Exploration of medical applications of electrical capacitance tomography

A thesis submitted to The University of Manchester for the degree of Doctor of Philosophy
in the Faculty of Engineering and Physical Science

2015

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Abstract

With the advantages of low cost, high imaging speed, non-intrusive and non-invasive, electrical capacitance tomography (ECT) becomes one of the most maturely developed industrial tomography modalities. However, ECT had not been considered for medical applications before this work. This thesis is to explore medical applications of ECT, especially for root canal treatment (RCT) and revision total hip replacement (THR).

A dental ECT system based on impedance analyser has been designed for RCT for two purposes: (1) to visualise the tooth surface in real time and (2) to determine the position of an endodontic file. To adapt the limited space in oral cavity, a miniature two-plate dental ECT sensor with either $2 \times 2$ or $2 \times 3$ array has been designed and fabricated. The sensor has a similar performance to the conventional ECT sensor and can provide good image quality. By registering and fusing with the radiograph based on a Major-axis method, a real-time image with high resolution can be obtained. A piecewise linear function has been used to locate the axial position of the apex of an endodontic file. The results show that high accuracy can be achieved near the ‘End Point’ as it is one of the reference points, determined by the sudden change in capacitance when a grounded metallic file touches the root apex (conductive media or solution).

For revision THR, a conventional 8-electrode single plane sensor has been used, generating real-time 3D images of a metallic rod using a model based method. By this method, the 3D image reconstruction is simplified to estimate the cross-sectional and axial portions of the rod in the sensing area and to draw an image of the rod with prior knowledge. A high accuracy can be achieved with the maximum absolute error of 0.13 cm in estimating cross-sectional position using a weighted mean method and 0.4 cm in estimating axial position by the linear function based on the relative change in capacitance between file and electrodes. A preliminary experiment has been carried out to generate an electrical impedance tomography (EIT) image of a metallic object in conductive solution with high permittivity. Using the impedance analyser based system, the EIT image can be obtained with a conventional ECT sensor and the result is promising, providing the possibility of obtained a real-time EIT 3D image of a milling/drilling tool during revision THR.
Declaration

No portion of the work referred to in this thesis has been submitted in support of an application for another degree or qualification of this or any other university or other institution of learning.
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Acknowledgements

First of all, I would like to express my sincere gratitude to my supervisor, Prof Wuqiang Yang, for his invaluable guidance and continued encouragement throughout my PhD study. I would like to thank him for giving me the opportunity to attend several interesting conferences and introducing me to wonderful people in the field of electrical capacitance tomography. I also would like to thank my co-supervisor, Prof Hugh Devlin and my advisor Prof Keith Horner, for their support and advices in medical field.

I would like to gratefully acknowledge the wonderful people at The University of Manchester, especially those in the SISP group. I would like to thank Dr Jiangtao Sun, Dr Paul Wright and Mimi Faisyalini Ramli for their interesting and technical discussion. I would like to express my special thanks to the technicians in workshops for assisting the construction of the system, and thank the staff of Joule library at the University of Manchester for offering their support in providing the reading materials during my PhD study. I am grateful to my friends for their help and care during my stay in the UK.

I would like to thank UMIP for collaboration and financial support, and thank Lucid Innovation Ltd. for designing and manufacturing the electro-mechanical rig.

Finally, I am forever indebted to my family for their love and support throughout my life. Also, I would like to thank my fiancée, Yin Wang, for his love, support and understanding.
Publications


Acronyms and Symbols

**Acronyms:**

<table>
<thead>
<tr>
<th>Acronym</th>
<th>Description</th>
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<tbody>
<tr>
<td>AHP</td>
<td>Aquasil Heavy-Polysiloxane</td>
</tr>
<tr>
<td>BEM</td>
<td>Boundary Element Method</td>
</tr>
<tr>
<td>BRF</td>
<td>Band Rejection Filter</td>
</tr>
<tr>
<td>CBCT</td>
<td>Cone-bean Computed tomography</td>
</tr>
<tr>
<td>CC</td>
<td>Correlation Coefficient</td>
</tr>
<tr>
<td>CDJ</td>
<td>Cementodentinal Junction</td>
</tr>
<tr>
<td>CEJ</td>
<td>Cementoenamel Junction</td>
</tr>
<tr>
<td>DAC</td>
<td>Digital-to-analogue Converter</td>
</tr>
<tr>
<td>DDS</td>
<td>Direct-digital-synthesiser</td>
</tr>
<tr>
<td>DUT</td>
<td>Device Under Test</td>
</tr>
<tr>
<td>ECT</td>
<td>Electrical Capacitance Tomography</td>
</tr>
<tr>
<td>EIT</td>
<td>Electrical Impedance Tomography</td>
</tr>
<tr>
<td>EMT</td>
<td>Electromagnetic Tomography</td>
</tr>
<tr>
<td>ERT</td>
<td>Electrical Resistance Tomography</td>
</tr>
<tr>
<td>ET</td>
<td>Electrical Tomography</td>
</tr>
<tr>
<td>FDM</td>
<td>Finite Difference Method</td>
</tr>
<tr>
<td>FEM</td>
<td>Finite Element Method</td>
</tr>
<tr>
<td>FPCB</td>
<td>Flexible Printed Circuit Board</td>
</tr>
<tr>
<td>GPIB</td>
<td>General Purpose Interface Bus</td>
</tr>
<tr>
<td>GUI</td>
<td>Graphical User Interface</td>
</tr>
<tr>
<td>LBP</td>
<td>Linear Back Projection</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
</tr>
<tr>
<td>NI</td>
<td>National Instrument</td>
</tr>
<tr>
<td>PET</td>
<td>Positron Emission Tomography</td>
</tr>
<tr>
<td>PGA</td>
<td>Programmable-gain Amplifier</td>
</tr>
<tr>
<td>RCT</td>
<td>Root Canal Treatment</td>
</tr>
<tr>
<td>ROI</td>
<td>Region of Interest</td>
</tr>
<tr>
<td>SCPI</td>
<td>Standard Commands for Programmable Instruments</td>
</tr>
<tr>
<td>SNR</td>
<td>Signal-to-noise Ratio</td>
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THR  Total Hip Replacement
TV  Total Variation
VISA  Virtual Instrument Software Architecture
X-CT  X-ray Computed Tomography

**Symbols:**

- **A**: overlapping area
- **B**: susceptance
- **C**: capacitance
- **C_{ab}**: capacitance between \(a\) and \(b\)
- **C_{Ei}**: measured capacitance between file and \(i^{th}\) electrode
- **C_{Eiopp}**: measured capacitance between \(i^{th}\) opposing electrode pair
- **C_{f}**: feedback capacitance
- **C_{H}**: capacitance when the sensor filled with the high permittivity materials
- **C_{i,j}(HIGH)**: capacitance between \(i^{th}\) and \(j^{th}\) electrodes when the sensing area contains the high permittivity material only
- **C_{i,j}(LOW)**: capacitance between \(i^{th}\) and \(j^{th}\) electrodes when entire sensing area contains the low permittivity material only
- **C_{i,j}(n)**: capacitance between the \(i^{th}\) and \(j^{th}\) electrodes when pixel \(n\) contains the high permittivity material and all other pixel with low permittivity material
- **C_{L}**: capacitance when the sensor filled with low permittivity materials
- **C_{M}**: measured capacitance
- **\overline{C_{M}}**: mean of measured capacitance \(C_{M}\)
- **C_{s}**: standing capacitance
- **D**: dissipation factor
- **E_{i}(x,y)**: potential distribution when an excitation voltage \(V_i\) is applied to electrode \(i\) while other electrodes remain grounded
- **Est\_P**: estimated position (axial)
- **G**: conductance
- **G_{cali}**: calibrated value (3D positioning)
- **G_{mea}**: measured value from the 3D simulation model
$G_{ref}$ reference value from 2D simulation model  
$H_{CUR}$ High current terminal (impedance analyser)  
$H_{POT}$ High potential terminal (impedance analyser)  
$I_{DUT}$ current in the DUT (impedance analyser)  
$Image$ grey scale of the target image  
$\overline{Image}$ mean of grey scale of $Image$  
$K$ factor associated with the geometry of the pair of electrodes forming the capacitor  
$K_g$ gain of the differential amplifier  
$L$ length of cylinder (cylinder and plane)  
$L_e$ depth of the electrode  
$L_g$ width of the gap  
$L_{CUR}$ Low current terminal (impedance analyser)  
$L_e$ depth of the electrode  
$L_g$ width of the gap  
$L_{POT}$ Low potential terminal (impedance analyser)  
$L_{root}$ length of tooth root  
$N_f$ number of frames  
$N_p$ number of pixels  
$P$ projection operator  
$Q$ quality factor  
$R$ resistant  
$Reg\_image$ grey scale of registered image  
$\overline{Reg\_image}$ mean of grey scale of $Reg\_image$  
$R_f$ feedback resistance  
$S$ sensitivity distribution  
$S^{-1}$ inverse of sensitivity distribution  
$S^T$ transpose of sensitivity distribution  
$S_{i,j}(x,y)$ sensitivity distribution between electrodes $i$ and $j$  
$V$ potential difference between two electrodes  
$V_{DUT}$ voltage drop across the DUT  
$V_i$ excitation voltage  
$X$ reactance
<table>
<thead>
<tr>
<th>Symbol</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>$Y$</td>
<td>admittance</td>
</tr>
<tr>
<td>$</td>
<td>Y</td>
</tr>
<tr>
<td>$Z$</td>
<td>complex impedance</td>
</tr>
<tr>
<td>$</td>
<td>Z</td>
</tr>
<tr>
<td>$d$</td>
<td>distance between cylinder and plane</td>
</tr>
<tr>
<td>$d_x$</td>
<td>height of insert object</td>
</tr>
<tr>
<td>$e_1$, $e_2$</td>
<td>offsets of op-amp 1 and 2</td>
</tr>
<tr>
<td>$e_k$</td>
<td>error vector at $k^{th}$ iteration</td>
</tr>
<tr>
<td>$f$</td>
<td>frequency</td>
</tr>
<tr>
<td>$g$</td>
<td>normalised permittivity vector</td>
</tr>
<tr>
<td>$\hat{g}$</td>
<td>estimated permittivity vector</td>
</tr>
<tr>
<td>$\bar{g}$</td>
<td>mean of true permittivity vector $g$</td>
</tr>
<tr>
<td>$g^T$</td>
<td>transpose of normalised permittivity vector</td>
</tr>
<tr>
<td>$\tilde{g}$</td>
<td>mean of estimated permittivity vector $\hat{g}$</td>
</tr>
<tr>
<td>$\hat{g}_k$</td>
<td>estimated permittivity vector at $k^{th}$ iteration</td>
</tr>
<tr>
<td>$l$</td>
<td>length of electrode</td>
</tr>
<tr>
<td>$l_x$</td>
<td>length of the object in sensing area</td>
</tr>
<tr>
<td>$n$</td>
<td>number of electrodes</td>
</tr>
<tr>
<td>$n \cdot j$</td>
<td>normal component of the current density</td>
</tr>
<tr>
<td>$p(x,y)$</td>
<td>area of the pixel at $(x, y)$</td>
</tr>
<tr>
<td>$r$</td>
<td>radius</td>
</tr>
<tr>
<td>$w$</td>
<td>width</td>
</tr>
<tr>
<td>$w_{(i,j)}$</td>
<td>weight or grey level in the pixel $(x_i, y_i)$</td>
</tr>
<tr>
<td>$\alpha_k$</td>
<td>step length at $k^{th}$ iteration</td>
</tr>
<tr>
<td>$\varepsilon$</td>
<td>permittivity</td>
</tr>
<tr>
<td>$\varepsilon_0$</td>
<td>permittivity of vacuum</td>
</tr>
<tr>
<td>$\varepsilon_{air}$</td>
<td>permittivity of air</td>
</tr>
<tr>
<td>$\varepsilon_r$</td>
<td>relative permittivity</td>
</tr>
<tr>
<td>$\varepsilon(x,y)$</td>
<td>permittivity distribution</td>
</tr>
<tr>
<td>$(\bar{x}, \bar{y})$</td>
<td>estimated position of the cross-sectional centre (3D positioning)</td>
</tr>
<tr>
<td>$\theta$</td>
<td>phase angle</td>
</tr>
<tr>
<td>$j$</td>
<td>current density</td>
</tr>
<tr>
<td>$\lambda$</td>
<td>normalised capacitance</td>
</tr>
</tbody>
</table>
$\lambda^T$  transpose of normalised capacitance
$u_\lambda$  identity vector
$\omega$  angular frequency
$\Phi(x,y)$  potential distribution
$\Gamma$  electrode surface
$\nabla$  gradient operator
Chapter 1: Introduction

This thesis is to explore medical applications of electrical capacitance tomography (ECT), especially for endodontic therapy and revision total hip replacement (THR). This chapter aims to introduce the research background, by briefly describing relevant human anatomy, and providing current techniques and the need of the new imaging techniques for these treatments. The aims and objectives of this research are presented in a later section of this chapter and the structure and organisation of the thesis is given in the final section.

1.1 Electrical capacitance tomography (ECT)

ECT has been developed for past 3 decades and is one of the most mature tomography techniques. While a lot of effort has been made on research into ECT for industrial applications, in particular for the oil and gas industry (Xie et al., 1992; Yang, 2010; Yang and Liu, 1999), ECT has not been considered for medical applications. The advantages of ECT over other tomography techniques are non-radiation, non-intrusive and non-invasive, high speed and low cost (Yang, 2010). Considering the fast imaging speed of ECT (typically 100 frames/second) (Liu et al., 2004), it is possible to use ECT for real-time imaging during the treatment. Information from an X-ray or X-CT can be fused with the images obtained by ECT, to improve the accuracy. The fundamental of ECT, including the hardware design and image reconstruction methods, will be introduced in the later chapter. ECT for medical imaging presents potential challenges but would contribute to clinical medicine. The new system will be not only used for treatment, but also contribute to physician training.

1.2 Human physiology

1.2.1 Dental anatomy

Human oral cavity is a complicated structure that consists of teeth, tissues, nerves, blood vessels and bone. A tooth includes two main parts: crown and root. For a developed permanent tooth, the crown of a tooth is defined as the area above the cementoenamel junction (CEJ) and is covered in enamel, as shown in Figure 1.1. The majority of the crown is composed of dentin with a pulp chamber in the centre. The root of a tooth is anatomically designated by the area below CEJ and is attached to the
surrounding gingival tissue and alveolar bone by fibrous attachment. Clinically, the tooth root is covered with cementum. As a result, the root is part of a tooth not visible in the oral cavity. Dentin is the majority constituents of the root, which normally contains single or multiple pulp canals. Canines and most premolars, except for maxillary first premolars, normally have one root, while maxillary first premolars and mandibular molars usually have two roots and maxillary molar have three (Sarkar and Rao, 2002).

**Tooth enamel**

Tooth enamel together with dentin, cementum and pulp is one of the four major tissues, which compose a tooth in oral cavity. Anatomically, tooth enamel is supported by underlying dentin and it covers the crown, thus it is the only visible dental tissue of a tooth in most cases. As the hardest and most mineralised tissue in the human body, tooth enamel is composed of 96% inorganic material and 4% organic material and water (Reyes-Gasga et al., 1997). Enamel varies in thickness over the surface of the tooth, normally thickest at the cusp, approx. 2.5 mm and thinnest at its border with cementum at CEJ (Franz, 2012).

**Dentin**

Dentin is normally covered by enamel on the crown and cementum on the root and surrounds the entire pulp (root canal). The majority of dentin is the mineral hydroxyapatite, approx. 70%, the other are 20% of organic material and 10% of water (Miller, 2012). Unlike enamel, dentin continues to form throughout life and can be initiated in response to stimuli, such as tooth decay or attrition. Generally, dentin can be differentiated into three types by appearance and stage of development. Primary dentin is the outermost layer and forms most of the tooth. Secondary dentin forms more slowly than primary dentin and is deposited after the formation of root. Tertiary dentin is created by replacement odontoblasts in response to stimuli, e.g. carious attack, abrasion, erosion, trauma, moderate-rate dental caries, and some operative procedures (Murray et al., 2000; Franz, 2012).
**Dental pulp**

Dental pulp is the innermost part of the tooth, rich in soft tissue with blood vessels and nerves. Due to continuous deposition of dentin, the pulp becomes smaller with age. The pulp in the crown of the tooth is known as the coronal pulp. The other pulp extends from the cervical region of the crown to the root apex, so called radicular pulp. The pulps are not always straight but vary in shape, size and number. Maxillary central incisor has shovel-shaped coronal pulp with 3 short horns and normally triangular in cross section, while cuspid has the longest pulp with elliptical cross section.

Apical foramen is anatomically defined as the opening of the radicular pulp into the periapical tissue. The average size of apical foramen is 0.3 to 0.4 mm in diameter (Franz, 2012). The narrowest part of the canal called apical constriction is just short of the apical foramen, as shown in Figure 1.2. The other landmark region is called cementodental junction (CDJ), which is defined as the line between dentin and cementum where pulp ends and periodontal tissue starts (Alothmani et al., 2013). Many reports (Nekoofar et al., 2006; Gordon and Chandler, 2004) equate CDJ with the apical constriction, but they do not always coincide that the apical constriction is also located in dentin or less frequently in cementum (Alothmani et al., 2013). The apical constriction is commonly considered as the ideal termination for endodontic therapy, however, some researchers suggested that sometimes no apical constriction presents, particularly with apical pathosis and root resorption (Simon, 1994; Coolidge, 1929). Clinically, it is difficult to locate either apical foramen or constriction. Usually, the termination point of endodontic therapy is estimated to be 0.5 to 1 mm short of the radiographic apex (Wu et al., 2000). Multiple foramina may exist, separated by a portion of dentin and cementum or by cementum only. Most infections or inflammations spread through the apical foramen from the pulp to periapical tissue, i.e. pulpitis, causing extreme sensitivity and/or pain.

![Figure 1.2 Concept of the apex (Wu et al., 2000)]
1.2.2 Thigh anatomy

Thigh is anatomically defined as the area between the pelvis and the knees, which is part of the lower limb. From outermost to central region, the thigh consists of skin, fat, muscle, blood vessels and nerves, and bone. The hip is the biggest and most stable joint in the human body. It is a typically ball-and-socket joint, formed from the femoral head and a deep cavity, with raised bone margins-the acetabulum (see Figure 1.3).

![Figure 1.3 3D-CT scans of a normal hip: (a) with and (b) without the femoral head (Hartofilakidis et al. 2014)](image)

The femur head, that forms two-thirds of a sphere, is covered by articular cartilage, except of the fossa for the attachment of ligamentum teres. The hip joints are also surrounded by cartilage that is to help support the joints and prevent bones from rubbing against each other. Clinically, the hip has greater range of motion, with flexion 120-130°, extension 10-20°, abduction 20-30°, adduction 30-40°, external rotation 30-40°, and internal rotation 20-30° (Hartofilakidis et al., 2014). Functionally, the main purpose of the hip joints is to support the upper body when a person is standing, walking, and running, and to deal with certain movement, e.g. bending and stretching.

Femur

The human femur can be divided into three parts: upper femur, shaft and distal portion. The upper femur includes the head, neck and extends to the lower limit of the lesser

![Figure 1.4 Structure of femur](image)
trochanter. The shaft is anatomically defined as a portion of the femur between the lesser trochanter and the distal portion. The lower-most 6 inches of the femur, so called distal portion, is part of the bone, in which the shaft gradually increases in size to form the articular surface of the lower end of the femur (Koch, 1917). Figure 1.4 shows the structure of the femur. The shaft consists of a thick cylindrical shell of compact bone with the centre occupied by fragile trabeculae, which support the marrow in the cavity but not normally affect the strength of the femur.

1.3 The treatments

1.3.1 Endodontic therapy

A tooth is infected for many reasons, such as poor oral hygiene, having a diet high in carbohydrates, or having a weakened immune system. Once infection occurs, dentists and patients would face a quandary between securing the infected tooth by endodontic therapy, so called root canal treatment (RCT), and replacement with an implant. Normally, the patients would receive primary endodontic treatment before tooth extraction while nonsurgical retreatment is regarded as the first choice when primary treatment fails (Ruddle, 2004). Surgical treatment will possibly cause a new or recurrent infection (Caliskan, 2005). Elemam and Pretty (2011) compared the success rates of these treatments. They indicated that the overall success rate for primary endodontic, nonsurgical retreatment, and surgical treatment were 86%, 78%, and 63%, respectively, while implants was 91%. Although implant treatment shows a higher success rate, most patients desired to retain the tooth. Thus, conventional RCT, especially for primary endodontic treatment, would be still the first choice for treatment and implant treatment can often be an alternative method (Elemam and Pretty, 2011; Fabbro et al., 2007; Torabinejad et al., 2009).

RCT is to eliminate infection and protect a decontaminated tooth. Traditionally, a dentist first uses a radiograph of teeth, showing the inner structure, to determine the working length of a fitted endodontic instrument (i.e. a file, as shown in Figure 1.5) and to preliminarily assess the tooth root. Then the dentist drills into the centre of the tooth...
and cleans the infected tissue in root canal(s) with the file. However, the traditional RCT is often difficult to perform. One challenge is the invisible RCT process, causing problems, e.g. under-filling, flush-filling or overfilling, and resulting in treatment failures (Sundqvist et al., 1998). Particularly, overfilled teeth may result in necrosis of cementum, periodontal ligament and alveolar bone due to the toxic filling materials (Lin et al., 1992). Consequently, to determine the working length is essential for the success of the treatment. Currently, the working length can be measured by two ways. The traditional and still most reliable method is using radiographs. A file is placed in each canal and a radiograph is taken for determining how depth the files should go. The second way is to utilise an electronic apex locator, which is useful when the end of the root on the radiograph is not actually in the position where the apical foramen or constriction is.

1.3.2 Total hip replacement (THR) and revision total hip surgery

THR is a highly age-related surgery and normally carried out in adults aged between 60 and 80. Common reasons related to a damaged hip joint, including osteoarthritis, rheumatoid arthritis and hip fracture. In a conventional THR procedure, the damaged femoral head is removed and replaced with a metal stem placed into the hollow centre of the femur. The femoral stem can be cemented to fit into the bone. A metallic or ceramic ball is placed on the upper part of the stem. This ball replaces the damaged femoral head which has been removed. The damaged cartilage surface of the acetabulum is also removed and replaced with a metal shell. The shell is normally held in place by press-fit for an initial fixation and by pegs or screws, for additional attachment. A polyethylene, metal or ceramic liner is inserted between the new ball and the shell to allow a smooth gliding surface (Learmonth et al., 2007). However, many patients need a revision surgery, normally 10 years after replacement. A hip revision surgery is taken for three major purposes:

(1) To relieve pain in the affected hip,
(2) To restore the patient’s mobility,
(3) To remove a loose or damaged prosthesis before irreversible harm to the joint.

Clinically, the revision surgery is more difficult to perform than the primary one, with a lower successful rate and higher risk (Mahomed et al., 2003). Generally, 3 steps are performed in a revision THR:
(1) Removal of THR components, possibly causing damage to the bone,
(2) Replacing any bone loss,
(3) Placing a new prosthesis.

The difficulties in a revision THR are bone loss or the poor integrity of the remaining bone stock, related to the Step 1 when removing the former THR components (Sheth et al., 2013). According to a survey in 2009 (Springer et al., 2009), the survival rate of primary total hip replacement is over 90% at 15-year follow-up, but the overall survivorship of revision THR is 85% at 10 years and 72.6% at 15 years. The reasons of failure related to a revision THR are aseptic loosening, instability osteolysis/wear, infection and periprosthetic fracture, where aseptic loosening is accounted for the majority of failures (Springer et al., 2009; Mahomed et al., 2003).

To date, the only imaging technique for either THR or revision THR is conventional radiograph or 3D X-CT, normally only for diagnosis and treatment planning (Taylor et al., 1999; Sheth et al., 2013). However, facing the same problem with RCT, neither radiograph nor 3D X-CT can provide a real-time guidance for THR/revision THR. Interestingly, it has been noticed that the procedure in revision THR is highly similar to RCT, only different in drilling/milling tool and operation region of human body.

1.4 Current techniques

1.4.1 Medical imaging techniques

Various imaging techniques have been investigated and developed for medical applications, such as X-ray, X-ray Computed Tomography (X-CT), ultrasonography, Magnetic Resonance Imaging (MRI) and Positron Emission Tomography (PET). Those image techniques, with specific parameters and different principles, are mutually complementary and are used for different applications. For dental imaging, the associated imaging techniques include X-ray (Webber and Grondahl, 1982), X-CT, cone-bean computed tomography (CBCT) (Vandenberghe et al., 2010, Gahleitner et al., 2003) and ultrasonography (Culjat et al., 2003). For THR/revision THR, X-CT is mostly used for diagnosing and surgical planning (Taylor et al., 1999; Sheth et al., 2013).
X-ray

The history of medical imaging can be traced back to 1895, when the first X-ray was discovered and quickly used for medical imaging by a German physicist Wilhelm Conrad Rontgen. X-ray imaging uses a single kilo-voltage beam from an X-ray generator to pass through a target and the beam is attenuated by tissues. The resultant image can be produced where chemical reaction occurs on a radiographic film (Webber and Grondahl, 1982). In a human body, different density leads to a marked difference in attenuation. For example, bone causes significant attenuation, and soft tissue leads to less attenuation and air or lung has minimal attenuation. An X-ray image represents the information of attenuation as an image of material distribution. From X-ray images, it is easy to discriminate between the high attenuation e.g. with bone from the low attenuation e.g. tissue. However, from X-ray images it is difficult to differentiate different soft tissues. Traditionally, X-ray could only provide 2D images, which would cause a blurred image due to the 3D structure of the human body. Therefore, X-CT has been developed to improve the image quality and to produce 3D images.

Dental radiograph, as shown in Figure 1.6, plays the most significant role in an RCT process, essential in producing a surgical plan before treatment. The radiograph film type varies for different purposes. A standard X-ray film with an intensifying screen has limited use in dental imaging, because it is too big for most uses except for extra-oral views. Non-screen film, which provides good radiographic detail, is typically used for extra-oral views and limited in intra-oral views, depending on the size of film and the mouth. One specialised dental radiograph film is called dental film, which has six speeds, labelled from A to F. Only D, E, F are utilised in practice and E and F are the most preferred speed, due to the reduced radiation exposure to both operatives and patient (Syriopoulos et al., 2001; Farman and Farman, 2000; Geist and Brand, 2001; Woolhiser et al., 2005). The challenge of dental radiograph is the radioactivity and low imaging speed, producing an off line image only.

Figure 1.6 Pre-operative dental radiographs
**X-CT**

Different from the traditional X-ray imaging, X-CT uses multiple pencil beams of kilo-voltage photons to pass through a target from multiple angles. On the opposite side of the target, a dosimeter is used to measure the amount of received ionising radiation, which allows determination of the attenuation of individual beams as they pass through the target (Kalender, 2006). Each part of the target is a 3D pixel, called a ‘voxel’ with the information of width, height and depth. Each beam passes through a number of voxels as it traverses the target. The attenuation of the beam can be considered as the sum of attenuation in each voxel. A computer is then used for image reconstruction using the attenuation information from each beam, resulting in a high resolution 3D image of the target (Vandenberghe et al., 2010). Similar to traditional X-ray, X-CT has limitation when different soft tissues need to be differentiated. With multiple beams of kilo-voltage photons, a relatively high amount of ionising radiation would be produced. As a result, X-CT is never considered for real-time imaging.

Despite their limitation, traditional X-ray imaging and X-CT are still the most popular medical imaging techniques, because of high accuracy and reliability.

**CBCT**

CBCT is a relatively new technique, which has been developed for 10 years. It consists of a 2D detector and an X-CT generator where the X-rays are divergent, forming a cone beam (Vandenberghe et al., 2010, Bian et al., 2010). CBCT is mostly used for dental applications, particularly in treatment planning and diagnosis in implant dentistry. During a CBCT scan, the scanner rotates around the patient and hundreds of images are obtained and a 3D image reconstructed. CBCT provides an undistorted view of the 3 dimensions of the target that can be used to accurately visualise the teeth, root orientation and anomalous structures. However, there are some disadvantages of CBCT, such as the increased susceptibility to the lack of appropriate bone density determination. Beside, 3D imaging based on X-rays is radioactive, time consuming and also high treatment price.

**Ultrasonography**

Ultrasound is a sound wave with a frequency higher than the upper limit of human hearing. The wave is produced and detected by ultrasound devices, e.g. piezoelectric.
For some materials, the shape changes when an electric current is applied. On the contrary, a change in shape can provoke an electric current to form. An ultrasound transducer utilises a piece of piezoelectric material between two electrodes. When an oscillating current is applied, ultrasound waves can be generated by the rapid vibration of the piezoelectric material (Coatney, 2001).

Ultrasound waves are attenuated through absorption and scattering when they are passed through a target. Waves may be reflected or refracted passing through different media. With either reflection or scattering, the returned ultrasonic waves cause the piezoelectric material to vibrate, generating an electric signal, which is converted into an image. Ultrasound is a relatively safe imaging technique compared with X-CT. With its low cost and non-invasion nature, ultrasonography is widely used. However, ultrasound waves are unable to penetrate bone and are completely aborted by air. Thus it is limited to use ultrasound for evaluating either dental or thigh structures (Culjat et al., 2003, Lees and Barber, 1968). Table 1.1 is a comparison of medical imaging techniques.

