International Journal of Engineering Science Invention (IJESI)
ISSN (Online): 2319 – 6734, ISSN (Print): 2319 – 6726
www.ijesi.org ||Volume 8 Issue 03 Series. III || March 2019 || PP 53-58

A kV CBCT Tool for Adaptive Radiotherapy

Sudhir Kumar[1], Rajiv Singh[2]
[1]*(Goldfield Institute of Medical Sciences, Chhainsa, Ballabgarh, Faridabad, Haryana 121004)*
Corresponding Author: Sudhir Kumar

Abstract :*Megavoltage cone-pillar CT MV CBCTis utilized for three-dimensional imaging of the patient life structures on the treatment table preceding or soon after radiotherapy treatment. To utilize MV CBCT pictures for radiotherapy portion imaging purposes, solid electron thickness ED distributions are required. Understanding scatter, beam hardening, softening effects impacts bring about measuring antiquities in MV CBCT pictures and manipulates the CT number to ED transformation.A method based on transmissionimages is presented to correct for these effects without using prior knowledge of the object'sgeometry. The scatter distribution originating from the patient is calculated with pencil beam scatterkernels that are fitted based on transmission measurements. The radiological thickness is extractedfrom the scatter subtracted transmission images and is then converted to the primary transmissionused in the cone-beam reconstruction. These corrections are performed in an iterative manner,without using prior knowledge regarding the geometry and composition of the object. The methodwas tested using various homogeneous and inhomogeneous phantoms with varying shapes andcompositions, including a phantom with different electron density inserts, phantoms with largedensity variations, and an anthropomorphic head phantom. For all phantoms, the cupping artifactwas substantially removed from the images and a linear relation between the CT number andelectron density was found. After correction the deviations in reconstructed ED from the true valueswere reduced from up to 0.18 ED units to 0.02 for the majority of the phantoms; the residualdifference is equal to the amount of noise in the images. The ED distributions were evaluated interms of absolute dose calculation accuracy for homogeneous cylinders of different size; errorsdecreased from 6.5% to below 1% in the center of the objects for the uncorrected and correctedimages, respectively, and maximum differences were reduced from 14% to 1.9%, respectively. Thepresented method corrects the MV CBCT images for cupping artifacts and extracts reliable EDinformation of objects with varying geometries and composition, making these corrected MVCBCT images suitable for accurate dose calculation purposes.*

Keywords *-CBCT, IVDT, electron density, radiotherapy, IMRT*

Date of Submission: 08-03-2019 Date of acceptance: 28-03-2019

I. INTRODUCTION

Accurate positioning of the patient and verification of the treatment delivery have become more important, since a small misalignment of the patient setup can have large consequences on the effectiveness of the treatment. Online imaging prior to treatment is currently used for geometric verification of the patient position, but these images can also be used for a verification of the dose delivered in the actual patient anatomy at the time of treatment. To perform this dosimetric verification, several methods are proposed in the literature to reconstruct the 3D patient dose distribution based on portal images and an electron density ED map of the patient (1).

CBCT can be divided into two categories depending on the energy of the photons, kV CBCT and MV CBCT. To use CBCT for dose calculation purposes, the scanner must, preferably, be attached to the treatment unit to avoid differences in patient position between the CBCT acquisition and treatment. Whereas for kV CBCT an additional kV source and detector are needed, MV CBCT can be performed with the linear accelerator linac in combination with an electronic portal imaging device EPID sharing the same isocenter at the linac as the actual treatment. A disadvantage is that the contrast in the MV CBCT images is lower compared to that in the kV CBCT images. In order to use MV CBCT images for 3D dose calculations, the images must be calibrated to electron density ED. This is not straightforward. The Feldkamp cone-beam algorithm,14 used to reconstruct the 2D portal images to a 3D image, assumes a linear relation between log attenuation and object electron density with thickness, which is valid in the case of a monoenergetic MV photon beam and no scatter radiation. However, in MV CBCT this linear relation is disturbed by three factors. First, the scatter radiation originating from the object contributes to the signal in the portal images. Second, the energetic spectrum of the treatment beam results in beam hardening as the beam travels through the object and a shift in the beam spectrum toward lower energies for offaxis locations, referred to as beam softening. Third, the response of the EPID to radiation depends on the energy of the photons incident on the detector. If the portal images are not corrected before

