Ultrasound 3D Positioning System for Surgical Instruments

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To my family

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Introduction

1

1.1 Medical-technical overview

The high-technology developments in the last decades in all areas of science were a great stimulus for the evolution of minimal invasive surgery (MIS). Therefore MIS is receiving more and more attention because it offers new ways for diagnosis and treatment[1.1].

Minimal-invasive laparoscopic procedures offer important advantages over the traditional open surgery, although they cannot always provide the optimal treatment for different diseases. It is very common in today’s medicine to limit the use of so called invasive surgery, which are more harmful to the patient compared to MIS and therefore lengthen the healing process. The success minimal-invasive procedure as a surgical technique derives from its ability to offer the surgeon a view inside the patient's body through small incisions. This way the discomfort to the patient as well as the pain are reduced, while the damage to healthy tissue and the risk of infection is minimised. Therefore the post-operative benefits for the patient include less trauma, shorter hospitalisation and a faster return to normal activity.

The technical evolution, especially in the last years, added new alternatives to laparoscopic procedures, enabling the minimal-invasive trend to reach different medical areas, such as endoscopy, neurology and radiology [1.2].

A number of medical instruments which enable interventionists to treat diseases with minimal trauma to the patient is already available on the market and new interventions tools are being optimised. All these procedures need a way of monitoring the medical instruments once they are inserted into the body. An endoscope is used to show the surgeon what is happening inside the body, however this view is very limited and gives no information on instrument position outside the camera view. Even the use of 'standard' imaging techniques such as X-ray, ultrasound, MRI and CT scans before or during the operation do not solve this problem satisfactory. To be able to work, the surgeon builds a three dimensional view in his head and tries to track the instruments inside. However, the position errors accumulate forcing surgeons to occasionally retract their instruments and /or the endoscope and ‘reset’ this mental model. Tracking the instruments becomes more difficult during the operation due to blood that obscures the few indicators which are used as guidance such as colour, texture and shape of the organs.

1.2 Motivation and objectives

To alleviate the problem described before and because at present there is no suitable medical device that renders the information with respect to the position of the laparoscopic instruments, it was decided to develop a system that would add more information about the position and orientation of laparoscopic instruments and to reduce the number of retractions of instruments. The device should record the position of the instruments during the surgical procedure, being especially useful to train less experienced young doctors.

Discussions with an interventional laparoscopic surgeon at the Erasmus Medical Centre in Rotterdam, Prof. Jaap Bonjer and a representative of Olympus Multinational, producer of laparoscopic instruments, underlined the main concern of the medical profession regarding interventional procedures.

Some of requirements define the technical parameters (positional information, precision in millimetres range, error) which are directly related to the performance of the system. The system should be able to detect the location of the tip of the instrument with a maximum position error of 9-10 mm. Other requirements represent the safety conditions which should be respected during the interaction processes: doctor-system, doctor-system-patient and system-sterilisation equipment. The last requirements lead to the evident conclusion that a wireless system is most suitable. The wireless solution will eliminate the risk that the system may impede the surgeon's actions which would reduce his efficiency.

In our attempt to fulfil the demands imposed by the clinicians, a ultrasound sensor system was developed. The sensor used to locate the laparoscopic tools consists of an ultrasound transmitter triggered by a RF receiver. Two pair of sensors placed outside to the human body, on the laparoscopic tools, determine the unknown position and orientation of the instrument tip inside the body. To detect the position of the sensors, an array of ultrasound transmitters placed above the surgical table is used[1.3].

This thesis presents the aspects related to the selection of the sensor, the methods and the measurements carried out by means of a patient model. Also in the last chapters of the thesis possible improvements are presented in order to extend the use of this system in microsurgery [1.4].

1.3 Organization of the thesis

The thesis is structured in eight chapters, which are correlated to each other. While the first chapter contains the present introduction, in the second chapter the medical background is presented. A short history of laparoscopic intervention is given and it is explained why minimal invasive techniques may be an alternative to open surgery. Special medical tools are designed for establishing the diagnosis and used for surgical interventions. A number of laparoscopic interventions are mentioned and short descriptions of the operating processes are discussed. In the second part of this chapter the focus lies on image viewing and the limitations during the surgical procedure. A brief illustration was given of how current and potential future techniques aim to overcome these limitations.

Chapter III starts with the problem definition and gives an overview of devices for instrument localisation. These include magnetic, optical (visible and infrared), acoustic and electromagnetic (radio-wave) solutions. The advantages and drawbacks of the presented systems are discussed and compared to the requirements for the developed system. In the second part of the chapter, the ultrasound sensor system is presented. Aspects related to the sensor location are discussed and the options for the placement of the ultrasound receivers in the surgical room are illustrated.

In Chapter IV the ultrasound transmitter-receiver sensor is presented. This chapter starts by presenting the consideration for choosing a specific type of ultrasound transducer. The optimal solution is presented and also the aspects regarding the stability of sensor related to variations of temperature, air flow and humidity are described. Possible methods to determine the position and orientation of ultrasound sensors are presented, the basic principles of Time of Flight and Phase Shift are described. Time of flight (TOF) method was preferred for distance estimation and different techniques and measurements based on this principle are described.

In Chapter V the system based on the TOF is presented. The electronic blocks used to trigger the ultrasound transmission and the hardware and software methods implemented to process the information derived from the acoustic sensors are described. The complete set-up of the ultrasound wireless positioning system is presented as well. Furthermore the influence of temperature and air flow are investigated and the limitations derived from those factors are presented.

Chapter VI and Chapter VII are dedicated to possible solutions to overcome the drawbacks and the limitations described in the last part of the Chapter V. In Chapter VI the phase shift method is proposed and the measurements indicated a good stability for the distortions introduced as air flow turbulence and multipath transmission.

Chapter VII discusses solutions (two frequency method, double way measurement) which may compensate entirely for airflow influences. The solutions were tested in a new method set-up which allows micrometer resolution for the distance calculations and makes the method suitable for microsurgery applications.
In the last chapter of the book the conclusions of the presented work are drawn and some aspects that need further research are presented. These aspects refer to work that needs to be carried out prior to testing of the system in a hospital.

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Medical Background 2

2.1 Minimally invasive surgery - an alternative to traditional open surgery

In recent years minimally invasive surgery (MIS) has gained more and more popularity among medical staff and has become a widely used surgical technique for diagnosis and treatment of many human diseases and injuries.

The goal is to reduce damage to surrounding healthy tissue caused in reaching the surgical point in open surgery. Small perforations replace the relatively large incisions in open surgery. These holes serve as entry points for optical and surgical instruments, or laparoscopic instruments. The major gain is fast post-operative patient recovery. [2.1][2.7]

2.1.1 History

The earliest recorded references to laparoscopy date back to the time of Hippocrates[2.1]. He describes rectal examination using a speculum. Hippocrates also advised about injecting a large volume of air into the intestines through the anus to treat intestinal obstruction.

Hippocrates also advocated the insertion of a suppository that was 10 digits long (1 digit or zebo = 18.7 mm). These descriptions suggest that Hippocrates was well aware of ileus with intestinal obstruction and he thought that there were several possible aetiologies, including faecal impaction, intussusceptions, and sigmoid volvulus. Moreover, Hippocrates treated these life-threatening conditions with minimally invasive approaches.

Sixteen centuries later, in 1585, Aranzi was first to use a light source for a laparoscopic procedure, focusing sunlight through a flask of water and projecting it into the nasal cavity.

The term trochar was coined in 1706 and is thought to be derived from *trochartor troise-quarts*, a three-faced instrument consisting of a perforator enclosed in a metal cannula.

Philip Bozzini built an instrument that could be introduced into the human body to visualise the internal organs. He called this instrument the "LICHTLEITER". Bozzini used an aluminium tube to visualize the genitourinary tract. The tube, which was illuminated by a wax candle, had fitted mirrors to reflect images.

In 1853 French surgeon Antoine Jean Desormeaux first used Bozzini's "Lichtleiter" on a patient. He is considered by many as the "Father of Endoscopy".

Desormaux (1867), Kusmaul (1868), Commander Pantaleoni (1869) and Dimitri Ott (1901) performed a number of experiments which would mark the beginning of laparoscopic techniques.

In 1901 the German surgeon Georg Kelling performed the first experimental laparoscopy in Berlin, using a cystoscope to view the abdomen of a dog, after first insufflating it with air.

H.C. Jacobaeus (1911), from Stockholm, coined the term "laparothorakoskopie" for a specific surgical procedure on the thorax and the abdominal area. He inserted a trocar into the body cavity directly without creating a pneumoperitoneum (the peritoneal cavity inflated with gas is termed a pneumoperitoneum) [2.15].

In 1918 O. Goetze developed an automatic pneumoperitoneum needle characterised for its safe introduction to the peritoneal cavity.

In Switzerland in 1920 Zollikofer discovered the benefit of using CO2 for insufflation rather than filtered atmospheric air or nitrogen[2.4].