Table 1.1 Comparison of medical imaging techniques

<table>
<thead>
<tr>
<th>Imaging Techniques</th>
<th>Image Resolution</th>
<th>Radiation</th>
<th>Limitation</th>
<th>Price</th>
</tr>
</thead>
<tbody>
<tr>
<td>X-ray imaging</td>
<td>Relatively low</td>
<td>Yes</td>
<td>2D only, less differentiation between soft tissue</td>
<td>Low</td>
</tr>
<tr>
<td>X-CT</td>
<td>High</td>
<td>Yes</td>
<td>less differentiation between soft tissue</td>
<td>Relatively low</td>
</tr>
<tr>
<td>CBCT</td>
<td>High</td>
<td>Yes</td>
<td>Lack of appropriate bone density determination</td>
<td>High</td>
</tr>
<tr>
<td>Ultrasonography</td>
<td>Relatively low</td>
<td>No</td>
<td>Not for bone or gas</td>
<td>Low</td>
</tr>
</tbody>
</table>

1.4.2 Other techniques

Apex locator

A new instrument to help determine the working length of RCT is the electronic apex locator. The principle of the apex locator is based on an experimental concept that at the apex point, a consistent electrical resistance value can be obtained between a root canal instrument in a pulp and an electrode applied to the oral mucous membrane (Nekoofar
et al., 2006, Gordon and Chandler, 2004). The apex locator is a device to find the apex location according to this consistent value.

Electronic apex locator has been explored since 1969. The first generation called Root Canal Meter was developed by Onuki Medical Co. in Japan (Gordon and Chandler, 2004). It uses a sine wave signal of 150 Hz to measure resistance. The first device suffered from a medical safety hazard due to the high excitation current. Then the current was reduced below 5 µA with the later version (Kobayashi, 1995). However, these devices were found unreliable compared with radiographs, with many of the readings being significantly different from the accepted working length (Gordon and Chandler, 2004).

The second generation uses impedance measurement (including resistance and capacitance) rather than only resistance to measure within the canal. The distance to apex in different canal conditions is measured with different frequencies. However, the second generation apex locators present incorrect readings, and could only be used in dry canals (Gordon and Chandler, 2004).

The third generation apex locators are similar to the second generation that they use multiple frequencies to determine the distance of the file apex from the end of the canal. Those devices explore microprocessors and can implement mathematical algorithms to achieve accurate readings. As an example, the Root ZX (J. Morita USA) can overcome the erroneous reading with electrolytes, introducing a ratio method and developing a self-calibrating Root ZX (Kobayashi and Suda, 1994, Ebrahim et al., 2007). This device works based on the principle that two electric currents with different frequency will have measurable impedance, which can be measured and compared as a ratio regardless of the type of electrolyte in the canal. A report from Kobayashi and Suda (Kobayashi and Suda, 1994) showed that the ratio of different frequencies has definitive value, and the ratio rate of variation did not change with different electrolytes in the canal. The change in capacitance at the apical constriction is the basis for the operation of the Root ZX. This device became the benchmark in development of apex locators, and gives a combined accuracy of 90% to within 0.5 mm of the apical foramen or the CDJ, depending on the reference point used (Pagavino et al., 1998). Practically, the Root ZX is the mostly commonly used apex locator to locate the root apex.
The fourth generation utilises two separate frequencies: 400 Hz and 8 kHz, similar to the third generation. The Bingo 1020 (Forum Engineering Technologies, Rishon Lezion, Israel) claimed to increase the measurement accuracy and the reliability based on the root mean square values of the combination of using only one frequency at a time (Apex Locator Bingo – ‘1020’, 1999). Some in-vitro studies (Kaufman et al., 2002; Tinaz et al., 2002) found that the Bingo 1020 performs as reliable as the Root ZX in-vitro and is easier for a beginner to use in RCT. Another apex locator to the market in 2003 is developed by the Element Diagnostic Unit and Apex Locator (SybronEndo, Anaheim, CA, USA). This device does not process the impedance information as a mathematical algorithm. Instead, it takes the resistance and capacitance measurements and compares them with a database to determine the distance to the apex of the root canal. It utilises a composite waveform of two signals, 500 Hz and 4 kHz. The signals are converted into an analogue signal via a digital-to-analogue converter (DAC). This resultant analogue signal is then amplified and goes to terminal circuit, which is assumed to be a resistor and capacitor in parallel. This circuit model is used to generate the database of the location (Nekoofar et al., 2006).

Electronic apex locators, based on electrical property measurement of the practical and anatomic termination point-CDJ, is still in development because no individual technique is truly satisfactory in determining the working length of an endodontic file. Those apex locating devices cannot actually locate the dental instrument but only estimate the apex location roughly.

1.5 Aims and objectives

Dental radiographs are the only reference for the dental use while for a revision THR surgery, the only reference is X-CT before treatment. Although, some electrical technologies (such as apex locator) have been involved in RCT to determine the working length before treatment, there is no reliable method for a visualised surgery and to navigate the surgery tool during RCT or revision THR. The conventional imaging techniques such as X-ray or X-CT, considering its radiation, are never considered to use in real-time, which causes a blind procedure during treatment. Therefore, RCT is the most painful dental procedure and revision THR is a lengthy operation (a minimum of 2-3 hours) and associated with a high risk. A real-time imaging and accurate positioning of a surgery tool, either endodontic file or drilling/milling tool, is demanded for an
efficient operation to reduce the risk of damage to the remaining tissues. A new concept is to use electrical capacitance tomography (ECT) to visualise tooth surfaces or anatomy of thigh and to position the surgery tool during treatment.

This thesis aims to explore the medical application of ECT and develop an imaging system based on capacitive arrays for particular medical applications. The research objectives are:

- Designing capacitive sensor arrays
- Designing imaging system to obtain capacitance measurement from sensors
- Investigating the possibility of using capacitive sensor to locate metallic objects
- Evaluating the performance of the system
- In-vitro experiment based on the fabricated ECT system

The novelty of this research includes

- Initial exploration of ECT for medical application
- Special design of sensor array for particular applications
- Exploring the possibility of ECT for detecting material with a frequency-dependend permittivity
- Image fusion between ECT and high resolution image, e.g. radiograph
- Detecting metallic object by capacitance measurement
- Positioning metallic object based on special sensor array

1.6 Thesis organisation

Chapter 2 reviews the fundamentals of ECT, including the principles of operation, the measurement hardware and the algorithm for image reconstruction.

Chapter 3 describes an ECT system based on an impedance analyser. The individual parts of the system are introduced, including the design of sensors, the impedance analyser and the multiplexer unit. The software design is also introduced in this chapter, including instrument control, image reconstruction and image processing.

Chapter 4 concentrates on the ECT system for the use in RCT. In this chapter, the design and performance of a dental capacitive sensor will be detailed. Simulation and in-vitro laboratory experiment are described to evaluate the system performance of a
single tooth. A laboratory test of detecting metallic object is also presented in this chapter. The principle of using ECT for metallic object detection is introduced and ideas of positioning a metallic file are explained in the later section of this chapter.

Chapter 5 investigates the possibility of using ECT for revision THR, which is similar to RCT in principle but different in sensor design and metallic rod positioning. The chapter will focus on the determination of the length of the metallic rod by ECT system.

Chapter 6 gives conclusions and future work.
Chapter 2: Fundamentals of electrical capacitance tomography (ECT)

2.1 Principle of ECT and overall ECT system

ECT, electrical resistance/impedance tomography (ERT/EIT) and electromagnetic tomography (EMT) are three branches of electrical tomography (ET). ECT has been developed for many years and becomes the most mature technique among various industrial tomography modalities. The basic concept of ECT is to measure variations in capacitance between electrode pairs and to generate a cross-sectional image, representing the permittivity distribution and therefore the material distribution (Yang and Peng, 2003). Unlike the conventional tomography techniques, e.g. X-CT, ECT exhibits its outstanding advantages of non-radioactive, non-intrusive and non-invasive, high imaging speed and low cost (Liu et al., 2004). A lot of effort has been made on research into ECT for many industrial applications, including oil pipelines (Xie et al., 1992, Yang et al., 1995, Li et al., 2013), wet gas separators (Yang et al., 2004), fluidised beds (Liu et al., 2002, 2005; Warsito and Fan, 2003, 2005), pneumatic conveyors (Yang and Liu, 2000) and gas/solid cyclones (Wang et al., 2006).

The basic formula giving the capacitance between two parallel electrodes is:

$$C = \frac{\varepsilon_0 \varepsilon_r A}{d}$$

(2.1)

where $\varepsilon_0$ and $\varepsilon_r$ are the permittivity of vacuum and relative permittivity of inner material, respectively, $A$ is overlapping area and $d$ is the distance between those two electrodes. If the electrodes are not exactly in parallel, Eq. 2.1 can be rewritten as:

$$C = \varepsilon_0 \varepsilon_r K$$

(2.2)

where $K$ is a factor associated with the geometry of the pair of electrodes forming the capacitor.

The measurement strategy with an ECT sensor can be described as follows. For an n-electrode ECT sensor, electrode 1 is selected as the source and the remaining n-1 electrodes for detection. Then electrode 2 is selected as the source and the other n-2 for detection. The sequence is repeated until capacitance is measured from all possible...
electrode pairs. Consequently, there are $\frac{n(n-1)}{2}$ independent capacitance measurements to be taken for each frame of image.

A typical ECT system consists of a capacitive sensor, a capacitance measuring unit and a host PC. Figure 2.1 shows an ECT system with an 8-electrode circular sensor. The function of the measuring unit is to measure the capacitances between electrode pairs via multiplexer switches so that the electrode pairs can be selected one by one. The host PC has functions of controlling the measuring unit, receiving the data from the measuring unit, data analysis, image reconstruction by an algorithm and showing the permittivity distribution of the inner space of the sensor.

![Figure 2.1 Typical ECT system with an 8-electrode circular sensor](image)

**2.2 ECT sensor**

For any imaging system, it is crucial to design a suitable sensor for a specific application. Yang (2010) provided an exhaustive introduction to the design of ECT sensors. An ECT sensor, as shown in Figure 2.2, normally includes following parts:

1. multiple electrodes,
2. an insulating frame,
3. outer shielding screens, to reduce external noise,
4. coaxial cables and connectors.

Some ECT sensors also include driven guard electrodes at both ends of the measurement electrodes, attempting to enable a design of shorter electrodes (Reinecke and Mewes, 1994). The number of electrodes is essential for an ECT sensor design, which determines the number of independent measurements that directly affects the
image resolution. In principle, a higher image resolution can be achieved with more independent measurements, thus more electrodes would be preferred for an ECT sensor. However, too many electrodes would result in (1) too small capacitance to be measured, (2) difficulty in solving the inverse problem for image reconstruction, (3) complicated and expensive hardware and (4) slower data acquisition rate (Yang, 2010). Peng et al. (2012) found that with the increase in the number of electrodes, the sensitivity matrices become more ill-conditioned and no further improvement in the image quality can be obtained if the number of electrodes is over 12. Consequently, the number of electrodes in an ECT sensor is typically 8 or 12.

A typical ECT sensor adopts a circular shape with electrode length between 5 and 10 cm (Yang, 2010). The electrode can be placed either internally or externally, depending on different demand. Generally, an external electrode is desired by industry, i.e. to mount the electrodes outside an insulating frame. With internal electrodes, although the media can be measured directly without the effect from the insulating frame, the ECT sensor is no longer non-invasive. The diameter of an ECT sensor can be down to 6 mm (Yang, 2010) or up to 1 m (Isaksen et al., 1994; Wang and Yang, 2011). Considering that the relative image resolution of a large ECT sensor is similar to a small ECT sensor, the absolute image resolution of a large ECT sensor is worse than that of a small ECT sensor. Consequently, the diameter of an ECT sensor is usually between 2.5 cm and 10 cm.

For some specific applications, e.g. circulating fluidised beds or industrial boilers, square ECT sensors are needed. The first square ECT sensor with 12 electrodes was reported by Yang and Liu (1999). The experimental results indicated an improved image quality from the square sensor, compared with a circular sensor, because the square sensor trends to show a higher sensitivity in the central region, which is the weakest area in the entire sensing space. Liu et al. (2000) also presented a square ECT sensor with 12 electrodes outside and 4 electrodes inside. The sensor with internal

![Figure 2.2 Cross-sectional view of a typical ECT sensor with 12 electrodes (Yang, 2010)](image-url)
electrodes can effectively enhance the sensitivity in the central region, producing improved images compared with the square sensor without internal electrodes. However, this sensor with internal electrodes may disrupt the flow thus has limited use. As the author’s previous work, a 16-electrode parallel ECT sensor was fabricated for a feasibility study (Ren and Yang, 2012). This sensor has 8 electrodes on each side and a fixed distance (30 mm) between opposite electrodes, providing incomplete measurement data. The sensor was used to produce images of three plastic bars by means of an ECT system based on an impedance analyser.

Some non-conventional ECT sensors are in conical shape, e.g. for imaging a cyclone or a fluidised bed dryer (Wang et al., 2006). Due to the special geometry, the shape of electrodes for a conical ECT sensor is no longer rectangular but trapezium. In terms of measured capacitances and sensitivity distribution, there is no apparent difference between the conical sensor and conventional circular sensor if the length of electrodes is larger than the largest diameter of cross section. Short electrodes may cause 3D effects to the conical shape, resulting in a more complicated design of sensor than the conventional one.

Although the existing ECT sensors provide good results to show the inner material distribution, the sensor with enclosed structure limits the geometry of the sensing area and cannot be used for medical applications, which are mostly associated with an irregular geometry, such as jaw anatomy in dental imaging. To date, little work has been done on ECT sensors with an open sensing area. For dental imaging in this thesis, it is essential to design an open structure ECT sensor to adapt to the special geometry of the jaw in oral cavity.

Currently, many efforts have been made on design of a 3D ECT sensor to produce a 3D image (Warsito and Fan, 2001, 2003, 2005; Marashdeh et al., 2008; Soleimani et al., 2007; Li and Holland, 2013). Unlike the conventional ECT sensor with single plane of electrodes, the 3D sensor normally involves multiple planes of electrodes, producing a 3D image by either interpolating independent 2D images from different planes or directly reconstructing an image based on a 3D sensitivity map, neither of which can provide real-time 3D images due to the huge number of measurements and complex computation caused by the complicated 3D sensor structure.
2.3 Hardware system

For an ECT system, it is crucial to measure variances of capacitance efficiently and precisely, which is a primary determinant for high quality images. However, a practical ECT system may be affected by three main sources of stray capacitance: (1) the screened cable, (2) the switches, used to select electrode mode (excitation, detection or grounded) and (3) the outer shielding screen. Additionally, the movement of cable and the change of ambient temperature may vary the stray capacitance. Thus, a stray-immune capacitance measuring circuit is desired for an ECT system (Yang, 1996a). To date, the commonly used capacitance measurement circuit for ECT system includes (1) charge/discharge circuit, (2) AC-based circuit, and (3) impedance analyser based system, where AC-based circuit is most suitable for the use in a practical ECT system.

2.3.1 Charge/discharge ECT system

The charge/discharge capacitance measuring circuit is shown Figure 2.3, where the stray capacitances are labelled as $C_{s1}$ and $C_{s2}$. The circuit works by repeatedly altering the status of CMOS switches S1 to S4 to charge and discharge the unknown capacitance $C_x$, forming a square wave and resulting in two phases: the charge phase and the discharge phase. During the charge phase, S1 and S4 are closed while S2 and S3 are open. The charge flows from the voltage source $V_c$ through the $C_x$, converted to voltage by the charge op-amp 1. The switch S2 and S3 are closed while S1 and S4 are open during the discharge phase. The charged measured capacitance $C_x$ starts to discharge and the right side of $C_x$ draws current from op-amp 2. A differential amplifier is to sum those signals and output the voltage as a function of the measured capacitance:

$$V_3 = K_g(V_2 - V_1) = 2K_g f V_c C_x R_f + K_g (e_2 - e_1)$$  \hspace{1cm} (2.3)

where $K_g$ is the gain of the differential amplifier, $f$ is frequency, $R_f$ is feedback resistance, $e_1$ and $e_2$ are the offsets of op-amp 1 and 2, respectively. Note that the offset problem from the op-amps can be relived as $e_1$ and $e_2$ can neutralise each other to some extent.
Neither $C_{s1}$ nor $C_{s2}$ would affect the capacitance measurement, because (1) $C_{s1}$ does not produce current flowing through $C_x$ as it is driven by $V_c$ directly or connected to the ground alternatively, and (2) $C_{s2}$ is kept at virtual earth by the two op-amps. Therefore, the charge/discharge circuit is stray-immune (Huang et al., 1988, 1992). The charge/discharge circuit is simple and low cost, but it suffers from charge injection problem by the switches, and drift problem because of using DC amplifiers (Yang, 1996a, 1996b).

### 2.3.2 AC-based ECT system

An AC-based circuit is most suitable for an ECT system as the circuit is less sensitive to stray capacitance in principle (Yang, 1996a). With its high signal-to-noise ratio (SNR), low drift and wide bandwidth, an AC-based ECT system is most commonly used in many applications (Johansen et al., 1996; Mewes and Reinecke, 1996; Yang and York, 1999).

Figure 2.4 shows a simplified block diagram of the AC-based capacitance measurement circuit, where direct-digital-synthesiser (DDS) is used to generate a sine wave signal as the excitation source.

The excitation voltage $V_i$ is applied to $C_x$, producing an AC input current. An operational amplifier with a feedback capacitance $C_f$ and a feedback resistance $R_f$ converts into an AC voltage $V_0$, which can be formulated as
Figure 2.4 AC-based capacitance measuring circuit (Yang, 1996a)

\[ V_0 = -\frac{j\omega C_x R_f}{j\omega C_f R_f + 1} V_i \]  

\[ (2.4) \]

where \( \omega \) is the angular frequency of excitation voltage.

When the selected feedback capacitance and resistance satisfies \( \frac{1}{\omega C_f} \ll R_f \), the capacitance becomes dominant, Eq. 2.4 can be simplified as

\[ V_0 = -\frac{C_x}{C_f} V_i \]  

\[ (2.5) \]

The output AC signal can be amplified further via an AC programmable-gain amplifier (PGA), which is constructed from operational amplifiers and CMOS switches, providing two or more gains, e.g. 10 and 100, to adapt to a large range of capacitance values, ensuring that the demodulated AC signal is in the order of volts, otherwise, the weak demodulated signal may result in serious noise and errors produced by the following circuit (Yang and York, 1999). The output from the AC PGA is still an AC signal and is demodulated by a demodulator, e.g. phase sensitive demodulator (PSD). The demodulated signal normally contains two components: DC and AC signals. To eliminate the AC signal, a low-pass filter with a low cut-off frequency, e.g. 5 kHz, is used. The filtered DC signal \( V_d \) is used to represent the measured capacitance.

As seen in Figure 2.4, two stray capacitances, \( C_{s1} \) and \( C_{s2} \), are raised between the measurement electrode and the earth. \( C_{s1} \) is directly driven by the excitation voltage, which has no effect on the voltage applied to the measuring capacitance. Also, the invert input of the operational amplifier is held at virtual earth by the feedback and the potential difference across \( C_{s2} \) is near zero which does not influence the capacitance measurement. Hence, the AC-based capacitance measurement circuit has less sensitivity.
to the stray capacitance, providing good results for capacitance measurement and is widely used for ECT applications (Yang, 1996a; Yang and York, 1999).

2.3.3 Impedance analyser based ECT system

While most ECT systems employ either charge/discharge or AC-based circuit as the capacitance measuring circuits, ECT systems based on an impedance analyser are not common. Impedance analysers are designed to measure electrical properties, such as capacitance, resistance and conductance (Agilent, 2000). Because of its high resolution, high accuracy and good reliability, the impedance analyser has excellent performance on electrical property measurement and has been used as capacitance measuring unit for some ECT applications (Chrondronasios, 2003; Hu et al., 2008; Hu, 2009).

The main difficulty with the impedance analyser based ECT system is the limited measurement channel and the coupling capacitance if external switches are added. To extend measurement channels of the impedance analyser, an impedance analyser based ECT system usually includes a multiplexer used as switches between the measurement terminals of the impedance analyser and the electrodes of an ECT sensor. Each electrode can be controlled independently and connected to either the High terminal or the Low terminal of the impedance analyser, or remain floating. Although an ECT system based on an impedance analyser is limited in acquisition rate and measuring channel (Hu et al., 2008), the system is ideal for developing a laboratory prototype because of the high accuracy and resolution. As the impedance analyser can provide multiple electrical property measurement, the impedance analyser based system can be used for both ECT and ERT/EIT. In this thesis, an ECT system based on impedance analyser HP4192A is utilised, providing a high accuracy and resolution in capturing small change in capacitance, which is one of the challenges in ECT for medical imaging, particularly in dental imaging. The detail of the system, including each hardware component and their software control, will be introduced in Chapter 3.

2.4 Image reconstruction algorithms

Two basic problems need to be solved for an ECT system. One is the so-called “forward problem”, which is to calculate capacitance of electrode pairs with a known sensor structure and a known permittivity distribution. The other problem is the “inverse
problem”, in which the permittivity distribution is determined from the measured capacitance.

2.4.1 Forward problem

There are three purposes of solving the forward problem: (1) to evaluate an ECT sensor design, (2) to obtain a sensitivity matrix and (3) for iteration. A sensitivity matrix is calculated (or measured) either from capacitances between electrode-pairs or directly from electric fields (Process Tomography Ltd., 2009). Generally, the second method is preferred as it is easy to implement by simulation and provides promising results.

In the first method, the inter-electrode capacitance is measured when each pixel contains the high permittivity material in turn with all other pixels hold the low permittivity material. The sensitivity coefficient at the $n^{th}$ pixel between the electrode pair $i$ and $j$ can be calculated by:

$$ S_{i,j}(n) = \frac{C_{i,j}(n) - C_{i,j}(LOW)}{C_{i,j}(HIGH) - C_{i,j}(LOW)} $$

where $C_{i,j}(n)$ is the capacitance between the $i^{th}$ and $j^{th}$ electrodes when pixel $n$ contains the high permittivity material and all other pixel with low permittivity material, $C_{i,j}(LOW)$ and $C_{i,j}(HIGH)$ are the capacitance between $i^{th}$ and $j^{th}$ electrodes when entire sensing area contains the low and high permittivity material only. Practically, it is difficult and time-consuming to generate a sensitivity matrix by measuring capacitances from different electrode pairs with change in permittivity in each pixel. Therefore, the first method is rarely used for calculating the sensitivity matrix in ECT.

In the second method, the sensitivity distribution is derived by the potential distribution directly from the electric field. The relationship between capacitance and permittivity distribution can be expressed as

$$ C = \frac{Q}{V} = -\frac{1}{V} \int_{\Gamma} \epsilon(x,y) \nabla \phi(x,y) d\Gamma $$

where $V$ is the potential difference between two electrodes, $\epsilon(x,y)$ is the permittivity distribution in the sensing area, $\phi(x,y)$ is the potential distribution and $\Gamma$ is the electrode surface (Yang and Peng, 2003).
To calculate a sensitivity map of an ECT sensor, an electrostatic field can be assumed because the excitation frequency is low, usually below 1 MHz. According to the Maxwell’s equation, the internal electric charge of the electrostatic field is zero. Therefore, the ECT sensor model can be described as

$$\nabla (\varepsilon(x,y) \nabla \phi(x,y)) = 0$$  \hspace{1cm} (2.8)

The electrical potential distribution is then derived as

$$E(x, y) = -\nabla \phi(x, y)$$  \hspace{1cm} (2.9)

To calculate the potential distribution, the sensing area is divided into many pixels, e.g. 60 × 60 grid with 3600 pixels. The sensitivity distribution between electrodes $i$ and $j$ is calculated based on the potential distribution, which can be expressed as

$$S_{i,j}(x, y) = -\int_{p(x,y)} \frac{E_i(x, y)}{V_i} \cdot \frac{E_j(x, y)}{V_j} \, dx \, dy$$  \hspace{1cm} (2.10)

where $E_i(x,y)$ is the potential distribution when an excitation voltage $V_i$ is applied to electrode $i$ while other electrodes remain grounded, and $p(x, y)$ is the area of the pixel at $(x, y)$ (Li and Yang, 2008). Normally, the potential distribution is solved by Finite Element Method (FEM) or Finite Difference Method (FDM), in which FEM is most preferred as it is more suitable for sensors with complex geometry.

From Eq. 2.2, the capacitance is related to the permittivity distribution. Therefore, the capacitance between the electrode pairs can be considered as a function of permittivity distribution and Eq. 2.2 can be rewritten as:

$$C = \xi(\varepsilon)$$  \hspace{1cm} (2.11)

The change in capacitance is therefore in response to a perturbation of the permittivity distribution, which is formulated as:

$$\Delta C = \frac{d \xi}{d \varepsilon} (\Delta \varepsilon) + O((\Delta \varepsilon)^2)$$  \hspace{1cm} (2.12)

where $\frac{d \xi}{d \varepsilon}$ is the sensitivity of capacitance versus permittivity distribution and $O((\Delta \varepsilon)^2)$ is the higher order terms.
Usually, $\Delta \varepsilon$ is small, thus the higher order can be ignored, and the Eq. 2.12 can be simplified and linearised as:

$$\Delta C = s \Delta \varepsilon, \quad \text{where} \quad s = \frac{d \xi}{d \varepsilon}$$

(2.13)

An ECT system is normally calibrated by measuring two capacitances $C_L$ and $C_H$, with the sensor filled with the low and high permittivity materials, respectively. All measured capacitance $C_M$ are then normalised to $\lambda$ with the value between 0 and 1.

$$\lambda = \frac{C_M - C_L}{C_H - C_L}$$

(2.14)

The linearised and discrete form of the forward problem can be expressed in a normalised form

$$\lambda = Sg$$

(2.15)

where $\lambda$ is the normalised capacitance formed as an $M \times 1$ vector, $g$ is a normalised permittivity vector which contains $N$ elements, and $S$ is an $M \times N$ Jacobian matrix which gives normalised capacitance distribution for each electrode-pair.

### 2.4.2 Inverse problem

The main task of the inverse problem is to determine the permittivity distribution $\varepsilon(x,y)$ from the measured capacitances. However, the inverse problem for ECT is difficult to solve, because of (1) a nonlinear relationship between the permittivity distribution and capacitance, (2) the limited number of independent measurements, which causes an under-determined problem, and (3) ill-posed and ill-conditioned problem.

In the past 3 decades, many algorithms were reported for ECT, including non-iterative algorithms such as linear-back projection (LBP) (Xie et al., 1992) and Tikhonov regularization (Tikhonov and Arsenin, 1978), iterative algorithms like the Newton-Raphson method (Hansen, 1998), Landweber iteration (Yang et al., 1999; Liu et al., 1999), total variation method (TV) (Wang et al., 2007), and some unconventional method like model-based iteration algorithms (Isaksen and Nordtvedt, 1993; Isaksen, 1996; Banasiak and Soleimani, 2010; Ren et al., 2014). To select a proper image reconstruction algorithm, the main criteria are image speed and accuracy. For most
practical ECT applications, LBP is commonly used, considering its simplicity and high speed (Liu et al., 2004).

**LBP**

Based on the linearised relationship between the measured capacitance and the image, LBP is used to determine the relationship between the permittivity vector and the measured capacitance vector (Xie et al., 1992). If the inverse of the normalised capacitance matrix $S$ exists, the permittivity vector can be solved by

$$g = S^{-1} \lambda$$  \hspace{1cm} (2.16)

Unfortunately, the inverse of $S$ does not exist because it is a non-square matrix. The permittivity vector can be obtained by assuming that the inverse relationship is true. This method is so-called LBP. If $S$ is considered as a linear mapping from the permittivity vector space to the capacitance vector space, $S^T$ can be considered as a related mapping from the capacitance vector space to the permittivity vector space. An approximated solution can be expressed as

$$\hat{g} = S^T \lambda$$  \hspace{1cm} (2.17)

A normalised form of Eq. 2.17 can be given by

$$\hat{g} = \frac{S^T \lambda}{S^T \mu_{\lambda}}$$  \hspace{1cm} (2.18)

where $\mu_{\lambda}$ is an identity vector.

LBP algorithm is still widely used for in time image reconstruction, though it can only provide low quality images.

**Landweber iteration algorithm**

Because all electrical tomography techniques suffer from the ‘soft-field’ problem, they present more challenges than ‘hard-field’ techniques, e.g. X-ray. To improve the image quality, iterative algorithms have been developed to solve the non-linear relationship between the permittivity distribution and the capacitance. The principle of iterative algorithm for ECT system is based on calculating the capacitance according to the current image. A new image is generated using the discrepancy between the measured
capacitance and the calculated capacitance. The process is repeated until a pre-set discrepancy is achieved. Among various iterative algorithms, Landweber iteration is simple in principle, efficient in iteration and has superior performance in image reconstruction (Yang and Peng, 2003), which is selected to solve the inverse problem in this thesis.

The Landweber iteration method was firstly designed for solving the ill-posed problem of the Fredholm integral equation of the first kind, and now it is widely used in optimisation theory (Landweber, 1950; Yang et al., 1999; Liu et al., 1999). The goal of the Landweber iteration method is to minimise \( \frac{1}{2}||Sg - \lambda||^2 \) and to find a proper permittivity vector \( g \). The function of minimisation related to \( g \) can be given by

\[
f(g) = \frac{1}{2}(Sg - \lambda)^T(Sg - \lambda) = \frac{1}{2}(g^T S^T S g - 2 g^T S^T \lambda + \lambda^T \lambda)
\]  

(2.19)

The gradient of \( f(g) \) can be expressed as

\[
\nabla f(g) = S^T (Sg - \lambda)
\]  

(2.20)

The gradient descent method chooses the direction in which \( f(g) \) decreases most quickly for the next iteration. Therefore, this direction should be opposite to the gradient of \( f(g) \) at the current point. The iteration process can be presented as

\[
\hat{g}_{k+1} = \hat{g}_k - \alpha_k \nabla f(\hat{g}_k) = \hat{g}_k - \alpha_k S^T (S \hat{g}_k - \lambda)
\]  

(2.21)

where \( \alpha_k \) is the step length for the \( k^{th} \) iteration.

