Chapter 1

Introduction

Medical imaging technologies, due to its impact on medicine and biology, has been selected as one of the greatest achievements of twentieth century by the National Academy of Engineering. In the last several decades, the advancement of medical imaging has taken a quantum leap. Some of the medical imaging techniques are X-ray, Computed Tomography (CT), Ultrasonography (USG), Magnetic Resonance Imaging (MRI), Positron Emission Tomography (PET) and Single Photon Emission Tomography (SPECT) etc.. Only a few devices among the medical equipments can match the popularity of the computed tomography scanner (or CT scanner). This invention undoubtedly started a revolution in diagnostic technology by obtaining a very clear anatomical image without violating the outer surface of the body, i.e. non-invasively.

The word *tomography* is derived from the Greek *tomos* (part) and *graphein* (to write). So, *tomography* means imaging by sections through the use of any kind of penetrating wave. In CT, the penetrating wave is called X-ray, which is an electro-magnetic wave. So, the origin of CT owes a great deal to accidental discovery of X-radiation (or X-rays) by the German Scientist Wilhelm Conrad Rontgen on 8th November 1895. The first CT scanner was developed in 1971, which is based on the mathematical theory devoloped by Czech mathematician Johann Radon in 1917 [1, 2]. Radon demonstrated mathematically that an object could be replicated from an infinite set of its projections.

The two people recognised as the fathers of CT are: Allan Nacleod Cormack and Godfrey Newbold Hounsfield. Both were awarded Nobel Prize for Physiology or Medicine in 1979. Cormack had solved the problem of image reconstruction of X-ray projections theoretically and experimentally confirmed the results of his research [3, 4]. Cormack, a theoretical physicist, was not concerned about the practical applications of his research. Hounsfield, employed that time at the Central Research Laboratories of EMI Ltd, constructed the first CT scanner and patented his device in 1968 [5]. Independent of Cormack, he developed a different approach to the problem of image reconstruction initially using gamma radiation and using the power of computers available at that time to carry out the complicated calculations. In this way, the concept of computed tomography came into picture. Later, along with neurologist James Ambrose, an improved prototype scanner, the EMI Mark I was developed replacing the source of gamma radiation with an X-ray tube in 1971 [6, 7]. In the first scanner, Hounsfield had used algebraic reconstruction technique (ART) algorithm. The scanner was able to produce the image of the head only due to its small size of the opening. The basic parameters of the first scanner were: scan time of about 4.5 minutes, reconstruction time of about 20 minutes and image matrix of size 80 $\times$ 80 pixels. By the end of 1973, first commercial CT scanner, EMI CT 1000 model was developed with scan time reduced to 20s and reconstructed image of size 320 $\times$ 320 pixels [5].

The formation of a CT image is a two stage process. The first stage i.e. scanning, produces data commonly known as sinogram. The second stage i.e. reconstruction, processes the acquired sinogram data and forms a digital image [8, 9]. There are various methods of scanning and the reconstruction procedure depends upon the particular scanning technique used in a CT scanner. The sampling geometry of CT scanners can be described in three configurations: Parallel-beam, Fan-beam and Cone-beam. Parallel-beam reconstruction belongs to the first approaches used for CT. Both the source and detector are placed on a circle. The negative logarithm of the projection data is back-projected along the ray direction. The projection data are smoothened and processed by a special filter. Fan-beam CT reduces scanning times since it captures a full scan line in parallel. Never the less, the parallel-beam reconstruction technique may be used as well for fan-beam CT, since individual parallel rays can be selected from each fan and considered as a parallel projection. During 1990s, the spiral CT was invented by Kalender [10]. Spiral CT uses fan-beam scanning method but the table is shifted with constant speed through the gantry so that the X-ray source follows a screw line around the patient. For spiral CT the projection lines do not lie in parallel planes as for fan-beam CT. Linear interpolation of the captured data on parallel planes allows use of the fan-beam reconstruction method. In recent years, CT scanners with several detector lines have been introduced reducing the overall scanning time for whole body scans. Despite the divergence of the rays in different detector rows, standard reconstruction techniques are used. The latest commercial CT scanners have two-dimensional detector arrays with up to 64 rows, allowing the coverage of a large area of the body within a short scan time. Along with efficient image reconstruction algorithms, hardware developments have played a very important role in the development of CT. Since the introduction of the first clinical scanner, tremendous advancement has been made in CT technology. The slice data acquisition time and the reconstruction time has been reduced drastically due to the change in scanning methods as well as due to the evolution of various reconstruction algorithms. Figure 1.1 shows the reported scan time over 40 years of CT history [11]. The reduction in scan time follows an exponential relationship over time. The CT scanners have doubled their acquisition power every 2.3 years over the last 40 years.

