Recent advances in Optical Computed Tomography (OCT) imaging system for three dimensional (3D) radiotherapy dosimetry

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Abstract. Radiotherapy delivery techniques for cancer treatment are becoming more complex and highly focused, to enable accurate radiation dose delivery to the cancerous tissue and minimum dose to the healthy tissue adjacent to tumour. Instrument to verify the complex dose delivery in radiotherapy such as optical computed tomography (OCT) measures the dose from a three-dimensional (3D) radiochromic dosimeter to ensure the accuracy of the radiotherapy beam delivery to the patient. OCT measures the optical density in radiochromic material that changes predictably upon exposure to radiotherapy beams. OCT systems have been developed using a photodiode and charged coupled device (CCD) as the detector. The existing OCT imaging systems have limitation in terms of the accuracy and the speed of the measurement. Advances in on-pixel intelligence CMOS image sensor (CIS) will be exploited in this work to replace current detector in OCT imaging systems. CIS is capable of on-pixel signal processing at a very fast imaging speed (over several hundred images per second) that will allow improvement in the 3D measurement of the optical density. The paper will review 3D radiochromic dosimeters and OCT systems developed and discuss how CMOS based OCT imaging will provide accurate and fast optical density measurements in 3D. The paper will also discuss the configuration of the CMOS based OCT developed in this work and how it may improve the existing OCT system.

1. Introduction
Approximately one-third of the population will develop cancer during their lifetime and about half of these are curable with advances in cancer treatment techniques. Patients are treated using surgery, radiotherapy or chemotherapy or in combination of several treatment techniques. The latest National Cancer Registry reported that the number of patients diagnosed with cancer in Malaysia is increasing;
about 18000 new cancer cases were diagnosed and registered in 2007 [1]. The number is expected to rise steeply by half-fold by 2030 according to a World Health Organization report in 2012 [2].

Modern radiotherapy is one of the main treatment technique to cure cancer. The technique delivers high radiation dose to the tumour whilst minimising dose to the healthy tissues. Development of advanced radiotherapy techniques such as Intensity Modulated Radiation Therapy (IMRT) has improved the ability to better target the radiation by delivering higher and more conformal radiation dose to the target, further improving the use of radiotherapy as a cancer treatment technique and increasing cancer cure rates [3]. However, the complexity of the advanced radiotherapy delivery can severely compromise the dosimetric accuracy of the beam delivery, hence, verification of the treatment is important to ensure treatment success. Radiotherapy treatment needs highly sensitive dosimetry systems with high spatial resolution to accurately map doses to tissues. According to the International Commission of Radiation Units (ICRU) recommendations, the accuracy of the dose delivered to the target should be within ± 5% [4]. To produce at least 95% accurate results, the standard deviation of the dosimeter factors should be less than 2% [5].

IMRT delivers the focused radiation beam by collimating the radiation with a multi leaf collimator (MLC) to match the shape of tumour region. This is to ensure radiation is delivered mostly to the target and minimising exposure to surrounding healthy tissue and other organs at risk. The field shape and dose rate also changes dynamically around the patient. These advanced treatments are prone to delivery errors due to their complex nature. The rapid advances in radiotherapy delivery techniques has outpaced the dosimetry system to accurately monitor the delivery. To achieve precise and accurate measurements of the associated absorbed dose distribution of the treatment, the medical physicist utilises dosimetric devices such as ionisation chambers, film, thermoluminescence detectors and diodes. Each of these devices is either a 1D or 2D measurement device, while conversely the patient and the delivery of radiotherapy is a 3D problem. These complex radiotherapy techniques require verification of the 3D dose distribution. The Monte Carlo simulation is well known to be capable of providing dose deposited in a medium; however, the end result is only as good as the model assumptions. Therefore, an experimental 3D verification is of significant importance in the delivery of dose in radiotherapy treatment. In this respect, there is a need for an advanced dosimetric system that can quantify and verify the accurate dose distribution as well as being capable in recording the integrated absorbed dose in 3D for radiotherapy dosimetry application.

