are uterine artery and inferior hypogastric plexus. [Crocco, 1988; Hamm, 2007]

Figure 2.6: In this figure is represented the mesosalpinx (a) and the mesometrium (b) (images from Hamm, 2007).

Normal Anatomy of the Uterus

The uterus is composed of three distinct anatomic regions, namely the corpus, the isthmus (lower uterine segment) and the cervix, Figure 2.7.

The wall of the uterine corpus differs from that of the cervix in that it mostly consists of myometrium, the strong muscle coat forming the mass of the organ. The uterine cavity is only a thin cleft and is lined by endometrium. Functionally, the endometrium consists of basal and functional layers. In women of reproductive age, the uterus usually is 6-9cm long and weighs 40-60 g.

The isthmus of uterus forms the junction between the corpus and cervix, is only about 5mm high and is less muscular than the corpus. [Crocco, 1988; Hamm, 2007]
The uterine cervix consists of the supravaginal cervical canal and the vaginal portion that project into the vagina.

The uterus is supplied with blood through the uterine and ovarian arteries. The uterine arteries course to the organ through the cardinal ligaments and are divided into an ascending and a descending branch. Lymphatic drainage from the corpus is through the broad ligament into parametria, Figure 2.8, and iliac lymph nodes.

The normal topography of the uterus is primarily ensured by the parametria, which is a kind of suspension system mainly consisting of connective tissue. The parametria contain large amounts of fatty tissue.

Most of the uterus is covered by peritoneum, that contributes only little support but ensures adequate mobility of the uterus relative to the urinary bladder and rectum, which is necessary to adjust to the variation in bladder filling and especially during pregnancy. [Crocco, 1988; Hamm, 2007]

Figure 2.7: Anatomic draft of the uterine cervix (coronal view, image from Hamm, 2007).

Figure 2.8: Anatomic draft, which presents the strong relation between uterine arteries and ureter (indicated by an arrow, image from Hamm, 2007).
Ovaries and Fallopian Tubes

The female adnexal structures are located in the lesser pelvis and include the fallopian tubes, the ovaries and ligamentous attachments, Figure 2.10.

The Fallopian tubes are 8 to 15 cm long paired tubular structures at the superior aspect of the broad ligament. They extend from the uterus to the ovaries and are composed of the intramural portion, the isthmus.

The ovaries are typically located in the ovarian fossa close to the lateral pelvic side walls, Figure 2.11. Adult ovaries measure approximately 3-5 cm in length, 1.5-3 cm in width, and 0.5-1.5 cm in thickness. The ovary is encapsulated by thin fibrous layer, the tunica albuginea. [Crocco, 1988; Hamm, 2007]
composed of follicles, corpora lutea, fibroblasts and smooth muscle cells, Figure 2.11. [Crocco, 1988; Hamm, 2007]

The normal fallopian tube contains a small amount of intraluminal fluid that is dispersed within multiple unfolding of the fallopian tube mucosa. [Crocco, 1988; Hamm, 2007]

**Vagina**

The vagina, Figure 2.12, is a fibromuscular sheath like structure connecting the external genitals with the uterus. It is lined with nonkeratinizing squamous epithelium and is 8-12cm long. The epithelium consists of up to 30 cell layers in women of reproductive age but of only a few layers in childhood and after menopause.

The vagina protects the internal genital organs against ascending infections and receives the penis in copulation. The vagina is supplied with blood through the descending branch of the uterine artery. [Crocco, 1988; Hamm, 2007]
Figure 2.12: Vagina - Images in sagittal (a) and axial (b) orientation (images from Hamm, 2007).

### 2.2.4 - Pelvic Girdle

Pelvic Girdle is a basin-shaped ring of bones connecting the vertebral column to the femurs and is composed of two coxae (hip) bones. The coxae bone consists of three separate parts: the ilium, the ischium, and the pubis, Figure 2.13.

The adult human bony pelvis is composed of four main elements: the right and left os coxae the sacrum and coccyx. The sacrum and coccyx are parts of axial skeleton and are actually variably fused vertebrae. [Crocco, 1988; Hamm, 2007; White, 2005]

Figure 2.13: Pelvic Girdle (image from White, 2005).
2.3 – Summary

The pelvic floor is subdivided according to a clinical discern: an anterior, middle and a posterior compartment.

The anterior compartment is constituted by the bladder, urethra, puboviscal ligament, pubic bone, urethral sphincter, and the tendinous arch of the pelvic fascia. Other structures that can happen are striated muscle like levator ani muscle, the pubococcygens and the puborectalis muscle (ventral parts). Important vessels, nerves and lymphatics of the anterior compartment are: inferior vesical artery, branches to the ureter, superior vesical artery, vesical lymph nodes, internal iliac lymph nodes, internal iliac artery, inferior hypogastric plexus and paravesical fat pad.

The middle compartment presents the uterus, vagina, ovaries, ovarian fossa, uterine artery, and inferior hypogastric plexus.

In the posterior compartment can be found skeletal elements (sacrum and the coccyx), muscular elements (levator ani muscle, pubococcygeus muscles, anal sphincter), digestive system (the only organ is the anorectum), and ligaments (uterosacral ligament, sacrospinous ligament). The important vessels, nerves and lymphatics are superior rectal artery, rectal nerves, rectal lymph nodes, inferior hypogastric plexus, superior hypogastric plexus, common iliac artery and internal iliaccutery.

All this compartments are protected and supported by the pelvic girdle.
III – Medical Imaging
3.1 – Introduction

The biological signals are transformed into images via medical imaging modalities.

In this chapter, will be realized a presentation of medical imaging story. This fact will allow knowing the origin and the development performance of medical imaging over the years. This information is very important to understand what were the main facts and aspects that happened to make the evolution of medical imaging possible.

However, it will be studied later magnetic resonance imaging, since the analysis and processing will be performed on this type of medical images.
3.2 – History of Medical Imaging

Radiograph was the first published medical image of the hand of Wilhelm Conrad Roentgen’s wife in December of 1895. This discovery could have profound impact in medicine. The first clinical use of x-rays occurred only two months later, in February of 1896. The static and dynamic techniques were developed. A static technique refers to an image taken at a single point on time. A dynamic technique refers to a series of images acquired over time. Planar radiographs, for many decades, were the only medical images being produced. Radiography was extended into transmission computed tomography or cross-sectional imaging. The first true CT scanner was produced by Godfrey Hounsfield in 1972. Godfrey used mathematical method for image reconstruction developed a decade earlier by Allan Cormack of the United States. Many radiologists consider CT scanning to be the most important development in medical since Roentgen’s original discovery.

Nuclear medicine arose from the discovery of radioactivity by Antoine Henri Becquerel in 1896. Radionuclides were used in cancer therapy rather than in medical imaging.

George de Hevesy, in 1923, introduced a concept of using radioactive tracers to study physiology.

Early studies with radiotracer used conventional non imaging radiation detectors to roughly determine amounts of radioactivity in various body regions. The first imaging system in nuclear medicine, the rectilinear scanner was started developed by Benedict Cassen at UCLA, in 1949. Technetium – 99m, which is the element of the most commonly used radionuclide in nuclear medicine, was discovered in 1937 by Perrier and Emilio Segre.