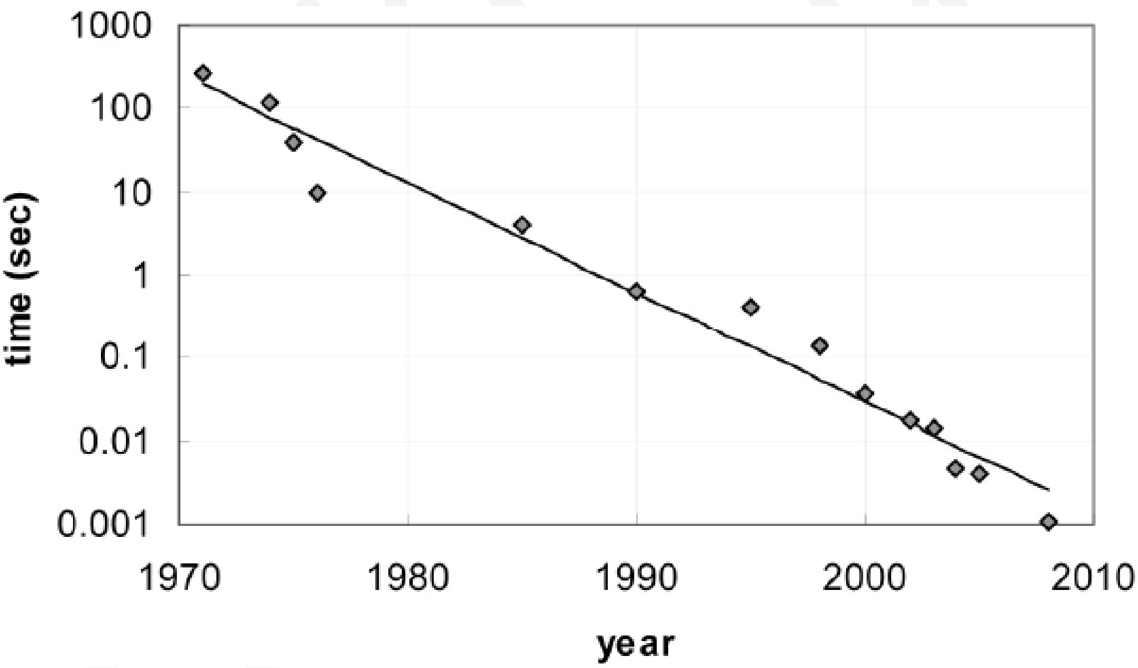

Figure 1.1: Scan time per slice as a function of time in logarithmic scale

During CT data acquisition, the X-rays pass through the slice of the body being examined. Each sample acquired in a CT system is equal to the sum of the image values along a ray pointing to that sample. There are various approaches to calculate the slice image given the set of its views. These are called CT image reconstruction algorithms. These reconstruction algorithms are generally based on mathematics of

Radon Transform. The practical reconstruction algorithms use either a direct or indirect method to implement inverse Radon transform. Several techniques such as Algebraic Reconstruction Techniques (ART) [12, 13], Filtered back-projection (FBP) [8, 14, 15] and Direct Fourier reconstruction (DFR) [16, 17] are applied to reconstruct the image.

Although many types of methods may be used for the reconstruction, the most widely used reconstruction algorithms are part of the FBP family. These algorithms consist of successive 1-D filtering followed by a "back-projection" integral. The FBP algorithms are the most popular because they allow for both fast computation and high accuracy.