In Eq. 2.21, the initial \( \hat{g}_0 \) can be obtained from a simple algorithm such as LBP. However, a problem with the Landweber iteration is its poor convergence property. To improve its convergence efficiently, a projected Landweber iteration is proposed, which is given by

\[
\hat{g}_{k+1} = P(\hat{g}_k - \alpha_k S^T (S \hat{g}_k - \lambda))
\]  

(2.22)

where \( P \) is a projection operator, which is defined as

\[
P[f(x)] = \begin{cases}
0 & \text{if } f(x) < 0 \\
f(x) & \text{if } 0 \leq f(x) \leq 1 \\
1 & \text{if } f(x) > 1
\end{cases}
\]  

(2.23)
It is notable in Eq. 2.21 that the step length $\alpha_k$ is associated with the efficiency of convergence. Normally, a fixed step length is utilized, typically chosen between 1 and 2 (Yang and Liu, 1999, Yang et al., 1999). Liu et al. (1999) suggested an optimal method to calculate $\alpha_k$ during an iterative process, providing a changing step length. It is assumed that in step $k$, an error vector between the measured capacitance and calculated capacitance is

$$e_k = S\hat{g}_k - \lambda$$

(2.24)

An updated step length is used in each step. In step $k+1$, the error vector becomes

$$e_{k+1} = S\hat{g}_{k+1} - \lambda = \alpha_k S S^T e_k - e_k$$

(2.25)

A function relative to $\alpha_k$ which describes the norm of the error vector

$$f(\alpha_{k+1}) = ||e_{k+1}||^2 = ||\alpha_k S S^T e_k - e_k||^2$$

(2.26)

The minimum $f(\alpha_{k+1})$ can be obtained when its deviation is zero. Therefore, the step length $\alpha_k$ is determined as

$$\alpha_k = \frac{||S^T e_k||^2}{||S S^T e_k||^2}$$

(2.27)

The step length is recalculated at each iteration step, corresponding to the error between true and estimated capacitance value. In principle, the efficiency of iteration can be enhanced with the optimised step length, which has been confirmed by the experimental results (Liu et al., 1999).

Model-based image reconstruction algorithm

Due to the limited number of measurements, all ECT systems face seriously ill-posed and ill-conditioned inverse problem with the conventional image reconstruction methods. To solve the problem, a model or shape based method has been developed for ECT. The concept of this method is to generate a regularisation or constraint model based on the prior knowledge during the reconstruction, resulting in a reduced number of unknowns, aiming to improve the image quality and to accelerate the imaging speed.

The first model-based iterative method for ECT was proposed by Isaksen and Nordtvedt (1993) based on a conventional 8-electrodes circular sensor. As shown in Figure 2.5, a few parameters are used to categorise and describe some typical distributions, such as
annular distribution (Figure 2.5 (a)), stratified (Figure 2.5 (b)) and the other oil/gas distribution (Figure 2.5 (c)). Therefore, the reconstruction process is simplified to optimise these parameters from the estimated models instead of calculating all pixels of the entire imaging area. When prior information on distribution is available, the method can provides promising results, which is a significant step for the ECT towards a quantitative process tomography.

The model/shape based method has potential use for 3D ECT imaging due to the simplified calculation in solving the forward and inverse problem. A shape based reconstruction method was suggested by Banasiak and Soleimani (2010). The method was implemented on a four-plane ECT sensor, where 8 electrodes on each plane. The shape-based approach utilises a level set algorithm that the inverse problem is solved only for the interface neighbourhood between two materials rather than the entire region of interest as in a conventional method. However, due to the complex computation process, the imaging time varies from 1 to 8 h with different stage level set method. Ren et al. (2014) used boundary element method (BEM) to solve the forward problem and the Levenberg-Marquardt method was used to reconstruct the optimal inclusion surface shape with a fast Jacobian matrix calculation method based on the reciprocity theorem. The results are promising based on a true 3D cubic ECT sensor with 24 electrodes (4 on each side). Generally, the model/shape based method can be adopted to reduce the number of unknowns and improve the image quality. However, for some complex structure of ECT sensor, such as 3D ECT sensor, the forward problem solving by FEM or BEM still faces time consuming problem during iteration, although the results were promising with the prior knowledge.
In this thesis, a 3D ECT imaging method will be investigated based on a conventional single-plane ECT sensor. With the simple structure ECT sensor and prior knowledge, the image speed would be faster than the methods discussed above, providing a potential use for positioning the milling/drilling tool in revision total hip replacement (THR) surgery.

2.5 Summary

This chapter reviewed the fundamental of an ECT system, including the hardware system and image reconstruction algorithms, which is associated with the work in this thesis. The design of ECT sensor was discussed and a few typical sensors were presented, such as the conventional circular sensor, unconventional sensor with square or conical shape or parallel structure, and 3D ECT sensor. None of those provides an open sensing area to adapt to the special geometry, which is common in medical application. In terms of hardware system, three commonly used capacitance measuring system were introduced, where the impedance analyser based ECT system is selected in this thesis due to the high accuracy and high resolution in capacitance measurement given by the impedance analyser. The detail of this system, including the hardware components and software controls, will be detailed in Chapter 3. Two problems associated with ECT were addressed. The sensitivity matrix can be calculated by two methods: (1) capacitance measured between each electrode-pair and (2) directly from the electric field, where the second method is most common and usually solved by either FEM or FDM. For solving inverse problem, LBP and Landweber iteration, which are the most commonly used algorithms, were introduced and selected to solve the inverse problem in this thesis. An updated step length was proposed for Landweber iteration, to improve the efficiency of iteration. To relief the ill-conditioned problem in inverse problem of ECT, a model/shape based image reconstruction algorithm was discussed. The method reduces the number of unknown elements and has potential use in 3D ECT imaging, but still time-consuming if a sensor with complex structure is utilised.

According to the review, to design an ECT sensor with open structure is essential in exploration of medical application of ECT, because most anatomies of a human body are associated with an irregular geometry. Additionally, due to the low image speed and low accuracy of the 3D ECT image caused by complex sensor structure and ill-posed
and ill-conditioned inverse problem, a real-time 3D reconstruction method based on simple sensor structure and prior knowledge is of particular interest in the thesis, which has potential use of ECT for some medical applications, such as positioning drilling/milling tool during revision THR.
Chapter 3: ECT system for medical application

This chapter describes an impedance analyser based ECT system for medical application. As introduced in Chapter 2, an ECT system consists of a capacitive sensor, a capacitance measuring unit and a host PC. The impedance analyser is designed to measure multiple electrical properties. Because of its high resolution, high accuracy and good reliability, the impedance analyser has excellent performance on electrical property measurement and has been used for many applications (Hu et al., 2008; Chrondronasios, 2003). In this thesis, an ECT system based on impedance analyser is used as the first trial for medical imaging. A multiplexer has been developed based on reed relays, providing minimum standing capacitances. By means of the impedance analyser and the multiplexer, the resolution of the system is 10 times better than conventional AC based ECT system (Chrondronasios, 2003, Yang and York, 1999).

3.1 Overall system

The developed ECT system, as shown in Figure 3.1, consists of four units: (1) an ECT sensor, (2) an impedance analyser (HP4192A), (3) a multiplexer and (4) a control PC. A GPIB-to-USB converter (NI GPIB-USB-HS) is used for communication between the HP4192A and the host PC. The multiplexer is controlled by the host PC via an RS232-to-USB converter. Between the multiplexer and the HP4192A, BNC connectors with shielding cables are used to avoid external noise. The impedance analyser is used to generate a source signal and to measure impedance via its UNKNOWN terminals. The multiplexer, as the name implies, is to configure the status of channels. In this system, each electrode is controlled by the multiplexer independently and three statuses can be provided: (a) excitation, connected to the source from the H Terminal of the impedance analyser, (b) detection, connected to the L Terminal of the impedance analyser, and (c) grounded, not connected to the impedance analyser. By changing the status of each electrode, independent measurements can be taken by different combination of electrode pairs. The measured capacitance data acquired by the impedance analyser can be transferred to the host PC, producing an image of the permittivity distribution. The following sections will introduce each part of the system and its software development.
Three-dimensional diagram showing the connectivity of devices in an ECT system.

Figure 3.1 Overview of dental ECT system based on an impedance analyser

3.2 Design of ECT sensor

As discussed in Chapter 2, the shape of ECT sensor can be varied for specific purposes, aiming to fit the geometry of the sensing space. While typical ECT sensors are circular because the sensors are usually mounted on pipeline used for flow measurement, the sensor for root canal treatment (RCT) employs a miniature two-plate structure, and the sensor for revision total hip replacement (THR) is designed in circular or conical shape.

3.2.1 ECT sensor for RCT

To generate a tooth image, sensor design is important. Theoretically, a bigger size of electrode provides a larger capacitance, corresponding to Eq. 2.1. However, due to the limited space in oral cavity, the overall size of the sensor for RCT is restricted and should be much smaller than a conventional ECT sensor. Moreover, conventional enclosed ECT sensor structure is not appropriate as an enclosed structure cannot fit the jaw morphology. Therefore, the sensor designed for RCT employs a miniature two-plate sensor with open structure. This structure enables the sensor to fit the specific anatomy structure in oral cavity. Prototype sensors, as shown in Figure 3.2 (a) and (b), were fabricated based on flexible printed circuit broad (FPCB) for experiment. With FPCB, the sensor can withstand slightly twisting and bending and the ultra-thin laminate of FPCB saves space in oral cavity and makes patients less painful. The fabricated sensor contains a pair of sensor plates. As the sensor employs an open sensing space, the
sensor plates can be placed on each side of the gum inside the mouth. Each sensor plate consists of a sensor head and a tail. Two sensor head designs are envisaged: a $2 \times 2$ array and a $2 \times 3$ array, configured as below:

- Overall sensor size: 21 mm in length and 18 mm in depth
- Shielding layer size: 21 mm in length and 18 in depth
- Electrode size: 9 mm in length and 7.5 mm in depth for $2 \times 2$ array, or 9 mm $\times$ 4.7 mm for $2 \times 3$ array
- Gap (between neighbour electrode): 1 mm
- Sensor tail length: 254 mm.

(a) $2 \times 2$ sensor array  
(b) $2 \times 3$ sensor array

Figure 3.2 Fabricated sensors for RCT

The FPCB allows the sensor pads to connect to external SMB connectors (soldered to the flexi circuit tail). Signals within a multi-layer FPCB may be effectively shielded using copper floods to the top and bottom layers. This determines the layer count as 3. Inner signal shielding is provided using traces between each signal. These traces are bonded to the outer shielding layers throughout by way of HDI through vias. The sensor pads in the sensor head are exposed on the front side (the side in contact with the gum) and shielded with a copper flood on the rear side.

In principle, the sensor with either $2 \times 2$ or $2 \times 3$ array can provide proximate 3D image if different electrode pairs are combined, i.e. a cross-sectional image can be generated by the electrodes in a row while a longitudinal image can be obtained by electrodes in a column. However, a small size of electrodes causes problems that the standing capacitance may dominate the measured capacitance. To obtain a larger capacitance, adjacent electrodes are excited or detected simultaneously. The performance of the two-plate sensors will be presented in Chapter 4.
The FPCB sensor with tail is proposed to be mounted to a mechanical structure, which locates the sensor heads in a parallel configuration with a subject’s gum interposed between the sensor heads. Once located the sensor heads shall be retained so that their position is in relation to each other and the sensor can be fixed on the measuring object(s). The materials present in the electronics assembly and any trace materials remaining from the manufacturing process must not pose a medical hazard and must be hypo-allergenic.

**A clamp-on sensor**

As an initial design of the clamp, Figure 3.3 shows a prototype that enables to fix the two-plate ECT sensor heads on the gum. The clamp is designed to hold a side of parallel sensor in a fitted cup. By bending and twisting the tail of FPCB sensor, the sensor head together with the tail can be clung on the clamp. The clamp employs a rod structure between the cups (sensor holder) and the clamp body, providing a flexibility of the sensor to adapt the unparalleled gum.

![Design of the clamp](image1)

**Figure 3.3 Design of clamp as dental ECT sensor**

As shown in Figure 3.3 (a), with a torsion spring applying a claping force, the spring loaded clamp is fitted onto the gum with sensors in place. A dental dam is fitted around the tooth, creating a seal that covers the clamp and sensors. The blue spring clip secures...
the dental dam onto the tooth, while extended wings on the clip prevent the clamp assembly from moving upwards. The wings are shaped to interface with the sensor cups to help prevent accidental movement after the dental dam has been fitted. Moreover, the clamp assembly is designed with an additional axis of rotation to accommodate angular irregularities in the gums.

**Test rig**

As a preliminary stage, an electro-mechanical rig has been made by Lucid Innovation Ltd. to replace the clamp. The rig aims to position the dental ECT sensor on either simulant materials or organic tissue. The rig consists of a clamp assembly and an SMB connector assembly, as shown in Figure 3.4. The clamp employs threaded adjustors (screws) to secure the position of FPCB sensor heads onto the surface of the test block. A spheroid structure on the adjustors head is utilised to connect clamp plates so that the clamp plates can be rotated to adapt to different angles between the surfaces of the test block. The connector assembly supports either $2 \times 2$ or $2 \times 3$ SMB outputs with plastic panels placed inside, providing location and support for the SMB connectors. The connector array can be changed easily by unscrewing the top screw and lifting out of the location slot.

![Figure 3.4 Electro-mechanical rig](image_url)
3.2.2 ECT sensor for revision THR

The shape of ECT sensor for revision THR can be either cylinder or cone to fit the anatomy of thigh. For simplicity, a conventional 8-electrode circular ECT sensor is utilised as a preliminary trial. The electrodes are mounted on the outer side of the wall. Because the fringe effect influences the measured capacitance, a few papers (Yang 2010; Sun and Yang, 2013) suggested employing grounded guards at each end of the sensor to reduce the fringe effect. In the practical sensor, grounded guards with a length of 2.2 cm are placed at both ends of the electrodes with a distance of 0.5 cm. A grounded shield is embedded to avoid external noise. The fabricated sensor, as shown in Figure 3.5, was embedded on a beaker made in glass, configured as below:

- Inner diameter: 10 cm
- Thickness of the wall: 0.2 cm
- Shield diameter: 11.6 cm
- Electrode length: 6 cm
- Length of the grounded guard: 2.2 cm
- Length of the shield: 11.6 cm
- Length of the wall: 13.6 cm
- Material of the electrode and the shield: Copper.

To fit the thigh anatomy, in the future design, the sensor for revision THR can be in conical shape with open structure, fabricated based on FPCB, forming a strap to adapt different diameter of thigh.

3.3 HP 4192A impedance analyser

HP4192A impedance analyser, as shown in Figure 3.6 (a), uses an auto-balancing bridge for measuring complex impedance with excitation amplitude up to 1.1 Volts in the frequency range of 5 Hz to 13 MHz (Agilent, 2000). The instrument allows the measuring impedance with a precision of 5 digits. For capacitance measurement, the maximum resolution is 0.1 fF.
Figure 3.6 HP4192A impedance analyser and its measurement principle (Agilent, 2000)

As shown in Figure 3.6 (b), HP4192A generates a sine wave with programmable amplitude and frequency as an excitation source to a multiplexer via UNKNOWN terminals—High current terminal (H\textsubscript{CUR}), High potential terminal (H\textsubscript{POT}), Low current terminal (L\textsubscript{CUR}) and Low potential terminal (L\textsubscript{POT}). The current terminals are to inject a test signal current flowing through the unknown device under test (DUT), I\textsubscript{DUT}, while the potential terminals are to detect the voltage drop across the DUT, V\textsubscript{DUT}. Therefore, the complex impedance Z can be obtained by $Z = \frac{V_{DUT}}{I_{DUT}}$. Two measurement modes can be selected in order to measure different impedance parameters. The HP4192A measures R+jX (impedance) in equivalent series circuit mode and admittance G+jB in parallel circuit mode. Other parameters are calculated from impedance or admittance by relative equations (Hu et al., 2008; Agilent, 2000).
In general, with a proper selection of measurement options, 11 impedance parameters can be measured, i.e. capacitance (C), inductance (L), phase angle (\( \theta \)), magnitude of impedance (|Z|), absolute value of admittance (|Y|), resistant (R), conductance (G), reactance (X), susceptance (B), quality factor (Q) and dissipation factor (D). The measured parameter value together with its unit can be shown in either DISPLAY A or DISPLAY B.

The data acquisition speed depends on the selected integration time. The integration time also has an influence on the measurement resolution and accuracy. There are three modes: HIGH SPEED, NORMAL or AVERAGE. A higher data acquisition rate may be achieved by the loss of the measurement accuracy, e.g. when the HIGH SPEED mode is chosen, the measurement repeatability drops and the precision of display decreases by one decimal. In this system, NORMAL mode is selected as a default setting for a compromise between the data acquisition rate and the accuracy. The integration time for each capacitance measurement is approx. 200 ms when the exciting frequency is over 1 kHz (Agilent, 2000). Obviously, the overall acquisition time of the system also depends on the number of electrodes. A sensor with more electrodes conduces to a larger number of independent measurements, resulting in a longer time for data acquisition for each frame.

Before measurement, the suggested warmup time for HP4192A is 30 min or longer to ensure the measurement accuracy. The self-calibration/test can be completed by pressing the BLUE key and the SELF TEST key from the front-panel, or via HP-IB remote control (program code S1). When no errors are detected, PASS is shown on DISPLAY A, otherwise, a corresponding error is displayed on DISPLAY A.

### 3.4 Multiplexer

Due to limited measuring channel of an impedance analyser, the impedance analyser based ECT system normally contains an external multiplexer. The multiplexer is used to connect measurement terminals of the impedance analyser to the ECT sensor, configuring the measuring channels (electrodes). The fabricated multiplexer, as shown in Figure 3.7, utilises reed relay switches to achieve minimum stray capacitance. There are 16 channels in the multiplexer, connecting to a maximum of 16 electrodes. Each channel is controlled individually by 3 relays and 2 control bits given by the host PC.
which allows 3 operating settings for each terminal (electrode): excitation, detection or grounded. As shown in Figure 3.8, each electrode is controlled by 3 relay switches in two ‘T’ configurations. This configuration ensures the open switch to be taken to the ground rather than to generate a stray capacitance in parallel, thus obtaining the minimum stray capacitance across the analyser measurement terminals.

Figure 3.7 16-channel multiplexer

A UCN5818 (Allegro Microsystems) serial-to-parallel converter is used to receive the serial data from the host PC. 32-bit parallel data can be latched to control the relays. The multiplexer is connected to the UNKNOWN terminals of impedance analyser that the impedance analyser can measure the capacitance between the selected channels. The stray capacitance is acquired from measurements of the impedance for all electrode combinations with the multiplexer and impedance analyser connected to known impedance in the form of a capacitor and a resistor in parallel. The measured stray capacitance in the order of 0.07 pF was achieved due to the configuration of the switching circuit (Chrondronasios, 2003). A low stray capacitance benefits to achieve high sensitivity, particularly for the miniature electrode as the measured capacitance variations are expected to be in the order of fF.

Figure 3.8 Switches for each electrode channel
3.5 Software development

The system is controlled by the host PC using MatLab GUI. MatLab provides a programming platform for data analysis, algorithm development, numerical computation and data presentation. Various toolboxes are available in MatLab, supporting hardware control and a number of functions, such as signal and image processing, control design, test and measurement. For the laboratory use, MatLab is an ideal tool to implement system optimisation.

3.5.1 Impedance analyser control

Instrument Control Toolbox enables MatLab to control instruments directly via test-based Standard Commands for Programmable Instruments (SCPI) commands over commonly used communication protocols such as General Purpose Interface Bus (GPIB) and Virtual Instrument Software Architecture (VISA). With this toolbox, data can be easily read and written between MatLab and the connected electronic devices. MatLab supports various GPIB hardware vendors, i.e. Agilent hardware, CONTEC hardware and National Instrument (NI). A GPIB object can be successfully created if the agreement can be satisfied. To meet the agreement, information from the vendors is provided such as vendor name, board index and instrument address. The impedance analyser is controlled by the host PC via the GPIB-USB-HS converter from the NI Cooperation Ltd. As SCPI commands (GPIB) is applied in the functions in Instrument Control Toolbox, the impedance analyser HP4192A can be fully controlled using simple pre-set commands in MatLab.

The HP4192A can be programmed into various measurement modes, obtaining different impedance parameters. The output signal to UNKOWN terminals, including the amplitude and frequency, can be programmed and fully controlled by the host PC. Some typical remote program codes are given in Table 3.1. When default remote program code ‘F0’ is selected, the data format obtained from impedance analyser is described in Figure 3.9. Data shown in DISPLAY A and DISPLAY B on the impedance analyser can be obtained by selecting the related information according to the data format (Agilent 2000).
### Table 3.1 Remote program code and parameter setting

<table>
<thead>
<tr>
<th>HP 4192 remote control</th>
<th>Program Code</th>
<th>Program code</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>DISPLAY A</td>
<td>C (Capacitance)</td>
<td>A4</td>
</tr>
<tr>
<td>DISPLAY B</td>
<td>Q (Quality Factor)</td>
<td>B1</td>
</tr>
<tr>
<td></td>
<td>D (Dissipation Factor)</td>
<td>B2</td>
</tr>
<tr>
<td></td>
<td>R/G (Resistance/Conductance)</td>
<td>B3</td>
</tr>
<tr>
<td>TRIGGER</td>
<td>HOLD/MANUAL</td>
<td>T3</td>
</tr>
<tr>
<td>Parameter</td>
<td>SOPT FREQ</td>
<td>FR</td>
</tr>
<tr>
<td></td>
<td>OSC LEVEL</td>
<td>OL</td>
</tr>
</tbody>
</table>

| XXXX : NNN.NNE ± NN, XXXX : NNN.NNE ± NN | Status of DISPLAY A | Function of DISPLAY A | Deviation measurement mode of DISPLAY A | Value of DISPLAY A | Unit of DISPLAY A | Comma | Status of DISPLAY B | Function of DISPLAY B | Deviation measurement mode of DISPLAY B | Value of DISPLAY B | Unit of DISPLAY B | Data Terminator |

Figure 3.9 Data format (Agilent, 2000)

#### 3.5.2 Multiplexer Control

The multiplexer is controlled by altering the status of two serial pins: DTR and RTS (Chondronasios, 2003). The purpose of the multiplexer is to realise various measurement modes to satisfy the different requirements. For each channel of the multiplexer, there are 2 control bits corresponding to the operation mode. As seen in Figure 3.8, control bit 1 is connected to relay 2 while relay 1 and 3 are controlled by bit 2. The relay connects to pin 1 when the control bit is ‘1’ and pin 2 when the control bit is ‘0’. There are three possible statuses for each electrode: (a) grounded when both
control bits are ‘0’, (b) excitation when bit 1 is ‘1’ and bit 2 is set to ‘0’ or (c) detection when bit 2 is ‘1’ with bit 1 either ‘0’ or ‘1’.

For a 16-channel unit, 32-bit parallel control is required. As it is a one-way communication, a stream of 32 serial bits is sent from the host PC when a change of electrodes status is required. An output control function is written in MatLab to set the status of serial pin RTS and DTR directly.

3.6 Summary

This chapter introduces the design of an ECT system for medical application based on an impedance analyser. Different from the conventional ECT sensor for industrial flow measurement, a miniature two-plate ECT sensor is required for RCT, to fit the limited and special geometry in the oral cavity. The two-plate ECT sensor contains two separate sensor plates with 4 or 6 electrodes on each plate, forming a $2 \times 2$ or $2 \times 3$ sensor array. This specific sensor structure allows both cross-sectional and longitudinal images, resulting in a proximate 3D image. The performance of this specific sensor will be detailed in the following chapter. For revision THR, a conventional 8-electrode circular sensor is utilised and the feasibility study of using ECT sensor for revision THR will be introduced in Chapter 5. Due to the comparatively small electrodes size, a multiplexer with low standing capacitance is used. This multiplexer utilises 3 relays for each electrode, allowing three status of the electrode: excitation, grounded and detection. The impedance analyser HP4192A is used for capacitance measurement with a maximum resolution of 0.1 fF, which can satisfy the measurement of a small variation in capacitance due to the miniature electrode on the fabricated ECT sensor for RCT.
Chapter 4: ECT for root canal treatment

Root canal treatment (RCT) is a common endodontic therapy to deal with infected teeth. In a traditional RCT procedure, radiograph is the only reliable reference for the dentist to determine the working length of the endodontic instrument. A radiograph is normally taken before the surgery, but the cleaning process is nearly ‘blind’. Currently, apex locators are used to assist dentists to locate the cementodentinal junction (CDJ) (Gordon and Chandler 2004). However, the apex locators have some limitations such as low accuracy and requiring dry working environment in the root canal (Popescu, 2010). Moreover, the apex locators can only give an alarm when the apex of the endodontic instrument is close to the CDJ, and cannot navigate the dentist during surgery as a visual tool for the endodontic therapy. This chapter describes a dental ECT system, which has been developed to assist the dentists for RCT. The ECT system is used for two purposes, one is to visualise the tooth surface and the other is to determine the location of the endodontic instrument. The following section will focus on the discussion of the performance of the fabricated dental ECT sensors and the possibility of using ECT to detect a metallic object based on simulation models and experiments.

4.1 Two-plate dental ECT sensors

4.1.1 Simulation models

To fit a jaw anatomy, a dental ECT sensor employs a miniature two-plate structure, to fit an open sensing area. Because of the open sensing space of the two-plate ECT sensors, the performance of the sensors is affected by many factors. Five of these are considered: (1) boundary setting of the open boundaries (2) the effective sensing area, (3) the distance between two plates, (4) the angle between two plates, and (5) the number of electrodes.

For simulation, 2D FEM models of two-plate ECT sensors were generated in COMSOL 3.5, where the ‘In-Plane Electric Currents’ Mode is selected and an 1 V peak to peak sine wave with a excitation frequency of 1 MHz is applied to electrodes. A sensor employs a wall and an earthed shielding layer, corresponding to the fabricated sensor. For simplicity, two sensor plates are placed in parallel. The sensors are shown in Figure 4.1 adopting 4, 6 or 8 internal electrodes. The configuration of sensors is listed below:
• Length of shielding layers and wall: 21 mm
• Length of electrodes: 9 mm (Sensor 1), 6 mm (Sensor 2), 4 mm (Sensor 3)
• Gap between adjacent electrodes: 1 mm
• Thickness of wall, electrode and shielding layers: 0.2 mm
• Material of shielding layers and electrode: copper
• Material of wall: FR4 ($\varepsilon_r = 4.8$).

Figure 4.1 2D FEM models and boundary settings for parallel two-plates ECT sensors

The testing object(s) with high permittivity ($\varepsilon_r = 3$) is placed in a low permittivity ($\varepsilon_r = 1$) background. A sensitivity matrix is calculated for each sensor with the sensing area only filled with the low permittivity material.

Two indicators were suggested in (Yang and Peng 2003) to evaluate the results: (1) the capacitance residual and (2) the correlation coefficient between the true image and the estimated image. The capacitance residual is normally selected as the stopping criteria of an iteration algorithm, while the correlation coefficient is associated with the image quality. It should be noticed that capacitance is treated as a vector, so that its norm is used to calculate the residual.

\[
\text{Capacitance residual} = \frac{\|\lambda - S\tilde{g}\|}{\|\lambda\|} \times 100\% \quad (4.1)
\]

\[
\text{Correlation coefficient} = \frac{\sum_{i=1}^{N_p}(g_i - \bar{g})(\tilde{g}_i - \bar{\tilde{g}})}{\sqrt{\sum_{i=1}^{N_p}(g_i - \bar{g})^2 \sum_{i=1}^{N_p}(\tilde{g}_i - \bar{\tilde{g}})^2}} \quad (4.2)
\]

where $\bar{g}$ and $\bar{\tilde{g}}$ are the mean of true permittivity vector $g$ and the estimated permittivity vector $\tilde{g}$, respectively and $N_p$ is the number of pixels. In this thesis $N_p$ is 3600, referring to a sensing area with $60 \times 60$ grids.
A. Boundary setting of the open boundaries

Figure 4.2 (a) shows the potential distributions and electric fields of the 4-electrode two-plate ECT sensors when electrode 1 is excited with all the other electrodes grounded. For a conventional circular or square ECT sensor with closed sensing area, the sensing area is covered by the sensor and the boundary of the sensing area can be simply set as ‘Continuity’ in simulation models. However, for the two-plate sensor, it is crucial to consider the boundary condition of the open boundaries (the boundaries marked in red colour in Figure 4.2 (a)) those are not covered by the sensor. Due to the open sensing area of the two-plate ECT sensor, the electric field is expanded to a larger sensing space. Thus, in principle, an extend image area is more in accordance with the real situation. However, a large image area may cause a low resolution in the region of interest (ROI) and redundant computation compared with the model with a small and ROI related sensing area. In simulation models, two boundary conditions are considered, ‘Port’ \((I = Y \cdot V)\) and ‘Electric insulation’ \((n \cdot j = 0)\), where \(I\) is current, \(V\) is potential, \(Y\) is admittance and \(n \cdot j\) is normal component of the current density. With the boundary setting of ‘Port’, the potential contours are completed and the electric field is expanded in the model with either the extended or small sensing area. However, with ‘Electric insulation’, the side electrodes show incomplete potential contours within the selected image area and electric field is blocked by the insulated boundaries. By comparing two boundary settings, with ‘Electric insulation’, the model with the small sensing area shows similar potential contours and electric fields to those of the extended sensing area with the boundary setting as either ‘Electric insulation’ or ‘Port’.

Figure 4.2 (b) shows the images of a circle (diameter of 10 mm and permittivity of 3) placed in the centre of the image area in the 4-electrode sensor with the same ROI but different sensing areas and open boundary settings. The images are reconstructed by Landweber with 100 iterations, representing the normalised permittivity distribution. With the open boundaries setting as ‘Port’, the images suffer from significant distortion caused by the open fringes. On the other hand, the models with ‘Electric insulation’ at the open boundaries show promising images, providing large correlation coefficients between the reconstructed images and the true distributions. Apparently, with either ‘Port’ or ‘Electric insulation’, the images by the models with the small sensing area present higher image contrast and larger correlation coefficient than the extended sensing area.
Figure 4.2 Contours of electric potential, electric fields and reconstructed images with different boundary conditions

**B Effective sensing area**

In a conventional ECT system, a fixed sensitivity matrix is usually used for image reconstruction because the traditional ECT sensor, either circular or square, employs a centrosymmetric sensing area. Consequently, the effects of the sensitivity matrix on the reconstructed image were rarely discussed in the past. However, for a two-plate sensor with an open sensing area, it is essential to choose a proper sensing area and to generate a sensitivity map to produce high quality images.