An optical computed tomography (OCT) is a 3D imaging system consists of a light source and an image sensor to detect the light and mapped the optical density of a dosimeter in 3D. The dosimeter is a radiochromic polymer material whose chemical properties change predictably upon exposure to x-rays. Light transmitted through the object is refracted and recorded by the image sensor as optical density. Current commercial OCT system uses a charged coupled device (CCD) as the image sensor [6-9]. CCD has matured and can provide good image quality, but the chip has limitation in providing on-pixel image processing that is useful in an OCT imaging system. The conventional CCD based OCT’s scanning process of the radiosensitive polymer is also time consuming, thus, limit the usefulness of this instrument as a rapid system for complex radiotherapy dose verification.

This paper will review development of radiochromic dosimetry system and OCT imaging system for radiotherapy dosimetry. The work investigates application of complementary metal oxide semiconductor (CMOS) image sensor to improve the optical computed tomography (OCT) system. The aim is to improve the imaging speed of the CCD-based OCT.

2. Review of 3D Radiochromic Dosimeter

Three dimensional (3D) radiochromic dosimeter is based on chemical reactions in the dosimeter material which are induced by ionising radiation, that enables the dosimeter to record the radiation dose distribution in 3D. Radiation sensitive dosimeter were first suggested by Day and Stein in 1950 when radiation produced colour changes in gels containing dyes such as methylene blue and phenol-indo-2:6-di-chlorol. Several studies have been reported using gel dosimetry for 3D verification in
radiotherapy. The development of 3D dosimetry started from use of Fricke Gels [10], Polymer Gels [11] and recently with the 3D radiochromic plastic material of PRESAGE™ [12].

The Fricke Gel dosimeter was introduced by Gore et al, in 1984, based on chemical dosimeters with a gelling agent to stabilise or fix the chemical reaction as a result of absorbed radiation [10]. The dosimeter is based on oxidation of ferrous ions (Fe+2) to ferric ions (Fe+3) as an aerated acid solution of ferrous sulphate is irradiated. The absorbed dose can then be estimated by measuring the concentration of ferric ions before and after irradiation. Gore et al (1984) proposed that nuclear magnetic resonance (NMR) relaxation measurements of a standard Fricke solution could be used to determine the absorbed dose of ionising radiation [13]. Ferrous sulphate solutions were subsequently incorporated into a gel matrix in order to stabilise the irradiated magnetic resonance imaging (MRI) absorbed dose signature spatially. Longitudinal and transverse relaxation rate images could then be used to map the spatial distribution of absorbed radiation dose [14]. Several studies have concerned the application of Fricke gel dosimeters in radiotherapy. This combination of chemical dosimeter, gelling agent and evaluation technique was the first truly 3D dosimeter to be used to investigate radiotherapy dose distributions. However the diffusion of ferric ions (Fe+3) throughout the gel give rise to significant blurring of the observed dose distribution [15]. The dosimetric properties of the Fricke gel dosimeters are strongly dependent on gel preparation. Since the dosimeter is based on chemical dosimeter principles, thus the chemical component will affect the radiation chemical yield.

The use of polymer gels for radiation dosimetry based on radiation polymerisation has also been proposed. Early study of polymer gel dosimetry was made in 1954 by Alexander et al [16] and Mesrobian et al [17], describing the effects of polymerisation in acrylamide following irradiation. The polymer gel dosimeters are made up from radiation sensitive chemical interaction with ionising radiation. Thus, a number of reactions are initiated that lead to the conversion of monomers to polymers, resulting in the formation of an insoluble polymer network. Polymerisation leads to a change in the optical properties of the dosimeter, and it therefore becomes opaque [11]. The rate of polymerisation is presumed to depend on the rate of formation of the initiating fragments, the intrinsic reactivity of the monomer and the concentration of radicals and monomers present in the solution [18]. Therefore, the process of polymerisation can be controlled by the type of monomer used, monomer concentration and dose rate [19]. Further investigation included studies by Hoecker in 1958, examining the dosimetric characteristics of radiation-induced polymerisation in liquids [20]. The application of polyacrylamide for gamma dosimetry was proposed by Boni in 1961 [11]. Several studies have reported changes in NMR transverse and longitudinal relaxation rate in irradiated polymer gel dosimeters [21]. This system was named BANANA (bis, acrylamide, nitrous oxide and agarose) [23]. This formulation did not suffer the same drawback that Fricke gels did with continual post-irradiation diffusion [11] and showed stability to prevail several months post-irradiation. The polymerisation reaction occurs by cross-linking of the monomers induced by free radical products of water radiolysis. This gradually increases the rate of water proton spin relaxation, up to doses of 15 Gy [23].