CBCT reconstruction, the MV CBCT images will experience cupping artifacts that can lead to errors in dose calculation (2). A number of different methods have been proposed to reduce the cupping artifacts in MV CBCT images in order to obtain reliable electron densities. Chen et al. (3) derived a spatially dependent cupping correction function with the MV CBCT of a large water cylinder. However, details about the exact method they used are not provided. In a more detailed paper by Morin et al. (4) of the same group, the cupping artifact of a 16 cm diameter water cylinder was characterized using multiple ellipsoid shapes of different sizes centered around the same point in space. The ellipsoid shapes are associated with different correction factors. The CT numbers of the voxels that are positioned on the surface of one ellipsoid are multiplied with the same correction factor to account for the cupping artifact. The remaining cupping artifact of the MV CBCT of a head-and-neck patient was smaller than 5%. In order to correct cupping artifacts of MV CBCTs of different body parts centered on different off-axis locations, multiple phantoms need to be scanned and different sets of ellipsoid shapes need to be acquired. Spies et al. (5) used Monte Carlo-based scatter kernels to remove the scatter from the MV CB portal images and applied a linear quadratic model accounting for pixel-to-pixel variations in sensitivity and beam hardening. A CCD camerabased EPID was used with a total dose of 636 cGy. The overall cupping artifact reduced from 30% to 8%, although discrepancies for cortical bone were still large, up to 17%. Dose calculations using kV CB CT scans have also been described.18–20 Yoo et al. (6) recommended mounting a bowtie filter to improve image quality, yielding dosimetric results equal to those based on CT plans. Other groups reported on correction of cupping artifacts mainly to improve image quality. Jarry et al. (7) corrected the cupping artifact for a kV CBCT bench-top system by using Monte Carlo generated scatter distributions to take into account the contribution of object scatter radiation in the portal images. For kV cone-beam data acquired with a micro-CT scanner, Kachelriess et al. (8) developed a cupping artifact correction method that passed raw projection image data through a single pixel-independent polynomial before CBCT reconstruction. Their results showed cupping-free images after correction, but these authors did not report on the feasibility of their method to correct cupping artifacts of objects with dimensions comparable to human patients. The method and scanner were used for imaging purposes and no efforts were made in quantifying the conversion from CT number to electron density, which is needed in order to use the images for dose calculations. Correction procedures designed for kV cone-beamCT scans are not guaranteed to be accurate for MV CBCT scans, because MV photon beams have completely different scatter properties compared to the kV photon beams. In addition, the attenuation of MV beams is a function of the electron density, whereas for kV photon beams the attenuation also depends on the atomic composition (9). The goal of the present study is to use the MV CBCT system to obtain accurate electron density distributions that can be used for dose calculations. For this purpose, an automated pre-reconstruction cupping correction method is developed that corrects the cupping artifacts in the MV CBCT images and converts the CT numbers to electron density distributions. Various homogeneous and inhomogeneous phantoms are used to experimentally verify the presented cupping correction algorithm as per existing recommendations (9). We compared our method with previously described methods by Kachelriess et al. (8) and Spies et al., (5) and differences and similarities in these approaches are discussed. The MV CBCT scans were acquired using a linear accelerator equipped with a commercially available low-dose CBCT acquisition mode and an amorphous silicon EPID that is used in clinical routine for patient imaging.

The planning CT scan should preferably not be used to provide the ED map, because differences in patient position and anatomy other than rigidbody transformations can occur between the time of the planning CT scan and treatment fractions, due to, for example, organ motion and weight loss (9). Online anatomical information of the patient on the treatment table is required. A first method was accomplished using helical tomotherapy with on-board imaging capabilities. Kapatoes et al. (8) reconstructed the 3D dose distribution with tomotherapy treatment based on transmission measurements and a CT image of the patient in treatment position. A second option is to use a CT scanner in the treatment suite that shares the same couch as the treatment unit. In this approach kV CBCT scans used for adaptive planning are acquired with an on-board imager and CBCT images are used for daily setup registration (9).This approach is superior as it helps to obtain online anatomical information of the patient on the treatment table by combininga large number of portal images, measured from different angles around the patient. If we hybridize this clinical implementation with 3D image reconstruction as suggested by Sorensen et al (10) and Wong et al (11) then we get a 3D working artifact of complex anatomy.