Between 1920 and 1980 other famous figures such as Heinz Kalk, John C. Ruddock, Janos Veress, Richard W. Telinde, Raoul Palmer, Kurt Semm, Dr. de Kok and Patrick Steptoe marked significant milestones in the history of laparoscopic intervention

In 1983 the first laparoscopic appendectomy was performed by Semm, a German gynecologist.

In 1985 Erich Mühe in Germany carried out the first documented laparoscopic cholecystectomy.

The first video-assisted laparoscopic cholecystectomy is attributed to Phillippe Mouret, Lyon, France[2.2].

An important step was made in 1994 when a robotic arm was designed to hold the telescope, the objective being to improve safety and reduce the need for a skilled camera operator.

Over the past decade laparoscopic techniques have undergone substantial development. The future for laparoscopic intervention looks promising.

2.1.2 Overview of laparoscopic interventions

This section offers insight into the modern approach of performing laparoscopic interventions. The general approach of using this technique that allows surgeons to operate without opening up the patient's body is also known by several other names, the most common being minimally invasive surgery (MIS) or less invasive surgery. Other names such as can be found in medical literature include "endoscopy" and "keyhole surgery". Whatever their name, these techniques are practised in a range of medical disciplines such as general, paediatric, thorax, orthopaedic and vascular surgery as well as in urology and gynaecology. MIS is known as 'laparoscopy' or 'thoracoscopy' depending on whether the surgical procedure is performed on the stomach or the chest respectively.

The main advantage of this new technique is that there is no need to make a large incision. Instead, the surgeon operates through 3 or 4 small openings about the size of buttonholes, while viewing the patient's internal organs on a TV monitor. Without a large incision the patient suffers less pain and recovery time is shorter. Rather than a six to nine inch incision, the laparoscopic technique is carried out

through only four small openings - all less than half an inch in diameter.
Instead of five to eight days' hospitalisation and four to six weeks' painful recovery at home, most patients now leave hospital within a day or two and return to normal activities within a week.

A small video camera and a few customised instruments allow surgeons to perform surgery with minimal tissue injury.

Figure 2-1: *Insertion of the instruments*

The camera and instruments are inserted into the abdomen or chest through small skin incisions, allowing the surgeon to explore the whole cavity without the need to make large standard incisions through the skin and the peritoneal muscle.

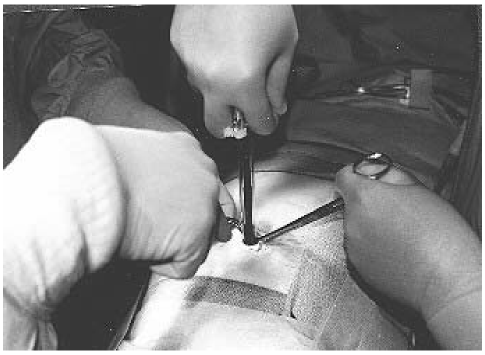

Figure 2-2: *The laparoscopic intervention*

After the incision is made to the umbilical area, a special (Veress) needle is inserted through the hole to commence insufflation. A pressure regulator CO2 insufflator is connected to the needle. The pressure obtained should not be beyond 15mmHg. After satisfactory insufflation the needle is removed and a 10mm trocar is inserted through the existing umbilical incision. The surgeon will access the operative area through the trocar using customised laparoscopic instruments. The trocar will be described in more detail in the next subsection.

This means of intervention reduces recovery time due to its minimal tissue damage.

2.1.3 Laparoscopic tools

This section presents a number of commonly used laparoscopic instruments.

As mentioned before, at the start of a laparoscopic intervention a small incision is made in the body wall and a hollow needle is inserted to insufflate CO_2. The needle is connected to a CO_2 pressure regulator. After inflating the surgery area, the next step is to replace the needle with a tubular pipe-like port shown in Figure 2-3, called a trocar.

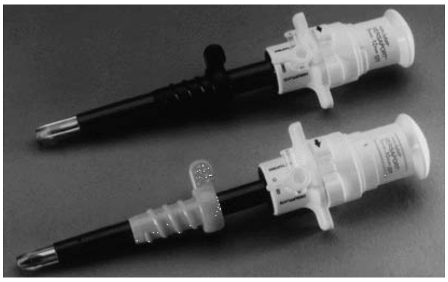

Figure 2-3: *Trocars*

Through the trocar a small optical telescope (called an endoscope - see Figure 2-4) is inserted into the body and connected to a video camera which images the internal view of the body cavity. The image is displayed on a monitor and provides the surgeon with information about the operating area.

Figure 2-4: *Endoscope*

In addition to the endoscope, at least two further trocars are used for inserting special surgical instruments.

Figure 2-5 shows an example of the placement of trocars for the instruments and the camera with the endoscope.

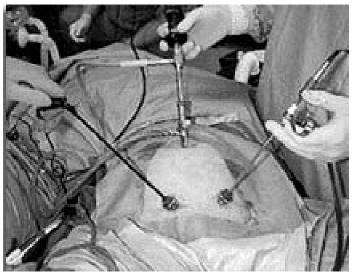

Figure 2-5: *Placement of trocars, endoscope and two more instruments.*

Because regular instruments for open surgery are unsuitable for use in laparoscopy, specially shaped instruments are used.

Figure 2-6 shows examples of scissors (a) and an electrical electrode (b) modified for laparoscopy.

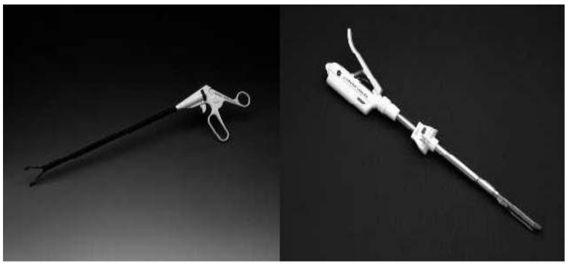

Figure 2-6: *a)scissors. b)electrical electrode used in laparoscopy.*

One of the conclusions shows that the methodology of performing laparoscopic surgery is quite different from that of open surgery. Surgeons are required to develop special skills to manipulate new instruments in laparoscopic procedures.

The application of these techniques will be described in the next subchapter.

2.1.4 Medical diseases and alternative treatments

Laparoscopic techniques can be divided into diagnostic and operative procedures. The differences lie in the methodology and the final level of investigation or the final level of disease treatment.

Diagnostic procedure

As a diagnostic procedure, laparoscopy is useful for taking biopsies of abdominal and pelvic growths, as well as lymph nodes. It allows the doctor to examine the abdominal area, including the female organs, appendix, gallbladder, stomach and liver.

Laparoscopy is used to determine the cause of pelvic pain or gynaecological symptoms that cannot be confirmed by a physical examination or ultrasound. For example, ovarian cysts, endometriosis, ectopic pregnancy and blocked fallopian tubes can be diagnosed using this procedure. It is an important tool when trying to determine the cause of infertility.

Another important aspect of diagnostic laparoscopic procedure are techniques used for injury-induced trauma. Injuries to solid abdominal viscera, liver and spleen cysts and also rectal injuries can be diagnosed with laparoscopic techniques. Chest injuries are successfully diagnosed and subsequently treated by performing video-assisted thoracoscopy (VATS) [2.13].

Operative procedures

After a positive laparoscopic diagnosis the surgeon is required to treat the established disease. When surgical intervention is required medical staff must decide whether to opt for laparoscopic or open surgery.

For complete removal of cancerous tumours, surrounding tissues and lymph nodes laparoscopic surgery is used on a limited basis. Otherwise this type of operation is widely used for non-cancerous conditions that previously required open surgery. Some of them are presented further:

Tubal ligation. In this procedure, the fallopian tubes are sealed or cut to prevent subsequent pregnancies.

Ectopic pregnancy. If a fertilised egg becomes embedded outside the uterus, usually in the fallopian tube, an operation must be performed to remove the developing embryo. This is often done with laparoscopy.

Endometriosis. A condition in which tissue from inside the uterus is found outside the uterus in other parts of (or on organs within) the pelvic cavity. This can cause cysts to form. Endometriosis is diagnosed with laparoscopy, and in some cases cysts and other tissue can be removed during laparoscopy.

Hysterectomy. In some cases the procedure of removing the uterus can be performed using laparoscopy. The uterus is cut away with the aid of laparoscopic instruments and is then removed through the vagina.

Ovarian masses. Tumours or cysts in the ovaries can be removed using laparoscopy.

Appendectomy. In the past, removing an inflamed appendix required open surgery. It is now routinely performed with laparoscopy.

Cholecystectomy. Like appendectomy, gallbladder removal once required open surgery. It can now be performed with laparoscopy in some cases. [2.3]

The above procedures have the advantage of involving less pain, less risk, less scarring and faster recovery in comparison with open surgery. These factors are determinant for laparoscopy as an interventional surgical method.