In 1992 a new gel dosimetry formulation was proposed, which was based on the polymerisation of acrylamide and N,N’-methylene-bis-acrylamide (bis) monomers infused in an aqueous agarose matrix [22]. This system was named BANANA (bis, acrylamide, nitrous oxide and agarose) [23]. This formulation did not suffer the same drawback that Fricke gels did with continual post-irradiation diffusion [11] and showed stability to prevail several months post-irradiation. The polymerisation reaction occurs by cross-linking of the monomers induced by free radical products of water radiolysis. This gradually increases the rate of water proton spin relaxation, up to doses of 15 Gy [23].

In 1994 Maryanski et al [24] refined the BANANA formulation by replacing agarose with gelatine and the first of the BANG gels was born, now consisting of bis (3%), acrylamide (3%), nitrogen and gelatine (5%). In 1994 this formulation was patented by MGS Research, Inc., Madison as a registered trademark and became commercially available and known as BANG® [24]. The BANG gel, readout using MRI facilities resulted in having a lower background for the non-irradiated gel, and a more transparent medium in which the irradiated region is clearly visible [25]. In 1996, a new BANG type gel formulation, which contained bis, acrylic acid, sodium hydroxide, water and gelatine was described by Maryanski et al [26]. The gel differs from BANG mainly in the substitution of acrylic acid for acrylamide and was named BANG-2. This also became commercially available under the
MGS Research Inc. Madison, later BANG - 3 and - 4 were developed and also became commercially available.

A manufacturing procedure for BANG gel was later described by Baldock et. al., [21], and is now most commonly referred to in the literature as PAG, which is an acronym for polyacrylamide gel. Further revision of the characteristics of polymer gel dosimeters were suggested with alternative monomers such as, N-vinylpyrrolidone argon (VIPAR) gels [27], sodium methacrylate [28], 2-hydroxyethyl methacrylate (HEMA) and 2-hydroxyethyl acrylate (HEA) [29]. However, the PAG formulation consisting of co-monomers N, N’- methylene-bis-acrylamide and acrylamide has been the most characterised from in the literature, being studied by numerous research groups, and is known to have desirable qualities for radiation dosimetry, such as reliability, stability and reproducibility [30].

Polymer gel dosimetry systems can also suffer major limitations. Oxygen inhibits the polymerisation reaction as it scavenges free radical initiators (such as OH* and H*) that are produced during water radiolysis [31] and therefore, the gels should only be manufactured under a oxygen free atmosphere (typically nitrogen rich), otherwise referred to as an hypoxic environment. While several groups have performed characterisation of the gels, polymer gel dosimeters have not yet been widely accepted in routine clinical radiotherapy environments. One reason for this is the laborious and complex manufacturing process involved, due in large part to the procedure for eliminating oxygen from the gel system.

Further development in polymer gel dosimetry occurred when Fong et al [32] described a method for developing normoxic gels. This was a significant advancement for polymer gel dosimetry as normoxic gels can be made under normal atmospheric conditions, which does away with the time-consuming process of oxygen elimination during the manufacture process of polymer gels. This meant that polymer gels could be manufactured on the bench top. Perhaps the most widely known normoxic formulation is MAGIC gel [32]. The principle behind the MAGIC gel is in the ascorbic acid oxygen scavenger, commonly known as vitamin C. Ascorbic acid binds free oxygen contained within the aqueous gelatine matrix into metallo-organic complexes and this process is initiated by copper sulphate. The normoxic gel has been reported to be tissue equivalent and has been shown to offer superior dose sensitivity and dynamic range using MRI evaluation when compared with other polymer gel formulations.