To investigate the effect of the effective sensing area, 3 FEM models with different sensing area are generated based on 4-electrode two-plate sensor, where the two sensor
plates are placed in parallel. A bar (diameter of 10 mm and permittivity of 3) is placed in the centre of the sensing area. The space between the bar and sensor plates is filled with air. In all models, the distance between the two sensor plates is fixed to 20 mm. Model I has a slightly smaller sensing area with 19 mm in width, which merely covers the electrodes. The sensing area in Model II has the same width as the shielding layer and the overall size is 21 mm × 20 mm. The largest sensing area is used in Model III with a size of 30 mm × 20 mm. Figure 4.3 shows the models, where the boundary condition of the open boundary in the sensing area is set to ‘Electric insulation’. Note that the sensitivity map shown in Figure 4.3 is the total sensitivity map, which is the sum of the sensitivity maps of all electrode pairs.

Figure 4.3 Impact of sensing area on reconstructed image
The reconstructed images, as shown in Figure 4.3, are generated by LBP and Landweber iteration. The results show that the sensitivity matrix as well as the reconstructed image is highly sensitive to the width of the sensing area. Theoretically, due to the open structure of the two-plate sensors, a larger sensing area is more accordant with the actual situation, because the ‘soft’ electric field would spread to a larger space and affect the neighbouring area, which is so-called fringe effect. However, a noise, i.e. large negative sensitivity, is observed at the edge of the sensing area where the shielding layer is not covered, as shown in Model III. The noise causes a distortion at its effective area, generating bright spots at the four corners in the reconstructed image. Distortion is also shown in Model I, where the width of the effective sensing area is slightly narrower than the shielding layers. A significantly high positive sensitivity is presented at the edge of the sensing area. For an ECT sensor, a higher sensitivity in the centre (the weakest area) is desirable to provide a good image resolution (Yang and Liu, 1999). However, the high sensitivity at the edge results in a relatively low sensitivity in the centre. Therefore, low contrast images are generated by either Model I or Model III compared with Model II.

Figure 4.3 also compares the images by LBP and Landweber iteration. From the reconstructed images, only Model II provides an improved image by using Landweber iteration. The correlation coefficients (CC) calculated by Eq. 4.2 are given below the images in Figure 4.3. In general, a high correlation coefficient refers to a good image. With high sensitivity showing at the edge, there is no obvious improvement of image quality by Landweber iteration in Model I and Model III. In the Model I, a higher correlation coefficient is obtained by LBP and the correlation coefficient reduces by 0.09 when using Landweber iteration. A significant enhancement of image contrast is presented in Model II and the correlation coefficient rises from 0.72 by LBP to 0.80 by Landweber after 100 iterations.

Figure 4.4 shows the capacitance residual with the three models, with 100 iterations. The relative capacitance residual is normally used as the stopping criterion in an iteration algorithm, although it is not directly associated with the image quality. Model I gives the largest relative capacitance residual at the beginning, and quickly converge to 50% in 5 iterations. The capacitance residual of Model II starts from a lower value and it rapidly decreases to 1% with 60 iterations and remains a small value of capacitance
residual afterward. Although the smallest starting capacitance residual can be found in Model III, the capacitance residual shows no apparent change within 100 iterations, which explains why there is no obvious difference between the reconstructed images by LBP and by Landweber. By comparing these three models, Model III needs fewest iterations (approx. 2) till its capacitance residual becomes stable, while Model II requires largest number of iterations but it reaches the smallest error between estimated and measured capacitance vectors.

![Figure 4.4 Capacitance residual against number of iterations](image)

Although the effect by selecting image area for sensitivity map is rarely discussed in the past, it is one of the key factors to obtain satisfactory images for an ECT sensor with open sensing space. From Figure 4.3 and 4.4, it can be concluded that to enhance the sensitivity in the central region and optimise the image quality, the width of its sensing area should be the same as the electrical shielding layer.

C Distance between two sensor plates

A two-plate sensor has a flexible distance between the sensor plates. From Eq. 2.1, the change in distance between two sensor plates would affect the measured capacitance and hence change the sensitivity distribution in the sensing area. Five distances were selected for the simulation study: 10, 15, 20, 25 and 30 mm. Figure 4.5 compares the sensitivity maps and the images generated by the 4-electrode two-plate sensor (plates are placed in parallel) against different distance between the two plates. According to the results from previous discussion, the width of sensing area is fixed to 21 mm, which is equal to the length of shielding layers, to obtain a relatively high image quality. In
each selected distance, two permittivity distributions are investigated. In the first distribution, annular distribution is adopted. A rectangular/square object with permittivity of 3 is placed in the centre of sensing area. The width of the object is fixed to 15 mm, which is 6 mm less than the width of the sensing area and height is changed with the distance, which keeps 4 mm less than the distance between the parallel plates. The other distribution employs a circle with a fixed diameter (10 mm) and permittivity of 3. The images are generated by LBP and Landweber iteration, respectively.

![Figure 4.5 Sensitivity maps and reconstructed images with different distance between two plates](image)

Figure 4.5 shows the total sensitivity maps and the reconstructed images of two distributions with different distance between two plates. A smaller distance between two plates shows less sensitivity between the adjacent electrodes and relatively high sensitivity in the central region. Note that there is a high negative sensitivity surrounding the electrode area when the distance increases to 25 mm. The negative sensitivity increases and the sensitivity between adjacent electrodes become smaller when the distance further increases. A larger distance results in a higher contrast between the object and background with annular distribution. However, with a fixed size circle placed in the central region, the image contrast reduces as the distance between two sensor plates increases, because the sensitivity in the central region is attenuated with a larger distance between two sensor plates. By comparing the images
reconstructed by the two reconstruction methods, Landweber iteration enhances the contrast in the centre area and gives improved images compared with LBP.

Figure 4.6 compares the capacitance residual with different distance between the two sensor plates. With the rectangular/square object (annular distribution), a smaller distance between plates gives a larger starting capacitance residual, but the capacitance residual sharply drops and converges to lower capacitance residual with less iterations. On the contrary, with a large distance, 25 mm and 30 mm, a low capacitance residual is obtained at the beginning and it reduces slowly afterward. The capacitance is still unstable after 100 iterations when a large distance between two plates is placed. Differently, when imaging a circular object with fixed diameter, the largest starting capacitance residual can be found when the distance between two parallel sensor plates is 20 mm. However, similar to the results of the annular distribution, by reducing the distance between two plates, the capacitance residual would become stable and reach the lowest value quickly.

The image quality is evaluated by correlation coefficient. Table 4.1 shows the correlation coefficient with the two distributions and with different distance between two sensor plates. Generally, the correlation coefficient is higher in the image of circular object with a fixed diameter of 10 mm than that of rectangular/square object. By LBP, the correlation coefficient decreases by increasing the distance between two plates for both distributions. With all selected distances, the correlation coefficient is improved by Landweber iteration, except for imaging circular object with a distance of 10 mm. With rectangular/square object, a large value of correlation coefficient is
obtained by Landweber iteration when the distance is close to the length of the shielding layer, i.e. 20 mm and 25 mm. However, when imaging the circular object by Landweber iteration, a higher correlation coefficient can be achieved with a smaller distance between two plates.

Table 4.1 Correlation coefficient with different distances between two plates

<table>
<thead>
<tr>
<th></th>
<th>Correlation Coefficient</th>
<th>Distance (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>10</td>
</tr>
<tr>
<td>Rectangular/square object</td>
<td>LBP</td>
<td>0.63</td>
</tr>
<tr>
<td></td>
<td>Landweber</td>
<td>0.65</td>
</tr>
<tr>
<td>Circular object</td>
<td>LBP</td>
<td>0.85</td>
</tr>
<tr>
<td></td>
<td>Landweber</td>
<td>0.83</td>
</tr>
</tbody>
</table>

In general, a smaller distance between two plates leads a larger sensitivity in the central region while a larger distance gives a higher contrast with an annular distribution. In terms of image quality, a higher correlation coefficient can be achieved by Landweber iteration. An object with a fixed shape is placed in the centre gives less image contrast, because the increase in the distance will reduce the sensitivity in the central region. With a smaller distance, less iteration is needed to stabilise the capacitance residual. On the other hand, a small distance between two plates limits the space in the imaging area. To compromise the sensing space and the image quality, the distance between the two plates should be equal to the length of the shielding layer.

**D Angle between two plates**

Due to the flexibility of placing the sensor, the two plates may not be exactly in parallel. Theoretically, the rotation of the sensor plates would change the distance between opposing electrodes, influencing the sensitivity distribution in the image area. This section discusses the rotation angle between plates from 0 to 90 degrees with a step length of 15 degrees. The distance between the centres of two plates is fixed to 20 mm. A bar (\(\varepsilon_r = 3\)) with 6 mm in diameter is placed in the centre of the image area.

Figure 4.7 shows the images and correlation coefficient generated by LBP and Landweber iteration, respectively. Note that 0 degree refers to the two plates placed in parallel while 90 degrees denotes that they are perpendicular. Apparently, the image quality is improved by Landweber iteration. Although the smallest correlation
coefficient is obtained by LBP with 30 degrees, the correlation coefficient is improved by 0.12 by Landweber iteration after 100 iterations. As shown in Figure 4.7, with a small rotation angle, e.g. below 30 degrees, there is no obvious change in image quality, particularly for the images generated by Landweber iteration. The image contrast is enhanced by a further increase in the rotation angle. By LBP, the largest correlation coefficient can be found when the plates placed in 90 degrees. With a large rotation angle, e.g. 75 and 90 degrees, the correlation coefficient reaches 0.7 with 100 iterations by Landweber iteration.

![Image 4.7 Images and correlation coefficients by LBP and Landweber with different rotation angle](image)

Figure 4.7 Images and correlation coefficients by LBP and Landweber with different rotation angle

Figure 4.8 shows the capacitance residual with different rotation angles in 100 iterations. With a smaller rotation angle, the capacitance residual converges more quickly and reaches a smaller value after 100 iterations. With the sensor plates placed in parallel, the capacitance residual rapidly drop from 45% and the value remains below 1% after 50 iterations. A notably low starting relative capacitance residual is shown with two sensor plates placed perpendicularly. However, the convergence process is slow when rotation angle is 90 degrees, with the smallest change in relative capacitance residual after 100 iterations, indicating longer time to meet the stopping criteria.
From the above results, it is clear that there is no obvious change in image quality with a small rotation angle (below 30 degrees). However, the effect of the rotation angle becomes greater with a larger angle. The model with larger rotation angle shows a higher contrast between the testing object and the background but comparatively lower convergence with Landweber iteration.

Figure 4.8 Relative capacitance residual with different rotation angles against numbers of iteration

E Number of electrodes

The image quality directly depends on the number of electrodes. Theoretically, more electrodes provide more independent measurements, possibly resulting in a higher resolution of the reconstructed image (Peng et al 2012). As an initial study of two-plate ECT sensors, 3 different numbers of electrodes are investigated for simulation: 4, 6 and 8. According to the results from previous discussion, the sensing area of those sensors is fixed to $21 \times 20 \text{ mm}^2$. For simplicity, the two plates are placed in parallel.

Figure 4.9 shows images of 8 permittivity distributions by LBP and Landweber iteration. By LBP (see Figure 4.9 (a)), the sensor with 8 electrodes generates higher contrast images. However, no sensor model could produce clear images by LBP when two or more objects are placed in the sensing area. In general, images by Landweber iteration show higher image contrast between testing object(s) and background than LBP. With Landweber iteration, the sensors with 6 and 8 electrodes produce high resolution image with multiple objects. However, the image of stratified 1 and 2 suffers from large distortion between electrodes when either a 6 or 8-electrode sensor is used. Moreover, distortion is observed in the gap between electrodes with a 6 and 8-electrodes sensor,
forming a bright ring or dot in the effective area. In principle, the two-plate sensor is highly asymmetry due to the open structure and the incomplete measurement. This is confirmed by the reconstructed images. With either LBP or Landweber iteration, the contrast is enhanced when the object(s) is placed close to electrodes, e.g. Single bar 2 and Two bars 2.

Table 4.2 lists correlation coefficients with different numbers of electrodes by LBP and Landweber iteration. By LBP, the correlation coefficient increases when the number of electrodes increases from 4 to 6. However, a further increase in the number of electrodes has no obvious improvement on correlation coefficient. Instead, the correlation coefficient drops with the sensor with 8 electrodes for some permittivity distributions, such as Single bar 2, Two bars 2, Three bars and Four bars. With 6 and 8

Figure 4.9 Permittivity distributions and reconstructed images
electrodes, the correlation coefficient of stratified distribution is larger with LBP than Landweber iteration. This phenomenon is confirmed by the images shown in Figure 4.9, where large distortion appears near electrodes by Landweber iteration. For other permittivity distributions, Landweber iteration provides larger correlation coefficients than LBP.

<table>
<thead>
<tr>
<th>Correlation Coefficient</th>
<th>No. of electrodes</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>4</td>
<td>6</td>
<td>8</td>
<td></td>
</tr>
<tr>
<td>LBP</td>
<td>Landweber</td>
<td>LBP</td>
<td>Landweber</td>
<td>LBP</td>
</tr>
<tr>
<td>Stratified 1</td>
<td>0.87</td>
<td>0.94</td>
<td>0.90</td>
<td>0.93</td>
</tr>
<tr>
<td>Stratified 2</td>
<td>0.85</td>
<td>0.92</td>
<td>0.81</td>
<td>0.94</td>
</tr>
<tr>
<td>Single bar 1</td>
<td>0.36</td>
<td>0.54</td>
<td>0.61</td>
<td>0.58</td>
</tr>
<tr>
<td>Single bar 2</td>
<td>0.37</td>
<td>0.61</td>
<td>0.75</td>
<td>0.58</td>
</tr>
<tr>
<td>Two bars 1</td>
<td>0.45</td>
<td>0.52</td>
<td>0.58</td>
<td>0.52</td>
</tr>
<tr>
<td>Two bars 2</td>
<td>0.29</td>
<td>0.49</td>
<td>0.76</td>
<td>0.48</td>
</tr>
<tr>
<td>Three bars</td>
<td>0.04</td>
<td>0.25</td>
<td>0.42</td>
<td>0.23</td>
</tr>
<tr>
<td>Four bars</td>
<td>0.11</td>
<td>0.19</td>
<td>0.22</td>
<td>0.17</td>
</tr>
</tbody>
</table>

Compared with LBP, the image quality by Landweber iteration has notable improvement when imaging multiple objects. As shown in Table 4.2, more electrodes also contribute to a larger correlation coefficient, indicating a higher image quality, especially for the distribution of multiple objects, e.g. Four bars.

4.1.2 Performance of fabricated sensors

An Experiment has been carried out by means of the developed impedance analyser based ECT system. Figure 4.10 shows the ECT system in operation. In principle, the fabricated sensor with either 2 × 2 or 2 × 3 array can provide both cross-sectional and longitudinal images by grouping electrodes in a row and in a column respectively. Due to the miniature size of individual electrodes, the measured capacitance of the fabricated sensor is much smaller than that from the conventional ECT sensors. The system may suffer from undesired standing capacitance from the cable and multiplexer. To obtain an effective capacitance value, a measurement strategy of multiple-electrode excitation is used instead of the conventional single-electrode excitation. By single-electrode excitation, there are two planes for either cross-sectional or longitudinal view when the 2 × 2 sensor is used, while the 2 × 3 sensor forms three planes for cross-sectional view.
and two planes for the longitudinal view with the single-electrode excitation. The concept of multiple-electrode excitation is to excite multiple electrodes simultaneously. By combining different electrodes, an ECT image of either cross-sectional or longitudinal view can be generated. Table 4.3 details the use of electrodes for the two excitation strategies, where the sensor configuration is followed by Figure 4.11.

Figure 4.10 Experimental set-up

![Experimental set-up](image)

**Figure 4.10** Experimental set-up

![Dental ECT sensors](image)

**Figure 4.11** Dental ECT sensors

Figure 4.12 shows the measured capacitance by $2 \times 2$ sensor with 7 frequencies varying from 10 kHz to 10 MHz. For an image of either cross-sectional or longitudinal view, four electrodes (or combined electrodes) are used, resulting in 6 independent capacitance measurements. Due to the open structure and large distance between the two sensor plates, the capacitances measured by adjacent electrode pairs are significantly larger than the other pairs. By comparing Figure 4.12 (a) and (b), as two electrodes excited simultaneously (multiple-electrode excitation), the resultant capacitance is doubled. Note that there is no obvious change in measured capacitance.
with frequency varying from 10 kHz to 5 MHz. For either single-electrode excitation or multiple-electrode excitation by $2 \times 2$ sensor, larger capacitance can be found when the exciting frequency is 500 kHz and 1 MHz. By comparing the two excitation strategies, the multiple-electrode excitation strategy presents lower sensitivity to the change in frequency when the frequency varies from 10 kHz to 5 MHz. However, when the frequency increases to 10 MHz, with both excitation strategies, the measured capacitance suffers from undesirable disturbance. One possible reason for the disturbance is resonance since the whole system can be equivalent to a RLC network.

Table 4.3 Excitation strategies and the excited electrodes

<table>
<thead>
<tr>
<th></th>
<th>$2 \times 2$ array</th>
<th>$2 \times 3$ array</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Cross-sectional</td>
<td>Longitudinal</td>
</tr>
<tr>
<td>Single-electrode excitation</td>
<td></td>
<td></td>
</tr>
<tr>
<td>First plane</td>
<td>E1, E2, E5, E6</td>
<td>E1, E3, E5, E7</td>
</tr>
<tr>
<td>E1-3*, E2-4, E5-7, E6-8</td>
<td></td>
<td>E1-2, E3-4, E5-6, E6-8</td>
</tr>
<tr>
<td>Second plane</td>
<td>E3, E4, E7, E8</td>
<td>E2, E4, E6, E8</td>
</tr>
<tr>
<td>Third plane</td>
<td>E5, E6, E11, E12</td>
<td></td>
</tr>
</tbody>
</table>

*En-m means electrode n and m are excited simultaneously

Figure 4.13 shows the measured capacitance by single-electrode and multiple-electrode excitation with the $2 \times 3$ sensor array. With single-electrode excitation, three planes can be obtained for the cross-sectional view and two planes for the longitudinal view. By comparing three planes for a cross-sectional image, the first plane generates the largest capacitance while the smallest capacitance is obtained by the electrodes in the middle plane. For longitudinal view, the measured capacitance of the first plane is slightly larger than that of the second plane. With multiple-electrode excitation, the measured capacitance is triple of that by single-electrode excitation for cross-sectional view and twice of measured capacitance by the single-electrode excitation for longitudinal view. In the frequency range from 10 kHz to 5 MHz, the measured capacitance is not sensitive to the excitation frequency, particularly with multiple-electrode excitation. However,
when the frequency increases to 10 MHz, the system suffers from undesirable disturbance, showing abnormal capacitances obtained by difference electrode pairs. The sensor with multiple-electrode excitation has more serious disturbance than the single-electrode excitation. Similar to the $2 \times 2$ sensor, a relatively large capacitance can be obtained by 500 kHz and 1 MHz with either single-electrode or multiple-electrode excitation.

Figure 4.12 Capacitance with different excitation strategy ($2 \times 2$ sensor array)
Figure 4.13 Capacitance with different excitation strategy (2 × 3 sensor array)
Signal-to-noise ratio (SNR) is used to evaluate the performance of the sensor as well as the impedance analyser based ECT system. Hu and Yang (2006) defined the SNR of an ECT system calculated by raw data, which can be expressed as

\[
SNR = 20 \log_{10} \left( \frac{\text{Signal}}{\text{Noise}} \right) = 20 \log_{10} \sqrt{\frac{\sum_{i=1}^{N_F} C_{M_i}^2}{\sum_{i=1}^{N_F} (C_{M_i} - \overline{C_M})^2}}
\]  

(4.3)

where \( N_F \) is the number of frames, \( C_M \) is the measured capacitance from an electrode pair and \( \overline{C_M} \) is the mean of \( C_M \).

Figure 4.14 shows the SNR of the fabricated sensor with 2 × 2 array. Note that Ei_opp is electrode i and its opposing electrode (opposing pair), while Em_Em denotes electrode m and electrode n. The SNRs are calculated by 10 frames of the capacitance measured by HP4192A with empty space. A sine wave of 1 V peak to peak from HP4192A is applied as the source signal. Four frequencies are used in the experiment, varying from 10 kHz to 10 MHz. With either single-electrode or multiple-electrode excitation, the highest SNR can be obtained with the excitation frequency of 1 MHz. There is no obvious difference in SNR between the two excitation strategies. Additionally, compared with the opposing electrode pairs, the adjacent electrode pairs show higher SNR with all selected frequency.

Figure 4.15 shows the SNR of 2 × 3 sensor with two excitation strategies. By exciting a single electrode, the SNR increases in frequency range from 10 kHz to 1 MHz, reaching a peak around 65 dB in opposing electrode pair and 70 dB in adjacent pairs. When the frequency increases to 10 MHz, an average SNR loss of 5 dB is observed in both opposing and adjacent electrode pairs. Multiple-electrode excitation shows similar performance of SNR to the single-electrode excitation. The highest SNR can be obtained with excitation frequency of 1 MHz. Similar to single-electrode excitation, the SNR decreases when the excitation frequency increases to 10 MHz, while the loss of SNR by multiple-electrode excitation is higher than single-electrode excitation, particularly for the opposite electrodes.
Figure 4.14 SNR of two-plane sensor (2 × 2 sensor array)

(a) Single-electrode excitation

(b) Multiple-electrode excitation

Figure 4.15 SNR of two-plane sensor (2 × 3 sensor array)

(a) SNRs by single electrode excitation

(b) SNRs by multiple electrodes excitation
To achieve a high SNR, 1 MHz is selected as the excitation frequency in the experiment. As the $2 \times 2$ sensor can only provide 6 independent measurements for either cross-sectional or longitudinal view, the two-plate ECT sensor with $2 \times 3$ sensor array is used in the experiment, allowing two measurement modes to represent the 4 or 6-electrode sensor by combining different electrodes. To obtain a larger capacitance, multiple-electrode excitation is used. A 4-electrode two-plate ECT sensor by multiple-electrode excitation strategy can be obtained by combining electrodes in a column, e.g. electrode 1, 3, 5 or electrode 2, 4, 6. To evaluate the 6-electrode sensor, electrodes in a row can be excited simultaneously, such as electrode 1 and 2, 3 and 4. A plastic square frame was placed between two sensor plates, ensuring a distance of 25 mm between them. Blu-Tack with permittivity of 7 is used as the testing object(s). Blu-Tack is reshaped into multiple rods with a diameter of approx. 6 mm and length of 25 mm. Those rods are placed in the square frame to form different distributions. A sensitivity matrix with 3600 pixels is calculated by COMSOL.

Figure 4.16 shows the normalised capacitances and the reconstructed images by 4-electrode and 6-electrode sensors (based on $2 \times 3$ sensor array) by multiple-electrode excitation. With the 4-electrode sensor, good images are obtained with stratified distributions, particularly for the image reconstructed by Landweber iteration. With a single bar placed in the centre of the image area, LBP gives a blur and low contrast image. The contrast between testing object and background is enhanced by Landweber iteration. With multiple objects, the 4-electrode sensor gives a low resolution image and cannot separate the objects, giving an unclear boundary between them.

The image resolution is improved by increasing the number of electrodes, and hence the independent capacitance measurements. Images shown in Figure 4.16 (d) are generated by the 6-electrode sensor with multiple-electrode excitation. With the stratified distributions, the increase in the number of electrodes does not give obvious improvement in image quality. The image distribution of stratified 1 by 6-electrode sensor is even worse than that by 4-electrode sensor. However, with single and multiple objects in the sensing area, the images by 6-electrode sensor presents higher image resolution and larger contrast between the testing object(s) and the background. In accordance with the simulation results, high sensitivity exists nearby the electrodes, and hence two bars placed near the electrodes can be seen clearly in the image (Two bar 2).
while the bars placed in the centre (Two bar 1) cannot be shown clearly. As shown in Figure 4.16 (d), the 6-electrode sensor can identify 3 and 4 objects near the electrodes.

(a) Normalised capacitances by 4-electrode sensor

(b) Normalised capacitances by 6-electrode sensor

(c) Images by 4-electrode sensor

(d) Images by 6-electrode sensor

Figure 4.16 Normalised capacitances and reconstructed images
The experimental results show satisfactory images in accordance with the simulation results. Similar to the simulation, more electrodes give more independent capacitance measurements and hence enhance the ability to separate multiple objects, particularly the objects near the electrodes. Unlike simulation, the experimental results show that the increase in the number of electrodes causes distortion to the image of stratified distribution.

4.2 Generating tooth image

To show the feasibility of visualising the tooth surface by ECT, 2D and 3D simulation models were built in COMSOL 3.5 and in-vitro experiment was carried out based on a human premolar. One challenge of applying ECT to medical imaging is the frequency-dependent permittivity of the organic tissues. Generally, the permittivity of bio-tissues decreases as the excited frequency increases (Gabriel et al., 1996b). According to Rivas et al. (2008), the permittivity of dentine shows dramatic change with the excited frequency, dropping from a high value in order of $10^4$ at 10 Hz to a very low value in order of 10 at 1 MHz and remaining at approx. 10 when the frequency further increases. As a reference, the permittivity of water retains 80 in the frequency range, which is considered as the highest permittivity in common industrial material.

4.2.1 Configuration of simulation models

Figure 4.17 shows 2D and 3D simulation models of the $2 \times 2$ and $2 \times 3$ sensors. For each sensor array, two 2D FEM models are built, referring to the cross-sectional and longitudinal view. The design of the models, including the size of electrode and shielding layers, is in accordance with the configuration of the fabricated sensors that were described in Chapter 3. The distance between the parallel sensor plates is fixed to 20 mm in all models. In the model of cross-sectional view, the sensing area is 21 mm in length and 20 mm in depth, surrounded by 4 electrodes. The overall size of the sensing area is $18 \times 20 \text{ mm}^2$ in the longitudinal view model. The electrodes are excited in turn by a source of 1 V with 1 MHz. The permittivity of the tooth is set to 10, which is the permittivity of tooth dentine when frequency is over 1 MHz (Rivas et al., 2008). In the 3D models, as shown in Figure 4.17 (b), the distance between two plates is 20 mm, which is the same as the 2D models. To consider the fringe effect, the sensing area in 3D models is extended to 100 mm in depth and 40 mm in width.
4.2.2 2D models

To simplify the simulation, a rounded rectangular/square object was used to represent a tooth in 2D models. The object is given an overall size of \(8 \times 8\) mm\(^2\) for the cross-sectional view model and \(8 \times 15\) mm\(^2\) for the longitudinal view model, as shown in Figure 4.18 and Figure 4.19 respectively. In the 2D simulation models, the fringe effect is ignored. Three tooth models with different root length are located in 5 cross-sectional positions. Consequently, 5 positions in cross-sectional view and 3 positions in longitudinal view are investigated. Figure 4.18 shows the cross-sectional image of \(2 \times 2\) and \(2 \times 3\) sensors as they share the same model for the cross-sectional view, representing the normalised permittivity distribution in the sensing area. For the model of the cross-sectional view, there is no difference in the model design with different root length as it presents in 2D. To generate images, the image area is divided into \(60 \times 60\) grids, resulting in 3600 pixels. The LBP and projected Landweber iteration are used for image reconstruction.

![Figure 4.17 Simulation models of dental ECT sensors](image)

(i) Cross-sectional model

(ii) Longitudinal model

\((2 \times 2)\)

(iii) Longitudinal model

\((3 \times 2)\)

(a) 2D models

(i) \(2 \times 2\) sensor

(ii) \(2 \times 3\) sensor

(b) 3D models

Figure 4.17 Simulation models of dental ECT sensors
According to Figure 4.18, with either LBP or Landweber iteration, a high image contrast can be obtained when the tooth is close to the electrode (see Position V). By comparing the two reconstruction methods, an image by Landweber iteration shows higher image contrast and larger correlation coefficient. The largest enhancement in correlation coefficient can be achieved when the tooth placed in central position (Position I).

Figure 4.19 shows images by the models of longitudinal view, where the three tooth models are labelled as ‘Short’, ‘Normal’ and ‘Long’, with root length of 10 mm, 16 mm and 20 mm, respectively. The tooth models with different root length are placed in three positions, corresponding to the positions in a cross-sectional view (note that Position I, II and III share the same model of longitudinal view). Similar to the model of cross-sectional view, a high image contrast is obtained when the tooth is placed close to the electrodes. Comparatively, the model with $2 \times 3$ sensor array provides higher image contrast than the $2 \times 2$ sensor. With either $2 \times 2$ or $2 \times 3$ sensor, the reconstructed images can differentiate the ‘Short tooth’ from the ‘Normal tooth’ and the ‘Long tooth’. However, it is difficult to verify the root length by image with the models of ‘Normal tooth’ and ‘Long tooth’. According to Figure 4.19, Landweber iteration has no apparent improvement to show the different root length compared with LBP. Additionally, with $2 \times 3$ sensor array, noise appears in the gap between the electrodes, shaping as an uncompleted ring and being enhanced by Landweber iteration, particularly in the model of ‘Normal tooth’ and ‘Long tooth’.
Table 4.4 lists the correlation coefficient, corresponding to the images of longitudinal view in Figure 4.19. With LBP, the tooth with longer root length shows larger correlation coefficient. By comparing two reconstruction methods, LBP presents higher correlation coefficients than Landweber iteration with the 2D longitudinal view image. As indicated in Table 4.4, the highest correlation coefficient can be achieved with Position III of the longitudinal view image with all tooth models, corresponding to Position V in cross-sectional view, where the tooth is placed near the electrodes.
In summary, images generated by 2D models show the possibility of generating cross-sectional and longitudinal views of a single tooth by a two-plate ECT sensor. A sensor with either $2 \times 2$ or $2 \times 3$ array shows highly asymmetry of sensitivity due to the open structure, resulting in a high image contrast with the tooth near the electrodes and low contrast when the tooth is far away. Landweber iteration improves the image quality of the cross-sectional view, but it is not constructive for generating longitudinal images. A sensor with $2 \times 3$ array provides higher image contrast in longitudinal view to some extent, due to the larger number of electrodes. However, neither of the sensor arrays can differentiate between the ‘Normal tooth’ and the ‘Long tooth’ by the image based on 2D models.