Ultrassound technology progressed through the 1960s from simple A-mode and B-mode scans to today’s M-mode and Doppler two dimensional (2-D) and even three-dimensional (3-D) systems.

Nuclear Magnetic Resonance was first described by Felix Bloch and Edward Purcell, they shared the 1952 Nobel Prize in Physics. In 1971, Raymond Damadian published a paper suggesting the use of magnetic resonance in medical imaging [Prince, 2005].
3.3 – Imaging Systems

The detection of different physical signals arises from four processes: transmission of x-rays through the body (in projection radiography and CT), emission of gamma rays from radiotracers in the body (in nuclear medicine), reflection of ultrasonic waves within the body (in ultrasound imaging), and precession of spin systems in a large magnetic field (in magnetic resonance imaging).

The imaging systems have a large variety of modalities. Projection radiography includes the following modalities:

- Routine diagnostic radiography: chest x-rays, fluoroscopy, mammography, and motion tomography.
- Digital radiography: includes all the scans in routine radiography, but with images that are recorded digitally.
- Angiography (the systems are specialized for imaging the body’s blood arteries and vessels): angiography and angiocardiography.
- Neuroradiology: specialized x-ray systems for precision studies.
- Mobile x-ray systems: designed for operating rooms or emergency vehicles.

Computed Tomography (CT) uses x-rays, which traveling in a 3-D cone beam, restricted in their geometric spread. They are collimated to travel within an approximate 2-D “fan beam”. Tissues create shadows in the x-ray beam that culminate in a 2-D cross-section of the body and a large number of small detectors are detected and collected for many angular orientations. These measurements are called projections and an image of the cross-section is computed from these projections.

Projection CT includes the following modalities:

- Single-slice CT (Standard).
- Helical CT.
- Multi-slice CT.

In helical CT, the x-ray tube and detectors continuously rotate around in a large circle, at the same time the patient is moved in a continuous motion through the circle’s center.
This technique is important due to ability to rapidly acquire 3-D data (whole body scan in less than a minute).

The multi-slice CT has several rows of detectors used to rapidly gather a cone of x-ray data, comprising a 2-D projection of the 3-D patient.

Nuclear medicine imaging can only be made with an appropriate introduction of radioactive substances. These drugs are labeled as radionuclides that emit gamma rays and radionuclides are called radiotracers.

Nuclear medicine imaging is an example of functional imaging method, whereas standard CT and MRI are anatomical or structural imaging methods.

Nuclear medicine includes:
- Conventional radionuclides imaging or scintigraphy - utilizes a special 2-D gamma ray scintillation detector.
- Single-photon emission computed tomography (SPECT) – utilizes a special 2-D gamma ray scintillation detector.
- Positron emission tomography (PET).

The special 2-D gamma ray scintillation detector is called Anger camera, which is designed to detect single x-rays or gamma rays.

Conventional radionuclide imaging combines the effects of emission with the effects of attenuation of the rays by intervening body tissues. This modality produces images that are 2-D projections of 3-D distribution of radiotracers.

SPECT produce images of slices within the body by rotating the Anger camera around the body and using CT methods to reconstruct images.

In PET, radionuclide decay produces a positron, which immediately annihilates to produce two gamma rays flying off in opposite directions.

Ultrasound imaging generates repetitive bursts of high-frequency sound using electrical-to-acoustical transducers.

Ultrasound imaging systems offer several imaging modalities:
- A-mode imaging: generates a one-dimensional waveform and can provide very detailed information about rapid or subtle motion.
- B-mode imaging: ordinary cross-sectional anatomical imaging and can give rise to images with different appearances.
- M-mode imaging: generates a succession of A-mode signals and the image is not anatomical but is important for measuring time varying.

- Doppler imaging: uses the property of frequency or phase shift caused by moving objects to generate images that are color coded by their motion.

Magnetic Resonance (MR) creates images using the property of nuclear magnetic resonance. The nucleus of the hydrogen atom in the presence of a strong magnetic field tends to align itself with the field. The human body includes a vast number of hydrogen atoms and alignment results in a net magnetization of the body.

The most general categories of operation are the following:

- Standard MR: time series of different excitation pulses.

- Echo-planar imaging (EPI): utilizes specialized apparati to generate images in real time.

- Magnetic resonance spectroscopic imaging: images other nuclei besides the hydrogen atom.

- Functional MRI (fMRI): uses oxygenation sensitive pulse sequences to image blood oxygenation in the brain [Prince, 2005].
3.4 – Magnetic Resonance Imaging

Magnetic resonance imaging is a noninvasive medical imaging technique. The tremendous clinical utility of Magnetic Resonance Imaging (MRI) is due to the great variety of mechanisms that can be exploited to create image contrast. If magnetic resonance images were restricted to water density, MRI would be considerably less useful, since most tissues would appear identical. Fortunately, many different MRI contrast mechanisms can be employed to distinguish different tissues and disease processes.

**Fundamentals of MRI**

Magnetic resonance imaging exploits the existence of induced nuclear magnetism in the patient. Most commonly protons ($^1$H) are imaged although carbon ($^{13}$C), phosphorous ($^{31}$P), sodium ($^{23}$Na), and Fluorine ($^{19}$F) are also of significant interest. Field strength is usually to be of 0.2 and 1.5T, but for spectroscopic and functional imaging is necessary higher field strength.

The ratio of the induced magnetization to the applied fields is only $4 \times 10^{-9}$.

The key idea is to measure the moment while it oscillates in a plane perpendicular to static field. [Bloch et al, 1946; F. Bloch, 1946] The torque always points perpendicular to the magnetization and causes the spins to oscillate in perpendicular plane to the static field. The frequency of rotation $\omega_0$ is proportional to the field:

$$\omega_0 = -\gamma B_0$$  \hspace{1cm} (3.1)

Where $\gamma$, the gyromagnetic ratio, is a constant specific to the nucleus, and $B_0$ is the magnetic field strength.

A complex molecule is placed in a highly uniform magnetic field, is produced microscopic a microscopic field variations within the molecule so that geometrically isolated nuclei rotate about distinct fields. Magnetic environment produces a peak in the spectra of the received signal.

Nuclear magnetic resonance signal from a human is due predominantly to water protons. Since these protons exist in identical magnetic environments, they all resonate at the same frequency.

The MR signal is simply proportional to the volume water. The most effective non-uniform field is a linear gradient where the field and the resulting frequencies vary linearly with distance along the object. Fourier analysis of the signal obtains a map of the spatial distribution of spins, k-space analysis of MRI [Twieg, 1983; Ljunggren, 1983].
**K-space Analysis of Data Acquisition**

In MRI, it is received a volume integral from an array of oscillators. During signal reception, the applied magnetic field points in the z direction. Hence a spin at position \( r=(x,y,z) \) has a unique phase \( \theta \) that describes its angle relative to the \( y \) axis in the \( xy \) plane.

\[
\theta(r, t) = -\gamma \int_0^t B_z(r, \tau) \delta \tau \quad (3.2)
\]

Where \( B_z(r,t) \) is the \( z \) component of the instantaneous, local magnetic flux density.