De Deene et al [33] have investigated the role of the different chemical constituents of the MAGIC gel. The sensitivity of MAGIC gel to ionising radiation was optimised by maximising the amount of oxygen scavenger and monomer required, while minimising the amount of overall ingredients. Further studies were performed on various oxygen scavengers and it was found that tetrakis (hydroxymethyl) phosphonium chloride (known hence forth just as tetrakis) was effective at scavenging oxygen. The rate of effective oxygen scavenging was found to be dependent upon its concentration, so that a greater concentration of tetrakis corresponded with a decrease in sensitivity to ionising radiation. However, due the increased effectiveness of the tetrakis a reduced concentration was required compared to other oxygen scavengers. Tetrakis was used to replace ascorbic acid and copper sulphate in the MAGIC formulation and a new formulation consisting of methacrylic acid, gelatine and tetrakis, named MAGAT was proposed [33]. Other normoxic gel formulations were also developed such as MAGAS (methacrylic acid gelatine and ascorbic acid), and PAGAS (polyacrylamide gel and ascorbic acid) gels [34]. This work was critical in further developing the potential of the normoxic dosimeter and due to its considerable advantages over current gel dosimeters it was proposed that this required further development. Following the development of normoxic polymer gels, a number of studies were performed on their characterisation. Studies included the ultrasonic properties of MAGIC gel [35], manufacturing methodology studies for MAGAS gel [36] and depth dose studies for MAGAS and MAGAT gels. The radiological water equivalence of MAGIC, MAGAS and MAGAT gels was investigated and studies were also performed on the R2 Dose response optimisation and stability of MAGAT gel. Further radiological attenuation properties of MAGAT and PAGAT gels were also investigated. From these studies, it became apparent that problems exist with the stability of some of
the normoxic gels when using oxygen scavengers. It was also clear that normoxic polymer gels responded to diagnostic doses in the same way as radiotherapy doses.

A relatively new class of polymer dosimeter has been proposed to overcome the radiochromic diffusion problem, the dosimeter being known as PRESAGETM (Heuris Pharma, Skillman, NJ, USA) [6, 12, 37]. PRESAGETM is a solid dosimeter which is based on clear polyurethane doped with leuco-dye leucomalachite green. The components of the dosimeter include an alkyl diisocyanate prepolymer, a hydroxyl reactive polyol along with a catalyst, which polymerises into optically clear polyurethane. The leuco dyes have a maximum absorbance at a wavelength of 633 nm and therefore is compatible with a He-Ne laser-based optical scanning system. It has been suggested that PRESAGETM has a number of potential advantages over both conventional polymer and Fricke gels. PRESAGE™ is a robust solid that does not require a container and therefore makes optical matching much easier. Irradiated regions exhibit negligible diffusion of the coloured medium and provided the sample is kept away from light while scanning, it appears highly stable. Initial studies of PRESAGETM have been reported to show very promising results. Further development and optimization techniques have since been published for radiotherapy, brachytherapy and microbeam radiotherapy [38-42]. It is clear that PRESAGETM has advantages over normoxic polymer gels in that they are better suited for evaluation optically and has demonstrated stability. PRESAGE™ dosimeters has a nominal elemental composition of C (63.27%), H (9.1%), N (4.91%), O (21.24%), Cl (2.7%) and Br (0.92%) with an empirical formula of C421H722N28O106Cl6Br giving an electron density of \( Z_{\text{eff}} \) = 9.33 and a mass density of 1.08 gcm\(^{-3}\) measured by manufacturer using gravimetric analysis [12]. The polymeric matrix is formed in two steps, the first step is prepolymer processes and the second step is integration with the leuco dye, a free radical initiator and catalyst. The PRESAGE™ dosimeter is in the form of cylinders with various sizes i.e. (i) diameter 60 mm and height 60 mm, (ii) 22 mm diameter and 60 mm height, (iii) 18 mm diameter and 60 mm height and (iv) 9.7 mm diameter and 60 mm height (Figure 1) [40, 43].