Table 4.4 Correlation coefficient of the 2D longitudinal image

<table>
<thead>
<tr>
<th>Correlation coefficient</th>
<th>Short tooth</th>
<th>Normal tooth</th>
<th>Long tooth</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>I</td>
<td>II</td>
<td>III</td>
</tr>
<tr>
<td>$2 \times 2$ LBP</td>
<td>0.66</td>
<td>0.62</td>
<td>0.82</td>
</tr>
<tr>
<td>Landweber</td>
<td>0.62</td>
<td>0.61</td>
<td>0.82</td>
</tr>
<tr>
<td>$2 \times 3$ LBP</td>
<td>0.73</td>
<td>0.72</td>
<td>0.87</td>
</tr>
<tr>
<td>Landweber</td>
<td>0.72</td>
<td>0.72</td>
<td>0.86</td>
</tr>
</tbody>
</table>

4.2.3 3D models

Unlike the 2D models that ignores fringe effect, 3D models extend the sensing area to $100 \text{mm} \times 40 \text{mm} \times 20 \text{mm}$. A cube and a cone are combined to form a 3D tooth model, representing the crown part and the root part respectively. The cube that represents the crown part is $7 \text{mm}$ in depth and $8 \text{mm} \times 8 \text{mm}$ in cross-section. Similar to 2D models, 3 tooth models with different root length are used, labelled as ‘Short tooth’, ‘Normal tooth’ and ‘Long tooth’ with the root length of $10 \text{mm}$, $16 \text{mm}$ and $20 \text{mm}$, respectively.

To generate cross-sectional and longitudinal images, multiple-electrode excitation is used. For each tooth model, 5 positions are investigated, corresponding to the positions used in 2D models. Figure 4.21 and Figure 4.22 shows the cross-sectional and longitudinal images with $2 \times 2$ sensor array by LBP and Landweber iteration, respectively. Compared with the image by 2D model, the image by 3D model presents
lower image contrast. The image contrast can be slightly improved by Landweber iteration, except Position V.

Figure 4.20 3D models with single tooth

Figure 4.21 Images by LBP with multiple-electrodes excitation (2 × 2 sensor array)
Figure 4.22 Images by Landweber and 3D models with multiple-electrodes excitation

(2 × 2 sensor array)

From Figure 4.21 (a), the model for 2 × 2 sensor enables to identify the 5 cross-sectional positions. In either cross-sectional or longitudinal view, the reconstructed image shows higher image contrast with longer root length. The image contrast can be improved by Landweber iteration in the cross-sectional image but the contrast reduces in longitudinal images. Due to the conical shape, the longitudinal image by 3D models shows shorter root length than that by 2D models. Neither LBP nor Landweber iteration can differentiate the ‘Normal tooth’ and ‘Long tooth’ based on the longitudinal image, which agrees with the results by 2D models.

Figure 4.23 and Figure 4.24 shows images with 3D 2 × 3 sensor model by LBP and Landweber iteration, respectively. The resolution of the longitudinal image is improved by increasing the number of electrodes, particularly in the Position V, where the tooth is
close to the electrodes. Unlike \( 2 \times 2 \) sensor or the 2D model of \( 2 \times 3 \) sensor that cannot differentiate the ‘Normal tooth’ from the ‘Long tooth’, the 3D \( 2 \times 3 \) model can separate the two tooth models by images, although the reconstructed image cannot present a correct root length. Apparently, the image contrast is enhanced by increasing the length of the root and the root length of ‘Long tooth’ model is slightly longer than the ‘Normal tooth’. However, when the image reconstruction is implemented by Landweber iteration, the longitudinal images based on 3D \( 2 \times 3 \) model suffer from noise, shaping as a half ring near the electrode, similar to the images in 2D models.

Figure 4.23 Images by LBP and 3D models with multiple-electrodes excitation (\( 2 \times 3 \) sensor array)
Figure 4.24 Images by Landweber and 3D models with multiple-electrodes excitation (2 × 3 sensor array)

Table 4.5 lists the correlation coefficient of the 3D models based on Eq. 4.2. Although for the cross-sectional image, the correlation coefficient is improved by using Landweber iteration, Landweber reduces the correlation coefficient of the longitudinal image in most cases. Only in Position V based on 2 × 3 sensor, the correlation coefficient is slightly enhanced. In Figure 4.23 and Figure 4.24, the image contrast increases by tooth model with a longer root length, but the longer length of root does not achieve higher correlation coefficient. The highest correlation coefficient can be obtained by the ‘Normal tooth’ model.

Theoretically, a 3D model, which tallies with the actual situation, gives more details and higher accuracy than a 2D model. From the reconstructed images, the image contrast by the 3D model is lower than that by the 2D model, due to the funnel-like shape of the 3D
tooth model. With the ‘Long tooth’, image generated by 3D model shows a shorter root length than 2D. A sensor with $2 \times 3$ array has superior performance in a longitudinal view image than $2 \times 2$ sensor, because the number of electrodes increases. In general, with either a 2D or 3D model, Landweber iteration improves the image quality for a cross-sectional image but it reduces the image quality when generating a longitudinal image in most cases.

Table 4.5 Correlation coefficient of the 3D models

<table>
<thead>
<tr>
<th>Correlation coefficient</th>
<th>(a) Short tooth</th>
<th></th>
<th></th>
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<td>V</td>
<td>I</td>
<td>II</td>
<td>III</td>
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<tr>
<td>Cross-sectional</td>
<td>LBP</td>
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<td>0.45</td>
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<td>0.52</td>
<td>0.63</td>
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<td>0.51</td>
<td>0.41</td>
<td>0.50</td>
<td>0.58</td>
<td>0.65</td>
<td>0.52</td>
</tr>
<tr>
<td>Longitudinal</td>
<td>LBP</td>
<td>0.63</td>
<td>0.66</td>
<td>0.62</td>
<td>0.60</td>
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<td>0.74</td>
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<td>0.62</td>
<td>0.64</td>
<td>0.54</td>
<td>0.60</td>
<td>0.66</td>
<td>0.71</td>
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<table>
<thead>
<tr>
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<tr>
<td>Cross-sectional</td>
<td>LBP</td>
<td>0.67</td>
<td>0.43</td>
<td>0.48</td>
<td>0.52</td>
<td>0.55</td>
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<tr>
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<td>0.48</td>
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<td>0.61</td>
<td>0.66</td>
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<tr>
<td>Longitudinal</td>
<td>LBP</td>
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<td>0.44</td>
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<td>0.67</td>
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<table>
<thead>
<tr>
<th>Correlation coefficient</th>
<th>(c) Long tooth</th>
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<td>III</td>
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<tr>
<td>Cross-sectional</td>
<td>LBP</td>
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<td>0.49</td>
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<tr>
<td></td>
<td>Landweber</td>
<td>0.61</td>
<td>0.51</td>
<td>0.50</td>
<td>0.56</td>
<td>0.61</td>
<td>0.66</td>
<td>0.53</td>
</tr>
<tr>
<td>Longitudinal</td>
<td>LBP</td>
<td>0.51</td>
<td>0.65</td>
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<td>0.52</td>
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<td>0.35</td>
<td>0.37</td>
<td>0.66</td>
<td>0.51</td>
<td>0.49</td>
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</tbody>
</table>

4.2.4 In-vitro experiment

An initial in-vitro experiment was carried out with the fabricated flexible printed circuit board (FPCB) sensors by means of the impedance analyser based ECT system, to image
a single tooth. A sine wave of 1 V is applied by the impedance analyser (HP4192A) as the source. In principle, higher frequency results in a lower permittivity of the tooth dentine (Rivas et al., 2008). However, with the impedance analyser based ECT system, the best SNR can be achieved when the excitation frequency is 1 MHz, which is discussed in previous sections. Two frequencies, 1 MHz and 10 MHz, are selected in the experiment. Parallel-circuit mode is selected as the measurement mode for the impedance analyser.

![Premolar and XYZ stand](image)

Figure 4.25 Premolar for in-vitro experiment and XYZ stand

To compromise the image speed and accuracy, ‘Normal’ mode is used as the default speed mode. Figure 4.25 (a) shows the human premolar used in the in-vitro experiment. The premolar has an irregular conical shape with 2 root canals. The total length of the premolar is 21 mm with a crown of 6 mm and a root of 15 mm. Figure 4.27 shows the in-vitro experiment with the single premolar in 5 positions, controlled by a XYZ stand shown in Figure 4.25 (b). With all positions, the cementoenamel junction (CEJ) of the premolar is located at the top edge of the sensor, thus the length of the premolar in the sensing area is 15 mm. A frame is used to fix the distance between two plates to 25 mm.

According to the simulation results, to obtain higher image quality, Landweber iteration is used for obtaining a cross-sectional image and LBP for a longitudinal image. From Figure 4.26 (a), there is no obvious difference in reconstructed images between the two excitation frequencies with a 2 × 2 sensor. Only in Position III, where the premolar is placed on the right-hand side, noise shaped as a half circle is shown in the left-hand side when an excitation frequency of 10 MHz is used. With an excitation frequency of 1 MHz, the 2 × 3 sensor array provides promising results, showing higher image contrast.
and clearer border between the premolar and the background compared with 10 MHz. By comparing Figure 4.26 (a) and (b), the sensor with the $2 \times 3$ array can provide superior results for longitudinal view compared with the $2 \times 2$ sensor. With the $2 \times 3$ sensor array, the length of the root shown in the reconstructed image is close to the actual length of the premolar.

Figure 4.26 Images by in-vitro experiments with single premolar with different positions.
Figure 4.27 shows the normalised capacitance, using the results from the 3D model as a reference. The ‘Normal tooth’ 3D model with multiple-excitation measurement strategy is selected, corresponding to the actual length of the premolar. Due to the unavoidable noise from the outer environment and the stray capacitance from the instruments and cables, the normalised capacitance by the experiment cannot exactly fit the simulation results. Comparatively, with an excitation frequency of 1 MHz, the normalised capacitance is closer to the simulation results than 10 MHz.

Figure 4.27 Normalised capacitances of in-vitro experiment and 3D simulation

Figure 4.28 shows the reconstructed images of the premolar in a dental impression material, i.e. Aquasil Heavy-Polysiloxane (AHP). The material can be used as the medium between the sensor plates and jaw, to ensure the sensor plates placing in parallel and to avoid the air gap between the sensor and the gum that would affect the resultant images significantly. To investigate the relative permittivity of the AHP, a simple parallel capacitor is used. The capacitor consists of a plastic block with a central hollow of 15 mm in diameter and two square copper sheets with a size of 25 mm × 25 mm. The sample AHP is shaped by the central hollow of the plastic block. Capacitance is measured with the solidified AHP block by the impedance analyser.
HP4192A. By measuring the plastic block with and without the AHP sample, the permittivity of the plastic block can be cancelled. The relative permittivity can be calculated by the area of the cross-section of the hollow and the distance between the parallel copper sheets based on Eq. 2.1.

![Figure 4.28 Images of premolar in Aquasil Heavy-Polysiloxane (AHP)](image)

**Figure 4.28 Images of premolar in Aquasil Heavy-Polysiloxane (AHP)**

Figure 4.29 shows the derived relative permittivity of the AHP corresponding to the excitation frequency varying from 1 kHz to 10 MHz. The relative permittivity of the AHP remains 3.2 from 1 kHz to 1 MHz and suddenly increases to 4.1 when the excitation frequency approaches 10 MHz, as a reference the permittivity of polysiloxane is around 2.3 to 2.8 (Mark, 2009). In Figure 4.28, the premolar is fixed in the centre of the AHP block with an overall size of 24 mm × 15 mm × 27 mm. Landweber iteration is used for the cross-sectional image and LBP is for the longitudinal images. From the reconstructed image, the region of the premolar shows higher grey level, confirming that the permittivity of the premolar is higher than the AHP.

![Figure 4.29 Relative permittivity of AHP](image)
In summary, as an initial exploration of ECT for medical imaging, the research focuses on visualising a single first premolar. The results from simulation models and experiments are consistent, showing the possibility of using ECT to generate images of a single tooth with different positions and with different root length. By comparing the two sensor arrays, the $2 \times 3$ sensor performs a higher resolution image of longitudinal view, because of the larger number of electrodes compared with the $2 \times 2$ sensor. Landweber iteration is adopted for cross-sectional image and LBP is for longitudinal image, to obtain high image quality.

4.3 Image registration and fusion

Normally, an image from an imaging technique contains limited information. Image fusion is the process of combining relevant information from two or more images into a single image with more information. One typical application of image fusion is X-ray and MRI for diagnosis. MRI can provide both an anatomical and functional image used for cancer detection. Due to the poor spatial accuracy, MRI is commonly fused with CT images to obtain good accuracy (Wong and Bishop 2008). Similarly, although ECT provides a real time image, the resolution of the image is low, which cannot provide the inner structure of the tooth and hence the position of the root canal. To improve the image quality, radiograph of high resolution can be used as a reference image and fused with the ECT images. By the original dental radiograph, it is easy to extract teeth from the background by segmentation methods, e.g. thresholding. However, the common segmentation method usually fails to discriminate teeth from bones due to the similar intensity between them, thus it is difficult to register the ECT image to the target tooth directly from the original radiograph. To overcome this problem, image pre-processing is required to enhance the image before image registration and fusion.

4.3.1 Image enhancement

Image enhancement is a process of adjusting a digital image and converting an original image into an enhanced image, which is more suitable for display or further analysis. The purpose of image enhancement is to remove noise, brightening an image and identifying the key features. It is a significant pre-processing procedure for further registration and fusion. Many image enhancement methods have been investigated. In general, these methods can be subdivided into enhancement in the spatial domain and in
the frequency domain (Kenney et al., 2001). Spatial domain enhancement manipulates pixel intensity directly, including histogram equalisation, median filter, threshold-based method, and top-hat and bottom-hat filter (Analoui, 2001a; Said et al., 2004). In the frequency domain method, the Fourier transform of the original image is modified using a filter, such as a low pass, high pass, band pass and notch (Analoui, 2001b).

**Histogram equalization**

Histogram equalization is a widely used method for image enhancement to adjust the image using a histogram. This method is to increase the global contrast of images by transforming the values in an intensity image, generating an image with its histogram approximately matching a specified histogram (Analoui 2001a).

The upper left figure in Figure 4.30 is an original histogram of the bitewing view of a dental radiograph. It is obvious that the histogram of the original histogram concentrates in the middle intensities. Histogram equalisation helps to effectively spread out the concentrated frequent intensity values to enhance image. The enhanced histogram and the resultant image is shown in the lower figures in Figure 4.30.

**Median filtering**

The median filter is a nonlinear digital filtering method in the spatial domain, normally used to reduce noise (Kenney et al. 2001). The median filter uses a pixel with its neighbouring pixels to determine a new grey scale for the pixel. A median value is calculated by comparing the value of chosen pixels and selecting the middle pixel value. The median value then replaces the original value of the central pixel. By replacing the original value to the median value, the median filter can effectively reduce noise, particularly for the salt-and-pepper noise that represents as dark pixels in bright regions or bright pixels in dark regions.
Threshold-based method

Threshold-based method is a method that is normally used for image segmentation. It is also used for image enhancement in some applications (Pal and Pal 1993). During the threshold process, a threshold is selected and each pixel in an image is compared with the threshold. If the grey scale of the selected pixel is higher than the threshold, the target pixel is given a value of ‘1’, showing a white colour, while the background pixel, whose grey scale is lower than the threshold value, is given by ‘0’ with black colour. A resultant binary image can be created by colouring each pixel either white or black.

Butterworth filter

The Butterworth filter is a signal processing filter in frequency domain. In this method, an original image is presented in the frequency domain using Fourier transform. The resultant image is then multiplied by a filter and transforms back to a spatial domain image by Fourier inversion, generating an enhanced image (Analoui 2001b). The concept of a blurred image is the lack of its high frequency components. The detail in the image is lost if the low frequency component is missing. On the contrary, if the high frequency component is increased, the image can be sharpened, resulting in an image with clearer boundary. To keep detail together with a sharpened boundary, it is necessary to keep both high and low frequency components, thus a Butterworth band
rejection filter (BRF) is preferred. Figure 4.31 shows the enhanced image presenting a clear boundary of the tooth. However, part of inner structure of the tooth is filtered.

![Original Image](image1.png) ![Image in frequency domain](image2.png)  
![BRF](image3.png) ![Filtered Image](image4.png)

Figure 4.31 Image enhancement using Butterworth band rejection filter

**Top-hat and bottom-hat filter**

Said *et al.* (2004) used a top-hat and bottom-hat filter to enhance dental radiographs. The results indicated that the top-hat and bottom-hat filter benefits to enhance teeth region apart from the air and bones, using as the pre-process before image segmentation (Said *et al.*, 2004). In mathematical morphology, top-hat filter is used to extract small element and details from given images, normally used for light objects on a dark background. On the contrary, the bottom-hat filter is to remove objects from an image, used for dark objects on a light background. To enhance the image, both filters are used on the original radiograph. The enhancement process is

1. Pre-enhancement by using histogram equalisation method
2. To generate top-hat and bottom-hat filtered images separately based on the pre-enhanced image
3. To sum the top-hat filtered image and the original image
4. To subtract the bottom-hat filtered image from the results of (3).

In Figure 4.32, the gum/bone and background areas can be suppressed by the top-hat and bottom-hat filter. Then the resultant image is processed for segmentation by a
threshold-based method and converted into a binary image for the image registration. By comparing thresholding I and II in Figure 4.32, the later image, which is the thresholded image of the enhanced image using top-hat and bottom-hat filter, has superior performance of extracting tooth from the background and bone than the former one which is segmented from the pre-enhanced image.

![Image enhancement using top-hat and bottom-hat filter](image)

Figure 4.32 Image enhancement using top-hat and bottom-hat filter

Figure 4.33 compares the original dental radiograph with the enhanced images by the enhancement methods introduced above. The original image has low contrast and suffers from noise from surrounding bones and tissues. Compared with the original image, the enhanced images have higher image contrast and clearer boundary of the teeth. By comparing the five image enhancement methods, top-hat and bottom-hat filtering method has the best performance with eliminated regions of bones and background, retaining the inner structure of the tooth simultaneously. The threshold-based method cannot show the inner structure of the tooth, however, it provides a clear boundary of the tooth, extracting teeth from the background.

A region of interest (ROI) is selected to identify the target tooth, as shown in Figure 4.34. The ROI is enhanced by a proper image enhancement method (top-hat and bottom-hat filter), producing an image of enhanced tooth region and eliminated background. The target tooth is extracted by threshold-based method and prepared for registration and fusion.
4.3.2 Image registration and fusion

The purpose of image registration is to overlay two or more images in place. Image registration can be classified into feature-based method and intensity-based method (Maintz and Viergever, 1998). A feature-based method is used to find correspondence between image features such as points, lines and contours. With the feature by feature correspondence between images, a transformation is then determined to match the target image to a reference image. Different from the feature-based method that based on a point-to-point or line-to-line matching, intensity-based method focuses on entire images or sub-images to compare intensity patterns in images. To register a blurred image with a high resolution image, the intensity-based method is more suitable, because it is difficult to find the corresponding point or line from the two images.
Due to low resolution image generated by ECT, a conventional medical registration method is not appropriate. A simple intensity-based registration method is used based on a Major-axis method. The principle of this method is to find the centre of gravity and the major axis of the reference image and target image, matching the two centres to register the position and rotating the reference image in place based on the major axis. The method can be used when the target object has a major axis, thus it is not appropriate of using the Major-axis based method on a circle or square shape object.

In the Major-axis based image registration, the target image (i.e. ECT image) and reference image (i.e. X-ray image) are both enhanced and segmented into binary images. With the binary image, the target is extracted and its centre of gravity can be found. An ellipse can be drawn to encircle the object and the major axis of this ellipse is identified as the major axis of the target.

Figure 4.35 shows two types of teeth with their centre of gravity and major axis. The ‘H’ type tooth is the tooth with more than one root canal, normally is the molars. The ‘I’ type tooth represents the incisor or canine, which normally has one root canal. For image registration, the centre of gravity is used to position the reference image and the major axis to register the two images in place.

![Figure 4.35 Different types of tooth with its centre of mass and major axis](image)

Figure 4.36 shows image registration based on the Major-axis method. Three test images with different positions and orientations were drawn to evaluate the method. Correlation coefficient between registered image and the target image is used to evaluate the registration performance, which is transformed from Eq. 4.2, expressed as...
**Correlation coefficient**

\[
Correlation \ coefficient = \frac{\sum_{i=1}^{N_p} (Image_i - \overline{Image})(Reg\_image_i - \overline{Reg\_image})}{\sqrt{\sum_{i=1}^{N_p} (Image_i - \overline{Image})^2 \sum_{i=1}^{N} (Reg\_image_i - \overline{Reg\_image})^2}}
\]  

(4.4)

where \(\overline{Image}\) and \(\overline{Reg\_image}\) are the mean of grey scale of Image and Reg\_image, representing the target image and registered image, respectively and \(N_p\) is the number of pixels.

<table>
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<tr>
<th>Major Axis</th>
<th>Correlation Coefficient factor</th>
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<td>After registration</td>
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<td>0.41</td>
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<td>Test Image 3</td>
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<tr>
<td>ECT image</td>
<td>0.44</td>
<td>0.75</td>
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</table>

Figure 4.36 Image registration with Major-axis method
As seen in Figure 4.36, the correlation coefficient has notable improvement after image registration. The ECT image is used as the target image and registered with the ‘I’ type tooth. Before registration, the correlation coefficient is about 0.44 and it increases to 0.75 after registration. The registered image is then fused with the target image by adding the two images together. A real-time fused image can be shown, providing the real-time position of the target root canal.

Once the tooth image is registered with a reference image, the ECT system is then used to guide the dentist to position the endodontic file with the registered image as a background. The following section will focus on the use of ECT for detecting metallic object.

4.4 Dental file tracing

4.4.1 Principle of imaging metallic object by ECT

Conventionally, ECT is only considered to measure dielectric materials. According to Cheng (2008), a capacitive sensor is also sensitive to conductive materials, such as metallic object. Figure 4.37 shows a parallel capacitor model with an object (permittivity of $\varepsilon_2$) inside a sensing area filled with background material (permittivity of $\varepsilon_1$). It is assumed that the object is with same width ($w$) as electrode. Obviously, with all materials, the change in capacitance is related to permittivity of filling material ($\varepsilon_1$) and the inserted area ($l_x \cdot w$).

![Parallel capacitor model](Cheng, 2008)

With dielectric material in the sensing area, the capacitance between $a$ and $b$ ($C_{ab}$) can be considered as 3 serial capacitors with one capacitor in parallel. The derived
capacitance is related to the height of the object \((d_x)\). With the sensing space filled with a low permittivity material, the capacitance is proportional to the length of the object with high permittivity in the sensing area \((l_x)\). On the contrary, if the permittivity of the object is lower than the background material, the capacitance would decrease with an increase in \(d_x\). Unlike the dielectric material, the permittivity of conductive material, e.g. metallic object, is rarely discussed in literature. For conductive materials, there are three possible statuses: floating, grounded and active (applied voltage or injected current). With an inserted floating metallic object, the distance between two opposing electrodes becomes smaller due to the thickness of the metal, leading a positive change in capacitance. On the other hand, the grounded metallic object conduces to a negative change of capacitance. The reason for that is the loss of effective area, thus the change in capacitance is inversely proportional to the inserted length of the object \((l_x)\), as shown in Table 4.6.

Table 4.6 Equivalent circuits and equations of capacitance with different materials

<table>
<thead>
<tr>
<th>Equivalent Circuit</th>
<th>Capacitance</th>
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<tr>
<td>(C_{ab_{\text{dielectric}}})</td>
<td>(C_{ab} = C_{ab_{l}} + \frac{1}{\frac{1}{C_{ac}} + \frac{1}{C_{cd}} + \frac{1}{C_{db}}} )</td>
</tr>
</tbody>
</table>
| \[\begin{align*} &= \varepsilon_0 w \left(\frac{(l - l_x)\varepsilon_1}{d_0} + \frac{l_x}{\varepsilon_1 d_1 + \varepsilon_2 (d_0 - d_x)}\right) \\
&= C_{ab} + \frac{\varepsilon_0 \varepsilon_1 w l_x d_x}{d_0} \left(\frac{\varepsilon_2 - \varepsilon_1}{d_x\varepsilon_1 + \varepsilon_2 (d_0 - d_x)}\right) \quad (4.5) \end{align*}\] |
| \(C_{ab_{\text{conductive}}}\) | \(C_{ab} = C_{ab_{l}} + \frac{1}{\frac{1}{C_{ac}} + \frac{1}{C_{db}}} \) |
| \[\begin{align*} &= \varepsilon_0 w \left(\frac{(l - l_x)\varepsilon_1}{d_0} + \frac{l_x\varepsilon_1}{d_1 + d_2}\right) \\
&= C_{ab} + \varepsilon_0 \varepsilon_1 w l_x \left(\frac{1}{d_1 + d_2} - \frac{1}{d_0}\right) \quad (4.6) \end{align*}\] |
| \(C_{ab_{\text{grounded}}}\) | \(C_{ab} = C_{ab_{l}} = \varepsilon_0 \varepsilon_1 w \frac{l - l_x}{d_0}\) |
| \[\begin{align*} &= C_{ab} - \varepsilon_0 \varepsilon_1 w \frac{l_x}{d_0} \quad (4.7) \end{align*}\] |

The active metallic object can be regarded as an inner electrode. For simplicity, the capacitance between the metallic object and the electrode is considered as a model of a
cylinder with an infinite plane. Figure 4.38 shows a cylinder with length $L$ and radius $r$ located a distance of $d$ above the infinite plane. The capacitance is derived as (Baxter 1997),

$$
C = \frac{2\pi \varepsilon_0 \varepsilon_r L}{\ln \left( \frac{d + \sqrt{d^2 + r^2}}{r} \right)} = \frac{2\pi \varepsilon_0 \varepsilon_r L}{\text{acosh} \left( \frac{d}{r} \right)}
$$  \hspace{1cm} (4.8)

Apparently, if the distance $d$ and radius $r$ are fixed, the capacitance between the cylinder and plane is in proportional to the length of cylinder $L$.

![Figure 4.38 Cylinder and plane (Baxter 1997)](image)

Figure 4.39 shows the electric field of a rounded square tooth with a dielectric or a conductive file in the centre. With a dielectric file, the electric field lines go through the tooth and file, causing distorted electric field lines at the border. The conductive file blocks part of the electric field, changes the ‘route’ of the electric field line and thus influences the capacitance.

![Figure 4.39 Electric field when electrode 1 and 3 (or electrode 1, 3, 5 for the 2 × 3 sensor array) is active](image)

As can be seen in Figure 4.39 (b) and (c), there is no electric field line within the metallic object. With a floating metallic file (boundary is set as ‘continuity’), the
electric field charges the floating metallic object, causing an induced voltage. On the other hand, with the metallic object grounded (boundary is set as ‘grounded’), some electric field lines are connected to the ground via the metallic file, forming a loss of electric field line, which is in accordance with the equivalent circuit in Table 4.6.

4.4.2 Experiment of capacitance change with different files

To investigate the capacitance change corresponding to different materials, an initial experiment is carried out with three files: plastic, Gutta-percha and endodontic file. The Gutta-percha is a common material for RCT, used as the predominant material to fill the empty space inside the root of a tooth after it has undergone endodontic therapy (Friedman et al., 1975). The endodontic file has different standard, different in diameter and length to adapt various tooth roots, which has been introduced in Chapter 1. The endodontic file of 30 mm in length (Size 30) is used in the experiment and its boundary status can be fully controlled by the multiplexer.

In the experiment, the files were inserted into the sensing space with air, deionised water and saline solution (0.9% NaCl), as shown from Figure 4.40 to Figure 4.42, respectively. The capacitance is taken by opposing electrode pairs of the fabricated sensor with the $2 \times 2$ sensor array. The XYZ stand is used to control the position of the files in 3 dimensions. In this section, only Z direction is adjusted to move the files up and down to determine the length of the file in the sensing area. The axial origin is located at the top edge of the sensor. For each measurement, capacitance is obtained with the distance between the file apex and the axial origin varying from -7 mm to 23 mm, where the negative distance denotes that the apex of the file is above the axial origin.

With files in the empty sensing area, the capacitances between all opposing electrode-pairs decrease as the distance between apex of the file and the axial origin increases. Compared with dielectric files (see Figure 4.40 (a) and (b)), the conductive file, as shown in Figure 4.40 (c) and (d), presents larger and smoother capacitance change. Due to the fringe effect, the capacitance starts to change when the apex of the file is approx. 5 mm above the axial origin. Note that the dielectric files and the floating endodontic file cause negative capacitance change, which conflicts the result in Table 4.6. The reason for this might be the fringe effect caused by ‘soft field’ of the electric field and
the small size of the file (less than 1 mm in diameter) in comparison with the sensing area.