With increase in computing power, interest in fully 3D computed tomography has become a focus [18]. Of course, multiple 2D reconstructions may be registered to yield a single 3D volume. However, this technique is undesirable due to the necessarily long acquisition times to obtain many slices. Instead, research efforts have focused on direct 3D reconstructions. To allow for the rapid acquisition of large amounts of projection data, a three dimensional rectangular cone-shaped X-ray beam together with a 2D detector array directly opposite to the X-ray source has been employed. CT performed with such data is termed as "cone-beam" computed tomography. Although many types of scanning paths are possible, focus in the clinical setting has been on cone-beam CT using a helical scanning path for the X-ray source. The patient lies on a platform which translates through the rotating source/detector gantry. In this way, the X-ray source traces out a helix around the body of the patient.

The development of spiral cone-beam CT may be divided into the following two methods: (1) approximate reconstruction (2) exact reconstruction. The merits of the approximate cone-beam algorithms are not only in their computational efficiency but also in several aspects of image quality and radiation dose. Given a small or moderate cone angle like what is now used for medical CT, approximate cone-beam algorithms may be competitive against exact alternatives in terms of image resolution, image noise, temporal consistency and dose efficiency. The current practice is that all the major CT manufacturers still use approximate cone-beam algorithms as the working model.

In the approximate reconstruction method, a number of very successful algorithms were developed using the Feldkamp approach. In 1984, Feldkamp, Davis, and Kress proposed a FBP algorithm for cone-beam CT suitable for a circular scanning path [19]. The FDK (Feldkamp, Davis, and Kress) algorithm may also be easily extended to more general scanning paths, such as a helix, e.g. see Wang et al[20]. The FDK algorithm is fast and gives reasonable results in some circumstances. However, it is based on an approximate inversion formula (before discretization). Consequently, it is difficult to fully analyze its numerical properties and to predict reconstruction artifacts. Since the FDK algorithm was proposed, a great deal of effort has been invested in looking for faster and more accurate reconstruction methods. Of particular interest are methods based on theoretically exact inversion formulas which are of the FBP type. The FDK algorithm introduces image artifacts which increase in proportion to the cone-beam angle.

In the method of exact reconstruction with multiple turns, the exact reconstruction is achieved using the Radon inversion approach. The key idea is to recover the Radon transform from truncated data segments. The exact 3D reconstruction method presented by Grangeat [21] consists of two steps viz. conversion of cone-beam data to derivatives of Radon data followed by reconstruction of 3D object data from derivatives of Radon data. The use of Modified Fast Radon transform (MFRT) is well established for parallel-beam and fan-beam scanning methods [22]. In recent years, several exact algorithms have been developed. In 2002, Alexander Katsevich made a breakthrough in the field of exact algorithm [23, 24]. In these papers, he proved a theoretically exact reconstruction formula for helical cone-beam CT. Although Katsevich algorithm is FBP type, 3D CT image reconstruction is computationally demanding. To accelerate the Katsevich algorithm, some work has been reported using CPU or cluster by Deng et al. [25], Yang et al. [26] and Fontaine and Lee [27]. Recently, GPU is used to accelerate the algorithm by Yan et al. [28].

In both CT and MRI systems, cross-section slice images of patients are produced after the reconstruction. These slice images can be used to generate 3D models that can be viewed from any direction. The process of visualizing such models is referred to as volume rendering. In direct volume rendering method, shear-warp factorization technique is used to reduce the amount of computations [29, 30]. Hence, Affine transformation of volume arrays are important for implementing volume modeling [31]. Affine transform consists of four basic operations such as rotation, scaling, shearing and translation.

In CT imaging systems, the analysis of the interior of an organ is critical to detect eventual diseases or for surgical operations. The image registration is required to compare pre-contrast and post-contrast CT images for lesion detection [32]. The Affine transform is also an important operation in medical image registration and is iteratively used during registration procedure. Also, to facilitate an accurate diagnosis and for detection of abnormality of affected organs, the medical practitioner may require to view the acquired CT images in different orientations from various angles. The object is often mapped to a new coordinate system by performing Affine transform on it.