![Figure 1](image.png)

**Figure 1.** PRESAGETM of different diameter (a) 60 mm, (b) 22 mm, 18 mm and 9.7 mm [40, 43].

### 3. Review of 3D Optical Computed Tomography Imaging System

Optical Computed Tomography (OCT) is a bench top 3D imaging system for translucent object. The system is more accessible compared to a magnetic resonance imaging (MRI) system. This technique reconstructed 3D dose distributions in a radiochromic dosimeters such as polymer gel or PRESAGE™ [44] using back projection of the 2D images. The optical properties of the polymer gel and PRESAGE™ changes when irradiated. It becomes opaque and develops colour changes. This is due to the increasing in optical density in response to the absorbed dose delivered to the PRESAGE™ dosimeter.

Gore et al., described the first-generation OCT system in 1996. In this setup, a laser beam was stepped across a sample by mechanically moving two synchronous mirrors and the resulting beam is measured using and a photodiode detector. The first-generation OCT suffered from slow scanning speed due to the mechanical movement of the mirrors and only one path through the sample is measured at a time. A single slice with 128 x
128 pixels were acquired in about 12 minutes. The rapid movement of the laser beam requires more complex optics that could lead to artefacts or distorted images, which needs more post-processing. In terms of image quality, the laser scanners are provided signal-to-noise ratio is superior due to the high intensity of the laser beam. This system shown to be insensitive to geometrical distortion although refractive and scattering effects have been observed. The OCT was eventually developed into the first commercially available optical CT system, the Octopus scanner (MGS Research Inc., Madison, CT, USA), with a reported 30-minute scan time.

The second-generation OCT uses an incoherent broad light source and an image sensor such as Coupled Charged Detector (CCD) (Mark Oldham, 2006). The projection data for the optical CT technique acquired by passing a broad light beam through sample and collected by an image sensor. The pixelated detectors are based on CCD to obtain a whole 2D projection instantly at each rotation step, which render a significant speed advantage. The speed and resolution of this type of scanner is superior, but the sensitivity to refractive index inhomogeneities and stray photons from the dosimeter and the walls surrounding it can be a problem when the absorbed dose is reconstructed into a 3D map. The reason for this is that the entire projection is formed from one single exposure of the entire dosimeter [7].

CCD-based optical CT as a viable alternative to MRI for readout of 3D radiation dosimeters. Development of clinically applicable optical CT scanners has been moving in two different sub-classes with different geometries; the cone-beam and the parallel beam [7]. A commercially available cone-beam OCT scanner is the Vista (Modus Medical Devices Inc., London, ON, Canada) with a reported scan time of 5 minutes. Due to the geometry of the cone-beam CT, various scatter correction methods have been applied to counter the scatter issues in pixelated area detectors [45]. Cone beam OCT has requires correction from stray light artefacts.

At the University of Surrey, a group of researchers has developed an OCT based on the parallel beam geometry using tele-centric optics to create beams that travel parallel through the sample and are focused on the detector [7]. The parallel beam OCT has greatly reduced stray light artefacts due to rejection of light ray that deviate. Several studies have been performed using this system with PRESAGE™ dosimetry [7, 8, 38-40, 42, 43, 46-50]. Figure 2 shows some of reconstruction images of 9.7 mm diameter size of PRESAGE™ using CCD based OCT for characterisation study using Microbeam Radiation Therapy (MRT) facilities at ESRF, Grenoble. A subsection. The paragraph text follows on from the subsection heading but should not be in italic.