A coil large enough to receive a time-varying flux uniformly from the entire volume produces:

\[
s(t) \propto \frac{d}{dt} \int_V M(r)e^{-i\vartheta(r,t)} dr \quad (3.3)
\]

\( M(r) \) represents the equilibrium moment density at each point \( r \). In general, gradient is \( (xGx + yGy + zGz) \), these gradient field components can vary with time, so the total \( z \) field is

\[
B_z(r, t) = B_0 + G(t) r \quad (3.4)
\]

In the presence of this general time-varying gradient, the received signal is

\[
s(t) \propto \frac{d}{dt} \int_V e^{-iyB_0 \tau} M(r)e^{-iy \int_0^t G(t) \cdot r \cdot dt} dr \quad (3.5)
\]

The center frequency \( \gamma B_0 \) is always much larger than the bandwidth of the signal. Hence the derive operation is approximately equivalent to multiplication by \( -i\omega_0 \).

Then can now identify as the spatial Fourier transform of \( M(r) \) evaluated at \( k(t) \). That is, \( S(t) \) scans the spatial frequencies of the function \( M(r) \). This can be written explicitly as

\[
S(t) \propto m(k(t)) \quad (3.6)
\]

Where \( m(k) \) is the three-dimensional Fourier transform of the object distribution \( M(r) \). Thus we can view MRI with linear gradients as a “scan” of k-space or the spatial Fourier transform of the image. After the desired portion of k-space is scanned, the image \( M(r) \) is reconstructed using an inverse Fourier Transform [Bronzino, 2000].
2D Imaging

The two-dimensional Fourier transform imaging is to scan through k-space along several horizontal lines covering a rectilinear grid in 2D k-space. The horizontal grid lines are acquired using 128 to 256 excitations separated by time TR. The horizontal line scans through k-space are offset in ky, by a variable area y-gradient are called pulse, which happens before data acquisition starts. For each ky phase signal is acquired while scanning horizontally with a constant x gradient [Bronzino, 2000].

Resolution and Field of View

The fundamental image characteristics of resolution and field of view are completely determined by the characteristics of the k-space scan. The extent of the coverage of k-space determines the resolution of the reconstructed image.

Diffraction limits the resolution and the resolution is limited to the wavelength divided by the angle subtended by the receiver aperture. The ultimate resolution is approximately the wavelength itself, but this is true for imaging systems based on optics like ultrasound, and x-rays. MRI is the only imaging system where the resolution is independent of the wavelength, because no attempt is made to focus the radiation pattern to the individual pixel or voxel (volume element). In this system the wavelength is often many meters. The gradients create spatially varying magnetic fields and individual pixels emit unique waveform signatures. These signals are decoded and assigned to unique positions. The problem is isolating the signals from two transmitting antenna towers separated by much less than a wavelength so is possible to distinguish the two signals if the two antennas transmit different frequencies. Both signals are received with a wide-angle antenna and then distinguish the signals through frequency-selective filtering [Bronzino, 2000].

Noise sources

The dominant noise source is due to thermally generated currents within the conductive tissues of the body. These currents create a time-varying flux which induces noise voltages in the receiver coil. Other noise sources include the thermal noise from the antenna and from the first amplifier. These system are designed so that the noise is negligible compared with the noise from the patient.
The noise received is determined by the total volume seen by the antenna pattern $V_n$ (noise volume based on the distribution of thermally generated currents) and the effective resistivity and temperature of the conductive tissue. The standard deviation of the noise from conductive tissue varies linearly with $B_0$ [Hoult, 1979].

The noise is filtered by integration over the total acquisition time $T_{\text{acq}}$, which effectively attenuates the noise standard deviation [Bronzino, 2000].

**Contrast Mechanisms**

The most important contrast mechanisms exploit relaxation of the magnetization with the two types of relaxations which are termed spin-lattice relaxation. The first type is characterized by a relaxation time $T_1$, and the other spin-spin relaxation, characterized by a relaxation time $T_2$.

Spin-lattice relaxation describes the rate of recovery of the z component of magnetization toward equilibrium after it has been disturbed by pulses. The recovery is given by

$$M_z(t) = M_0(1 - e^{-t/T_1}) + M_z(0)e^{-t/T_1} \quad (3.7)$$

Where $M_0$ is the equilibrium magnetization. Differences in the $T_1$ time constant can be used to produce image contrast by exciting all magnetization and then imaging before full recovery has been achieved. This is illustrated on the left in Figure 3.1. The plots show the recovery of two different $T_1$ components. The short $T_1$ component recovers faster and produces more signals.

![Figure 3.1: The two primary MRI contrast mechanisms, $T_1$ and $T_2$. $T_1$, illustrated on the left, describes the rate at which the equilibrium magnetization is restored after it has been disturbed. $T_1$ contrast is produced by imaging before full recovery has](image-url)
been obtained. $T_2$, illustrated on the right, describes the rate at which the MRI signal decays after it has been created. $T_2$ contrast is produced by delaying data acquisition, so shorter $T_2$ components produce less signal (Image from [Bronzino, 2000]).

Spin-spin relaxation describes the rate at which the NMR signal decays after it has been created. The signal is proportional to the transverse magnetization and is given by

$$M_{xy}(t) = M_{xy}(0)e^{-t/T_2} \quad (3.8)$$

In addition to the intrinsic tissue contrast, artificial MRI contrast agents also can be introduced and are usually administered intravenously or orally. The most popular agents decrease both $T_1$ e $T_2$ [Bronzino, 2000].

**Hardware/ Instrumentation**

In this subchapter will be described the basic components and the operating principles of MRI scanners.

The main components of the MRI scanner are its magnet and radiofrequency system [Novelline, 2004].

**Fundamentals of MRI Instrumentation**

There are three types of magnetic fields (static fields ($B_0$), gradient fields and a radiofrequency (RF) fields ($B_1$)) that are required in MRI scanners. It also usually necessary to use coils or magnets that produce shimming fields to enhancement the spatial uniformity of the static field $B_0$. Most MRI hardware engineering is concerned with producing and controlling these various forms of magnetic fields [Bronzino, 2000].

**Static Field Magnets**

The main field magnet [Thomas, 1993] is required to produce an intense and highly uniform, static magnetic field over the entire region to be imaged. This field must be extremely uniform in space and constant in time. The spatial variation of it main field of a whole-body scanner must be less than about 1 to 10 parts per million (ppm) over a region approximately 40cm in diameter. The temporal drift of the field strength is normally required to less than 0.1 ppm/h.
The units of the magnetic field strength are the gauss (G) and the tesla (T), which is more recently adopted unit, but is a part of the SI system of units and, for this reason, is generally preferred. The tesla is a much larger unit than the gauss – 1T corresponds to 10000G. The range of the static magnetic fields of modern MRI scanners is 0.5 to 1.5 T. However, useful scanners have been built using the entire range from 0.02 to 8T. The signal to noise ratio (SNR) is one of the key parameters that determine the performance capabilities of a scanner, because it is the ratio of the NMR signal voltage to the ever-present noise voltages that arise within the patient and within the electronic components of the receiving system. Magnetic fields are produced by using either electric currents or permanently magnetized materials as sources [Bronzino, 2000].