2.1.5 Advantages of MIS versus open surgical intervention.

The final part of the last subchapter described laparoscopy as a preferred alternative to open surgery.

This affirmation is argued by the following reports. Laparoscopic surgery is becoming increasingly popular with patients because the scars are smaller and their period of recovery is shorter. After surgery patients usually experience pain around the incisions in their skin and often a sense of general discomfort in the abdominal area. Many patients have reported pain at their shoulder tip, which is due to the indirect effect of small amounts of carbon dioxide remaining in the abdomen. But pain relief is always given. Patients are fully recovered within 48 to 72 hours.

The risk that accompanies the general anaesthetic is extremely low if the patient is in good general health.

The risks of laparoscopy include accidental damage to the bowel or blood vessels within the abdomen or pelvis. These complications affect between one to two people per thousand cases and require immediate further surgery to correct any damage. One illustration of the advantages described before could be the treatment of colon cancer. As was presented in section 2.1.4, some surgeons still do not consider laparoscopic solutions as an alternative to cancer extirpation. But, there are the opposite affirmations, and one of them is that the procedure could be done by laparoscopy-assisted colectomy. According to MD Dmitry Oleynicov "I predict that much like laparoscopic cholecystectomy in the early 1990s, laparoscopic colectomy will be embraced and may become the new standard of care[2.9]. Patients recover more quickly and morbidity is lower.

Another factor which represents an advantage for laparoscopy versus open surgery is the lower cost resulting from faster patient recovery. [2.13]

All these arguments are making MIS more popular and indicate that the future of surgery looks set to make interesting new advances.

2.2 Image viewing and limitations during the surgical procedure

2.2.1 Introduction

The previous section presented the advantages to be gained from using laparoscopy procedures.

The price for these advantages is paid by the surgeon. The major limitation of this technique lies in the 2D image perception. Surgeons work with artificial two-dimensional video pictures on a monitor, which are acquired using the endoscope. This section describes the functionality of the endoscope and the principal limitations of using this instrument.

2.2.2 Endoscope and video monitoring systems

In the past, endoscopic procedures were performed without the aid of monitors. Surgeons visualised the body cavity directly through the eyepiece of the scope. This made performing procedures very difficult.

As new technology has developed, the endoscopic camera has had a significant impact on surgical techniques. It was now possible to achieve a good magnification of the image. The surgeon and the whole surgical team were able to view the image and in this way surgical intervention became more comfortable.

For the first time in 1956 in France, Soulas performed the first bronchoscopy. He used a television and a black and white camera connected to a bronchoscope.

Three years later, a laparoscopic procedure using a closed-circuit television program (the "Fourestier method") was demonstrated. A beam of light was transmitted along a quartz rod placed from the proximal to distal ends of the laparoscope.

In Australia in 1960 the first miniature endoscopic black and white camera was used.

Over the next forty years, endoscopic technology went from strength to strength. Special cameras built in different technologies NTSC (National Television System Committee), PAL (Phase alternation line) and SECAM (Sequential colour and memory) transmitted the images acquired by camera to the surgeon's eyes. The final image quality depends upon the number of resolution lines and pixels. The broadcasting standards for each are summarised in table 2.1.

Table 2-1.

SYSTEM	PAL	SECAM	NTSC
Number of lines	625	625	525
Visible lines (max)	575	575	486
Frequency Cycle/ sec	50	50	60
Frames/sec	25	25	30

Nevertheless there are certain drawbacks which have limited the performance of endoscopic systems. The resolution, number of pixels or the dot pitch represent the technical limitations that occur in the displaying process.

Aside from these limitations, there are disadvantages resulting from the 2D visualisation of the image. Only a 2D picture is shown on the monitor. The operative field is represented just by monocular depth cues. The ability to perceive the cue of a flat image is significantly reduced when the surgeon is confronted with a scene which has not been viewed before [2.14].

Another problem derives from the significant distance between monitor and surgeon. As a result the efficiency of the surgeon decreases. Apart from pictorial cues the picture can be disturbed by anti-cues. These may be effects from the monitor. Reflection resulting from glare is one of these anti-cues. Also, information about the resolution and the contrast of the picture which are parameters transmitted during the image acquisition process from the endoscope through the monitor represents a factor which may limit the quality of the image. Resolution and contrast may be measured on a specially designed optical bench and described as Modulation Transfer Function (MTF). For excessive glare the result is decreased contrast and resolution. Another aspect which influences the perceived image is barrel distortion. Field curvature occurs when there is improper focus of the centre from other parts.

The list of drawbacks does not end here. When a moving object is shown on a monitor, unless the speed with which it is moving is similar to the speed of pixels are refreshed, then jerky movements will occur. This effect is called "temporal aliasing". Fatigue and headache are the main consequences for the surgeon. Special software filters which can process the image or slow the object's movement could compensate for this problem.

Another problem is caused when surgeons must constantly look at the monitor and operate in a different direction. Spatial disorientation increases as the distance between surgeon and monitor increases. Optimum efficiency is obtained when the surgeon is able to look and operate in the same direction, as in open surgery. The medical term used here is the "gaze-down position". All these disadvantages show that 2D viewing systems are not the most suitable solutions for use in laparoscopic surgery. A possible option for overcoming the 2D visualisation problem is to develop 3D stereoscopic endoscope systems. To realise these systems different approaches have been considered. One solution is to use dual lens systems that capture slightly different images of the operating field. The systems provide independent images to the right and left eyes. As a result the right and left eye will capture different views of a single image. The second way to develop a 3D stereoscopic endoscope is to use a single optical channel which captures the image with a single objective lens placed at the distal end of the endoscope. After that the image is split into separate left and right eye images and acquired by a two CCD stereo camera [2.14]. Figure 2-7 describes the basic principles of these two 3D stereo endoscopes.

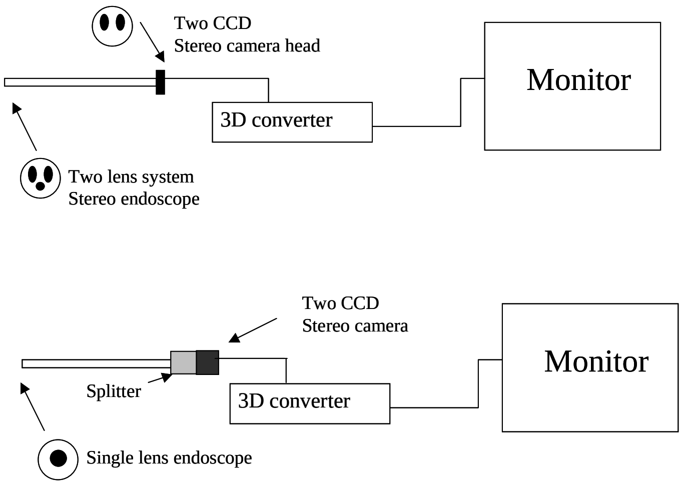

Figure 2-7: *3D stereo endoscope with two lens system or with single lens system*

In both systems the surgeon sees the images through the special liquid crystal glasses with shutter technology. The principle involves on-off switching of the transmission of images between eyes. When a picture is transmitted to one eye, the glass corresponding to the other eye will be turned off. Next, transmission will be activated for the previously blocked glass and the second one switched off. The surgeon's brain super-imposes these two pictures and the 3D picture is built.

A similar way to obtain a 3D visualisation is to use polarised glass. The shutter system is implemented in the monitor. The superimposition of the picture is also made by the brain.

Research has also shown that current 3D systems have many disadvantages. For example, visual cues are not similar to normal vision. The cues are unbalanced and disturb the sense of depth. This technique is harmful to the surgeon with prolonged use, gives incorrect depth perception and the consequences are headaches and eye strain.

Another limitation comes from the fact that a 3D effect requires very close distances and if placed further will not produce the desired effect.

In conclusion it has been shown that current 3D systems are not relevant achievements in the visualisation process for surgical interventions.

Looking further, new technological developments are being researched which bring new techniques aimed at overcoming current limitations. These systems are in different experimental phases and time will tell whether they will have the desired effect.

The next subsections will briefly describe some of these systems.

One of these involves the use of a head-mounted display which aims to normalise the visual-monitor axis. It consists of a monitor connected to the surgeon's head. The advantages include a lightweight device, a comfortable position, reduced mental stress and the fact that the display is cheaper than monitor systems, reduces eye strain and permits surgeons to view the abdomen directly.

However, this system also has its limitations, such as low definition and occasional nausea.

Another system was designed on the basis of the advantage obtained when the surgeon operates in the "gaze-down position". As mentioned before, the gaze-down position gives the surgeon alignment between direction of operation and the image displayed. This technique gives optimal efficiency. The 'view-site' project is based on the gaze-down position and uses a sterile screen placed above the patient's abdomen close to the surgical area. The position is comfortable for the surgeon but it does have some drawbacks. The system gives low resolution for the visualisation of the surgical area and the area of operational use will be significantly reduced [2.14].