4. Application of CMOS Image Sensor Technology In Optical Computed Tomography (OCT) Imaging System

The CMOS image sensor (CIS) is a solid state imager manufactured using the standard CMOS process. The CIS is relatively new technology compared to the CCD. Chronologically, CMOS imager was first introduced by Fossum in 1993 [51], more than 20 years after Smith and Boyle’s invention of the CCD, which over the past, has long matured, produces superior image quality and is widely used in both commercial and scientific application. The CIS has evolved since its first introduction as a Passive Pixel Sensor (PPS) into a competitive Active Pixel Sensor (APS) imaging technology due to technology scaling of CMOS and the feasibility of high numbers of pixels integrated on-chip [52]. The major breakthrough for the CMOS APS is the ability to integrate sensing with signal processing at pixel level. CISs allow row and column addressing for random pixel access and on-chip fast multiple region-of-interest read-out, which allows the user to trade resolution, or array size, with read-out speed. Some CMOS sensors are integrated with on-pixel intelligence such as in-pixel analogue-to-digital converters. These features were recognised as advantages over the mature CCD technology despite the read noise in CMOS being higher than CCD [53].

As the technology has developed, APS has progressed in quality and can now achieve superior performance comparable to CCDs with lower noise and higher spatial resolution. Furthermore, the fast frame rate achievable with the CIS gives an extra edge to the sensor compared to the slow frame rate of CCD. CIS has now gained increasing popularity as a competitive imaging technology to CCD and is now been adapted in high end consumer products (e.g. mobile imaging devices, i.e. mobile phones and digital cameras) as well as scientific applications.
Figure 2. Reconstruction image of 9.7 mm diameter size of PRESAGETM using CCD-based OCT. (a) Characterise the dose linear response, (b) characterise the spatial resolution of the OCT system and (c) visualise 3D dose deposition using microbeam radiation therapy (MRT) treatment [40, 43].

5. Description Of CMOS-Based Optical Computed Tomography (Oct) Imaging System Developed

In this study, a CMOS based OCT system was developed for imaging radiochromic dosimeter in 3D. A preliminary characterisation of the system was performed to study the performance of the image sensor and other components to acquire projection images of the 3D object, such as the field of view, the sensor frame rate and the speed of the stepper motor. Figure 3 displays the schematic diagram for the optical CT that is placed in a dark room.

The imaging system consists of a 10 Megapixels CMOS image sensor coupled with a lens, model M3520-MPV (Computar, NC, USA). The CMOS image sensor was housed inside a Basler ace camera enclosure, model acA3800-14um (Basler, Ahrensburg, Germany). The CMOS image sensor is a monochrome sensor with a resolution of 3840 by 2748 pixels (ON Semiconductor, Phoenix, AZ). The size of the sensor array is 6.4 mm × 6.4 mm. Each pixel is 1.67 µm × 1.67 µm and is readout using rolling shutter mechanism. The temperature of the camera housing is monitored by thermometer and was kept under 50°C. The sensor is illuminated with a 10 cm × 10 cm large array LED of 633 nm wavelength. The LED light is controlled by light intensity controller (LC-15-4CH-A1, TMS Lite, Penang, Malaysia) and connected to a 12V DC supply.
Figure 3. The schematic diagram of the in-house developed OCT (a) large array LED, (b) stepper motor, (c) CMOS image sensor, (d) dosimeter holder of 60 mm diameter, and (e) matching liquid tank.

A GUI was also developed using LabVIEW to capture and display the projection images of the dosimeter image. Using the GUI, the region of interest (ROI) and the frame rate can be controlled. The ROI of the sensor can be determined by adjusting the height and width of the pixels and x-, y-coordinates. The maximum ROI is 3840 x 2748 pixels and the minimum ROI that the sensor can changed is 64 x 64 pixels. With full resolution, the frame rate of the sensor is 8.6 fps and started to increase until 28.5 fps by decreasing the ROI. As shown in Figure 4, the complete GUI for displaying, capturing and saving the image were developed and the image is saved as RAW file and can be analysed by MATLAB. The ROI can be changed and the frame rate can be monitored as listed in Table 1.