![Data flows](image)

Figure 3.2: Digital and analog domains for MRI imaging. MRI involves the flow of data and system commands between these two domains (image from [Schenck and Leue, 1991]).

To produce the highly uniform magnetic field required for MRI, it is necessary to more or less surround the patient with a magnet. The main field magnet is the most important determinant of the cost, performance, and appearance of an MRI scanner. The main magnets are subdivided in four classes: the permanent magnets, electromagnets, resistive magnets, and superconducting magnets. All this magnets have been used in MRI scanners [Schenck and Leue, 1991].
Permanent Magnets and Electromagnets

The permanent magnets and electromagnets use magnetized materials to produce the field. In a permanent magnet, the patient is placed in the gap between a pair of permanently magnetized pole faces. With electromagnets the pole faces are made of soft magnetic materials, which become magnetized only when subjected to the influence of electric current coils that are wound around them. Electromagnets require the use of an external power supply, but permanent magnets don’t need [Bronzino, 2000].

For permanent magnets and electromagnets the magnetic circuit is completed by use of a soft iron yoke connecting the pole faces to one another. The materials of the permanent magnets for use in MRI scanners include high-carbon iron, alloys such as Alnico, ceramics such as barium ferrite, and rare earth alloys such as samarium cobalt [Bronzino, 2000].

The advantages of permanent magnets are:

- Production of a relatively small fringing field;
- Don’t require power supplies.
They tend to be very heavy (up 100 ton) and can produce relatively low fields, on the order of 0.3 T or less.

![Permanent Magnet Diagram](image)

Figure 3.4: Permanent magnet. The figure shows a schematic cross-section of a typical permanent magnet configuration. Electromagnets have a similar construction but are energized by current-carrying coils wound around the iron yoke. Soft magnetic shims are used to enhance the homogeneity of the field. (Image from [Schenck and Leue, 1991])

They are also subject to temporal field drift caused by temperature changes. If the pole faces are made from an electrically conducting material, eddy currents induced in the pole faces by the pulsed gradient fields can limit performance as well. For to make lighter-weight permanent magnet scanners was used recently alloy of neodymium, boron, and iron [Schenck and Leue, 1991].

**Resistive Magnets**

In the late 1970s and early 1980s was manufactured the first whole-body. This used four to six large coils of cooper or aluminum wire surrounding the patient. These coils are energized by powerful (40 to 100kW) direct-current (dc) power supplies. For prevent overheating is necessary use of cooling water flowing through the coils. The heat dissipation increases rapidly with field strength, and it is not feasible to build resistive magnets operating at fields much higher than 0.15 to 0.3T. The resistive magnets are seldom used except for very low field strength (0.02 to 0.06 T) applications [Bronzino, 2000].
Superconducting Magnets

The use of cryogenically superconducting magnets [Wilson, 1983] has been the most satisfactory solution to the problem of producing the static magnet field for MRI scanners. The property of exhibiting absolutely no electrical resistance near absolute zero has been known as an exotic property of same materials since 1911. These materials retain the ability to carry loss-free electric currents in very high fields [Schenck and Leue, 1991].

![Superconducting Magnet](image)

Figure 3.5: Schematic drawing of a superconducting magnet. The main magnet coils and the superconducting shim coils are maintained at a liquid helium temperature. (image from [Schenck and Leue, 1991]).

The figure 3.5 illustrates the construction of a typical superconducting whole-body magnet. Six coils of superconducting wire are connected in a series and carry an intense current. A current of 200A produces 1.5T magnetic field at the magnet’s center. The coils have a diameter of about 1.3m, and the total length of wire is about 65km. If the magnet wire has no such flaws, the magnet can be operated in the persistent mode. A constant persistent current flow indefinitely so long as the temperature of the coils is maintained below the superconducting transition temperature. This temperature is about 10K for niobium-titanium wire. Many magnets now make use of cryogenic refrigerators that reduce or eliminate the need for refilling the liquid helium reservoir.

Magnets have operated for years completely disconnected from power supplies and maintained their magnetic field constant to within a few parts per million. Superconducting magnets have become the most widely used source of the main magnetic fields for MRI scanners [Bronzino, 2000].
Magnetic Field Homogeneity

The manufacturing tolerances and field perturbations caused by extraneous magnetic field sources, such as steel girders in the building surrounding the magnet that produce additional inhomogeneity in the imaging region. These field imperfections are reduced by the use of shimming fields. One approach, uses additional coils which are designed to produce a magnetic field corresponding to a particular term in the spherical harmonic expansion. The magnetic field is carefully mapped, and the currents in the shim coils are adjusted to cancel out terms in the harmonic expansion to some prescribe high order – active shimming. The alternative approach utilizes small permanent magnets that are placed at the proper locations along the inner walls of the magnet bore to cancel out contaminating fields – passive shimming [Bronzino, 2000].

Fringing Fields

A strong magnetic field is produced by a powerful magnet in the region surrounding it as well as in its interior. The fringing field can produce undesirable effects such as erasing magnetic tapes. It is also a potential hazard to people with implanted medical devices.

Radiofrequency Coils

The radiofrequency coils are used for two essential purposes: transmitting and receiving signals at the resonant frequency of the protons within the patient [Schenck, 1993]. In the range of field strengths currently used in whole body scanners, 0.02 to 4T, the operating frequency ranges from 0.85 to 170.03 MHz. For the commonly used 1.5T scanners, the operating frequency is 63.86MHz.

Ideally, the radiofrequency field is perpendicular to the static field, which is in the z direction. The radiofrequency field can be linearly polarized in either the x or y directions.

The most efficient RF field results from quadrature excitation, which requires a coil that is capable of producing simultaneous x and y fields with a 90 degree phase shift between them.

There are three classes of radiofrequency coils: body coils, head coils, and surface coils. Head and body coils are large enough to surround the region being imaged. Body coils have a large enough diameter to entirely surround the patient's body (50 to 60cm). Head coils are designed only for head imaging (28cm). Surface coils are smaller coils designed to image a
restricted region of the patient’s anatomy. They come in a wide variety of shapes and sizes [Bronzino, 2000].
3.5 – Summary

Medical Imaging relies on noninvasive techniques to image body structures and function.

Each technique or method has a large variety of different imaging modalities.

The main imaging modalities are projection radiography, computed tomography, nuclear medicine, ultrasound imaging, and magnetic resonance imaging.

The signal of interest is defined by the modality and specific imaging parameters.

In general MRI has the following advantages:

- MRI does not use any other type of ionizing radiation. Alternatively, powerful magnetic fields and radio frequency are used to acquire images, so MRI provides a safer imaging.

- MRI has the ability to change the imaging plane by choosing radio frequency pulse, which also called ‘slice selection’. Therefore the patient's body does not need to be moved during the MRI scan.