A similar principle is used for another system, but instead of the monitor, a projector suspends the image in space. This technique is called the "suspended image system" (SIS). Use of this system requires a high precision retro-reflector and a beamer splitter. Resolution is optimised and the dimension of the image is close to its actual size. Moreover, the system improves the sense of depth. More work should be done on this system before it can really be considered as a viable alternative.

2.3 Expected future of minimally invasive surgery

As presented in the previous subchapter future developments are underway which aim to overcome current technical limitations faced by actual systems.

In the future it is expected that increased experience on the part of surgeons allied with further technical developments such as systems with tactile feedback and task-oriented robots will increase the range of procedures performed in laparoscopic interventions [2.5][2.6][2.8].

Therefore, further we have to ask ourselves whether the internet and laparoscopy might share a common future. Is there a future for cybersurgery?

According to R.K. Mishra, "the future remote handling technology will overcome the manipulative restriction in the current instruments...". It is likely that in 20 years' time some surgeons will operate via the internet using the latest high-tech tools to travel inside the human body [2.1].

In 1996 this idea came a step closer to reality. The first live interactive broadcast through the internet was realised. Video images were displayed in a 320x240 pixel window. Audio transmission was relayed separately. The quality of the transmission was good enough to allow satisfactory identification of anatomical structures [2.10].

At present, videoconferencing via the internet is often used. This method is viable for transmitting information in real time, allowing surgeons worldwide to work together during surgical procedures [2.11].

Transmission is possible using a low bandwidth internet connection and provides adequate image quality to support real-time surgical consultations. Because of this, a greater range of consulting procedures is available.

Considering the future improved surgical systems based on the new technological directions, the remote handling technology for the perspective of the long distance use and based on the Internet platform, will become reality [2.12].

2.4 Conclusions

The success of laparoscopy as a surgical technique derives from its ability to offer the surgeon a view inside the patient's body through small incisions in the stomach wall. Post-operative benefits for the patient include less trauma, shorter hospitalisation and a faster return to normal activity.

This chapter gave an overview of laparoscopic procedures and presented aspects related to this medical field. A brief introduction to endoscopic techniques and a description of the limitations of the visualisation process was given. A brief illustration was also given of how current and potential future techniques aim to overcome these limitations.

The next chapter will give a more detailed description of the commercial systems on the market and systems which are still in the development phase. The advantages and the drawbacks of each type of sensor (mechanical, optical, magnetic, acoustic and radio-wave) are also presented.

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3 3D localisation systems in MIS

3.1 Introduction

The present work focuses on the development of a 3D localisation system able to detect the position and orientation of laparoscopic instruments used in Minimally Invasive Surgery (MIS). The drawbacks specific to the acquisition process of 2D images during the endoscopic procedure were described in the previous chapter.

A new system is proposed which should overcome part of these drawbacks and will add more information about the instrument position. The system may be used for special training programmes for new surgeons. During these procedures the system will simplify the localisation process of some vague instrument positions and make it possible to record all phases of the laparoscopic procedure.

Different systems for similar purposes are being developed or already exist on the market. However, none of the existing systems fulfils the complete list of requirements for medical application. Therefore, in this chapter the problem definition is stated and an overview of existing devices is given, showing their advantages and drawbacks.

Finally, the solution to the medical problem is presented which will be investigated in more detail in the next chapters.

3.2 Problem definition

Discussions with an interventional laparoscopic surgeon at the Erasmus Medical Centre in Rotterdam, Prof. Bonjer and a representative of Olympus Multinational, producer of laparoscopic instruments, underlined the main concern of the medical profession regarding interventional procedures. The procedure itself is performed using long, thin instruments. An endoscope camera is used to show the surgeon what is happening inside the body, however this view is very limited and gives no information on instrument positions just outside the camera view. Even the use of 'standard' imaging techniques like X-ray, ultrasound, MRI and CT scans before or during the operation does not provide a satisfactory solution to this problem. As a consequence of losing the instrument view, surgeons often have to retract the instrument from the body and reinsert it.

One of the situations in which a surgeon may lose positional information on his instruments is when he involuntarily changes the position of his hand. For example, to manipulate instruments, the surgeon pushes pedals and his foot movement is transmitted through the body to the hand resulting in a different position of laparoscopic instruments. Because the surgery area is very small, uncontrolled movements of instruments of a few centimetres are critical during moments when the instruments are situated outside of the endoscope view. Surgeons cannot afford to risk continuing the procedure and instrument reinsertion is necessary to recalibrate the positional image. In this instance, a 3D localisation system can be very helpful to inform the surgeon about the position and orientation of the laparoscope. This will enable the surgeon to continue the procedure without having to recalibrate.

Another advantage suggested by Prof. Bonjer is the use of such a system for special training programmes for young surgeons. The system will store continuously the 3D position of the instrument. This information is useful after surgical intervention for the professor to analyse and to explain the most difficult sequences of the procedure. Moreover, in the previous chapter other drawbacks of the endoscopic system were identified. The endoscopic system gives the surgeon a poor estimation of the instrument cue (the positional information derived from the instrument's view on the monitor). Therefore, future 3D instrument localisation systems should provide clearer information about the laparoscope's cue.

Therefore the drawbacks characteristic of existing endoscopic systems and the possibility of gaining benefits which were described in the previous subsection 2.2.2 indicates the necessity to developing the actual system[3.12][3.13].

The first part of the research was to identify the requirements for such a system. The conclusions that arose during the discussions with the surgeons and completed with opinions from literature led to the main following specifications:

- *the system should supply additional and useful position information of the laparoscopic instruments to the surgeon in a clear and transparent way*
- *the system should by no means restrict the surgeon in his movements or sight*
- *it should be reliable and safe to both the patient and the medical personnel*
- *it should be simple and flexible to operate and easy to sterilise.*

The first requirement defines the technical parameters (positional information, precision, error) which are directly related to the performance of the system. The second, third and fourth requirements represent the safety conditions which should be respected during the interaction processes: doctor-system, doctor-system-patient and system-sterilisation equipment. The last three requirements lead to the evident conclusion that a wireless system is most suitable. The wireless solution will eliminate the risk that the system may impede the surgeon's actions which would reduce his efficiency.

The next step of the research work consisted of a brief identification and analysis of several systems with similar medical utilisation. This research helped to understand the advantages, disadvantages and limitations specific to these systems and to define the proposed system for laparoscopic intervention in more detail. Initially, a detailed literature study was conducted and it showed that a number of localisation and navigation systems exist in research programmes and even on the market.

As will be presented in the next section, most of the devices are mainly developed for use in MIS (minimally invasive surgery), but none of them meets all the requirements specified above.

3.3 Current methods and devices

In medical and in technical literature a multitude of systems for instrument localisation is presented. Instrument localisation implies the detection of the position and orientation of the tip of the instrument and its presentation in a visual, audio or any other relevant way. When an instrument is sequentially located during the medical procedure it is named the navigation instrument.

To localise the successive position of the instruments during navigation one can use various detection methods and visualisation techniques.

In laparoscopic intervention the main technique applied is endoscopy. As presented in chapter two, different drawbacks exist in the use of this technique. To support the surgeon in acquiring an accurate image of instrumental positions, various magnetic, optical, mechanical, radio-wave and ultrasound devices are being developed for MIS procedures.

Parameters such as resolution, accuracy and system responsiveness (additional characteristics of robustness, registration, and sociability are not considered here) should be respected in ranges related to the specific nature of a medical application.

- -***Resolution***. *Measures the exactness with which a system can locate a reported position. It is measured in terms of inch per inch or millimetre per millimetre for position, and degrees for orientation.*

- *-**Accuracy**. The range within a reported position is correct. This is a function of the error involved in making measurements and often it is expressed in statistical error terminology as degrees root mean square (RMS) for orientation and inches (mm) RMS for position.*

System responsiveness comprises:

- *- **Sample rate.** The rate at which sensors are checked for data, usually expressed as frequency.*
- *- **Data rate**. The number of computed positions per second, usually expressed as frequency.*
- *- **Update rate**. The rate at which the system reports new position coordinates to the host computer, also usually given as frequency.*
- *- **Latency**, also known as lag. The delay between the movement of the remotely sensed object and the report of the new position. This is measured in milliseconds (ms).*
- *-**Working volume** or **range**, which may be bound by intrinsic limitations such as mechanical linkage or signal strength. This is the volume in which a sensor accurately reports position. It is variously expressed in feet or meters, inches or feet in diameter, or as some portion of a geometric shape such as a sphere.*

To indicate what the studied devices have to offer in terms of the position detection method used, their precision, size and any harmful effect caused, the next subsections briefly present some existing magnetic, optical, mechanical, and ultrasound systems, describing their advantages and drawbacks. For devices still under development, the expected working principle is described.