II. MATERIALS AND METHODS

2.1 HARDWARE AND OPERATION

A Siemens Oncor medical linear accelerator with MV CBCT acquisition mode is used in combination with an OptiVue 1000 ST amorphous silicon EPID Siemens Medical Solutions, Concord, CA. The EPID has an active area of 4141 cm^2 with 1024 x 1024 pixels and a pixel resolutionof 0.4 mm. The Siemens COHERENCE THERAPIST WORKSPACE acquisition software was used for the measurement of conebeam portal images. To acquire a MV CBCT scan of an object, the object is irradiated with the 6 MV treatment beam in a low-dose-rate mode with a field size in the isocenter plane of 27.4 x 27.4 cm^2. The total dose at the EPID can be variedfrom 2 to 60 MUs. The source-to-detector distance SDD is fixed at 145 cm. During MV CBCT acquisition the gantry rotates continuously from 270° to 110° clockwise, acquiring 200 portal images with increments of 1°. The portal images are automatically corrected for individual pixel sensitivity, variation in intensity along the beam profile, the energydependent response of the EPID to the nonattenuated beam, dead pixels, and dark current. Dead pixels and dark current are corrected by the system interactively. The correction for individual pixel sensitivities, the variations in intensity along the beam profile, and the dependence of the response of the EPID on the energetic spectrum of the beam is performed in a single step using flood field images, also referred to as gain images.The reconstruction of the 2D portal images to a 3D volume is performed using a filtered back-projection method based on the Feldkamp algorithm implemented in a research version of the cone-beam reconstruction software provided by Siemens. A geometric calibration phantom is usedto determine the relation between the position of the voxels of the 3D image and the position of the pixels of the EPID for each gantry angle. The reconstructed volume has dimensions of 25.6 x 25.6 x 25.6 cm3 and contains voxels of 111 mm^3. To limit the amount of noise in the MV CBCT images, the majority of the scans is performed using the maximum dose of 60 MU. To illustrate the feasibility of cupping correction also with low-dose MV CBCT, additional MV CBCT images are made with a total dose of 8 MUs.

2.2 THE CLINICAL CUPPING CORRECTION MODEL

The cupping correction model subtracts the scatter radiation originating from the object, corrects for beam hardening and softening effects, and the energy dependent response of the EPID. The correction model consists of five steps. First, the raw portal images are filtered with a Perona-Malik diffusion filter and corrected for differences in accelerator output, which is measured with the monitor chamber for each projection. The beam intensity profile is then restored in the portal images of the object. Second, the scatter contribution is predicted and subtracted from the portal images to yield the primary signal. Third, the primary signal is converted to the radiological thickness i.e., water-equivalent thickness or path length using a mapping function that inherently takes into account beam hardening and softening effects and the energy-dependent response of the EPID. With the radiological thickness a new estimate of the scatter is calculated. In step 4, steps 2 and 3 are repeated in an iterative manner until convergence of the radiological thickness is achieved. In the fifth step, the final estimate of the radiological thickness is used to calculate the corresponding primary transmission of a monoenergetic beam, which is needed for cone-beam reconstruction. Regions of interest were contoured at the center of each of the phantom plugs and the mean HU values within the contours were noted. The electron density of each phantom plug was noted from specs and physical density corresponding to mean CT values was recorded and analyzed as the CT to density curve (9). This served as the baseline for iterative corrections on IVDT. The resulting 200 primary transmission images are used as input for the conebeam reconstruction algorithm, and the reconstructed CT numbers are converted to ED. The next sections describe these five steps in detail, whereas in Sec. 2.3 the derivation of the model parameters is described.For 3D set reconstruction we used approach of cupping iterative algorithm like Soresen et al (10) and Wang et al (11), to yield a 3D image.