- MRI has the ability to adjust the contrast of scans by changing the radio waves and magnetic fields, different structures and tissues can be highlighted.

MRI is not applicable for some patients who have received certain types of surgical clips, metallic fragments, cardiac monitors, or pacemakers.

Radiologists are trained to look for specific patterns, defined by the modality, specific imaging parameters and differences in the expected signal in health and disease.
IV – Image Analysis
4.1 – Introduction

The theme of the work is focused on analysis and image processing, for this the presented chapter is divided into 4 main sections. This starts with the description of the method that realizes the digitalization of the images, then will display the software used to make the practical work. Later can be seen a description of the algorithms of image enhancement and finally the practical implementation of algorithms.

The method that realizes the digitalization is an important part of this chapter because it becomes possible to know the characteristics of data that will be processed.

A brief overview of the software used must be made to demonstrate what the tools used during this work.

The practical work will be based on image enhancement and it is important to know which methods were used to perform this step.
4.2 – Digital Image

The main processes involved in creating a digital image from an optical image are sampling and quantization.

The sampling (4.1) is the process that defines time instants or locations (4.4) where the image values are recorded (4.2). This process also can be denominated by discretization. [Gonzalez, 2008]

The sampling operation is represented by:

\[
(4.1) \quad f(t,z) = f(t,z) \ast s_{2\Delta t}(t,z)
\]

\[
(4.2) \quad f(m,n) = f(t,z)
\]

\[
(4.3) \quad f(t,z)
\]

\[
(4.4) \quad s_{2\Delta t}(t,z) = \sum_{m=-\infty}^{\infty} \sum_{n=-\infty}^{\infty} \delta(t - m \Delta t, z - n \Delta z)
\]

The quantization is the discretization of the continuous image intensity in a finite number of levels. The number of picture elements (pixels) used to represent the image and the number of quantization levels used to represent pixel intensity is the main characteristics of an image. [Gonzalez, 2008; Gonzalez, 2004]

For defined an image is necessary two-dimensional function, f(x,y), where x and y are spatial coordinates. The amplitude of “f” is called intensity or gray level. A digital image is composed by a finite number of elements, each of which has a particular location and value. Pixels word is used to denote the elements of a digital image.

Image analysis is the area of image processing and computer vision.

In computer analysis has three types of computerized processes:

- Low-level processes – primitive operations such as image processing to reduce noise, contrast enhancement, and image sharpening.
- Mid-level processes – segmentation (portioning an image into objects), description of those objects, and classification (recognition) of individual objects. Outputs are attributes extracted from the input image.
- Higher-level – performing the cognitive functions normally associated with human vision “making sense” of an ensemble of recognized objects. [Gonzalez, 2004]
Color images are constituted by a combination of individual 2D images. The basic system is the RGB color system, which consists in three individual component images (red, green and blue). Many of the techniques developed for monochrome images can be extended to color images. [Gonzalez, 2008]

![Polychromatic Image](image)

In the MATLAB a digital image can be represented by a matrix:

$$f = \begin{bmatrix}
    f(1,1) & \cdots & f(1,N) \\
    \vdots & \ddots & \vdots \\
    f(M,1) & \cdots & f(M,N)
\end{bmatrix}$$

M – Represents the number of lines;
N – Represents the number of columns. [Gonzalez, 2004]
4.3 – MATLAB and the Image Processing

MATLAB is a language for technical computing, which integrates computation, visualization, and programming. MATLAB is a high performance language that uses an environment where problems and solutions are expressed in familiar mathematical notation.

Image processing Toolbox of the MATLAB uses: Math and computation, algorithm development, data acquisition, modeling, simulation, data analysis, visualization, scientific and engineering graphics, application development, etc.

![Image processing Toolbox](image)

Figure 4.2: Representation of Image processing Toolbox and its functions.

The basic data element that is used in MATLAB is an array that does not require dimensioning. MATLAB is the computational tool for research, development, and analysis, which is complemented by a family of application specific solutions called toolboxes.

The Image Processing Toolbox is a collection of functions that have the capacity for the solution of digital image processing problems.

The MATLAB software was used in the creation of the enhancement algorithms. [Gonzalez, 2004]
4.4 – Computational Vision

Computational vision is the capacity to describe or analyze digital images. For human, this process is relatively easy, because of his powerful vision system.

The image processing and the computational vision are constituted by the following areas:

- Image enhancement and restoration;
- Image segmentation;
- Image characterization;
- Image registration;
- Image classification.

In conclusion, the computer vision is subdivided in three steps: the processing, analysis and recognition. [Gonzalez, 2004]

**Image Enhancement**

Enhancement techniques basically are procedures designed to manipulate an image in order to take advantage of the psychophysical aspects of the human visual system.

Image enhancement can be obtained with a variety of operations. In the magnetic resonance imaging the result is an image with gray scale pixels.
Image enhancement deals with the improvement of visual appearance of the scene for better contrast and visibility of features of interest to be used by either a machine vision system or a human observer.

The enhancement system has all the following functions:

- Attenuate the effects of sub-sampling;
- Attenuate quantization effects;
- Remove noise and simultaneously preserve edges and image details;
- Avoid aliasing effects;
- Attenuate the blockiness effect;
- Improve image contrast;
- Enhancement special features to be more easily detected by a machine or a human observer.

For medical images, noise is always involved in the signal due to the limitations of imaging hardware and protocols. Noise reduction is one the most important objectives for medical image processing. Almost all model-free segmentation methods are sensitive to noise in images. Mode-based methods are more robust, too much noise still can lead to an unsatisfying segmentation result.

Usually, an image is smoothed to remove certain noise by applying a convolution operation onto the image with a certain smoothing kernel.

Image enhancement using local operators is a kind of operation that is also called neighborhood processing or spatial filtering.[Gonzalez, 2008; Gonzalez, 2004]

**Spatial filter**

For filter MR images can be used linear spatial filtering and nonlinear spatial filtering. Linear spatial filter uses the spatial convolution and is characterized by a kernel.

The average filters or arithmetic mean filters performs as a result an image smoothing. [Gonzalez, 2004; Gonzalez, 2008]
Gaussian filters are a class of smoothing filters where the kernel values have 2D Gaussian Shape. In image enhancement for to realize noise reduction in the magnetic resonance images is necessary utilize a Gaussian filter.

Usually, an image is smoothed to remove certain noise by applying a convolution operation onto the image with a Gaussian convolution kernel. Gaussian kernel is defined by equation:

\[ g(x, y) = \frac{1}{2\pi\sigma^2} e^{-\frac{x^2 + y^2}{2\sigma^2}} \] (4.5)

In digital image processing the Gaussian kernel is expressed in a discrete form with a certain size. The kernel is symmetric in both vertical and horizontal directions, and the sum of all kernel elements equals to 1, which guarantee the stability of convolution operation on image data.

The Gaussian smoothing method is that it also tends to blur the sharp boundaries in the image while removing the noise. Even smoothing does not obliterate boundaries of the image. It tends to distort the fine structure of the image and thereby changes subtle aspects of the anatomical shapes. [Feissel, 1984]
An anisotropic diffusion algorithm is used to reduce noise of MRI scans without removing significant features of edges of the structure.