1.1 Motivations of the Present Work

Motivation of the work reported in this thesis is briefly addressed here:

- The most time consuming and complex part of the reconstruction is backprojection. To accelerate the backprojection in the Katsevich algorithm a lot of efforts have been made using cluster of CPUs and GPU. As far as we know, the acceleration of the Katsevich algorithm on FPGA has not been reported in the literature so far. Therefore, a high speed, low power VLSI architecture is required that will provide reconstructions of high quality at low computational effort, with the ultimate aim of making an exact algorithm, i.e. the Katsevich algorithm, an efficient choice for routine clinical use in cone-beam CT machines.

- Generally, in the 3D imaging techniques (e.g. CT and MRI), data are acquired as a series of 2D slices. So, pair-by-pair registration of a series of 2D biological images enables the reconstruction of a 3D image. The registration between two consecutive slices has to be achieved in real time during the data acquisition, which demands real time implementation of the Affine transform. This transform is also used for better visualization of the CT images by medical practitioners in real time. So, for real time processing, a dedicated hardware involving parallelism and pipelining may be a good alternative.

1.2 Objectives of the Thesis Work

The objectives of this thesis are as follows.

- Analyze the Katsevich algorithm for cone-beam 3D CT image reconstruction.

- Design an architecture to implement the Katsevich algorithm for cone-beam CT in FPGA, maintaining the image quality.

- Develop a modified algorithm for faster implementation of the Affine transform.
- Design an efficient architecture to implement fast 2D Affine transform on FPGA for real time visualization as well as for image registration.
- Design and implement fast 3D Affine transform on FPGA for real time visualization.

Both back projection and Affine transform cores are synthesized using verilog and prototyped on Xilinx Field Programmable Gate Array (FPGA). During implementation, attempts are made to reduce the power consumption as well as the gate count and hence area on silicon.

1.3 Outline of this Dissertation

In this dissertation, VLSI architecture is proposed to implement the back projection algorithm of the cone-beam CT using Katsevich algorithm which may help in routine clinical use. Apart from that, an unified VLSI architecture for 2D and 3D Affine transform is proposed which is used for better visualization of CT images. This section provides the structure of this dissertation.

The history and evolution of CT is described in Chapter 1. Starting from the accidental invention of X-ray to first CT machine has been described in this chapter including a brief description about the evolution of the CT machine along with the reconstruction algorithm. Finally, objectives of the research work carried out in this dissertation are presented.

Overview of the CT machine is covered in Chapter 2. Detailed discussion about the various blocks of the CT machine is covered. Emphasis has been given to describe the various scanning methods and reconstruction algorithms. The reconstruction algorithm of the cone-beam CT using approximate and exact algorithms are being introduced in this chapter.

Due to the complexity of the Katsevich algorithm, it has not been included in current CT machines till now. But lot of research is going on to implement this algorithm and to make it feasible to include in current CT machines. VLSI architecture for the backprojection of the Katsevich based reconstruction algorithm is presented in Chapter 3. The use of CORDIC technique to reduce the mathematical complexity is explained. The detailed parallel architecture of the algorithm and its implementations in FPGA are discussed.

In CT machines, after reconstruction 2D slices of the image are obtained. For better visualization, 3D volume rendering of the slices are required. 2D and 3D Affine transforms play an important role for this operation. Chapters 4 and 5 describe the VLSI architectures of 2D and 3D Affine transforms respectively. An modified algorithm applicable for both 2D and 3D Affine transforms are proposed and mathematically proved. The proposed algorithm is implemented in MATLAB to analyse the image quality. The performance analysis of the proposed algorithm with the conventional algorithm is also presented. The pipelined VLSI architectures of both 2D and 3D Affine transforms are described considering the parallelism of the algorithm and are implemented in FPGA.

Finally, Chapter 6 summarizes the contributions of this dissertation. Possible directions for further research works are also presented in this chapter.