2.3 ITERATIVE CORRECTION ALGORITHM

The portal images are corrected by the system automatically for the nonflat beam intensity profile by multiplication with a gain image. The first step of the correction model is to restore the beam intensity profile BP(x,y)in the measured portal images of the object and of the open beam, which is needed for accurate scatter calculations. The analysis of CT to density curve generated from Yadav et al (9)can be described mathematically by

$$M(x,y,\theta) = I(x,y,\theta) .BP(x,y) \quad (1)$$

To calculate the thickness map, the primary transmission must be known, which depends on the scatter contribution as shown in equation (2). Also, the transmission can be derived from this rule as shown in equation (3).

$$TP(x,y,\theta, tx,y,\theta) = TP(x, y, \theta) - TS(x, y, \theta, t\theta) \quad (2)$$

However, the scatter contribution depends on its turn on the thickness map. Therefore, the thicknessmap is

$$T_P(x,y,\theta,t_{x,y,\theta}) = f(t_{x,y,\theta},(x,y)) = \exp(-\sum_{m=1}^{M} \alpha_m(x,y)\cdot t^m_{x,y,\theta}). \quad (3)$$

extracted from the portal images in an iterative manner according to the following steps:

Step (i): Restore the beam profile in the filtered portalimages with and without object in the beam that arecorrected for differences in accelerator output using Equation (1).

Step (ii): Calculate the transmission using Equation (2).

Step (iii): Calculate the scatter component with the nth estimate of the thickness,

$$T_S(x,y,\theta,t_\theta^{(n)}) = \int\int_{x',y'\in\text{field}} \frac{M_{\text{open}}(x',y',\theta)}{M_{\text{open}}(x,y,\theta)} K(r_{(x-x',y-y')},t^{(n)}_{x',y',\theta})dx'dy' \quad (4)$$

Step iv: Calculate the primary transmission for this thickness estimate,

$$T_P(x,y,\theta,t^{(n)}_{x,y,\theta}) = T(x,y,\theta) - T_S(x,y,\theta,t_\theta^{(n)}). \quad (5)$$

Step v: Calculate a new estimate of the thickness map based on the primary transmission,

$$t^{(n+1)}_{x,y,\theta} = f^{-1}(T_P(x,y,\theta,t^{(n)}_{x,y,\theta}),\ (x,y)). \quad (6)$$

This approach is iterated over and over until the convergence criteria for equation (1) are met. In practice this correction algorithm is applied to approximately 350 portal images of the entity under scan by the reconstruction software developed using python.

2.4 CALIBRATION OF CT NUMBERS TO ELECTRON DENSITY

After cone-beam reconstruction the CT numbers are calibrated to electron density. For this purpose, a single linear fit is used that converts the reconstructed gray-scale values CT numbersinto electron density values (9).

2.5 VERIFICATION OF THE MODEL

In order to verify the accuracy of our model in terms of ED calibration and dose calculations, MV CBCT images are made for a large number of phantoms and reconstructed ED distributions are used for dose calculations. The next sectionsdescribe the phantoms that were used and the dose calculations that were performed for this verification.

2.6 HOMOGENEOUS PHANTOMS

First, two PMMA polymethyl methacrylatecylinders filled with water were used. The first cylinder has a radius R of 10 cm, a length of 30 cm, and a wall thickness of 5 mm. The second cylinder has a radius of 5 cm, a length of 30 cm,and a wall thickness of 3 mm. The cylinders are referred to as the large and small cylinder, respectively. An IMRT headand- neck phantom CIRS, Norfolk, VAwas used. This phantom is cylindrically shaped with a radius of 8 cm andcomposed of a material with water-equivalent ED. The cylinder contains five cavities that can accommodate rod inserts. For the homogeneous IMRT phantom, water equivalent inserts were used. All cylinders are positioned on the treatment table with the central axis coinciding with the axis of rotation of the linac y-axis. The z-axis is the vertical axis and the x-axis the cross-plane axis.

III. RESULTS

3.1 THE CUPPING CORRECTION MODEL

The pencil beam scatter kernels are fitted with high accuracy; the maximum difference between the fitted and measured transmission was smaller than 0.3%. With increasing phantom thickness, the kernels are more sharply peaked, reflected by the decreased width c2tof the kernels, while the on-axis absolute scatter is increased, indicated by the parameters. The maximum difference between the on- and off-axis locations in terms of the primary transmission can be up to 40% for large phantom thicknesses. The on-axis parameters were 0.051 cm^{-1} off-axis values range from 0.050 to 0.070 cm−1, 1.1710−4 cm^{-2} off-axis range −1.010to +1.510cm^{-2}, and −5.43.