This algorithm was proposed by Perona and Malik in 1990 [Perona, 1990]. In this method a gaussian smoothed image can be taken as a solution to heat diffusion equation with respect to a certain time value t. The heat diffusion equation is in the form of

\[
\frac{\delta g(x,y,t)}{\delta t} = \nabla \cdot \nabla g(x,y) \quad (4.6)
\]

\[
f(x,y) = g(x,y,0) \quad (4.7)
\]

\[
g(x,y,t) = G(\sqrt{2t}) \otimes f(x,y) \quad (4.8)
\]

Where \(f(x,y)\) is an input image, \(g(x,y)\) is a Gaussian smoothed image with a parameter of t, and \(G\) is a Gaussian kernel. The conductance term is included in the anisotropic diffusion. The conductance depends on the differential results of the images. With this term is possible controller the smoothing.

\[
c(|\nabla g(x,y,t)|) = e^{\frac{|\nabla g(x,y,0)|^2}{2\sigma^2}} \quad (4.9)
\]

In this equation \(c\) is the variable conductance term. If the edges of structure in the smoothed images have higher gradient magnitude than of non-edge region, the conductance will be reduced, and the smoothing effect will be limited. The time-dependant gradient magnitude, the conductance is depends on \(k\), which is the conductance parameter. This parameter controls the sensitivity of the process to the structure edge.

In the anisotropic diffusion filter, the \(t\) and \(k\) are the control parameters, while \(\sigma\) controls the Gaussian smooth kernel. In this process is necessary an iteration number, which is need to define when is used the anisotropic smoothing. [Perona, 1990]

Non linear spatial filters don’t use the spatial convolution, an example is median filter. This filter is utilized when is observed salt and pepper noise. When is realized this filter the pixel values contained in the predefined neighborhood are ordered, and choice the middle value.
The gradient or first derivative is calculated from smoothed MRI scans. The intensity of each pixel or voxel changes by providing two types of information, the magnitude and orientation. The magnitude of gradient tells how quickly the intensity changes. The direction tells in which orientation the intensity changes most rapidly. The sharp intensity changes are often across the edges of structures, so gradient magnitude is useful for extracting all of the possible structure edges.

The gradient function can be expressed as a continuous function with two variables.

\[
\nabla f = \text{grad}(f) = \begin{bmatrix} \frac{\partial f}{\partial x} \\
\frac{\partial f}{\partial y} \end{bmatrix} \tag{4.10}
\]

\[
\begin{align*}
M(x,y) &= \text{mag}(\nabla f) = \sqrt{g_x^2 + g_y^2} \\
\theta(x,y) &= \arctg\left(\frac{g_x}{g_y}\right) \tag{4.11}
\end{align*}
\]

Where \(f(x,y)\) is an input image, \(g_x\) is the result of the convolution between the input image and the sobel operator in the \(x\) direction, and \(g_y\) is the result of the convolution of \(f(x,y)\) with the sobel operator in the \(y\) direction.

The sobel operator consists of a pair of 3x3 kernels.

\[
G_x = \begin{bmatrix} 1 & 0 & -1 \\
2 & 0 & -2 \\
1 & 0 & -1 \end{bmatrix} \tag{4.13}
\]
Basic Intensity operations

This operations are characterized by a histogram, $h(g_k)$, of a digital image. Where $g_k$ is the grey level of $k$, $h(g_k)=n_k$, and $n_k$ is the number of pixels with grey level $g_k$.

The histogram equalization assume for a moment that intensity levels are continuous quantities normalized to the range $[0,1]$. The occurrence of gray levels can also be provided in terms of probability values. The normalized histogram of a digital image is:

$$p(g_k) = \frac{n_k}{n} \quad (4.15)$$

Where $n$ is total number of image pixels.

The histogram equalization can be considered as a discrete cumulative distribution function. In this function $T$ is the mapping function:

$$g_k = T(f_k) = \sum_{j=0}^{k} p(g_k) \quad (4.16)$$

This operation corresponding continuous operation gives rise to a uniform histogram.

The net result of this intensity level equalization process is an image with increased dynamic range, which will tend to have higher contrast. The values are approximations to the probability of occurrence of each intensity level in the image.

Other enhancement operation that results of a histogram modification is the histogram stretch. This operation is characterized by a gray-level scaling, when the higher and lower intensity, of the input image and of the histogram, is necessary to know. The histogram stretch results of:

$$g = \frac{g_{\text{max}} - g_{\text{min}}}{f_{\text{max}} - f_{\text{min}}} (f - f_{\text{min}}) + g_{\text{min}} \quad (4.17)$$

Where $g_{\text{max}}$ is the maximum value of the histogram result, $g_{\text{min}}$ corresponds to minimum value of the histogram, $f_{\text{min}}$ and $f_{\text{max}}$ are respectively the intensity values minimum and maximum. [Gonzalez, 2004; Gonzalez, 2008]
Non linear filters

In medical images, the noise signal can be observed as a higher frequency signal. So, a low pass filter can be used to remove or reduce noise [Qazi, 2008].

In general the frequency components can be expressed in low and high ranges. The high-frequency information belongs to sharp details, edges and noise, whereas low-frequency range components usually represent shapes and blurred structures in the image.

Frequency domain filtering methods process an image in the Fourier domain to emphasize or de-emphasize specific frequency components, but the convolution in the spatial domain is the same that the Fourier transform in the frequency domain. Therefore, can be obtained a low pass filter when is realized the convolution with the Gaussian kernel or mean kernel. [Gonzalez, 2008; Qazi, 2008]
4.5 – Practical Results

To study the best techniques of enhancement was realized the comparison between these techniques or methods that using magnetic resonance images of female pelvic cavity. Although this is a valid approach, also are observed some difficulties or restrictions like the noise observation. When performing the enhancement the noise is attenuated as well as the important components of the images. When is doing the edges enhancement the noise is amplified. So, is necessary have in consideration these relations.

Images Characterization

The methodology used for reading magnetic resonance images is an important part of the algorithms implemented in MATLAB.

The images that are used:

![Images from a sagittal T2 – weighted MRI (a and b. Image from a axial T1 – weighted MRI (c. (Images from [Sciarra, 2010])](image)

Figure 4.7: Images from a sagittal T2 – weighted MRI (a and b. Image from a axial T1 – weighted MRI (c. (Images from [Sciarra, 2010])

As mentioned previously, for magnetic resonance images enhancement can be used the following methods:

- Average filters;
- Gaussian filters;
- Anisotropic Diffusion filters;
- Gradient Magnitude;
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- Histogram Equalization;
- Histogram Stretch.