3.3.1 Magnetic devices

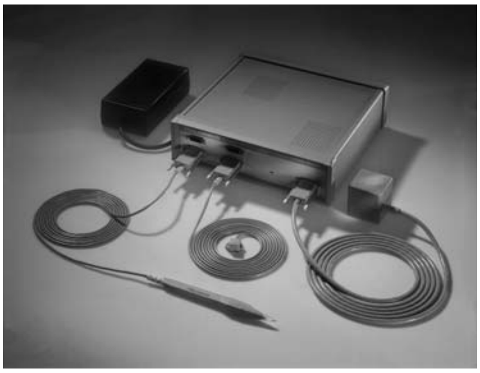

Figure 3-1: *Fastrak devices*

Fastrak

The Polhemus Fastrak was developed as a reworking of the vendor's Isotrak product. The Fastrak accurately computes the position and orientation of a small receiver as it moves through space. This device provides dynamic, real-time six degrees of freedom measurement of position (X, Y, and Z cartesian coordinates) and orientation. It is an accurate electromagnetic tracking system available on the market. The Fastrak system utilizes a single electromagnetic transmitter and can accept data from up to four receivers.

The transmitter and receiver contain enclosed electromagnetic coils that emit or detect the magnetic field. The transmitter is the system's reference frame for receiver measurements. The receiver's position and orientation is derived from the magnetic field measurements.

The use of advanced digital signal processing (DSP) technology provides an update rate of 120Hz (with a single receiver) and 4ms latency. The data is then transmitted over a high speed RS-232 interface at up to 115.2k baud or a via USB interface.

The accuracy reported is about 0.03 (0.7mm) inches RMS for a resolution of 0.002 (0.05mm) inches per inch (25mm).

FASTRAK uses low-frequency magnetic transducing technology in order to maintain a clear line of sight between receiver and transmitter.

Unfortunately, when transmitters or receivers are located close to large metallic objects such as desks or cabinets, the performance of the system is affected by ferromagnetic interference.

Figure 3-2: *LIBERTY device*

LIBERTY

LIBERTY is another Polhemus development in electromagnetic tracking. With a speed of 240 updates per second per sensor, LIBERTY makes it easy to track virtually anything that is non-metal.

The easy-to-use GUI interface allows four independent user-definable profiles for setting system parameters such as filtering, output formats, coordinate rotations, allowing multiple applications or users.

Like Fastrack, LIBERTY incorporates Digital Signal Processing (DSP) in connection with alternative magnetic currents (A/C) which provides the user with improved signal to noise ratios, increased range, stability, resolution and speed. Distortion Sensing is a

distinctive feature of the LIBERTY system. Each sensor is able to independently detect distortion within the environment, alerting the user to make appropriate changes as needed.
Specifications include a latency of 3.5 milliseconds, resolution of 0.015 in (0.38 mm) at 12 in. (30 cm) range and angular resolution of 0.0012°.

Patriot

The latest Polhemus product is the **Patriot**. This dual sensor has a range of five feet, resolution of 0.0015 in.(0.038 mm) and 0.01 degree and a static accuracy of 0.1 in (2.54 mm). RMS for the X, Y and Z position and 0.75 degrees RMS of orientation.

The latency is 17 milliseconds for both sensors and this represents a disadvantage over the Fastrak and the Liberty systems. The ferromagnetic sensitivity has distortive effects on measurements, a disadvantage also identified for the Fastrak and Liberty systems.

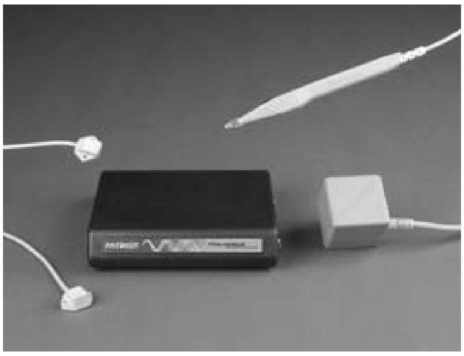

Figure 3-3: *Markers used by Patriot device*

Flock of Birds

Flock of Birds is a 6 DOF tracking system by Ascension Technology Corporation. It is intended for tracking human motions in character animation, medical, and virtual environment VE applications. In particular, Flock trackers are used for head tracking in flight simulators/trainers, head, hand, and body tracking in VE games and full body tracking for character animation, performance animation, virtual walk-through, and sports analysis. Flock of Birds has full 360° coverage without blocking or echoing problems and a fast measurement rate-up to 144 position and orientation measurements per second. It can simultaneously track up to 60 separate independent points out to 8 feet (2.4m) with the Extended Range Transmitter option. Each standard range transmitter allows operation within an approx. 3 foot (91cm) radius. The static accuracy position is about 1.8mm RMS and the angular accuracy 0.5 degree. Ascension claims it has the lowest lag of all trackers when tracking multiple points.

The Flock of Birds emitter radiates a sequence of DC pulses, in effect switching the emitted field off and on. This design is intended to reduce the effect of distorting eddy currents induced by changing magnetic fields in metallic objects. While it minimises the effect of conductive metals, sensitivity to ferromagnetic metals remains.

Figure 3-4: *Flock of birds device*

PC/BIRD

PC/BIRD is a new offering from Ascension Technology Corporation that uses the same patented pulsed DC magnetic technology employed in the Flock of Birds tracking system. Intended for use with PCs (an option which does not exist for Flock of Birds), this tracker is configured as an ISA-compatible board, a receiver that can be mounted on any non-metallic object, and either a standard or extended range transmitter.

Figure 3-5: *PC/Bird device*

With the standard range transmitter, PC/BIRD operates with a range of 4 feet (1.2 m), the extended range transmitter allows a range of up to 10 feet (3m). The static position accuracy is about 1.8 mm RMS and the angular accuracy 0.5 degree.

Measurements are made at the rate of up to 144 per sec. Additional cards and receivers may be used to track multiple objects simultaneously. An optional mouse, with three programmable buttons, is available for providing user inputs in 2D or 3D.

Accuracy varies from one location to another over this translation range and will be degraded if there are interfering electromagnetic noise sources or metal in the operating environment.

AURORA

The AURORA magnetic tracking system is developed by NDI. The system delivers real-time measurement. It is also flexible, accurate and reliable.

The sensors are as small as 0.9mm in diameter, and the system tracks up to ten sensors simultaneously with up to six degrees of freedom.

The sophisticated algorithms detect and indicate the magnetic field disturbances. Computer integration takes the form of the Application Programmers Interface and RS-232 communications.

The main disadvantage of the Aurora system lies in the difficulty of the calibration method used for the process of mapping the magnetic field. For optimal characterisation of the distortion in magnetic field lines, between 50 and 300 precalibration measurement points are necessary. Any metallic object will introduce changes in the magnetic field map. This disadvantage makes it difficult to use the system in the surgical environment.

Figure 3-6: *Aurora device*

Table 3-1. *Magnetic devices*

Product	Vendor	Dof	Accuracy	Frequency	No. sensor	Latency	Resolution	Working Volume
Fastrak	Polhemus	6	Linear Accuracy (RMS) 7mm Angular Accuracy (RMS) 0.15 °	120 Hz/ number of receivers	4,16	4 msec	0. 0002 mm/ mm,range 0. 025 °	3 - 9 m
Liberty	Polhemus	6	Accuracy(RMS)0.76 mm for x, y, and z, 0.0012 ° for receiver orientation	240 Hz/ sensor	8,12,16	3.5 msec	0 .00015 in (0.038 mm) at 12 in. (30 cm) range; .0012° orientation	90 cm
Patriot	Polhemus	6	Accuracy(RMS) 2.4mm x,y, and z, 0.75 ° for receiver orientation; < 5 ft range	60 Hz/ sensor	2	10 msec	0. 38 mm, range 0.01 °	1.5m
Aurora	Northern Digital INC	6	Accuracy positional 0.9-1.6 mm 3D angular 0.8 - 1.1 ° RMS	22-45 HZ	1,8,16	No specificate	No specificate	0.5m/0.5m /0.5m
Flock of Birds	Ascension Technology Corp.	6	Angular Accuracy (RMS) 0.5 ° Translation Accuracy (RMS) 1.8mm	Up to 144 Hz	4	7. 5 msec, 39-47 msec	0.5mm 0.1 ° RMS at 0.3m	1x2.4 m
PC/ BIRD	Ascension Technology Corp.	6	Translation Accuracy 2mm Angular Accuracy 1°°	Up to 144 Hz	4	10 msec ,	0.5mm 0.1 ° RMS at 0.3m	1.2x3 m

3.3.2 Optical and RF available methods and devices

Optical tracking method

A representative method for optical tracking, the phase-shift measurement (or phase-detection) ranging technique involves continuous wave transmission as opposed to the short pulsed outputs used in time of flight (TOF) systems. A beam of amplitude-modulated laser or RF is directed towards the target. A small portion of this wave (potentially up to six orders of magnitude less in amplitude) is reflected by the object's surface back to the detector along a direct path. The returned energy is compared to a simultaneously generated reference that has been split off from the original signal, and the relative phase shift between the two is measured to determine the round-trip distance the wave has travelled. For high-frequency RF or laser-based systems, detection is usually preceded by heterodyning the reference and received signals with an intermediate frequency (while preserving the relative phase shift) to allow the phase detector to operate at a more convenient lower frequency. The relative phase shift expressed as a function of distance to the reflecting target surface is[3.2]:

$$\Phi = 4\pi \frac{d}{l}$$

where Φ = phase shift, d = distance to target, l = modulation wavelength.