Table 1. The relationship between the region of interest and the frame rate. The coordinates are chosen from the centre of the sensor.

| Frame rate (fps) | Width (x) | Height (y) | Offset (x) | Offset (y) |
|------------------|-----------|------------|------------|------------|
| 8.6              | 3840      | 2748       | 0          | 0          |
| 10.8             | 2000      | 2000       | 920        | 374        |
| 13.0             | 1500      | 1500       | 1170       | 624        |
| 16.4             | 1000      | 1000       | 1420       | 874        |
| 18.3             | 800       | 800        | 1520       | 974        |
| 22.1             | 500       | 500        | 1670       | 1124       |
| 25.6             | 300       | 300        | 1770       | 1224       |
| 28.5             | 100       | 100        | 1870       | 1324       |
| 28.5             | 80        | 80         | 1880       | 1334       |

A rotary stage for the dosimeter is positioned between the camera and the LED. The rotary stage for the dosimeter consists of a stepper motor and a stepper motor controller (National Instrument, Austin, TX, USA). The stepper motor controller was connected to a 12 V power supply and a motion interface and was controlled from the PC via peripheral component interconnect (PCI) card. The diameter of dosimeter PRESAGETM is 60 mm (fixed diameter for this holder) and was placed inside a water tank (130 x 120 x 95) mm, made of acrylic. The stepper motor can hold a maximum weight of 4.4 kg which is sufficient for a 3D dosimeter. The motor rotates the dosimeter while the CMOS sensor captures the image for every step angle specified in the GUI. The minimum step angle of the motor is 1.8°, which means 200 steps per revolution for 360° rotation. The components of the rotary stage are integrated with the imaging system and controlled using LabVIEW (National Instrument, Austin, TX, USA). The step angle, time delay, velocity and acceleration of the motor can be controlled in the GUI as shown in Figure 4(b).
Figure 4. a) GUI for the CMOS image sensor, b) GUI for the stepper motor

| Working Distance (mm) | Height of CMOS (mm) | Horizontal View (mm) | Vertical View (mm) | Distance from the Ground (mm) |
|----------------------|---------------------|----------------------|--------------------|-------------------------------|
| 600                  | 35                  | 107                  | 77                 | 20                            |
| 650                  | 35                  | 115                  | 83                 | 12                            |
| 700                  | 35                  | 125                  | 90                 | 16                            |
| 750                  | 50                  | 135                  | 97                 | 39                            |
| 800                  | 50                  | 145                  | 103                | 30                            |
| 850                  | 50                  | 153                  | 110                | 37                            |
| 900                  | 50                  | 160                  | 115                | 36                            |
| 950                  | 50                  | 171                  | 122                | 30                            |
| 1000                 | 50                  | 181                  | 128                | 27                            |
| 1050                 | 58                  | 191                  | 135                | 11                            |

Table 2. The FOV is measured with different working distance and height of the CMOS sensor.

The sensor captures the images at every 1.8° of step angle, and requires 200 images for complete one revolution, 360°. For every rotation of the dosimeter, the sensor will capture the image and save it as RAW File. Later the image will be reconstructed and analysed using MATLAB. To obtain the right working distance, several distance and height of the sensor were tested. The FOV of the sensor at certain working distance was measured by using FOV chart and is shown in Table 2. The optimum distance between the imaging system and the rotary stage is 700 mm, where the field of view (FOV) can covers the whole dosimeter. The development of algorithm for image reconstruction and combination between sensor and motor GUI are of interest for further studies.

6. Conclusion
In this paper, an in-house CMOS base OCT has been developed and the setup has been described. The system can capture image at higher frame rate using ROI compared to existing CCD based OCT. Further experiments are needed to characterise the image quality of the system and application of the system for dosimeter readout. Future advances in CMOS image sensor would enable even faster scanning system with on-pixel image processing.
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**Acknowledgments**

We would like to acknowledge University of Surrey for their CCD based OCT facility. We would like to thank Universiti Sains Malaysia for their help to expand our study for the new development of CMOS based OCT (Short Term Grant 304.CIPPT.6312148). Thanks to Universiti Teknologi MARA to provide the financial support under BESTARI Grant 600-IRMI/MyRA 5/3/BESTARI (030/2017).