Average Filter

For realize the average filters was used the following algorithm:

```matlab
% read the image:
Img_mri = imread('mri_a.jpg');
% call the average filter:
[I, Ig1, Ig2, Ig3] = filtragem_media(Img_mri);
% Visualization of results for different kernel sizes:
Subplot 221, imshow(I), title('Original Image');
Subplot 222, imshow(Ig1), title('Average filter [3 3]');
Subplot 223, imshow(Ig2), title('Average filter [6 6]');
Subplot 224, imshow(Ig3), title('Average filter [12 12]');
```

The average filter function was created and is represented following:

```matlab
function [I, Ig1, Ig2, Ig3] = filtragem_media(Img_mri)
I=Img_mri;
H1=fspecial('average',[3 3]);
Ig1=imfilter(I,H1);
H2=fspecial('average',[6 6]);
Ig2=imfilter(I,H2);
H3=fspecial('average',[12 12]);
Ig3=imfilter(I,H3);
end
```

This function realizes three convolutions with different kernel sizes and was obtained these results, figure 4.8, from the image a):

![Figure 4.8: Results of the image after the passage of the filter. The first image represents the original image and any filtering was performed on this array.](image)

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The less noise can be obtained through the difference between the original and filtered image. The following algorithm shows the results for different kernels in order to verify the most effective.

```matlab
% I represents original image
% [Ig1, Ig2 and Ig3] are the result of the average filter
noise_1 = [-Ig1];
noise_2 = [-Ig2];
noise_3 = [-Ig3];
% Visualization of the results
subplot(131), imshow(noise_1), title('Noise Attenuated with kernel [3 3]');
subplot(132), imshow(noise_2), title('Noise Attenuated with kernel [6 6]');
subplot(133), imshow(noise_3), title('Noise Attenuated with kernel [12 12]');
```

The obtained images were:

![Images showing noise attenuation with different kernels](image)

Figure 4.9: Images that representing the attenuated noise.

With the analysis of noise images is observed that the use of the filter is affected by the kernel size. A great size of kernel attenuates the contours. This attenuation may hamper the next analysis processes of the contours, such as segmentation, which uses the edges as a starting point for carrying out this type of processing.

So, it appears that the kernel size [3 3] and [6 6] are more acceptable than the kernel [12 12]. The latter performs an exaggerated attenuation and causes a high blurring.

For the others two images was performed the same procedure and obtained the following results, figure 4.10 and figure 4.11, for the filtering.

The noise attenuation are represented in figures 4.12 (image b) and 4.13 (image c).
Figure 4.10: Results of Average filtering for the image b.

Figure 4.11: Results of Average filtering for the image c.
As in the image “a”, the images “b” and “c” contain a high attenuation of the contours for the average filter when the kernel presented a size greater than 3.

So, when this type of filters are used on medical images must take into account that the use of a small kernel is the best way to get good results without too much blur the edges.

**Gaussian Filter**

The second algorithm created to achieve enhancement of the images was the Gaussian filter.

For implementation of Gaussian filter was used the following code.
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The function called by “Filtragem_gaussiana” is presented as:

```matlab
function [I, Igl, Ig2, Ig3] = Filtragem_gaussiana(I)
I1=fspecial('gaussian',[3 3],0.5);
Igl=imfilter(I,I1);
I2=fspecial('gaussian',[6 6],1);
Ig2=imfilter(I,I2);
I3=fspecial('gaussian',[12 12],2);
Ig3=imfilter(I,I3);
end
```

This code shows the result of the filter kernels with different sizes and consequently with different standard deviations. The results obtained for the image “a” are shown in figure 4.14.

![Figure 4.14: Results obtained by filtering Gaussian on the picture “a”.

Original Image  |  Gaussian Filtering [3 3] 0.5
---|---
Gaussian Filtering [6 6] 1 | Gaussian Filtering [12 12] 2
To check if the noise, which was eliminated, also consists in major components of the image was held again the difference between the original image and the result of the filtering.

![Figure 4.15: Eliminated noise in the Gaussian filtering (image a).](image)

For the images B and C was done getting the same processing. As the result of Gaussian filtering are presented the figures 4.16 and 4.17. The eliminated noise was observed and is expressed in the figures 4.18 and 4.19.

![Figure 4.16: Result of Gaussian filtering on the image b.](image)
Figure 4.17: Result of Gaussian filtering on the image c.

Figure 4.18: Attenuated noise from the image b.

Figure 4.19: Attenuated noise from the image c.
When observing the results of Gaussian filtering it appears that this type of filter for medical images need a kernel greater than 6 and therefore the standard deviations must be higher than 1.

These aspects can be conclude because with observation of images for the kernels [3 3] and [6 6] the contours don’t are very attenuated.

### Anisotropic Diffusion Filter

To achieve the anisotropic diffusion filter was used the following algorithm:

```matlab
function diff_im = anisodiff2d(I, num_iter, delta_t, kappa, option)
%ANISO2D Conventional anisotropic diffusion
% DIFF_IN = ANISO2D2D(IM, NUM_ITER, DELTA_T, KAPPA, OPTION) performs
% conventional anisotropic diffusion (Perona & Malik) upon a gray scale
% image. A 2D network structure of 8 neighboring nodes is considered for
% diffusion conduction.

% ARGUMENT DESCRIPTION:
% I - gray scale image (IMN).
% NUM_ITER - number of iterations.
% DELTA_T - integration constant (0 <= delta_t <= 1/7).
% Usually, due to numerical stability this
% parameter is set to its maximum value.
% KAPPA - gradient modulus threshold that controls the conduction.
% OPTION - conduction coefficient functions proposed by Perona & Malik:
% 1 = c(x,y,t) = exp(-(module/kappa).^2),
% privileges high-contrast edges over low-contrast ones.
% 2 = c(x,y,t) = 1./|1 + (module/kappa).^2),
% privileges wide regions over smaller ones.

% OUTPUT DESCRIPTION:
% DIFF_IN - (diffused) image with the largest scale-space parameter.

% Convert input image to double.
I = double(I);

% PDE (partial differential equation) initial condition.
diff_im = I;

% Center pixel distances.
dx = 1;
dy = 1;
d1 = sqrt(2);

% 2D convolution masks - finite differences.
hN = [0 1 0; 0 -1 0; 0 1 0];
hS = [0 0 0; 0 -1 0; 0 1 0];
hE = [0 0 0; 0 -1 1; 0 0 0];
hW = [0 0 0; 1 -1 0; 0 0 0];
hNE = [0 0 1; 0 -1 0; 0 0 0];
hSE = [0 0 0; 0 -1 0; 0 1 0];
hSW = [0 0 0; 0 -1 1; 0 0 0];
hNW = [1 0 0; 0 -1 0; 0 0 0];
```

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```
\% Anisotropic diffusion.
for t = 1:num_iter
    \% Finite differences. [imfilter(.,.,'conv') can be replaced by conv2(.,.,'same')]
    nablaN = imfilter(diff_im,N,N,'conv');
    nablaS = imfilter(diff_im,S,S,'conv');
    nablaW = imfilter(diff_im,W,W,'conv');
    nablaE = imfilter(diff_im,E,E,'conv');
    nablaNW = imfilter(diff_im,NW,NW,'conv');
    nablaSE = imfilter(diff_im,SE,SE,'conv');
    nablaSW = imfilter(diff_im,SW,SW,'conv');
    nablaNE = imfilter(diff_im,NE,NE,'conv');