The phase shift between outgoing and reflected sine waves can be measured by multiplying the two signals together in an electronic mixer, then averaging the product over many modulation cycles. This integration process can be relatively time-consuming, making it difficult to achieve extremely rapid update rates.

Commercial optical systems available

OPTOTRAK 3020

The **OPTOTRAK 3020** by Northern Digital Inc. is an infrared (IR)-based, non-contact position and motion measurement system. Small IR LEDs (markers) attached to a subject are tracked by a number of custom designed sensors. The 3D positions of the markers are determined in real time; up to 256 markers can be tracked. The position sensor consists of three 1D charged coupled device (CCD) sensors paired with three lens cells and mounted in a 1.1m long stabilised bar. Within each of the three lens cells, light from the LED is directed onto a CCD and measured. All three measurements together determine the 3D location of the marker, which is calculated and displayed in real time.

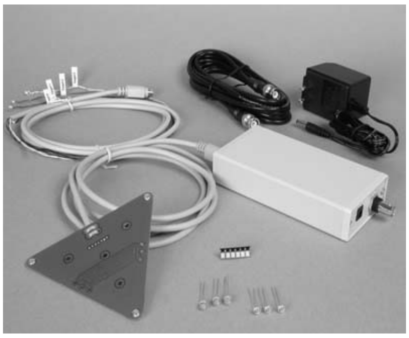

Figure 3-7: *Optotrack device*

The standard **OPTOTRAK 3020** system includes one position sensor unit, a kit of 24 markers, a system control unit, standard data collection, display, and utility software, together with cables and other hardware.

POLARIS

There are two variations of this new technology: the **passive POLARIS** and the **hybrid POLARIS**. Both of these systems combine a mix of features and capabilities providing solutions for tracking applications such as medical device tracking, robot calibration, and digitisation for part inspection.

Its compact size, friendly handling, real-time tracking algorithms and wireless reflective target based tools have made **POLARIS** a representative product in its market category.

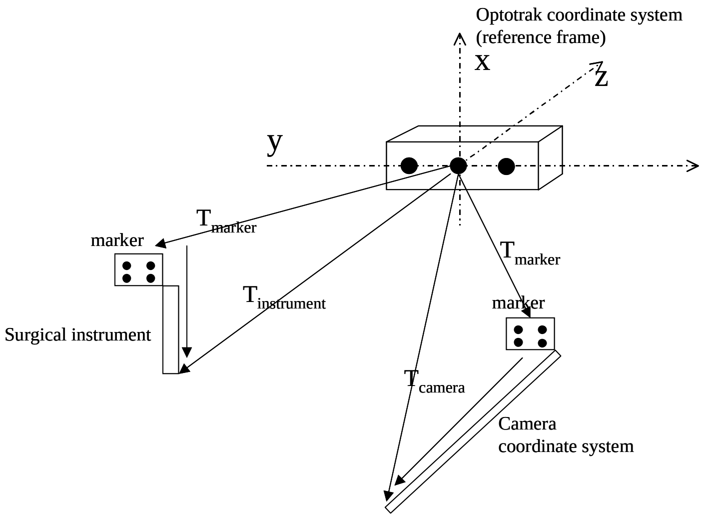

Figure 3-8: *The Optotrack setup*

- *the ability to track both active markers and passive targets enabling tool designs with high performance and good ergonomics*
- *advanced real-time tracking algorithms providing superior measurement accuracy,*
- *a small, lightweight position sensor that is easily mounted in virtually any indoor environment,*
- *an Application Programmer's Interface built into the POLARIS simplifying application development.*

The passive POLARIS is engineered to track the real-time position and orientation of passive reflective targets. These passive reflective targets may be retro-reflective spheres or discs. The innovative technology of the passive POLARIS removes the need for cable connections from the tools to the POLARIS and replaces the Tool Interface Unit (TIU) with a simple universal power supply.

Introduced with the passive POLARIS, the hybrid POLARIS has the capability of simultaneously tracking both active marker and passive target based tools.

Figure 3-9: *Polaris system*

MacReflex Motion Measurement System

The **MacReflex Motion Measurement System**, by Qualisys, Inc. also is designed to measure the 3-D motion of subjects in real-time. The system is comprised of (1) one or more MacReflex position sensors (a 3-D system uses from two to seven position sensors), (2) software to enable the user to set up and calibrate the field of view of the position sensors, and process the measured spatial coordinates of the target markers that are attached to the subject being tracked, (3) passive reflective target markers, (4) a calibration frame for 3-D measurements, and (5) a Macintosh computer system. The position sensor has two components: a CCD digital video camera, and a video-processor. The camera views up to 20 markers in real-time. It then sends the video image to the video processor which determines the centroid of each marker and determines its x, y coordinates. A program converts the x, y coordinates to enable calculation of position, displacement, velocity, acceleration, angles, angular velocity, and angular acceleration.

DynaSight

The Origin Instruments Corporation tracking product, **DynaSight**, is an electro-optical sensor with integrated signal processing that performs 3-D measurements of a passive, non-tethered target. A two-color LED on the front of the sensor indicates the tracking status to the user. In a typical application, the sensor is mounted just above the viewable area of a real-time graphics display. The sensor's field of view is a nominal 75° cone, and the sensor is pointed such that this field covers the comfortable range of head/eye positions for the user of the display. The sensor measures and reports on the 3D movements of a tiny target that is referenced to the user's forehead. The passive target itself can be mounted on eye glasses, stereoscopic goggles, or on the user's forehead. Larger high-performance targets are available that allow measurements at a sensor-to-target range of up to 20 feet.

The Active Target Adapter enables tracking of up to four active targets tethered to the Adapter. Five DOF are achieved with two targets, while 6 DOF can be achieved by tracking three or four active targets. DynaSight is the first in a new line of 3-D measurement products. It is planned that future systems will offer 6 DOF for HMDs using passive sensors and multiple sensors for networked operations in large virtual volumes.

Figure 3-10: *DynaSight Device*

RK-447 Multiple Target Tracking System

The RK-447 Multiple Target Tracking System, by ISCAN, Inc., is a video tracking system which can track up to 64 facial points at 60Hz with a latency of 16msec. It is a real-time digital image processor employing ISCAN's proprietary Simultaneous Multiple Area Recognition and Tracking (SMART) architecture. The ISCAN SMART processor computes the position and size of up to 256 areas that are within a particular range of intensity levels. Filtering the output of the SMART processor allows the complete system to specify targets of desired size, position, and intensity parameters from a field containing many potential targets.

After positioning the imaging sensor to include the desired field of view, the image grey level corresponding to the target may be selected. The areas of the video image whose intensity is within the grey level threshold setting are presented on the monitor as a bright overlay, letting the operator see precisely the video information being processed. For each threshold area, size and position data are computed and stored in a data table which may be accessed by an external computer.

The RK-447 Multiple Target Tracking System divides the image signal into a 512 (horizontal) by 256 (vertical) picture element matrix. As the targets' position and size data are automatically determined over the monitor image area, the data within the azimuth and elevation coordinate table correspond to the horizontal and vertical coordinates within the video matrix. These coordinate data are updated every 16 msec and are available for input to a computer. Parametric information may be input to the RK-447 to automatically limit the data set to targets within a particular size or position range.

Advantages and limitations of optical devices

The systems are developed for some specific medical applications and the possibility for extending the use of these for laparoscopic intervention is limited. The main limitation lies in accidental obturation situations when the surgeon or the instruments are interposed between optical markers and video camera. Then the optical transmission path is blocked. Also, the passive markers are large and heavy and the active markers are not wireless.