The influence of the scatter correction and of the conversion of the transmission to radiological thickness on the final solution was analyzed. The effect of the scatter correction on the ED is similar to the effect of applying a constant offset to the ED in the entire phantom, which results in a mean ED that is equivalent to the real ED but the cupping artifact is still present. The ED without the scatter correction shows hardly any cupping but a systematic lower ED of approximately 0.25. The influence of the number of iterations N was also analyzed. The iterative cupping correction model rapidly converges after 2–3 iterations to the final solution. This is shown in terms of the ED for a fixed number of iterations varying from 0 to 3. Increasing the number of iterations to as much as 20 does not affect the results. For N3 the mean and the standard deviation SDof the differences in ED in the entire phantom between two subsequent iterations are below 0.001 ED units.

3.1 HOMOGENEOUS PHANTOMS - MODEL VERIFICATION

The correction model is applied to the portal images of a PMMA cylinder filled with water with a radius of 10 cm. The window levels are set to the mean ED2 standard deviations SD. This allows a visual inspection of the homogeneity of the reconstructed images. If artifacts are removed the window level is determined solely by the level of noise. The cupping artifact is clearly visible in the uncorrected images the ED is 32% smaller in the center than near the wall of the cylinder, but when the cupping correction method is applied the variations in ED due to the cupping are smaller than the level of noise. The results expressed as a mean 1 SD of the ED in the entire phantom are 0.860.09 if the images are not corrected and 1.00x 0.03 if the correction model is applied. Due to the smaller dimensions of this phantom the cupping is less pronounced than in the larger cylinder, but it is still clearly visible.

3.2 INHOMOGENEOUS PHANTOMS

The accuracy of the cupping correction method was also assessed using five inhomogeneous objects. For the first object, Phantom A. The ED of the cork 0.19 and polystyrene 0.97was derived from the kV planning CTscan. Due to the cupping artifact, the ED is decreased in the polystyrene layer in the middle and in the cork regions, but this was corrected by the cupping correctionmethod. The bottom row shows absolute difference images between the reconstructed ED in the MV CBCT scans and the real ED. These images confirm that the cupping artifact was corrected using the method proposed here. The second inhomogeneous phantom Phantom B consisted of slabs of polystyrene and cork and a cylindrical phantom R=5 cm filled with water. The ED profiles and the absolute difference images show that theED was accurately reconstructed after the cupping correction method was applied. The third inhomogeneous phantom PhantomCwas a water cylinder with a radius of 10 cm that contained cylindrical inserts of varying ED. If uncorrected, the ED is underestimated up to 0.30 ED units in the water region and issystematically more than 0.19 ED units too low in the inserts with ED values corresponding to muscle, inner bone, and cortical bone. If the cupping correction method is applied,the ED deviates at a maximum 0.04 ED units from the real ED in water and in all inserts.

3.3 DOSE CALCULATIONS

Dose calculations are performed with the ED distributions of the water cylinders with radii 5 and 10 cm and with the IMRT phantom with water-equivalent inserts. Applying the correction model results in a decrease of the maximum difference from 17.6% to 0.6% compared to the reference dose distribution. The corresponding dose difference at the center of the cylinder reduces from7.1% to 0.4%. For the IMRT phantom the maximum and mean difference reduce from 5.3% and 1.0% to 2.0% and 0.9%, respectively.equal to the level of noise, compared to a maximum of 0.09 if the portal images are not corrected. In three of the inserts the standard deviation 1 SDin ED is 0.06, but this is merely caused by the increased level of noise in the small inserts rather than by residual cupping. The mean ED couldbe reconstructed for all phantoms with an error smaller than 0.04 ED compared to the real ED, except for the bone insert in the IMRT phantom ED=1.51 and for the lung insert ED=0.36 in Phantom C. Because the ED in the cork regions ED=0.19in Phantoms A and B could be reconstructed with a high accuracy, it is expected that the overestimation of the ED in the lung insert is not due to its low ED. It is more probable that its small size, the large gradients in ED near the wall of the insert, and its location in the center of the cupping where the ED is 0.3 units too low in the surrounding water are of importance(12). The CT numbers in the bone-equivalent region in the IMRT phantom were 7.2% too low, yielding an underestimation in ED of 0.14 units. Taking into account the underestimation in CT number of 4.4% in the entire phantom, the deviation in mean CT number in the bone insert is only 3.2% too low, which is comparable with the other inserts.

IV. CONCLUSION

In this study a model is proposed capable of correcting cupping artifacts and converting CT numbers to electron density for low-dose MV CBCT images for a large range of objects with varying shapes and compositions without using prior knowledge of the geometry.