    \% Diffusion function.
    if option == 1
        cN = exp(-(nablaN/kappa).^2);
        cS = exp(-(nablaS/kappa).^2);
        cW = exp(-(nablaW/kappa).^2);
        cE = exp(-(nablaE/kappa).^2);
        cNW = exp(-(nablaNW/kappa).^2);
        cSE = exp(-(nablaSE/kappa).^2);
        cSW = exp(-(nablaSW/kappa).^2);
        cNE = exp(-(nablaNE/kappa).^2);
    elseif option == 2
        cN = 1./1 + (nablaN/kappa).^2;
        cS = 1./1 + (nablaS/kappa).^2;
        cW = 1./1 + (nablaW/kappa).^2;
        cE = 1./1 + (nablaE/kappa).^2;
        cNW = 1./1 + (nablaNW/kappa).^2;
        cSE = 1./1 + (nablaSE/kappa).^2;
        cSW = 1./1 + (nablaSW/kappa).^2;
        cNE = 1./1 + (nablaNE/kappa).^2;
    end

    \% Discrete PDE solution.
    diff_im = diff_im + delte_t*((1/(dy^2)) * cN * nablaN + (1/(dy^2)) * cS * nablaS +
                                (1/(dx^2)) * cW * nablaW + (1/(dx^2)) * cE * nablaE +
                                (1/(dd^2)) * cNW * nablaNW + (1/(dd^2)) * cSE * nablaSE +
                                (1/(dd^2)) * cSW * nablaSW + (1/(dd^2)) * cNE * nablaNE);

    \% Iteration warning.
    fprintf('| iteration %d/n',t);
end
```
This filtering was implemented for the three images, considering the first option, obtained as a result the figures 4.20 (image a), 4.21 (image b) and 4.22 (image c).

![Image of anisotropic diffusion filtering results](image)

**Figure 4.20**: Result of anisotropic diffusion filtering for image “a”.
Figure 4.21: Result of anisotropic diffusion filtering for image “b”.

Figure 4.22: Result of anisotropic diffusion for image “c”.
The noise extracted from the images for this type of filtering was obtained by the same method. The results can be seen in the figures 4.23, 4.24 and 4.25.

![Figure 4.23](image1.png): Removed noise for the image “a”.

![Figure 4.24](image2.png): Removed noise for the image “b”.

![Figure 4.25](image3.png): Removed noise for the image “c”.
With the first option was observed the privilege of high-contrast edges at the expense of low-contrast, functioning as a high pass filter instead of the filters previously processed.

**Gradient**

As already stated this process is specific to enhancement the edges, because it corresponds to the derivate in the horizontal and vertical orientation.

The next algorithm is the gradient operation.

```
function mag = grad_func(I)
im = double(I);
w = fspecial('gaussian', [18 18], 2.5);
f = imfilter(im, w, 'corr');
[px,py] = gradient(f);
Ph=px.^2;
Pv=py.^2;
mag=sqrt(Ph+Pv);
end
```

The results observed for the three images are shown in figures 4.26, 4.27 and 4.28.

![Figures 4.26, 4.27 and 4.28: Results of the gradient magnitude for the image "a".](image-url)
The magnitude of the gradient should be held only after removing noise, because this operation is very much affected if don’t realize the removal noise. This erroneous result will interfere in steps towards. So, the next processing and analysis of images is particularly concerned.

**Histogram Equalization**

Histograms indicate the number of pixels corresponding to certain intensity, as previously stated.

To implement algorithm of this kind of histogram was performed the following code:
The results of this implementation are observed in figures 4.29, 4.30 and 4.31. In addition of the resulting images can be seen the resulting histograms.

Figure 4.29: Results for the image “a”.

```matlab
%% Read Image
I=imread('mri_a.jpg');

%%Function of the histogram equalization
G = histeq(I);

%%Visualization of the results
figure, subplot 221, imshow(I,[],), title('Original image'),
subplot 222, imshow(G,[],), title('Resulted Image'),
subplot 223, imhist(I), title('Histogram of the Original Image'),
subplot 224, imhist(G), title('Histogram equalization');
```
Figure 4.30: Results for the image “b”.

Figure 4.31: Results for the image “c”.
Through analysis of the images can be seen that the histogram equalization transform the intensity levels present in the original image in an equalized distribution of all levels between the maximum and the minimum value by using the cumulative function.

**Histogram Stretch**

The last designed algorithm for the observation of the enhancement was the histogram Stretching. The code is presented below.

```matlab
%Read Image
I = imread('mri_c.jpg');

I = rgb2gray(I);

%call function
stretch = imadjust(I);

%Visualization of the results
figure, subplot 221, imshow(I,[],), title('Original image'),
subplot 222, imshow(stretch,[],), title('Resulted Image'),
subplot 223, imhist(I), title('Histogram of the Original Image'),
subplot 224, imhist(stretch), title('Histogram Stretch');
```

For three original images were obtained results in the form of image and histogram. Figures 4.32, 4.33 and 4.34 show that.
Figure 4.33: Results of the histogram stretch from image "b".

Figure 4.34: Results of the histogram stretch from image "c".
After viewing the figure 4.32 it appears that the original image is already enhanced by the use of the histogram stretch, because the results are exactly alike.

**Discussion**

The realization of these methods allowed the observation of several facts.

To achieve the enhancement is necessary eliminate high frequency noise while maintaining the sharp edges. The performed algorithms show results that only satisfy one of the parties.

The average filter and Gaussian filter perform smoothing behave as low pass filters. These filters soften the contours. When compared, it appears that the Gaussian filter has a better behavior because no blurs the edges too much.

The gradient operation should not be used without a performing a prior filtration.

The anisotropic diffusion filter blurs areas of low contrast and enhanced the high contrast (edges). This filter works as a high pass filter.

The histogram equalization and histogram stretch don’t present significant changes in the resulting images.

Thus, propose the use of a band-pass filter, which initially performs a Gaussian filtering and then an anisotropic diffusion filter. With this band pass filter the high frequency noise is eliminated and edges are enhanced.
V – Final Considerations and Future Perspectives
5.1 – Final Considerations

The techniques of image enhancement have a high technological potential. These techniques are always used together with methods of analysis and quantification.

In this work was prepared a review of the state of the art with the description of the methodologies of image enhancement. The literature review was also performed to provide increased knowledge in this area.

The conclusion focuses on the fact that the high frequency noise needs to be removed and the contours remain highlighted. In order to respond to this problem proposed the creation of a band-pass filter (Gaussian filter followed by Anisotropic Diffusion).

Beyond the study of methods of image enhancement in this work were also analyzed anatomically organs of the pelvic cavity, as well as the principles of magnetic resonance images. Both studies were instrumental in the work developing.
5.2 – Future Perspectives

This document serves as an element of the draft framework paper. To conduct the dissertation I intend to perform the validation of the methods by using a larger number of images and to make a statistical analysis.

This work represents the first step to then be made other processes, such as segmentation, features extraction, alignment and 3D reconstruction.
References
Analysis of Structures in Medical Images


Analysis of Structures in Medical Images