Table 3-2. *Commercial optical systems*

Product	Vendor	Accuracy	DOF	Frequency	Latency	Resolution	Markers	Wireless markers	Working volume
Selspot II	Selcom AB	Not Given	6	10KHZ	Not Given	0.025% of milliard	up to 120 LED,up to 16 videocamera	no	up to 200 m
OptotraK 3020	Northern Digital INC	0.1 mm for *x* , *y* coordinates 0.15 mm for z coordinate	6	60HZ	Not Given	0.01mm at 2.25 m	active (Ired)	no	1.28 x 1.34 m at 2.25 m distance, 34o x 34o square
Polaris	Northern Digital INC	0.35 mm RMS[1]	6	60 HZ	Not Given	Not Given	Passive/active	yes/no	
Mac Reflex	Qualisys, Inc	Depends of no. of sensor 2-7 for 3D	6	50-200Hz	Not Given	Not Given	Passive max 20	yes	0.5m-30m indors
Dyna Sight	Origin Instruments Corporation	2mm,CRT-8mm DRT,0.05mm for 40cm range	6	64Hz	9-28 msec	0.1mm cross range,0.4 mm down range	Passive/active	yes/no	0.1-1.5 for 7mm target
RK-447 Multiple Target Traching System	Iscan ,INC	Not Given	6	60HZ	16 msec	Not Given	256	no	Not Given

3.3.3 Inertial available devices

The inertial systems use accelerometers and gyroscopes. Orientation of the object is computed by jointly integrating the outputs of the rate gyros whose outputs are proportional to angular velocity about each axis. Changes in position can be computed by double integrating the outputs of the accelerometers using their known orientation.

The advantages of these systems are: unlimited range, fast, no LOS (loss of sight) problems, no magnetic interference problems, senses orientation directly, small size, low cost. All these characteristics indicate a suitable system. But there are drawbacks which derive from the following reasons. Inertial sensors measure acceleration or motion rates, so their signals must be integrated to produce position or orientation. Noise, calibration errors and the gravity field impart errors on these signals, producing accumulated position and orientation drift. Position requires double integration of linear acceleration, so the accumulation of position drift grows as the square of elapsed time. Orientation only requires a single integration of rotation rate, so the drift accumulates linearly with elapsed time.

To compensate for these drawbacks there are solutions which combine different measurement technologies.

Hybrid systems use multiple measurement from inertial, magnetic and optical techniques and combine these to compensate for the shortcomings of each.

Active target magnetic and passive target optical, inertial sensors and active optic target, are some examples of hybrid systems.

The 3D measurement performance of inertial systems is limited by the condition that it must be used in combination with a hybrid system.

This solution is not convenient for the 3D laparoscopic system where one of the important selection criteria is the simplicity of the system.

3.3.4 Ultrasound methods and devices

Ultrasound is the propagated mechanical vibration of the molecules of a material in a solid liquid or gas. Depending on the direction of propagation there are two types of waves. Waves which have just one propagation direction are called ultrasound longitudinal waves. The field specific for this type of wave is the gas field. The molecules vibrate straight and back through the direction of propagation. The second type of wave has one more direction of propagation, the transversal one. The molecules vibrate perpendicular to the propagation direction of the waves. The last type of waves are specific for the propagation in solids and liquids.[3.20]

For propagation in solids and liquids the ultrasound waves used are in the range of the MHz spectrum. Waves propagated in the air or other gases are in the kHz range.

When ultrasound travels through a solid medium it is attenuated due to near divergence, (the intensity along the beam axis is reduced because of the spreading of acoustic energy over a large beam area), absorption (the energy is transferred from the beam to the body tissues and then eventually transformed into heat) and deflection (energy is deflected out of the beam). When an ultrasound wave propagates through different media, some of the energy is reflected and the remainder is transmitted through the media. The direction of the echo (the reflected wave) depends on the orientation of the sound wave with respect to the reflection surfaces.

Ultrasound has a non-ionising effect, which means that it has enough energy to dislodge orbiting electrons from atoms. The most common physical effect is heat generation.

The literature [3.22][3.23] confirmed no adverse effect due to the use of ultrasound in patients.

The imaging modality based on the propagation of ultrasound in tissue is called ultrasonography [3.21]. This technique takes various forms in practice, e.g. pulse echo ultrasound, Doppler ultrasound, sonomicrometry, which are presented in the next subsections.

SonoWand *a commercial navigation system based on ultrasound imaging*

SonoWand is a novel, single-rack intraoperative imaging system which can be used as a standard neuronavigation system, as a stand-alone ultrasound scanner, and more importantly, as a navigation system which integrates 3D ultrasound imaging with an optical tracking of the ultrasound probes.

Figure 3-11: *Sonowand device*

This novel design saves space in the operating theatre and enables the surgeon to plan and navigate using preoperative MR or CT images and also, in a similar manner, navigate directly by intraoperative 3D ultrasound.

Figure 3-12: *Sonowand marker*

The system supports a number of optical trackable instruments and accessories such as pointers, ultrasound probes and a reference frame.

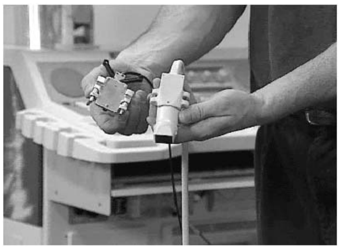

Figure 3-13: *Sonowand probe*

The ultrasound probe is equipped with a special tracking adapter which is factory-precalibrated in order to ensure high navigation accuracy.

Ultrasonography techniques

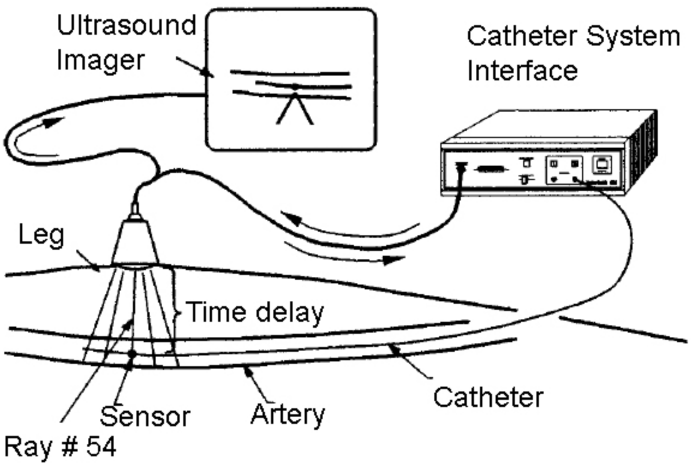

Figure 3-14: *The ultrasonic system presented by Vilkomerson*

Pulso-echo ultrasound for medical-tools localisation and navigation

The ultrasonic imaging systems presented by Vilkomerson et al. [3.17] uses a display to visualise the location of a medical instrument, an ultrasound transducer placed at the tip of the medical instrument, an ultrasound scanner on the patient's skin, and a signal processing device.

The working principle of the system is based on the pulse echo technique, by means of which internal anatomical images are generated by transmitting ultrasound waves via an ultrasound scanner to the body and processing their reflection. When an ultrasound transducer is placed on the tip of the instruments, it will provide an electrical signal each time it is struck by an ultrasound wave [3.14].

In ultrasound systems, the returning echoes from the transmitted pulses provide one line in an ultrasound image, which corresponds to the sequence of interfaces encountered by the ultrasound pulse as it propagates downward into the body of the patient. To detect the position of the transducer, one must know the line at which the transducer appears on the image and the position on that line. Since the transducer emits an electrical signal every time it is struck by an ultrasonic wave, its location can easily be detected, and, by means of the processing devices, it can be visualised on the display (for example as a coloured dot). By sequentially indicating the determined location on the ultrasound image it is possible to navigate the medical instrument to the intervention site[3.25].

When compared with other techniques, the use of this system is limited due to the poor imaging capabilities offered by ultrasound. The 2D images that are generated in the plane of the scanner are rather blurred views of the anatomical structures. Thus ultrasound systems based on the echo pulse method where the propagation is inside of the human body do not achieve the performance expected of a 3D localisation system for laparoscopic tools.

Doppler effect for MIS instrument position.

Beside the pulse echo techniques used by the above-presented systems, the Doppler effect may be used for instrument localisation. Let us assume that an ultrasound transmitter generates a frequency f;

it is travelling with speed U in the human body towards a receiver. The Doppler shift between the received echo and the reverberation in the direction of the receiver is as follows [3.15]:

$$\Delta f_r = \frac{2Uf\cos\phi}{c}$$

where Δf_r is the Doppler frequency shift, f is the incident frequency, U is the speed of transmitter, c the speed of sound and the ϕ the angle between the ultrasound beam and the transmitter direction.

The Doppler effect is applied in interventional radiology and cardiology as a non-invasive method for catheter tip localisation and for detection of faulty tip placement. Normally, the speed of sound in the tissue is in the range of 1500m/sec and the maximum speed of instruments is about 0.5m/sec. This means that the frequency shift will be in the kilohertz range.

Therefore, the human ear may act as a sophisticated Doppler signal processor in a large number of clinical applications, as shown in [3.24].

The Doppler receiver is placed on the human body close to the operating area. The catheter is normally insulated with a saline solution. When the catheter is moved into the vicinity of the receiver, the saline solution modifies the frequency shift. Depending on the position of the catheter a shift frequency is registered; the human ear could detect the frequency change and process the information relating to the proximity of the sensor to the Doppler receiver. Then the wrong placement of the instrument is detected. For laparoscopic intervention, a similar procedure may be developed, but the achieved accuracy for the laparoscopic 3D localisation will be less then required[3.19].