Figure 4.40 Files in empty sensing area

Figure 4.41 shows capacitance change when sensing area is filled with deionised water. The water level is 6 mm below the axial origin, drawn as a red line in the figures. With dielectric files, the change in capacitance is not affected by the deionised water, showing fluctuating decreasing trend, similar to the capacitance change with empty sensing space. However, with the metallic file, a capacitance change is notable when the file touches the liquid. Apparently, the change in capacitance with grounded endodontic file is larger than that with a floating file.

Similar results are obtained with the files in the saline solution, as shown in Figure 4.42. The difference between saline solution and deionised water is the different conductivity. Due to the high conductivity, a sudden change in capacitance occurs when the metallic file touches the saline solution, and the capacitance becomes stable when further increases the length of the file in the sensing area (with liquid).
Figure 4.41 Files in deionised water

(a) Gutta-percha file
(b) Plastic file
(c) Endodontic file (floating)
(d) Endodontic file (grounded)

Figure 4.42 Files in saline solution

(a) Gutta-percha file
(b) Plastic file
(c) Endodontic file (floating)
(d) Endodontic file (grounded)
In the oral cavity, the tissue and blood can be considered as conductive material and dentin is not conductive. During RCT, the root, which can be dried by paper-points, would be non-conductive. However, the root apex is rich in periodontal tissue together with blood and nerves, which are conductive. With conductive material or liquid, there would be a sudden change in capacitance, which can be considered as a reference point that the endodontic file reaches the root apex or CDJ.

4.4.3 Experiment on exploring relationship between file position and capacitance

According to Figure 4.41 and Figure 4.42, the grounded endodontic file causes a significantly large and regular capacitance change corresponding to the inserted length in the sensing area. To find the relationship between the position of a grounded metallic file and the capacitance change, a plastic cube with a size of 30 mm × 20 mm × 15 mm is used for locating files. Fifteen through holes are drilled on the cube, providing 15 cross-sectional positions of the file route. The cube is placed above saline solution so that the model can be equivalent to the root with the apex, where the border between cube and saline solution denotes the position of CDJ. The top of the cube is placed in the same level with the top edge of the sensor, regarding as the axial origin. The axial position (Z direction) of the file is controlled quantitatively by the XYZ stand. The apex of the file can be located from 7 mm above the axial origin (-7 mm) to 22 mm below the axial origin.

Figure 4.43 shows the sketch of the file in the sensor with the 2 × 2 array. The structure of the 2 × 2 sensor can be equivalent to a cube. The 8 electrodes can be regarded as the 8 corners of the equivalent cube. In principle, the position of the inserted file can be located by the 8 corners. In this section, two methods are explored for positioning the file by measuring capacitance. Method I is to measure the capacitance between opposing electrodes. By this method, only the position of X and Z direction can be obtained. Method II is to measure the capacitance between the file and each electrode, by which 3D position of the file can be derived.

An idea of using capacitance difference is proposed instead of the capacitance obtained by opposing electrode pairs directly, because the fringe effect caused by the ‘soft field’ results in disturbance at the fringe, causing a premature change in capacitance by a single opposing electrode pair. According to Figure 4.40 to Figure 4.42, the capacitance between the opposite electrodes starts to change when the file apex is 5 mm above the
sensing area, which makes confusion of locating the file. By utilising the difference in capacitances by two opposing electrode pairs, the effect from the fringe can be notably eliminated.

![Diagram](image.png)

(a) Sensor with file  
(b) Equivalent cube  
(c) Object and holes

Figure 4.43 Sensor with file and object with holes

Figure 4.44 shows the change in capacitance difference with different axial position of the grounded endodontic file (dental file) by Method I with the $2 \times 2$ sensor. Obviously, the capacitance difference of lower plane electrode pairs lags behind the upper plane. The capacitance differences of upper plane electrode pairs shows a decreased gradient when the capacitance differences of lower plane electrodes start to change.

Figure 4.44 (a) shows the capacitance difference between left and right electrodes pairs, providing the position along X direction. When the file in the left-hand side holes (Position (-5, 0) or (-2.5, 0)), the capacitance difference shows positive changes while the capacitance difference shows negative changes if the file is placed in the right-hand side holes. With the grounded file in Position (0, 0), there is no obvious change in capacitance difference between left and right opposing electrode pairs, indicating that the difference of left opposing electrode pairs is equal to that of right opposing electrode pairs, confirming the central position. Note that, with all positions, a sudden capacitance drop occurs in lower plane electrode pairs when the file touches the bottom (20 mm from the axial origin).

Figure 4.44 (b) shows the capacitance difference between electrode pairs in a column, referring to the axial position of the file. With the file in left-hand side holes, capacitance measured by the electrode pairs in left plane dominates the change in capacitance while the right plane electrode pairs become domination when the file placed in the right-hand side holes. The change in capacitance difference of the
electrodes in a column forms a parabola, that the capacitance difference decreases as the
distance between the file apex and the axial origin increases from 0 to 9 mm, while the
capacitance difference of the electrode in a column increases with further increasing the
distance of the file apex from the axial origin. This parabolic change in capacitance
difference is clearer by the dominated electrode pairs. Moreover, with all positions, the
right plane electrode pairs shows larger change in capacitance difference when the file
approaching the bottom saline solution.

\[(i) \quad C_{\text{Upper plane}} = C_{1.5} - C_{2.6} \]
\[(ii) \quad C_{\text{Lower plane}} = C_{3.7} - C_{4.8} \]

(a) Change in capacitance difference between opposing electrode pairs in a row (for X-
axis position)

(b) Change in capacitance difference between opposing electrode pairs in a column (for Z-axis position)

Figure 4.44 Change in capacitance difference against different position of the grounded
dental file (Method I) \((2 \times 2\) sensor)

Similarly, Figure 4.45 shows the change in capacitance difference corresponding to
different position of the file with the \(2 \times 3\) sensor array. A sudden change in
capacitance difference is shown in the lowest plane when calculating the capacitance
difference between electrodes in a row. The change in capacitance with lower planes
lags behind the upper planes, while the gradient of the change in capacitance difference
in the upper plane electrode pairs becomes smaller when the capacitance difference in the lower plane starts to change. With the file in left-hand side through hole, a positive change in capacitance difference is shown, because the file is close to the left electrodes, causing a larger change in the measured capacitance. With the file in central position, there is no apparent change in capacitance difference.

Figure 4.45 (b) shows the change in capacitance difference of the electrode pairs in a column corresponding to different file positions. Significant change in capacitance difference can only be shown with the lower planes in the capacitance difference of the electrodes in a column. The right-hand side electrode pairs show larger capacitance change when the file reaching the bottom. Although the two fabricated sensors shows similar results of change in capacitance difference, the $2 \times 2$ shows a slightly better performance in generating a smoother and more regular change in capacitance, particularly with the capacitance difference of electrode pairs in a column compared with Figure 4.44 (b) and Figure 4.45 (b).

\( C_{\text{upper plane}} = C_{1,7} - C_{2,8} \)

\( C_{\text{middle plane}} = C_{3,9} - C_{4,10} \)

\( C_{\text{lower plane}} = C_{5,11} - C_{6,12} \)

(a) Change in capacitance difference between opposing electrode pairs in a row (for X-axis position)
(b) Change in capacitance difference between opposing electrode pairs in a column (for Z-axis position)

Figure 4.45 Change in capacitance difference against different position of the grounded dental file (Method I) (2 × 3 sensor)

The other method is to measure the capacitance difference obtained by Method II, by which 3D information can be obtained rather than 2D data only by Method I. In this method, to obtain the X-axis position, capacitance difference can be calculated by left and right electrode. The Y-axis position is calculated by capacitance difference of front and back electrodes while Z-axis position can be derived from upper and lower electrodes.

Figure 4.46 shows capacitance difference by means of Method II with the 2 × 2 sensor. Note that the capacitance difference by Method II shows an opposite change compared with Method I, because the capacitance between file and electrode increases with the increase in the length of the file in the sensing area, which is confirmed by Eq. 4.8. As shown in Figure 4.46 (a), the capacitance difference calculated by the left and right electrodes presents a negative change when the file placed in left-hand side and a positive change when the file in right-hand side. Similar to Method I, a sudden change
in capacitance difference occurs with the lower plane electrodes when the file touches the bottom, showing as a significantly high relative value (the relative change is over 100%).

Figure 4.46 (b) presents the change in capacitance difference related to the position along Z direction. With all file positions, the capacitance difference begins to change when the file reaches the effective area, forming a parabolic trend when the capacitance difference is measured by electrodes in a column. Figure 4.46 (c) shows the capacitance difference between front and back electrodes. By comparing Position (0, -3) and (0, 3), opposite change in capacitance difference can be obtained before the file touching the bottom. The reason for this change is the different dominated electrodes. The front electrodes dominate the capacitance change with the file in Position (0, -3), while in Position (0, 3), the back electrodes become domination. With the other positions, the capacitance difference between front and back electrodes shows a negative change, because the through holes of the plastic cube is slightly close to the electrodes in the back plate (Electrode 5, 6, 7, 8).

\[
\text{(i) } C_{\text{upper front}} = C_1 - C_2 \\
\text{(ii) } C_{\text{upper back}} = C_5 - C_6 \\
\text{(iii) } C_{\text{lower front}} = C_3 - C_4 \\
\text{(iv) } C_{\text{lower back}} = C_7 - C_8
\]
(b) Change in capacitance difference between file and electrode (for Z-axis position)

(i) \( C_{\text{Right front}} = C_1 - C_3 \)

(ii) \( C_{\text{Right back}} = C_5 - C_7 \)

(iii) \( C_{\text{Left front}} = C_2 - C_4 \)

(iv) \( C_{\text{Left back}} = C_6 - C_8 \)

(c) Change in capacitance difference between file and electrode (for Y-axis position)

(i) \( C_{\text{Upper right}} = C_1 - C_5 \)

(ii) \( C_{\text{Upper left}} = C_2 - C_6 \)

(iii) \( C_{\text{Lower right}} = C_3 - C_7 \)

(iv) \( C_{\text{Lower left}} = C_4 - C_8 \)

Figure 4.46 Change in capacitance difference between file and electrode (Method II)

\((2 \times 2\) sensor\)
Figure 4.47 shows the change in capacitance difference between file and electrode with \(2 \times 3\) sensor by Method II. With the \(2 \times 3\) sensor array, only the first and third plane are used and the electrodes in the middle plane (Electrode 3, 4, 9 and 10) are connected to the ground, using as earthed screens. With the middle plane grounded, the capacitance change between the file and each electrode becomes small. Compared with the \(2 \times 2\) sensor array, there is no apparent change in capacitance difference of the lower plane (the third plane). Moreover, the change in capacitance difference along the \(Y\)-axis, is in accordance with the change with \(2 \times 2\) sensor array. Although \(2 \times 3\) sensor employs more electrodes, the change in capacitance difference shows more disturbances compared with the \(2 \times 2\) sensor.

\[
\begin{align*}
(i) \ C_{\text{upper \ front}} &= C_1 - C_2 \\
(ii) \ C_{\text{Upper \ back}} &= C_7 - C_8 \\
(iii) \ C_{\text{Lower \ front}} &= C_5 - C_6 \\
(iv) \ C_{\text{Lower \ back}} &= C_{11} - C_{12}
\end{align*}
\]

(a) Change in capacitance difference between file and electrode (for \(X\)-axis position)
(ii) $C_{Left\ front} = C_2 - C_6$

(b) Change in capacitance difference between file and electrode (for $Z$-axis position)

(i) $C_{Upper\ right} = C_1 - C_7$

(ii) $C_{Upper\ left} = C_2 - C_8$

(iii) $C_{Lower\ right} = C_5 - C_{11}$

(iv) $C_{Left\ back} = C_8 - C_{12}$

(c) Change in capacitance difference between file and electrode (for $Y$-axis position)

Figure 4.47 Change in capacitance difference between file and electrode (Method II) (2 × 3 sensor)

4.4.4 Positioning the endodontic file by ECT

An in-vitro experiment has been carried out to position the endodontic file. As a preliminary stage, it can be considered that the change in capacitance by either Method I or Method II is linear to the distance of the file apex from the axial origin. With the simplified linear relationship between the measured capacitances and the file position along the $Z$-axis (axial direction), only two reference points are needed for estimating...
all other positions. The two reference points are ‘Start Point’ and ‘End Point’, where the former one is obtained when the sensing area without the file and the later one is the sudden change in capacitance of the lower plane electrodes. With the reference points, the positions can be derived by the capacitance based on the proposed linear relationship between the Z-axis position and the capacitance.

To calculate the Z-axis position, a pre-measurement is required. In the pre-measurement, capacitances from either opposing electrode pairs (Method I) or between file and electrode (Method II) are recorded until the file reaching the root apex (sudden change in capacitance). The measured capacitance is labelled as $C_{Eopp}$ and $C_{El}$, obtaining by Method I and Method II, respectively, where $i$ is the electrode number. Additionally, there are some constants needed for calculating the position, including (1) the total length of the root ($L_{root}$), measured from high resolution image, (2) the width of the gap ($L_g$), and (3) the depth of the electrode ($L_e$). The estimating procedure adopts an iterative method based on the proposed linear relationship. Table 4.7 lists the procedure of estimation by the two methods.

Table 4.7 Iterative method for estimating distance of the file apex from axial origin

<table>
<thead>
<tr>
<th>Step</th>
<th>By Method I (capacitance of opposing electrode pair)</th>
<th>By Method II (capacitance between file and electrode)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Collecting the capacitances at the ‘Start Point’ ($C_{Eopp1}$) and the one before ‘End Point’ ($C_{Eopp2}$) from the pre-measurement, where $i$ is the electrode number ($i = 1$ to $4$ for $2 \times 2$ sensor, $1$ to $6$ for $2 \times 3$ sensor)</td>
<td>Collecting the capacitances at the ‘Start Point’ ($C_{El1}$) and the one before ‘End Point’ ($C_{El2}$) from the pre-measurement, where $i$ is the electrode number ($i = 1$ to $8$ for $2 \times 2$ sensor, $1$ to $12$ for $2 \times 3$ sensor)</td>
</tr>
<tr>
<td>2</td>
<td>Calculating the Z-axis position by the first plane: $Est_{P_{El1}} = \frac{C_{Eopp1} - C_{El1}}{C_{Eopp2} - C_{El1}} \times L_{root}$ ($i = 1, 2$ for both $2 \times 2$ and $2 \times 3$ sensor, $C_{Eoppm}$ is the current measured capacitance from electrode $i$)</td>
<td>Calculating the Z-axis position by the first plane: $Est_{P_{El1}} = \frac{C_{El1} - C_{El2}}{C_{El2} - C_{El1}} \times L_{root}$ ($i = 1, 2, 5, 6$ for $2 \times 2$ sensor and $1, 2, 7, 8$ for $2 \times 3$ sensor, $C_{Elm}$ is the current measured capacitance from electrode $i$)</td>
</tr>
<tr>
<td>3</td>
<td>Average the values of first plane: $Est_{P_1} = average (Est_{P_{El1}})$</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>If $L_{root} \geq (2 \times L_g + L_e)$, find the capacitance ($C_{El1}'$) of the second plane at</td>
<td>If $L_{root} \geq (2 \times L_g + L_e)$, find the capacitance ($C_{El1}'$) of the second plane at</td>
</tr>
</tbody>
</table>
capacitance \( C_{E(t)opp1}' \) of the second plane at the position when the \( \text{Est}_P_1 \) reaching \( 2 \times L_g + L_e \) (\( i = 3, 4 \) for \( 2 \times 2 \) and \( 2 \times 3 \) sensor) from the pre-measurement, otherwise, the \( Z \)-axis position of the file is the position is \( \text{Est}_P_1 \) and stop iteration

<table>
<thead>
<tr>
<th>Step</th>
<th>Calculation</th>
</tr>
</thead>
</table>
| Step 5 | Calculating the \( Z \)-axis position by the second plane:  
\[ \text{Est}_P_{E12} = \frac{c_{E(t)oppm} - c_{E(t)opp1}'}{c_{E(t)opp2} - c_{E(t)opp1}'} \]  
\( (L_{\text{root}} - 2 \times L_g - L_e) + 2 \times L_g + L_e \), \( (i = 3, 4 \) for both \( 2 \times 2 \) and \( 2 \times 3 \) sensor) |

Calculating the \( Z \)-axis position by the second plane:  
\[ \text{Est}_P_{E12} = \frac{c_{E(t)m} - c_{E(t)1}'}{c_{E(t)2} - c_{E(t)1}'} \]  
\( (L_{\text{root}} - 2 \times L_g - L_e) + 2 \times L_g + L_e \),  
\( (i = 3, 4, 7, 8 \) for \( 2 \times 2 \) sensor and \( 3, 4, 9, 10 \) for \( 2 \times 3 \) sensor) |

| Step 6 | Average values of the first and second planes:  
\[ \text{average} (\text{Est}_P_{E12}) \] |

(Following steps for \( 2 \times 3 \) sensor only)

| Step 7 |  
If \( L_{\text{root}} \geq (3 \times L_g + 2 \times L_e) \), find the capacitance \( C_{E(t)opp1}'' \) of the third plane at the position when the \( \text{Est}_P_2 \) reaching \( 3 \times L_g + 2 \times L_e \) (\( i = 5, 6 \) from the pre-measurement, otherwise, the \( Z \)-axis position of the file is \( \text{Est}_P_2 \) and stop iteration |

If \( L_{\text{root}} \geq (3 \times L_g + 2 \times L_e) \), find the capacitance \( C_{E(t)1}'' \) of the third plane at the position when the \( \text{Est}_P_2 \) reaching \( 3 \times L_g + 2 \times L_e \) (\( i = 5, 6, 11, 12 \) from the pre-measurement, otherwise, the \( Z \)-axis position of the file is \( \text{Est}_P_2 \) and top iteration |

| Step 8 | Calculating the \( Z \)-axis position by the first plane:  
\[ \text{Est}_P_{E13} = \frac{c_{E(t)oppm} - c_{E(t)opp1}''}{c_{E(t)opp2} - c_{E(t)opp1}''} \]  
\( (L_{\text{root}} - 3 \times L_g - 2 \times L_e) + 3 \times L_g + 2 \times L_e \) (\( i = 5, 6 \)) |

Calculating the \( Z \)-axis position by the first plane:  
\[ \text{Est}_P_{E13} = \frac{c_{E(t)m} - c_{E(t)1}''}{c_{E(t)2} - c_{E(t)1}''} \]  
\( (L_{\text{root}} - 3 \times L_g - 2 \times L_e) + 3 \times L_g + 2 \times L_e \) (\( i = 5, 6, 11, 12 \)) |

| Step 9 | Average values of three planes:  
\[ \text{average} (\text{Est}_P_{E13}) \] |

Figure 4.48 compares estimated position by the two methods described in Table 4.7, with the true position as a reference. For both sensor arrays, the estimated \( Z \)-axis position by Method II presents a lower average error to the true position than by Method I. The estimated position is also founded close to the true position when the apex of the file is close to the ‘End Point’ (apex of the root), as the ‘End Point’ is one of the reference points. Although the lowest average error of estimated position can be
obtained by the $2 \times 3$ sensor based on Method II, low accuracy appears near the ‘End point’. However, from Figure 4.48, the accuracy of the estimated data is still low due to the simplified linear relationship between capacitance and position. A more complex equation for describing the relationship would be required in the future.

![Comparison of true and estimated Z-axis position](image)

**Figure 4.48** Comparison of true and estimated Z-axis position

### 4.5 Software integration

MatLab graphical user interface (GUI) provides a user-friendly interface that allows users to develop codes with built-in components and icons. MatLab contains numbers of intuitive controls, e.g. pushbuttons, list boxes, sliders, menus and test box. For each control, a call-back function can be automatically created into an ‘m file’ and user can modify it to complete various functions.

A GUI shown in Figure 4.49 has been developed for RCT with the fabricated two-plate ECT sensor of either the $2 \times 2$ or $2 \times 3$ sensor array, integrating functions of (1) instruments configuration and data acquisition, (2) image reconstruction, (3) image pre-processing and registration, and (4) dental instrument positioning. The excitation signal generated by the impedance analyser can be fully programmed by the GUI with key-in frequency and amplitude, while the measurement mode of the impedance analyser can be selected in the GUI. Both cross-sectional and longitudinal ECT images can be shown in the ‘Images’ area. Landweber iteration is used for image reconstruction and the iteration number can be adjusted in the interface. An X-ray image can be loaded from a file and pre-processed as shown in Figure 4.49. The ROI with target tooth can be
segmented and enhanced, preparing for the image registration with ECT images. The registered and fused image is then prepared for detecting the dental instrument.

Figure 4.49 GUI for RCT

Figure 4.50 shows the interfaces for positioning the endodontic file. A pre-measurement interface is created to calculate the X-axis position and to find the two reference points. The X-axis position is calculated by the slope of the capacitance difference between the left and right electrodes. In fact, it is not essential to estimate a precise X-axis position, because the reference image (i.e. radiograph) would provide an accurate position of the root canal(s). The purpose of deriving X-axis position is to find the working root canal when multiple roots exist, such as molar. The demonstration shown in Figure 4.50 is to position file apex in the testing plastic cube. The ‘End Point’ is captured when the sudden capacitance change occurs as shown in Figure 4.50 (a), simultaneously, a warning message box pops up as a warning to avoid further inserting the file. The position of the file is then calculated based on the data captured in the pre-measurement. The real-time position of the endodontic file in the plastic cube can be shown as a red spot on the background image of the plastic cube, as shown in Figure 4.50 (b).
4.6 In-vitro experiment on pig’s jaw

The first organic tissue based in-vitro experiment is carried out on a specimen from lower jaw of a baby pig, as the permanent teeth are clearly seen under the baby teeth, showing in the radiograph (70kV, 0.36s) in Figure 4.51. The pre-molar is selected as the target tooth and the target root canal is the mesiobuccal root that is the right-hand side root with endodontic file showing in the radiograph. The total length of the mesiobuccal root is approx. 24 mm and dried by the paper-point. The dental impression material-AHP is filled between the jaw and sensor plates to eliminate the air in the sensing area and to secure the two plates placed in parallel. As a preliminary trial, the fabricated FPCB sensor with the 2 × 2 array is used in the in-vitro experiment.

Figure 4.51 In-vitro experiment on pig’s jaw
Figure 4.52 shows the cross-sectional and longitudinal images of the pig’s jaw by ECT system with the $2 \times 2$ sensor at 1 MHz. The cross-sectional image is generated by Landweber iteration while the longitudinal image is reconstructed by LBP. Due to the complex structure of the jaw and high permittivity of organic tissue, ECT images can only differentiate the tissue from the impression material (AHP) but it is difficult to separate teeth from the bone.

Figure 4.53 shows the relative change in capacitance and capacitance difference of opposing electrode pairs. The XYZ stand is used to adjust the position of the endodontic file quantitatively. According to Figure 4.53 (a), there are two sudden changes in capacitance, one is when the file apex is 5 mm from the root apex (19 mm) and the other is when the file is approaching the root apex (24 mm). The capacitance remains at a bottom value when the file apex beyond the root apex. As the endodontic file is located at mesiobuccal root, which is close to electrode 1 and 3, capacitance difference between left and right electrodes shows negative change in both upper and lower plane. Unlike the results based on the plastic cube, with the organic specimen, the four opposing electrode pairs show similar capacitance change. From the capacitance difference, it is difficult to determine the sudden capacitance change. The initial in-vitro experiment with organic tissue is failed for many reasons, (1) the high permittivity of organic tissue, (2) the highly asymmetric distribution, (3) the complex structure of tooth anatomy, (4) the lack of information obtained by current 2D radiograph.
Due to the limited space in oral cavity, the ECT sensor for RCT adopts a miniature two-plate structure. According to the simulation and experimental results, the two-plate ECT sensors perform like a conventional ECT sensor, but some specific issues need to be considered. For a two-plate ECT sensor, it is recommended to set the width of the sensing area and the distance between the two plates equal to the length of the shielding layer with the open boundaries setting as ‘Electric insulation’. Due to incomplete measurement data, the sensitivity in the central region of the two-plate ECT sensor is low. More electrodes can distinguish more objects. Multiple-electrode excitation with 1 MHz is suggested to provide stable measurements for the fabricated two-plate sensor.
with either $2 \times 2$ or $2 \times 3$ electrodes by means of impedance analyser based system. The sensor with the $2 \times 3$ array performs higher resolution image in longitudinal view than the $2 \times 2$ one due to the number of electrodes. The two-plate ECT sensor enables a sensing area with special geometry and can be used for dental imaging in RCT. Specifically, the miniature size allows the sensor to be placed in the limited space in oral cavity while the two plate structure enables the sensor plates to fix on the both sides of the gum and to fit the complex geometry of the jaw. The fabricated ECT sensors are based on the FPCB technique which is space-saving and cable-free.

Two purposes of using ECT for RCT are (1) to generate a real time image of the tooth and (2) to position the endodontic file in the root canal. Based on both simulation and experimental results, tooth images for a single premolar can be generated by ECT. Due to the specific structure of the dental ECT sensor, both cross-sectional and longitudinal (side view) image can be obtained. The reconstructed images can identify the cross-sectional position of the premolar and with $2 \times 3$ sensor array the root length of the premolar can be shown more accurately.

As an ECT system can provide low resolution image only, an X-ray image showing the inner structure of the tooth is used as a reference to be registered with the real-time ECT image, aiming to obtain high accuracy and fast imaging speed. Before registration, the original radiograph is enhanced by top-hat and bottom-hat filter and segmented by threshold-based method. A simple intensity-based registration method is used. This method is to register the centre of gravity and the major axis of two images, which is suitable for the practical situation that a high resolution image is registered with a low resolution image. By registering with the high resolution image (e.g. radiograph), the image gains both high speed and high accuracy, preparing to be used as the background image for the file positioning.

ECT can not only image dielectric material but also is sensitive to grounded metallic objects. The grounded objects would block part of electric field and make capacitance drop. A sudden change in capacitance occurs when the grounded endodontic file touches the conductive solution, which is regarded as the location of the root apex (or CDJ). Two methods have been introduced to find the relationship between the file position and the capacitance. One is to measure capacitance of opposing electrodes pairs (Method I) and in the second method, the capacitance is captured between file and each
electrode (Method II). From the experimental results, the capacitance is linear to the
distance of the file apex from the axial origin by both methods. With the simplified
linear relationship, the file position along the Z-axis can be estimated by two reference
points, which are ‘Start Point’ when there is no file in the sensing area and ‘End Point’
when there is a sudden change in capacitance. The position between the two reference
points can be estimated based on the proposed linear relationship. However, this linear
relationship shows low accuracy in estimating the positions because the relationship
between the capacitance and the position is not linear due to the ‘soft field’ and fringe
effect. A more accurate method of estimating the file position can be considered in the
future, which may involve more complex function to explain the relationship between
the capacitance and the file position.

Interfaces have been built in MatLab for RCT, integrating many functions such as (1)
instruments configuration and data acquisition, (2) image reconstruction, (3) image pre-
processing and registration and (4) dental file positioning. The main interface deals with
image display and calling the sub-interface. The sub-interface is to show the registered
image as the background and to position the file apex.

A preliminary in-vitro experiment has been carried out on a specimen of pig’s lower jaw.
The resultant ECT image can separate the jaw out from AHP. However, it is difficult to
differentiate the tooth from the bone and tissue due to the high permittivity of the
organic tissue and the complex structure of the jaw. By measuring the capacitance of
opposing electrode pairs, two sudden changes in capacitance occur when the endodontic
file inserted into the mesiobuccal root of a premolar. The capacitance remains at the
lowest value when the file is inserted beyond the root apex.
Chapter 5: ECT for revision total hip replacement

5.1 Introduction

An ageing population leads to a dramatic rise of total hip replacements (THRs). Due to mechanical failure or other reasons, many patients need revision surgery, normally 10 years after replacement. According to a survey (Verteuil et al., 2008), over 80,000 THRs were performed in the UK annually while more than 10% were revision. Technically, a revision THR is more complicated than the primary one. It is a lengthy operation (a minimum of 2-3 hours) and associated with a high risk. There are 3 steps for revision THRs: (1) removal of THR components, possibly causing damage to the bone, (2) replace any bone lost and (3) place a new THR. To minimise the damage and to enhance the efficiency, it is desirable to use special equipment to navigate the milling/drilling tools. Thus, a real-time 3D imaging and accurate positioning surgery tool is demanded for an efficient operation, in attempt to reduce the risk of damaging the bone. For those purposes, a preliminary study has been carried out to develop a real-time 3D visualisation approach for the revision THR based on ECT.

To date, most 3D ECT images are 2½ D rather than 3D, because they are generated by stacking 2D images with a conventional single-plane ECT sensor, plus time as the third dimension (Wang et al. 2003). To obtain a real 3D ECT image, a traditional method employs multiple planes of electrodes to form a 3D sensor. Research into 3D ECT has been reported by literature (Li and Holland 2013, Marashdeh et al. 2008, Warsito and Fan 2001, 2003, 2005), and some typical 3D sensors have been discussed in Chapter 2. Generally, 3D images can be obtained by two methods. The first method is to generate a few independent 2D images by different planes and to interpolate into a 3D image, which is similar to the concept of the X-CT. The other method is a ‘real’ 3D reconstruction, or ‘direct 3D imaging’. In this method, the sensors take advantage of the fringe effect in the axial direction, providing 3D measurement and the image is reconstructed based on a 3D sensitivity map (Yang, 2010).
Recently, research into 3D ECT focuses on 3D reconstruction methods. Chapter 2 has introduced some model/shape based reconstruction methods to reduce the number of unknowns during 3D image reconstruction and relieve the ill-posed inverse problem. However, the complex multi-plane 3D sensor and its forward problem solved by FEM or BEM still causes long time to produce a 3D image. This chapter introduces a method for real-time 3D imaging of a metallic cylinder, referring to the revision THR. According to Sun and Yang (2013), the fringe effect is associated with the axial position of an object in an ECT sensor, resulting in a different change in capacitance and thus different ECT images. The finding indicates a possibility to use a conventional single-plane ECT sensor to obtain an accurate 3D image with prior knowledge (e.g. shape and diameter of the milling/drilling tool). With a single-plane ECT sensor, the number of independent measurements is much less than a 3D sensor, giving a possibility of real-time 3D imaging. In this chapter, a metallic rod is placed in an 8-electrode single plane ECT sensor and the effect of the metallic rod is investigated by simulation models and experiments.