CBCT based imaging is a preferable option for the anatomical delineation of soft tissues due to its superiority in comparison to MVCT based imaging. From the perspective of the adaptive planning, this system improves the overall workflow with a quick review and analysis of the dynamic requirements (9).An advantage of the present model compared to other approaches is that noMonte Carlo simulations are needed to describe the scatter kernels, and that it can be easily implemented. If cupping artifacts are not corrected, errors in dose calculations up to 17% can occur. Correcting the portal images with the present model results in cupping-free images that are converted to electron density distributions and used for accurate dose calculations with errors in dose smaller than 2%.Thus, kV CBCT has potential to act as a valuable tool for adaptive radiotherapy and significantly helps in avoiding the excessive patient scanning which may lead to cumulative high doses in patients (9).

REFERENCES

[1]. R. J. W. Louwe, E. M. F. Damen, M. van Herk, A. W. H. Minken, O.Torzsok, and B. J. Mijnheer, "Three-dimensional dose reconstruction of breast cancer treatment using portal imaging," Med. Phys. 30, 2376–2389 (2003).

[2]. O. Morin, A. Gillis, J. Chen, M. Aubin, M. K. Bucci, M. Roach III, and J. Pouliot, "Megavoltage cone-beam CT: System description and clinical applications," Med. Dosim. 31, 51–61 (2006).

[3]. J. Chen, O. Morin, M. Aubin, M.K. Bucci, C.F. Chuang, and J. Pouliot, "Dose-guided radiation therapy with megavoltage cone-beam CT," Br. J. Radiol. 79, Spec. No. 1, S87–S98 (2006).

[4]. O. Morin, J. Chen, M. Aubin, A. Gillis, J. F. Aubry, S. Bose, H. Chen, M. Descovich, P. Xia, and J. Pouliot, "Dose calculation using megavoltage cone-beam CT," Int. J. Radiat. Oncol. Biol. Phys. 67, 1201–1210 (2007).

[5]. L. Spies, M. Ebert, B. A. Groh, B. M. Hesse, and T. Bortfeld, "Correction of scatter in megavoltage cone-beam CT," Phys. Med. Biol. 46, 821–833 (2001).

[6]. S. Yoo and F. F. Yin, "Dosimetric feasibility of cone-beam CT-based treatment planning compared to CT-based treatment planning," Int. J. Radiat. Oncol. Biol. Phys. 66, 1553–1561 (2006).

[7]. G. Jarry, S. A. Graham, D. J. Moseley, D. J. Jaffray, J. H. Siewerdsen, and F. Verhaegen, "Characterization of scattered radiation in kV CBCT images using Monte Carlo simulations," Med. Phys. 33, 4320–4329 (2006).

[8]. J. M. Kapatoes, G. H. Olivera, K. J. Ruchala, J. B. Smilowitz, P. J. Reckwerdt, and T. R. Mackie, "A feasible method for clinical delivery verification and dose reconstruction in tomotherapy," Med. Phys. 28, 528–542 (2001).

[9]. Yadav, Poonam, VelayudhamRamasubramanian, and BhudattPaliwal. "Feasibility study on effect and stability of adaptive radiotherapy on kilovoltage cone beam CT." Radiology and oncology 45 (3): 220-226 (2011).

[10]. S. P. Sorensen, P. E. Chow, S. Kriminski, P. M. Medin, and T. D. Solberg, "Image-guided radiotherapy using a mobile kilovoltage x-ray device," Med. Dosim. 31, 40–50 (2006).

[11]. J. R. Wong, L. Grimm, M. Uematsu, R. Oren, C. W. Cheng, S. Merrick, and P. Schiff, "Image-guided radiotherapy for prostate cancer by CTlinear accelerator combination: Prostate movements and dosimetric considerations," Int. J. Radiat. Oncol. Biol. Phys. 61, 561–569 (2005).

[12]. S. Midgley, R. M. Millar, and J. Dudson, "A feasibility study for megavoltage cone beam CT using a commercial EPID," Phys. Med. Biol. 43, 155–169 (1998).

Sudhir Kumar" A kV CBCT Tool for Adaptive Radiotherapy" International Journal of Engineering Science Invention (IJESI), Vol. 08, No. 03, 2019, PP 53-58