### 5.2 Simulation study

In Chapter 4, the capacitance change with dielectric and conductive material has been discussed theoretically and confirmed by the results from the experimental study based on the fabricated dental ECT sensors. This section will investigate the capacitance change between dielectric material and metallic material with different boundary conditions by simulation models. To investigate the effect of the metallic object, Figure 5.1 shows 2D and 3D FEM models generated in COMSOL 3.5, where the ‘In-Plane Electric Currents’ mode and the ‘Electric Currents’ mode are chosen, respectively. In both models, the parameters are set to be the same as the experimental sensor described in Chapter 3. In the 3D model, the length at both ends of the sensor is extended by 10 cm.

A cylinder of air and a cylinder of aluminium are placed in the sensor, both of them with a diameter of 2.6 cm. For the aluminium cylinder, three boundary conditions are
considered, (1) floating (continuity), (2) grounded and (3) active (applied voltage of 1V). The space between the test cylinder and the glass wall is filled with plastic beads ($\varepsilon_r = 2.1$), which is the high permittivity material. With the prior knowledge, only cross-sectional and axial positions are needed for generating a 3D image. The cross-sectional origin is located at the centre of the image area while the axial origin is the upper edge of the electrodes. Three cross-sectional position of the cylinder are investigated, which are centre (0, 0), near centre (1, 1) and near edge (2.5, 2.5) with unit of cm. The axial position of the cylinder apex varies from 2 cm above the axial origin to 8 cm below the origin.

A sensitivity distribution is generated by COMSOL 3.5, with the sensing area filled with the low permittivity material ($\varepsilon_{air} = 1$). A sine wave is applied with a voltage of 1 V and 1 MHz. For an 8-electrode sensor, 28 independent capacitance measurements can be taken for each frame. To eliminate the effect of the standing capacitance, the sensor is calibrated and capacitance normalised based on Eq. 2.14 by filling the sensing space with high and low permittivity materials. The sensing area is divided into $100 \times 100$ mesh grids, resulting in 10,000 unknowns. To solve the ill-posed problem, both LBP and Landweber iteration are used.

Figure 5.2 shows images reconstructed by the 2D models. Compared with the floating and the grounded metallic cylinders, images of the air cylinder are close to the true distribution shown in the 2D models. The grounded metallic cylinder causes a large effective area, indicating that an ECT sensor is highly sensitive to the metallic material.
connected to the ground. On the other hand, the sensor is less sensitive to a floating metallic object and the object forms higher grey level in its effective area than the background, denoting a higher permittivity value compared to the surrounding material, which is consistent with Eq. 4.6.

Figure 5.2 Reconstructed images by 2D simulation models

Figure 5.3 compares the average normalised capacitance and the average grey level of the reconstructed images by the 2D models. As seen from Figure 5.3 (a), the normalised capacitance with the grounded metallic cylinder (green) is highly different from the other two. The normalised capacitance shows a negative value when the grounded metallic cylinder is located in the centre and near centre. With the air or the floating metallic cylinder, the change in average normalised capacitance is not sensitive to different cross-sectional positions. Note that the average normalised capacitance with
the floating metallic object is over 1, referring to a higher measured capacitance compared to the capacitance of the sensing area filled with the high permittivity material only, which is in accordance with Eq. 4.6. As shown in Figure 5.3 (b), the grey level of the grounded metallic cylinder is far less than the other two and varies with different cross-sectional positions, increasing from 0.28 in the centre to 0.44 near edge. With either air or the floating metallic cylinder, however, a high value of average grey level can be obtained when the object is located in the centre. Similar to the change in the normalised capacitance, with the cylinder of air and floating metal, the average grey level is around 0.9 and is not sensitive to different cross-sectional position.

![Comparison of normalised capacitance and grey level](https://via.placeholder.com/150)

**Figure 5.3 Normalised capacitance and grey level (2D model)**

Figure 5.4 shows the reconstructed images by the 3D models, while Figure 5.5 shows the average normalised capacitance and the average grey level corresponding to the reconstructed images. With the 3D models, the capacitance change caused by different axial position of the test cylinder is investigated. In Figure 5.4, the cylinder is placed in the centre of the image area at three axial positions: 0, 3 and 6 cm. Due to the fringe effect caused by the ‘soft field’, the capacitance starts to change when the cylinder apex is located at the axial origin, form different grey levels to represent the location of the cylinder. With an increase in the distance of the rod apex from the axial origin, the grey level of the area associated with the cylinder is reduced and the effective area by the cylinder is enlarged, particularly with the grounded metal.
According to Figure 5.5, only floating metallic cylinder causes a positive change in average normalised capacitance. With either the air or the grounded metallic object, both normalised capacitance and grey level reduce as the distance between the cylinder apex and the axial origin increases. A significant drop occurs in both normalised capacitance and grey level with the grounded metallic cylinder, corresponding to the axial distance between the cylinder apex and the axial origin varying from 0 cm to 6 cm. This significant and regular change in either normalised capacitance or grey level provides a possibility to derive the axial position of the grounded metallic object based on the reconstructed images.

With the prior knowledge (i.e. the shape and the diameter of the cylinder), the procedure of generating a 3D image is simplified to obtain the cross-sectional and axial position of the cylinder in the sensing area. The following section will focus on estimating the cross-sectional and axial position of the cylinder by capacitance measurement based on the 3D models.
5.2.1 Estimation of cross-sectional position

To obtain the cross-sectional position of the cylinder, a weighted mean method is used, which can be specified in three steps:

1. Obtain the binary image by thresholding based on Otsu’s method (Zhang and Hu, 2008)
2. Find the largest target area and segment the target from the background (to eliminate the noise)
3. Estimate the position of the target centre \((\bar{x}, \bar{y})\) by calculating the weighted mean value of the segmented image based on the following equation:

\[
\bar{x} = \frac{\sum_{j=1}^{m} \sum_{i=1}^{n} W(i,j)x_i}{\sum_{j=1}^{m} \sum_{i=1}^{n} W(i,j)}, \quad \bar{y} = \frac{\sum_{j=1}^{m} \sum_{i=1}^{n} W(i,j)y_i}{\sum_{j=1}^{m} \sum_{i=1}^{n} W(i,j)}
\]

(5.1)

where \(W(i,j)\) is the weight or grey level in the pixel \((x_i, y_i)\), \(m\) and \(n\) are equal to 100, corresponding to 100 \(\times\) 100 grid with 10,000 pixels.

Table 5.1 lists the derived cross-sectional centre of the grounded metallic cylinder with different axial positions based on the images by LBP. As mentioned in previous section, three positions are investigated: centre \((0, 0)\), near centre \((1, 1)\) and near edge \((2.5, 2.5)\) with cm as the unit. The estimated cylinder centre appears a ‘centre effect’, i.e. the centre of the cylinder shifts slightly to the centre of the sensing area. When the cylinder
is close to the edge, the derived cylinder centre shows a complex and nearly-parabolic change corresponding to the axial position of the cylinder apex. When the cylinder is half inserted in the sensor, the derived centre is close to the true position while the estimated centre shows low accuracy if the cylinder apex is located 0 or 6 cm from the axial origin. The best accuracy is achieved when the apex of the cylinder is 2 cm below the axial origin.

Table 5.1 Cross-sectional centre of the grounded metallic cylinder with different axial positions based on images by LBP

<table>
<thead>
<tr>
<th>Coordinate</th>
<th>True centre</th>
<th>Axial distance of the cylinder apex from the origin (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>0  1  2  3  4  5  6</td>
</tr>
<tr>
<td>Centre</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>0</td>
<td>0.09 0.08 0.09 0.08 0.07 0.08 0.08</td>
</tr>
<tr>
<td>y</td>
<td>0</td>
<td>0.09 0.10 0.11 0.11 0.11 0.12 0.12</td>
</tr>
<tr>
<td>Near centre</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>1</td>
<td>0.70 0.75 0.77 0.80 0.79 0.80 0.81</td>
</tr>
<tr>
<td>y</td>
<td>1</td>
<td>0.70 0.77 0.80 0.81 0.82 0.83 0.84</td>
</tr>
<tr>
<td>Near edge</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>2.5</td>
<td>1.93 2.37 2.50 2.46 2.39 2.31 2.26</td>
</tr>
<tr>
<td>y</td>
<td>2.5</td>
<td>1.94 2.37 2.48 2.44 2.36 2.29 2.24</td>
</tr>
</tbody>
</table>

Table 5.2 shows the estimated cross-sectional centre based on the images by Landweber iteration. Generally, a higher accuracy of estimation is shown when the cylinder is placed in the centre or near centre than near edge. With different axial position, the derived cross-sectional centre of the test cylinder shows higher accuracy when the apex of the cylinder is located at the axial origin (0 cm) or fully inserted into the sensor (6 cm). A low accuracy is shown at 1 cm and 2 cm.

Comparing Table 5.1 and Table 5.2, it can be seen that the Landweber iteration improves the accuracy of the estimation in the cross-sectional centre to some extent. To find a proper number of iterations, a series of iteration numbers are investigated, varying from 0 to 1500. Figure 5.6 compares the average absolute error between the estimated centre and the true centre with different number of iterations, where Figure
Table 5.2 Cross-sectional centre of grounded metallic cylinder with different axial positions based on images by Landweber

<table>
<thead>
<tr>
<th>Coordinate</th>
<th>True centre</th>
<th>Axial distance of the cylinder apex from the origin (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>0</td>
</tr>
<tr>
<td>Centre</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>0</td>
<td>0.09</td>
</tr>
<tr>
<td>y</td>
<td>0</td>
<td>0.07</td>
</tr>
<tr>
<td>Near centre</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>1</td>
<td>1.14</td>
</tr>
<tr>
<td>y</td>
<td>1</td>
<td>1.13</td>
</tr>
<tr>
<td>Near edge</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x</td>
<td>2.5</td>
<td>2.66</td>
</tr>
<tr>
<td>y</td>
<td>2.5</td>
<td>2.70</td>
</tr>
</tbody>
</table>

As shown in Figure 5.6, the increase in the number of iterations has no obvious effect on estimation of the cross-sectional centre when the cylinder is placed in the centre of the sensing area. The average absolute error remains 0.1 cm with all iteration numbers with the cylinder placed in the centre. However, the number of iterations influences the estimation of the cross-sectional position when the cylinder is placed off-centre. A significant drop in the average absolute error occurs when the number of iterations changes from 0 to 10. By the Start point, the average absolute error increases when number of iterations is larger than 10 and it becomes stable after 50 iterations. When estimated by the End point, the average absolute error reduces when 10 iterations are given. However, the error fluctuates with the increase in number of iteration, particularly for the cylinder near edge. With cylinder placed near edge, the smallest average absolute error is achieved with 500 iterations. However, a further increase in the number of iterations does not show any improvement in the accuracy of the estimation, instead, results in a bigger error. Practically, a larger number of iterations results in longer time to reconstruct an image. The time spent by 1500 iterations is 10 times more than LBP. From the Figure 5.6 (b), Landweber with 100 iterations gives an average absolute error of 0.1 cm at three cross-sectional positions. To compromise the
imaging speed and accuracy, Landweber with 10 iterations is selected to estimate the cross-sectional centre with the Start point and 100 iterations for estimating with the End point.

![Figure 5.6 Average absolute error of the estimated cross-sectional centre with different number of iterations of Landweber iteration algorithm](image)

Table 5.3 shows the estimated cross-sectional centre by the Start point and the End point. Eight cross-sectional positions are evaluated. The cross-sectional centre is estimated by the Landweber iteration algorithm with 10 iterations with the Start point and 100 with the End point, calibrated by adding an offset of 0.1 cm. Estimated by the Start point, the largest error is found in the Position IV with the absolute error is 0.13 cm. A higher accuracy of the estimated centre is achieved by estimating with the End point at all selected positions. With the End point, the average absolute error between the true and estimated centre reduces to 0.04 cm and the largest error is shown in Position VII with the value of 0.08 cm.

Cross-sectional images can be redrawn with the estimated cross-sectional centre and the prior knowledge (shape and diameter). Figure 5.7 shows the cross-sectional image with and without prior knowledge, using the true distribution as reference. After using the prior knowledge, a more accurate 2D cross-sectional image can be obtained. High proximity between the true distribution and the images with prior knowledge can be found at all axial positions when the metallic cylinder is placed in the centre and near centre.
Figure 5.7 Cross-sectional image of the grounded metallic cylinder with and without prior knowledge with different cross-sectional and axial positions

Table 5.3 Estimating cross-sectional centre by calibration

<table>
<thead>
<tr>
<th>Position</th>
<th>I</th>
<th>II</th>
<th>III</th>
<th>IV</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coordinate</td>
<td>x</td>
<td>y</td>
<td>x</td>
<td>y</td>
</tr>
<tr>
<td>True</td>
<td>0</td>
<td>0</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Estimated by Start</td>
<td>0.00</td>
<td>-0.03</td>
<td>0.96</td>
<td>0.91</td>
</tr>
<tr>
<td>Estimated by End</td>
<td>-0.01</td>
<td>-0.03</td>
<td>0.99</td>
<td>0.98</td>
</tr>
</tbody>
</table>
5.2.2 Estimation of axial position

To estimate the axial position, three methods are considered: (1) the normalised capacitance, (2) the grey level, and (3) the change in capacitance between the metallic cylinder and each electrode. Figure 5.8 shows the normalised capacitance and the grey level of the reconstructed images with a grounded metallic cylinder at 3 cross-sectional positions. The axial origin is located at the upper edge of the electrodes. The axial position varies from -2 cm to 8 cm, where the negative position denotes that the apex of the cylinder is above the axial origin. Generally, a large value of average normalised capacitance and grey level can be achieved when the grounded metallic cylinder is placed near edge, which is consistent with the results of the 2D models. To reduce the effect from different cross-sectional positions, the values are calibrated by

\[
G_{cali} = \frac{(1 - G_{ref}) - (1 - G_{mea})}{(1 - G_{ref})}
\]  

(5.2)

where \(G_{ref}\) is the reference value (either the normalised capacitance or the grey level), which is calculated from the 2D simulation model, \(G_{mea}\) is the measured value from the 3D simulation model. After calibration, the changes in the average normalised capacitance at three cross-sectional positions are nearly overlapped. A similar phenomenon is also shown in grey level, as seen Figure 5.8 (d).
Figure 5.8 Normalised capacitance and grey level of grounded metallic cylinder with different axial position before and after normalisation.

Figure 5.9 shows the calibrated normalised capacitance and grey level selected from Figure 5.8 (c) and (d), respectively, with the axial distance of the cylinder apex from the origin varying from 0 cm to 6 cm. With all cross-sectional positions, the normalised capacitance decreases with the increase in the axial distance. The grey level sharply decreases until the cylinder is half inserted in the sensor and the decrease in grey level slows down when the apex of the metallic cylinder is approaching to the end of electrodes.

Apparently, the relationship between the axial position and either normalised capacitance or grey level is not linear. A second order polynomial and a third order polynomial are used to fit the trend of the normalised capacitance corresponding to different axial position of the cylinder, while the relationship between grey level and the axial position is fitted by three curves: (1) the second order polynomial, (2) the third order polynomial, and (3) the exponential curve.
Table 5.4 and Table 5.5 list the derived axial position of the cylinder based on the normalised capacitance by the second order and the third order polynomial, respectively. Both of the polynomials are derived by 7 reference points. Generally, a higher accuracy can be achieved by the third order polynomial than the second order polynomial. However, by both polynomial, a large error is shown at the Start point (0 cm) and near the End point, i.e. over 5 cm away from the axial origin. The second order polynomial causes larger average errors but obtains higher accuracy in estimating the axial distance at the End point (6 cm). The estimated axial distance is approx. 0.1 cm less than the true distance at the End point.

Tables 5.6 to 5.8 show the estimated axial position of the cylinder based on the grey level by the second order polynomial, the third polynomial and the exponential curves, respectively. Estimated by 7 reference points, the third order polynomial shows lowest error with an average value of 2.4%, which is 2.7% less than the second order polynomial and 3.8% less than the error by the exponential curves. Although the third order polynomial obtains the smallest average error, the derived axial distance shows a large error near the End point (e.g. from 5 cm to 6 cm) and the smallest error at the End point can be achieved by the second order polynomial. Compared with the aixal position derived by the third order polynomial, the derived axial position by the exponential curve shows smaller errors near the End point, although its average error is the largest among the three fitting curves. Obviously, the accuracy of the derived
polynomial and exponential curve is associated with the number of effective reference points. The reference points normally include the Start and the End point together with the positions in between, showing characteristic changes. According to the numbers of coefficients, there are at least 4 reference points needed for deriving a third order polynomial, 3 for a second order polynomial and 2 for an exponential curve. Practically, more reference points with characteristic changes contributes to a higher accuracy of estimation.

Table 5.4 Derived axial distance of cylinder apex from axial origin by second order polynomial and average error based on normalised capacitance

<table>
<thead>
<tr>
<th>True distance</th>
<th>Axial distance of the cylinder apex from the origin (cm)</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0</td>
<td>0.3</td>
</tr>
<tr>
<td>Centre</td>
<td>0.30</td>
<td>0.48</td>
</tr>
<tr>
<td>Near centre</td>
<td>0.30</td>
<td>0.44</td>
</tr>
<tr>
<td>Near edge</td>
<td>0.28</td>
<td>0.48</td>
</tr>
</tbody>
</table>

Table 5.5 Derived axial distance of cylinder apex from axial origin by third order polynomial and average error based on normalised capacitance

<table>
<thead>
<tr>
<th>True distance</th>
<th>Axial distance of the cylinder apex from the origin (cm)</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0</td>
<td>0.3</td>
</tr>
<tr>
<td>Centre</td>
<td>0.17</td>
<td>0.38</td>
</tr>
<tr>
<td>Near centre</td>
<td>0.15</td>
<td>0.33</td>
</tr>
<tr>
<td>Near edge</td>
<td>0.06</td>
<td>0.32</td>
</tr>
</tbody>
</table>

Table 5.6 Derived axial distance of cylinder apex from axial origin by second order polynomial and average error based on grey level

<table>
<thead>
<tr>
<th>True distance</th>
<th>Axial distance of the cylinder apex from origin (cm)</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0</td>
<td>0.3</td>
</tr>
<tr>
<td>Centre</td>
<td>-0.08</td>
<td>0.20</td>
</tr>
</tbody>
</table>

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Table 5.7 Derived axial distance of cylinder apex from axial origin by third order polynomial and average error based on grey level

<table>
<thead>
<tr>
<th>True distance</th>
<th>Axial distance of cylinder apex from origin (cm)</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Centre</td>
<td>0.08 0.31 1.00 1.28 1.99 2.28 3.00 3.25 3.91 4.43 4.90 5.37 5.59</td>
<td>2.0%</td>
</tr>
<tr>
<td>Near centre</td>
<td>0.09 0.27 1.00 1.29 1.99 2.31 3.02 3.29 3.95 4.32 5.03 5.27 5.89</td>
<td>1.4%</td>
</tr>
<tr>
<td>Near edge</td>
<td>0.15 0.36 1.00 1.31 2.01 2.30 3.02 3.31 3.95 4.41 5.30 5.60 6.41</td>
<td>3.7%</td>
</tr>
</tbody>
</table>

Table 5.8 Derived axial distance of cylinder apex from axial origin by exponential curve and average error based on grey level

<table>
<thead>
<tr>
<th>True distance</th>
<th>Axial distance of cylinder apex from origin (cm)</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Centre</td>
<td>-0.08 0.15 0.87 1.18 1.97 2.30 3.12 3.42 4.14 4.70 5.16 5.58 5.77</td>
<td>8.8%</td>
</tr>
<tr>
<td>Near centre</td>
<td>-0.01 0.17 0.90 1.20 1.95 2.30 3.10 3.40 4.13 4.51 5.17 5.37 5.77</td>
<td>7.3%</td>
</tr>
<tr>
<td>Near edge</td>
<td>0.28 0.46 1.02 1.30 1.97 2.27 3.00 3.29 3.94 4.39 5.17 5.40 5.92</td>
<td>5.6%</td>
</tr>
</tbody>
</table>

Figure 5.10 shows the relative change in capacitance between the excited metallic cylinder and each electrode at different cross-sectional and axial positions. The capacitance values are calibrated by a reference capacitance when the cylinder apex is 2 cm above the axial origin. The relative change in capacitance is extremely large and shows a nearly linear trend corresponding to the axial position varying from 0 to 6 cm. The relative change in capacitance reaches a peak value and becomes stable when the cylinder apex is located over 6 cm from the origin. The relative change in capacitance between the excited metallic cylinder and electrodes is associated with the cross-sectional position. As shown in Figure 5.10 (a), the relative capacitance changes are overlapped when the cylinder is placed in the centre. When the cylinder is placed near
electrodes 3 and 4, a huge capacitance change occurs between the cylinder and either electrode 3 or 4, which is approx. 6 times higher than the change between the cylinder and the other electrodes. The relative change in capacitance is averaged in Figure 5.10 (d). The green curve, referring to the relative change in capacitance when the cylinder is located near edge, is much greater than the other two curves.

![Figure 5.10 Relative change of capacitance between exited metallic cylinder and each electrode with different cross-sectional and axial positions](image)

With the proposed linear relationship between the relative change in capacitance and the axial position of the metallic cylinder, it is easy to estimate the axial distance of the cylinder apex from the the axial origin. In princile, the coefficents of a linear function can be easily derived by two known points.

Table 5.9 lists the aixal position calculated by the linear function, which is derived from the reference points of the relative capacitance changes at 1 cm and 2 cm. High accuracy can be obtained near the reference points, while the accuracy becomes lower when the derived axial distance is far away from the reference points. The estimated axial position becomes smaller than the true position when the axial distance of the
cylinder apex from the axial origin is larger than 4 cm. The driven End point is approx. 0.8 cm smaller than the true position. A large average error is shown when the cylinder is placed in the centre. The accuracy can be improved by choosing different reference points. Table 5.10 shows the derived axial distance by the linear function calculated by two reference points with its axial position at 1 cm and 5 cm, by which the accuracy of estimation at the End point can be improved. However, a large average error occurs when the cylinder is placed off-centre.

Table 5.9 Derived axial distance of cylinder apex from axial origin by linear function and average error (reference points at 1 cm and 2 cm)

<table>
<thead>
<tr>
<th>True distance</th>
<th>0</th>
<th>0.3</th>
<th>1</th>
<th>1.3</th>
<th>2</th>
<th>2.3</th>
<th>3</th>
<th>3.3</th>
<th>4</th>
<th>4.3</th>
<th>5</th>
<th>5.3</th>
<th>6</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Centre</td>
<td>0.08</td>
<td>0.37</td>
<td>1.00</td>
<td>1.31</td>
<td>2.00</td>
<td>2.30</td>
<td>2.99</td>
<td>3.22</td>
<td>3.82</td>
<td>4.12</td>
<td>4.61</td>
<td>4.79</td>
<td>5.15</td>
<td>5.4%</td>
</tr>
<tr>
<td>Near centre</td>
<td>0.07</td>
<td>0.27</td>
<td>1.00</td>
<td>1.29</td>
<td>2.00</td>
<td>2.30</td>
<td>2.98</td>
<td>3.22</td>
<td>3.79</td>
<td>4.10</td>
<td>4.60</td>
<td>4.77</td>
<td>5.11</td>
<td>4.8%</td>
</tr>
<tr>
<td>Near edge</td>
<td>0.05</td>
<td>0.26</td>
<td>1.00</td>
<td>1.29</td>
<td>2.00</td>
<td>2.29</td>
<td>2.97</td>
<td>3.22</td>
<td>3.73</td>
<td>4.13</td>
<td>4.74</td>
<td>4.91</td>
<td>5.28</td>
<td>4.3%</td>
</tr>
</tbody>
</table>

Table 5.10 Derived axial distance of cylinder apex from axial origin by linear function and average error (reference points at 1 cm and 5 cm)

<table>
<thead>
<tr>
<th>True distance</th>
<th>0</th>
<th>0.3</th>
<th>1</th>
<th>1.3</th>
<th>2</th>
<th>2.3</th>
<th>3</th>
<th>3.3</th>
<th>4</th>
<th>4.3</th>
<th>5</th>
<th>5.3</th>
<th>6</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Centre</td>
<td>-0.02</td>
<td>0.30</td>
<td>1.00</td>
<td>1.34</td>
<td>2.11</td>
<td>2.45</td>
<td>3.21</td>
<td>3.47</td>
<td>4.13</td>
<td>4.46</td>
<td>5.00</td>
<td>5.20</td>
<td>5.60</td>
<td>3.6%</td>
</tr>
<tr>
<td>Near centre</td>
<td>-0.03</td>
<td>0.18</td>
<td>1.00</td>
<td>1.32</td>
<td>2.11</td>
<td>2.45</td>
<td>3.20</td>
<td>3.47</td>
<td>4.10</td>
<td>4.45</td>
<td>5.00</td>
<td>5.19</td>
<td>5.57</td>
<td>6.6%</td>
</tr>
<tr>
<td>Near edge</td>
<td>-0.01</td>
<td>0.21</td>
<td>1.00</td>
<td>1.31</td>
<td>2.07</td>
<td>2.38</td>
<td>3.11</td>
<td>3.37</td>
<td>3.92</td>
<td>4.34</td>
<td>5.00</td>
<td>5.18</td>
<td>5.57</td>
<td>4.6%</td>
</tr>
</tbody>
</table>

In summary, with the prior knowledge, 3D imaging of a metallic cylinder in dielectrical materials is simplified to estimate the cross-sectional and axial positions of the cylinder. From the simulation result by 2D and 3D models, the cross-sectional position of the grounded metallic cylinder can be derived by the weighted mean method while the axial position can be estimated by (1) the normalised capacitance change, (2) the grey level of
the reconstructed image, and (3) the relative change in capacitance between the metallic cylinder and each electrode. Promising results can be obtained by the simulation models and the absolute error is within 0.2 cm in estimating cross-sectional position and 1 cm in axial position. The following section will focus on the experimental study of positioning a metallic rod in plastic beads.

5.3 Experimental study

In the experiment, the ECT system introduced in Chapter 3 is used together with a conventional 8-electrode sensor shown in Figure 5.1 (c). Figure 5.11 shows the experiment setup of an aluminium rod of 2.6 cm in diameter placed in a sensor filled with plastic beads. An XYZ stand is used to adjust the axial position of the rod. Capacitances are measured by the impedance analyser HP4192A via the multiplexer. A MatLab graphical user interface (GUI) realises the hardware control and image reconstruction. Similar to the simulation, the experiment aims to estimate the cross-sectional and axial position of the grounded metallic rod and to plot a 3D image of the rod with prior knowledge. In the experiment, two cross-sectional positions are investigated (near centre and near edge) and the axial position varies from 0 to 6 cm.

Figure 5.11 Experimental set-up
Figure 5.12 shows the cross-sectional images when the grounded rod is placed at 2 cross-sectional positions and 3 axial positions. With the rod placed near centre, the derived cross-sectional centre is less sensitive to different axial position of the cylinder. However, when the rod located near edge, the derived cross-sectional centre of the rod slightly shifts to the centre of image area when the rod apex approaches to the end of the electrodes (6 cm).

Figure 5.12 Cross-sectional images with and without prior knowledge

Figure 5.13 shows the normalised capacitance and the grey level (calibrated by Eq. 5.2) at the axial position from 0 cm to 6 cm with a step length of 0.5 cm. Although the experimental results suffer from many factors, including the noise from the environment and the standing capacitance from the sensor, cables and the multiplexer, the changes in normalised capacitance and the grey level are highly in accordance with simulation results.

According to the simulation results, the third order polynomial is the best fitting curve to estimate the axial position compared with the second order polynomial and the exponential curve. In the experiment, only the third order polynomial is chosen to fit the
change in either normalised capacitance or grey level corresponding to the axial position of the rod. Table 5.11 lists the axial distance of the rod apex from the origin derived by the third order polynomial based on normalised capacitance. The derived position shows high accuracy with an average error of 1.2%. Table 5.12 shows the derived axial position by a third order polynomial based on grey level of the reconstructed image with an average error of 1.4%. Comparatively, the position by the normalised capacitance achieves higher accuracy at the End point than by the grey level.

![Normalised capacitance and grey level against different axial position of aluminium rod](image)

**Figure 5.13** Normalised capacitance and grey level against different axial position of aluminium rod

### Table 5.11
Derived axial distance of rod apex from axial origin by third order polynomial function and average error based on normalised capacitance

<table>
<thead>
<tr>
<th>True distance</th>
<th>Axial distance of aluminium rod apex from origin (cm)</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0  0.5  1  1.5  2  2.5  3  3.5  4  4.5  5  5.5  6</td>
<td></td>
</tr>
<tr>
<td>Near centre</td>
<td>0.33 0.48 1.00 1.47 2.00 2.5 3.0 3.5 4.0 4.5 4.9 5.4 5.95</td>
<td>0.9%</td>
</tr>
<tr>
<td>Near edge</td>
<td>0.39 0.55 1.00 1.47 1.99 2.4 2.9 3.5 4.0 4.4 4.9 5.4 6.00</td>
<td>1.4%</td>
</tr>
</tbody>
</table>

### Table 5.12
Derived axial distance of rod apex from axial origin by third order polynomial function and average error based on grey level

<table>
<thead>
<tr>
<th>True distance</th>
<th>Axial distance of aluminium rod apex from origin (cm)</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0  0.5  1  1.5  2  2.5  3  3.5  4  4.5  5  5.5  6</td>
<td></td>
</tr>
<tr>
<td>Near centre</td>
<td>0.36 0.53 1.00 1.48 2.00 2.51 3.00 3.54 4.01 4.47 5.04 5.59 6.16</td>
<td>1.3%</td>
</tr>
<tr>
<td>Near edge</td>
<td>0.43 0.54 1.00 1.46 2.01 2.50 2.99 3.52 4.00 4.48 5.07 5.56 6.10</td>
<td>1.5%</td>
</tr>
</tbody>
</table>
Another method is to estimate the axial position based on the capacitance between the aluminium rod and each electrode. Figure 5.14 shows the relative change in capacitance between electrodes and excited rod at different axial position. Similar to the simulation, a linear relationship is shown between the relative change in capacitance and the axial distance of the rod apex from the axial origin. For the average change in capacitance, the capacitance of the rod placed near edge is slightly higher than near centre. There is a small change from 0 cm to 0.5 cm due to a mechanical false of the XYZ stand.

![Figure 5.14 Relative change in capacitance at different axial distance](image)

(a) Near centre  (b) Near edge  (c) Comparison of average relative change

A linear fitting line is generated by two reference points. As discussed in simulation, the selection of reference points affects the accuracy of the estimated axial position. Table 5.13 presents the derived axial distance of the rod apex from the axial origin by the linear function generated by the reference points at an axial distance of 1 cm and 2 cm. A high accuracy is obtained with an average error of 2%. The second linear function is generated by two reference points with the axial distance 1 cm and 5 cm.
away from the axial origin. Table 5.14 lists the derived axial positions by the second linear function. By increasing the distance between two reference points, the average error reduces to 2.0% when the rod is placed near centre and 1.7% near edge. However, the accuracy of the estimated End point by the second linear function is lower than by the first linear function.

Table 5.13 Derived axial distance of rod apex from axial origin by linear function and average error based on relative change in capacitance (1 cm and 2 cm)

<table>
<thead>
<tr>
<th>True distance</th>
<th>Axial distance of aluminium rod apex from origin (cm)</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Near centre</td>
<td>0.37 0.52 1.00 1.50 2.00 2.50 3.12 3.64 4.13 4.68 5.15 5.56 5.88</td>
<td>2.2%</td>
</tr>
<tr>
<td>Near edge</td>
<td>0.41 0.54 1.00 1.49 2.00 2.51 3.09 3.60 4.06 4.53 5.03 5.46 5.83</td>
<td>1.8%</td>
</tr>
</tbody>
</table>

Table 5.14 Derived axial distance of rod apex from axial origin by linear function and average error based on relative change in capacitance (1 cm and 5 cm)

<table>
<thead>
<tr>
<th>True distance</th>
<th>Axial distance of aluminium rod apex from origin (cm)</th>
<th>Average error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Near centre</td>
<td>0.40 0.54 1.00 1.48 1.96 2.44 3.04 3.55 4.02 4.55 5.00 5.40 5.70</td>
<td>2.0%</td>
</tr>
<tr>
<td>Near edge</td>
<td>0.41 0.54 1.00 1.48 1.99 2.50 3.08 3.58 4.03 4.50 5.00 5.43 5.79</td>
<td>1.7%</td>
</tr>
</tbody>
</table>

Once the cross-sectional and axial positions obtained, a 3D image can be drawn with prior knowledge. Figure 5.15 shows 3D images of the metallic rod by using the prior knowledge. The cross-sectional and axial positions of the rod are derived by the primary reconstructed image and the relative change in capacitance between rod and electrodes, respectively.

Similar to the simulation result, a 3D image is obtained by estimating the cross-sectional and axial position of the metallic object and high accuracy is achieved. There is no image reconstruction algorithm required in estimating the axial position by either
normalised capacitance or relative change in capacitance between the metallic object and electrodes. Therefore, the simplified 3D imaging process largely reduces imaging time because there is no complicated 3D reconstruction algorithm involved.

![Figure 5.15 3D image of metallic rod at different cross-sectional and axial position](image)

Although ECT can provide promising result in 3D imaging of a metallic rod in dielectric materials, due to very high permittivity of biology tissue, it is difficult to use traditional ECT to image metallic object in total hip. The following section focus on the feasibility of imaging metallic objects in high permittivity conductive materials by measuring both conductance and reactance, aiming to obtain electrical impedance tomography (EIT) images.

5.4 EIT for revision total hip replacement

From the previous discussion, it is possible to generate real-time 3D images with prior knowledge by a conventional single plane ECT sensor. However, according to the result
from Chapter 4, it is difficult for ECT to measure a small change in capacitance inside a human body because of its high permittivity, e.g. the permittivity of bone, muscle and skin are all over 100 (Gabriel et al., 1996a, 1996b). As a result, a further research will focus on imaging high permittivity conductive materials by measuring both resistive and reactive components to obtain an EIT image, the imaging principle of which is the same as ECT and also prior knowledge will be used.

To date, traditional EIT with current-injection and voltage-detection has been considered for some other medical applications, such as brain function imaging and breast cancer detection (Rebecca, 2011; Zou and Guo, 2002; Assenheimer et al., 2001; Brown, 2003). The main advantage of this design is the reduced effect from the contact impedance compared to a voltage source EIT system (York, 2001). Practically, this effect can be reduced by two possible solutions. One is to combine multiple-electrode impedance measurement methods to measure the contact impedance directly (Cardu et al., 2012). The other solution is to avoid direct contact of the electrodes and electrolyte and reduce the effect of the contact impedance by parallel or series resonance or by increasing the coupling capacitance between the electrodes and the electrolyte (Sun and Yang, 2013). Technically, a high-precision voltage-source EIT system is easier to implement with lower cost than the current-source system (Saulnier et al., 2006) and the design will not suffer from serious fringe effect as the traditional EIT approach with current-injection and voltage-detection does (Qureshi et al., 2013). With voltage excitation, EIT measurement can be made with a conventional ECT sensor (Sun and Yang, 2013). As the impedance analyser HP 4192A can not only measure capacitance but also provide resistance/conductance measurement, it is possible to generate EIT images based on the developed impedance analyser based ECT system.

Figure 5.16 shows the sensor with 8 inner electrodes, the configuration of which is listed below

- Inner diameter: 74 mm
- Electrode size: 65 mm × 25 mm
- Thickness of wall: 3 mm
- Distance between shield and electrodes: 11 mm
- Overall height of the pipe: 175 mm
- Length of grounded end guards: 20 mm
- Gap between electrodes and the end guards: 5 mm.

Figure 5.16 EIT/ECT sensor

In principle, to measure conductance, lower excitation frequency is preferred because the resistance/conductance becomes domination in an impedance measurement at low frequency and the capacitance becomes domination with high frequency and leads large systematic errors to the resistance/conductance measurement.

In the experiment, a frequency range from 5 kHz to 10 MHz is investigated with the voltage-source EIT system. The aluminium rod of 2.6 cm in diameter is placed in deionised water and saline solution, using a plastic rod with equal diameter as a reference. The conductivity of the saline solution of 0.46 S/m, corresponding to the average conductivity of the muscle within frequency range from 10 kHz to 1 MHz (Gabriel et al., 1996a).

Figure 5.17 shows the images when the plastic rod and the metallic rod are fully inserted in the EIT/ECT sensor with the sensing area of deionised water or saline solution. Four excitation frequencies are chosen, two of which are low excitation frequency 5 kHz and 10 kHz and the other two are high excitation frequency: 500 kHz and 1 MHz. Figure 5.17 (a) and (b) shows the images by measuring capacitance and conductance, representing the distribution of the normalised permittivity and normalised conductivity, respectively. Note that with deionised water, no image can be obtained at excitation frequency of 5 kHz as HP4192A cannot obtain the conductance data at this frequency due to the measurement failure of HP4192A (Agilent, 2000).
By measuring capacitance, there is no obvious difference with different frequency when the rod (either dielectric or conductive) placed in the deionised water. However, with saline solution, the images with low excitation frequency are highly different from the image with high excitation frequency. With the metallic rod in saline solution, the image suffers from notable distortion, showing an irregular shape associated with the rod in the centre. Images by conductance are highly consistent in all selected frequencies. Only with the grounded metallic rod in saline solution, the grey level of central region corresponding to the area affected by the metallic rod becomes larger with the increase in the excitation frequency, referring to a change in conductivity measured by HP4192A with different excitation frequency.

(a) By capacitance
Figure 5.17 Images of rods in deionised water and saline solution

Figure 5.18 shows ECT and ERT images of an off-centre grounded metallic rod in saline solution with different axial position at different excitation frequency. With low frequency, i.e. below 100 kHz, the images by conductance show an enlarged effective area of the grounded metallic rod corresponding to the increase in distance of the rod apex from the axial origin. A similar phenomenon is shown in the images by measuring capacitance at a high excitation frequency. However, with high frequency, e.g. 5 MHz and 10 MHz, the grounded metallic rod forms an area with high grey level, associated with a higher permittivity region compared to the surrounding materials, which is different from the images with low excitation frequency. With 1 MHz, the image by capacitance shows low grey levels in the centre when the rod apex is near the axial...
origin, which is similar to the images by capacitance with the low frequency. However, with an increase in distance between the rod apex and axial origin, a high grey level and enlarged effective area is shown, which is in accordance with the images with the high excitation frequency.

Figure 5.18 Images of off-centre grounded metal rod in saline solution (0.46 S/m) at different axial position and excitation frequency
Figure 5.19 shows the change in average capacitance and conductance between the metallic rod and electrodes corresponding to the different axial position of the metallic rod in the sensing area filled with saline solution with the conductivity of 0.46 S/m. The excitation frequency varies from 10 kHz to 10 MHz. Obviously, the capacitance or conductance shows a frequency-related change. In particular, the average conductance shows completely opposite changes with low and high excitation frequency, i.e. a positive change in conductance with low excitation frequency and a negative change with high excitation frequency. When the excitation frequency reaches 10 MHz, resonance occurs, resulting in significant distortion in the measured capacitance. In general, with the excitation frequency range from 50 kHz to 500 kHz, an increasing trend is shown in both capacitance and conductance associated with the increase in length of the rod in the conductive saline solution, which is similar to the change in capacitance of the rod in non-conductive materials as discussed in previous section, providing a possibility of 3D imaging of the metallic rod in high-permittivity conductive materials, e.g. human tissue, using the model based image reconstruction method with prior knowledge.

From Figure 5.19, with 10 kHz, the change in conductance is nearly linear to the axial distance of the grounded metallic rod from the axial origin, while the change in capacitance is linear to the axial position of the rod at 5 MHz. An idea of using multiple frequencies can be suggested for the future 3D imaging EIT system for the revision THR, in an attempt to measure capacitance and conductance at different excitation frequency and combine the data to produce an EIT image. With the linear relationship between conductance and the axial position of the rod at low frequency and between capacitance and the axial position of the rod at high frequency, the 3D image can be obtained by estimating the cross-sectional position based on the weighted mean method and the axial position based on two reference points.
Figure 5.19 Change in capacitance/conductance (average) between excited metal rod and electrode at different axial position and frequency
5.5 Summary

This chapter describes a promising method of generating real-time 3D images of a metallic cylinder with the prior knowledge by a conventional single plane ECT sensor, which has potential use for navigating milling toll in revision THR surgery.

An aluminium rod is inserted into plastic beads filled in the ECT sensor as the first physical simulation. It has been confirmed by both simulation and experimental results that the change in capacitance and its related image is associated with both the cross-sectional and axial positions of the rod, indicating the possibility of obtaining 3D position of the metallic object. With the prior knowledge (i.e. shape and the diameter), an image of the aluminium rod with accurate 3D position can be generated.

The procedure of obtain 3D image is simplified to obtain the cross-sectional and axial positions of the test object. To obtain the cross-sectional position, an initial image is obtained by Landweber iteration algorithm. The centre is derived by calculating the weighted mean of the image. The axial position is estimated by three possible methods: (1) normalised capacitance value, (2) grey level of reconstructed image, and (3) the average change in capacitance between the rod and each electrode. In general, the relationship between the axial distance and either normalised capacitance or grey level is not linear. Polynomial and exponential curves are used to fit the change. With 7 reference points, high accuracy is achieved, particularly by the third order polynomial. The average change in capacitance between the metallic rod and electrodes indicates a nearly-linear change with the axial distance, showing the possibility of using less reference points of deriving the trend line. With a linear relationship, only 2 reference points are needed and an average error of 4.8% is obtained with the simulation model and 2% in experiment. Once the cross-sectional and axial positions are obtained, an accurate real time 3D image with prior knowledge can be drawn, since there is no complex sensor design and 3D image reconstruction involved.
To overcome the problem of high permittivity associated with the organic tissue, an initial experiment has been carried out to generate an EIT image of a metallic object in conductive solution, corresponding to the revision THR. With the impedance analyser based system, the EIT image can be obtained based on a conventional ECT sensor with voltage excitation. The results show that with the grounded metallic rod in conductive solution, the change in either capacitance or conductance is highly frequency-related. As the organic tissue is a complex material that combines resistance and reactance, the resistance (conductance) dominates the measured impedance in low frequency and the reactance (capacitance) becomes domination in the impedance measurement with a high excitation frequency. In general, there are two possible EIT systems to generate a 3D EIT image. The first system uses a single excitation frequency with the range from 50 kHz to 500 kHz to obtain both capacitance and conductance data. The other system is based on multiple-frequency to obtain ERT and ECT images separately at low and high excitation frequency, respectively, and combine the ERT and ECT images into an EIT image. The capacitance or conductance measured between excited metallic rod in the saline solution and electrodes shows similar change with the rod in non-conductive materials, giving a possibility of obtain 3D images of milling tool during revision THR by model based image reconstruction method.
Chapter 6: Conclusions and future work

6.1 Conclusions

While ECT has been developed for 3 decades and becomes one of the most mature imaging modalities in industry, it has never been considered for medical applications. In this thesis, the feasibility of using ECT for two medical applications: RCT and revision THP, has been investigated by simulation and experiment.

To meet the requirement of measuring small change in capacitance in medical applications, e.g. RCT, the developed system takes advantage of an impedance analyser HP4192A with high accuracy in electrical property measurement. Due to the limited working channel of the HP4192A, a multiplexer based on relay switches, providing minimum standing capacitance, is used as a part of measurement unit to extend and configure the measuring channels. With the HP4192A and the multiplexer, the resolution of the system is 10 times better than conventional AC ECT systems (Chrondronasios, 2003). The conclusions will be given in the following two parts, corresponding to the two medical applications: RCT and revision THR.

6.1.1 ECT for RCT

The dental ECT system has been designed for RCT for two purposes: (1) to visualise the tooth surface in real time and (2) to determine the position of an endodontic file. To adapt the anatomy of the jaw and the limited space in oral cavity, a miniature two-plate dental ECT sensor has been designed and fabricated base on FPCB. A clamp-on sensor has been designed and a test rig has been made by Lucid Innovation Ltd. to replace the clamp for the preliminary experiment, aiming to position the dental ECT sensor on either simulant materials or organic tissue.

The performance of the two-plate ECT sensor has been introduced for the first time. Although this type of sensor provides incomplete measurement data only, due to its specific open structure, the sensor has a similar performance to the conventional ECT sensor and can provide good image quality. The performance of the sensor has been discussed based on simulation and experimental data. In the simulation study, the two-plate sensor has been investigated in 5 aspects, (1) the boundary setting of the open
boundaries, (2) the effective sensing area, (3) the distance between the two sensor plates, (4) the angle between the two sensor plates, and (5) the number of electrodes.

In general, the open boundary that is not covered by the sensor is set as ‘Electric insulation’. A too small or too large sensing area causes high sensitivity at the edges and relatively low sensitivity in the central region, resulting in a low contrast image in the centre and distortion at the edge. To eliminate the noise and improve the image quality, the width of the sensing area should be equal to the length of the shielding layer. According to the simulation results, the change in distance affects the sensitivity in the sensing area significantly. A smaller distance between the two sensor plates provides higher sensitivity in the central region but reduces the sensing space. To compromise the imaging space and the image quality, the distance between the two plates should be equal to the shielding layer. Due to the unfixed two-plate structure, it is possible that the two plates are not placed exactly in parallel. Fortunately, according to the simulation results, with a small rotation angle between the two plates, i.e. less than 30 degrees, the effect of the angle can be ignored. For an ECT sensor, it is crucial to determine the number of electrodes, because it is associated with the number of independent capacitance measurements and thus directly affects the resolution of the reconstructed image. In principle, a larger number of independent measurements help to obtain a higher image resolution. From the simulation results, the increase in the number of electrodes contributes to a higher image resolution to some extent, particular for the distributions of multiple objects.

Experiments have been carried out using the impedance analyser (HP4192A) based ECT system. Two sensors have been fabricated with either $2 \times 2$ or $2 \times 3$ sensor array on each sensor plate. For both sensors, two measurement strategies have been introduced, which are single-electrode excitation and multiple-electrode excitation. The results show that with multiple-electrode excitation, the sensor can provide stable and large capacitance data with the excitation frequency from 10 kHz to 5 MHz and the highest SNR can be obtained at 1 MHz. Seven permittivity distributions have been chosen to evaluate the performance of the fabricated sensor with $2 \times 3$ sensor array, which can provide 4 or 6-electrode sensors by different electrodes combinations. The 4-electrode sensor gives a higher image quality in stratified distribution than the 6-electrode sensor. With the other permittivity distributions, the 6-electrode sensor, that
provides more independent measurements, shows a larger image contrast and clearer
boundary of multiple objects, which is consistent with the simulation results.

To image the tooth by ECT, simulation study has been taken based on 2D and 3D
models generated in COMSOL 3.5. In the simulation models, single tooth models with
different root length are placed at different cross-sectional position. From the simulation
results, it can be concluded that (1) the two-plate ECT sensor can provide cross-
sectional and longitudinal views of the single tooth, (2) Landweber iteration gives high
image quality for the cross-sectional view and LBP provides high quality images and
high correlation coefficients for the longitudinal view, (3) the sensor with $2 \times 3$ array
has superior performance in longitudinal view images for determining the length of the
root in the sensing area. Experiments have been carried out using a single human
premolar. As all organic tissues associate with a frequency-related permittivity, a higher
excitation frequency, i.e. over 1 MHz, is used to reduce the permittivity of the tissue to
the range that ECT can measure, e.g. the permittivity of dentin is approx. 10 with the
excitation frequency over 1 MHz. The results from the experiment are highly consistent
with the simulation results with 1 MHz excitation frequency.

To solve the problem of the limited image resolution produced by ECT, image
registration and fusion has been considered. A high resolution image, i.e. radiograph, is
used to improve the image quality of the dental ECT image. A simple Major-axis
method has been introduced for image registration. By registering and fusing with the
radiograph, a real-time image with high resolution can be obtained.

Once the ECT image is registered with the radiograph, the resultant image can be used
as the background image for positioning an endodontic file. The principle of using ECT
for detecting an endodontic file has been introduced. The results show that the ECT
system is highly sensitive to a grounded metallic file. The change in capacitance is
associated with the length of the file in the sensing area, proving a possibility of
locating an endodontic file by measuring capacitance. A significant change in
capacitance occurs when a grounded metallic file touches conductive media or solution.
This sudden change in capacitance can be considered as a reference point, showing that
the grounded metallic file reaches the root apex. In principle, an ECT sensor with $2 \times 2$
or $2 \times 3$ array provides the position of the endodontic file by measuring capacitance
change. Two methods have been introduced to fit the relationship between the change in
capacitance and the file position. Method I is to obtain the capacitance by opposing electrode pairs, by which 2D positions can be estimated. Method II is to measure the capacitance between file and each electrode, providing 3D position. To eliminate the fringe effect, an idea of using the capacitance difference is introduced instead of the capacitance measured by individual electrode directly. According to the experiment results based on the file in the fabricated plastic cube, the measured capacitance by both Methods I and II is nearly-linear to the length of the file in the sensing area. The trend of the capacitance change corresponding to the axial position of the file apex is fitted by a piecewise linear function. With the linear relationship, the position of the file can be derived by the capacitance measurement and two reference points, which are the ‘Start Point’, obtained by the sensing area without the file, and the ‘End Point’, the sudden change in capacitance when the file touches the apex. By a pre-measurement, the two reference points can be obtained and the X-axis position can be derived by the capacitance difference between left and right electrodes, by which the target root can be selected according to the registered image. An iterative method has been used to estimate the axial position of the file apex based on the piecewise linear function. The results show high accuracy near the ‘End Point’ as it is one of the reference points. Comparatively, the 2 × 3 sensor with Method II shows a superior performance in estimating the axial position of the file.

A preliminary in-vitro experiment has been carried out on a specimen of pig’s lower jaw with the 2 × 2 sensor. Although the permittivity of the dentin is reduced to 10 with the excitation frequency of 1 MHz, the permittivity of periodontal tissue and bone is still too high at this frequency range (i.e. in the order of 10²). Due to the high permittivity and the complex anatomy of the jaw, from the reconstructed ECT image, it can only separate the jaw from the impression material-AHP, but cannot differentiate the tooth from the bone and tissue. By measuring the capacitance between opposing electrode-pairs, there are two sudden changes in capacitance obtained when the endodontic file is inserted into the mesiobuccal root of a premolar. When the apex of the file reaches the root apex, the capacitance drops to the lowest value and there is no further decrease in capacitance when the file apex inserted beyond the root apex.
6.1.2 ECT for revision THR

An initial study has been carried out on the research of ECT for revision THR. A conventional 8-electrode single plane sensor is used, generating real-time 3D images of a metallic rod with prior knowledge based on a model based method. In this method, the 3D image reconstruction is simplified to estimate the cross-sectional and axial positions of the rod in the sensing area and to draw an image of the rod based on the prior knowledge. In both simulation models and experiment, the metallic rod is inserted into the sensing area filled with plastic beads.

A weighted mean method has been introduced to estimate the centre of the cross-sectional position of the rod. The method has been evaluated by 8 cross-sectional positions. The cross-sectional centre of the rod can be derived based on the Start point or the End point, which are defined as the position when the rod apex is located at the axial origin and the position when the rod apex is 6 cm below the axial origin, respectively. After calibration, high accuracy in estimation of the cross-sectional centre can be achieved, with the maximum absolute error of 0.13 cm. Comparatively, the results by the End point shows higher accuracy with an average absolute error of 0.04 cm, although the estimation by the Start point achieves high accuracy in some evaluation points.

To estimate the axial position, three methods have been introduced: (1) the normalised capacitance, (2) the grey level of the reconstructed image and (3) the change in capacitance between the metallic cylinder and each electrode. Similar to the method in ECT for RCT, a pre-measurement is needed to find the reference points for deriving the fitting line. A third order polynomial and a second order polynomial are used to fit the relationship between the normalised capacitance and the axial position of the rod, while the relationship between the grey level and the axial position is fitted by a third order polynomial, a second order polynomial and an exponential curve. For all those fitting curves, 7 reference points are used and high accuracy can be achieved when the third order polynomial is used for fitting the relationship between the axial position of the metallic rod and either the normalised capacitance (average relative error of 2.2%) or grey level (average relative error of 2.6%). By estimating the position of the file apex in RCT application, a linear function is proposed to fit the trend of the axial position of the rod corresponding to the change in capacitance between the rod and electrodes. In
principle, only two reference points are needed for deriving a linear function. The accuracy in estimation varies with different reference points. A low relative error, i.e. 4.9%, can be achieved when one of the reference points are near the Start point and the other is close to the End point.

Experiments have been carried out using the developed ECT system. The experimental results are highly consistent with the simulation results. By using the third order polynomial, the derived axial position based on either normalised capacitance or grey level can obtain low value in average error, 1.2% and 1.4%, respectively. By estimating the axial position based on the linear function corresponding to the capacitance change between active metallic rod and each electrode, the average error is about 2%.

A 3D image can be drawn by means of the prior knowledge of the metallic rod (e.g. shape and diameter) and the estimated cross-sectional and axial positions. As there is no complex computation of 3D reconstruction involved, the 3D imaging speed by this model based method is high, much higher than generating a 2D image based on an iterative method with large number of iterations.

As discussed in Chapter 4, the high permittivity of organic tissue causes problems to obtain a good ECT image of the grounded metallic object(s) inside a human body. To solve this problem, a preliminary experiment has been carried out to generate an EIT image of a metallic object in conductive solution with high permittivity (conductivity of 0.46 S/m and permittivity of 80), corresponding to the revision THR. With the impedance analyser based system, the EIT image can be obtained with a conventional ECT sensor and voltage excitation. As the organic tissue is a complex material that combines resistance and reactance, the resistance dominates the measured impedance in low frequency and the reactance becomes domination in the impedance measurement with a high excitation frequency. In the experiment, with the grounded metallic rod in the saline solution, the change in either capacitance or conductance is highly frequency-related. With the selected frequency range from 10 kHz to 10 MHz, the change in conductance between the rod and electrodes is proportional to the axial distance of the rod apex from the axial origin with the excitation frequency from 10 kHz to 500 kHz. A similar change in capacitance is shown with the frequency range from 50 kHz to 5 MHz. The change in either capacitance or conductance with a grounded metallic rod in conductive solution is similar to the change in capacitance of the rod in dielectric
materials. Based on the results, it is possible to estimate the cross-sectional position of the rod based on the initial cross-sectional image and the axial position by the change in capacitance or conductance between the rod and electrodes, in attempt to obtain a real-time 3D image of the milling/drilling tool during revision THR by the model based 3D image reconstruction method.

6.2 Future work

The future work is addressed into the following parts:

- Further research on impedance measurement
- Positioning accuracy
- In-vitro and in-vivo experiment

6.2.1 Further research on impedance measurement

Although initial experiments have been carried out based on measuring both conductance and reactance with a metallic object in saline solution, the research into EIT for RCT and revision THR can be continued. Further research can focus on the EIT system using the same ECT sensor with voltage excitation and current detection, but with adaptive phase angle for measurement. According to the experiment results of EIT for revision THR, the change in capacitance or conductance of the grounded metallic objects is highly frequency-related, giving two possible EIT systems to generate a real-time 3D image. The first system uses a single excitation frequency range from 50 kHz to 500 kHz to obtain capacitance and conductance data. The other system is based on multiple-frequency to obtain ERT and ECT images at low and high excitation frequency, respectively, and combine the ERT and ECT images into one EIT image.

6.2.2 Positioning accuracy

According to the results of ECT for RCT and revision THR, it is possible to position a grounded metallic object by measuring capacitance. However, due to the fringe effect caused by the ‘soft’ electric field, the relationship between the axial position of the metallic object and the change in capacitance is complex and non-linear. In the thesis, a simplified piecewise linear function has been proposed for estimating the position of an endodontic file with either 2 × 2 or 2 × 3 two-plate sensor. In the future, a more
complex function would be derived to fit the non-linear relationship between the capacitance and the axial position of the metallic object.

For the dental ECT system, a piecewise second order or third order polynomial can be considered instead of the linear function. However, to derive a second or third order polynomial, more efficient reference points are required. Therefore, it is essential in the pre-measurement to find the reference points and the method of selecting the efficient reference points should be discussed in the future. Clinically, the root canal is not always straight and the periodontal tissue is highly inhomogeneous, which may affect the capacitance measurement significantly and present challenges in ECT for RCT. To solve the problem, system calibration is needed to reduce the effect of the inhomogeneous distribution of the tissue and more in-vitro experiments with different shape of root canal should be discussed in the future.

More effort should be made on the research into ECT for revision THR, although it is easier to implement ECT for revision THR than RCT because of the enlarged size in sensor and the sufficient number of electrodes in one plane. The thesis has shown promising results of generating a real-time 3D image for positioning a metallic rod based on a conventional ECT sensor, but the average error between the estimated position and the true position is still large due to the simplified function. Similar to the dental ECT system, a higher accuracy can be obtained by a more complex relationship to fit the trend of capacitance change corresponding to the axial position of the rod. Additionally, as EIT can be used instead of ECT in the future, the 3D position by measuring conductance is another issue to improve the accuracy of estimation.

6.2.3 In-vitro and in-vivo experiment

As a medical application, it is essential to implement both the in-vitro and in-vivo experiments to show the possibility of applying the new techniques for clinical use. Although an initial in-vitro experiment has been carried out based on a single premolar for dental ECT system and the results are promising, more effort is needed in imaging the tooth together with bone and tissue or clinical experiment, i.e. to operate the system on the human body. According to the initial experiment of using the dental ECT system on pig’s jaw, it is easy to differentiate the jaw from the background but difficult to determine tooth from the jaw. The reasons for this failure are the complex anatomy of jaw and the high permittivity of the bone and periodontal tissue at the selected
frequency range. Thus, as suggested in previous section, an impedance measurement is required in the future to measure both conductance and capacitance to generate tooth images. Another idea is to extend the depth of the dental ECT sensor to the crown part of the tooth, which is not covered by the bone and periodontal tissue. By this solution, ECT is only used to differentiate the crown part of the tooth and the air or impression material, similar to what has been down in the Chapter 4.

For revision THR, simulation study is needed to optimise the configuration of the sensor and the system. The electrical property of the materials used in the revision THR surgery should be measured before implementing the in-vitro and in-vivo experiments. To fit the specific anatomy of thigh, a sensor in conical shape with open structure is desired. The sensor may take advantage of FPCB, forming an adjustable strap to adapt different diameter of thigh. Another clinical problem should be considered in the future research that, as the femur cavity has been replaced by other artificial material (e.g. bone cement) in the primary THR surgery, during the revision THR, the bone cement would be removed by the milling/drilling tool, forming an empty bone cavity, the permittivity of which is significantly lower than the surrounding tissue. This may cause problem in positioning milling/drilling tool by the grey level of the reconstructed image since it is difficult to measure the capacitance or conductance change of the metallic object in an empty bone cavity that surrounded by the media with high permittivity and conductivity. Therefore, in practice, only by measuring the capacitance or conductance measured between the milling/drilling tool and electrodes, the position of the milling/drilling tool may be derived in the in-vitro animal experiment and in-vivo experiment on human body.
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