Sonomicrometry

Sonomicronometry is another medical application of ultrasound, where ultrasound is used to measure distance. Generally the transducers are made of piezoelectric materials that operate at frequencies of 1MHz and higher. In order to perform distance measurement between two transducers, one of them emits an ultrasound burst and the second receives the ultrasound signal. The counted time of the ultrasound transmission is a direct proportional representation of the distance between the transmitter and the receiver (Time of flight principle [3.4]).

The Sonometrics Corporation used the concept for heart -catheter navigation. The system uses piezoelectric crystals on the patient's chest and at the tip of the catheter and computes the location of the medical tools by processing the data from the ultrasound transducers. An important boundary is the homogeneity of the media. Elapsed distance is a product between time of flight and the speed of sound. The speed of sound has a constant value just in homogenous transmitted medium. In the Sonometrics Corporation application the method was feasible taking account of the fact that the intervention site is restricted to the heart volume and thus the distance transmitter receiver could be considered homogenous.

Unfortunately in laparoscopic intervention, the medium is not homogenous due to tissue structures and bone structures. Under these circumstances the system cannot be used in laparoscopic intervention.

3.3.5 Advantages and drawbacks

The previous section illustrated a number of devices employed to solve a range of medical problems that appear in different surgical interventions. It is the objective of this section to point out the advantages of existing systems and to present their shortcomings from the point of view of laparoscopic procedure.

Of course, each of these systems was developed to solve a very specific medical problem and on the whole they have all been successfully implemented and tested. It should therefore be understood that it is important to characterise existing medical guidance systems from the point of view of interventional laparoscopy and to clearly indicate the reasons why such systems or methods cannot be used as a solution to our medical problem.

At present, in interventional laparoscopy, the most frequently used method for instrument navigation is endoscopy. It offers high-contrast images of the organs and tissues and helps to assess the location of diseases. Unfortunately the method does not supply enough information about the position of laparoscopic instruments. For example, the depth of the instrument is difficult to estimate. 2D images limit the precision of the instrument localisation. Moreover, the instruments tracking procedure is difficult when the surgeon uses multiple instruments and some of them may be moved accidentally out of the endoscopic view area.

Literature studies showed that magnetic, optical, inertial and ultrasound systems for instrument location exist on the market or are still under development.

All these systems may supply additional information for the endoscopic method. The advantages and shortcomings of using those systems will be pointed out in the following subsections.

Fastrak, Liberty, Patriot, Flock of Birds PC/BIRD and Aurora are magnetic systems based almost on the same detection technique. The major drawbacks for these systems are ferromagnetic and eddy current interference in the presence of metal objects. Different algorithms are developed to reduce the resulting errors. However, the solutions are not simple and many calibration measurements are required to achieve better accuracy. These arguments exclude the possibility of developing a system based on this method.

Optical and RF methods used give the best resolution on the 3D detection process. Optotrak 3020 and Polaris systems, Mac Reflex device, DynaSight system and the RK-447 Multiple Target Tracking System provide optical devices for the medical market.

The systems are developed for some specific medical applications and the possibility for extending the use of these for laparoscopic intervention is limited. The main limitation lies in accidental obturation situations when the surgeon or the instruments are interposed between optical markers and video camera. Then the optical transmission path is blocked and require a system recalibration.

The ultrasonic pulse echo method, one of the well-known imaging techniques, means a rather limited interventional area. The limitation results to the homogenous medium. The variation of speed of sound makes it difficult to predict the instrument position. Even though pulse echo ultrasound scanners are rather inexpensive and easy to use, their application in laparoscopic surgery still remains limited.

Sonomicronometry relies on the same principle. Because the non-homogenous tissues in the body, the technique is excluded as a solution. As a conclusion to the presented systems and methods, Table 3-3 gives an overview of the most common characteristics of those devices and techniques.

From the previous table it can be seen that a variety of systems have been developed. Considering the advantages and drawbacks specific to each one, it was decided to continue the research to develop a completely new system to determine the position and orientation of laparoscopic tools.

The next subchapter describes the major reasons identified which supported the decision for one of the investigated fields: magnetic, optical and ultrasound.

3.4 Solution to the medical problem

The literature study on 3D localisation of position and orientation of laparoscopic instruments has shown that only few devices could solve partial the problems involved in laparoscopic intervention. A compromise solution is proposed in this section by developing a system that fulfils the requirements imposed by medical staff (a wireless system, a system position error not larger than 9-10mm) and add more information to the imaging provided by endoscopy.

As was explained in the previous chapter an endoscopic camera is used to show the surgeon what is happening inside, however this view is very limited and gives no information on instrument positions outside the camera view. Even the use of 'standard' imaging techniques such as X-ray, ultrasound, MRI and CT scans before or during the operation do not solve this problem satisfactorily.

The system to be developed should include a marker or transmitter on the laparoscopic instrument, endoscope and the trocar. The other part of the system will include the tracker for the markers. This piece of equipment must determine the position of the different markers/transmitters with respect to each other.

Table 3-3. *Overview of the strenghts and the weakness of different technologies*

Technology	Descriptions	Strengths	Weaknesses
Optical	Use a variety of detectors, from ordinary video cameras to LEDs, to detect either ambient light or light emitted under control of the position tracker. Infrared light is often used to prevent interference with other activities.	High availability Can work over a large area Fast No magnetic interference problems High accuracy	LOS necessary Limited by intensity and coherence of light sources Weight Expensive
Acoustic	Use three microphones and three emitters to compute the distance between a source and receiver via triangulation. Use ultrasonic frequencies (above 20 kHz) so that the emitters will not be heard.	Inexpensive No magnetic interference problems Lightweight	Ultrasonic noise interference Low accuracy since speed of sound in air varies with environmental conditions Echoes cause reception of "ghost" pulses LOS necessary
Magnetic	Use sets of coils (in a transmitter) that are pulsed to produce magnetic fields. Magnetic sensors (in a receiver) determine the strength and angles of the fields. Pulsed magnetic field may be AC or DC.	Inexpensive Accurate No LOS problems Good noise immunity Map whole body motion Large ranges - size of small room	Ferromagnetic and/ or metal conductive surfaces cause field distortion Electromagnetic interference from radios Accuracy diminishes with distance High latencies due to filtering
Inertial	Use accelerometers and gyroscopes. Orientation of the object is computed by jointly integrating the outputs of the rate gyros whose outputs are proportional to angular velocity about each axis. Changes in position can be computed by double integrating the outputs of the accelerometers using their known orientations.	Unlimited range Fast No LOS problems No magnetic interference problems Senses orientation directly Small size Low cost	Only 3 DOF Drift Not accurate for slow position changes

In addition, the systems must not interfere with the actions performed by the surgeon as part of the operation.

Therefore, the markers must be small and light, preferably wireless and easy to attach to different instruments.

To determine the nature of markers, different solutions were investigated.

The following energy domains are available for the 3D position measurements: magnetic, optical (visible and infrared) and acoustic Each has its own advantages and disadvantages. For example, magnetic fields may be disturbed by the presence of ferromagnetic materials and at higher frequencies by eddy currents. For optic and acoustic the propagation paths may be obstructed by personnel and/or instruments. The optic detection is accurate but the passive optical instruments are large and heavy and the tracker is expensive. The active optical markers consist of a wired solution. In case of ultrasound techniques the principle of determining the marker position is not accurate as for the optic detection but still there are advantages specific for this method. These are found in the low speed of wave propagation (comparing with light), price and dimensions of sensors. If there are methods to increase the accuracy of ultrasound detection all these differences are in favour of the acoustic domain. Therefore we use an acoustic system for measuring position and orientation. The marker is the acoustic transmitter itself.

3.4.1 Ultrasound markers setup for 3D detection of laparoscopic instrument

The process of developing the medical guidance system started to define the optimal placement for the ultrasound sensor.

Different requirements were considered with the following conclusion: the sensor should not disturb the surgeon in his movement. Also, the presence of the sensor should not stress the patient.

The placement of sensors inside the patient's body may lead to medical complications. Depending on the size, the shape and the materials of the sensor, it is difficult to keep the sensor sterile when the sensor is placed inside the body.

Therefore, the most suitable placement is outside the patient body attached to the laparoscopic instruments. The position of a single laparoscopic instrument is well described by two *x, y, z* measurements along the instrument. When the rotational position information is also required a third measurement point might be necessary.

Different options exist (with the restriction that sensors do not enter the patient's body). For example, one measurement can be done on the trocar *T(x, y, z)* and the other on the instrument handle *P(x, y, z)*. When the length of the instrument is known the exact informational position *X(x, y, z)* of the instrument tip is determined. The following rationalising way describes the localisation algorithm. Considering L the length of the instrument, *P(x, y, z*) the placement of the ultrasound transmitters on the tip of the laparoscop *T(x, y, z)*, the placement of the second ultrasound transmitter on the trocar and *X(x, y, z)* the unknown position of the instrument tip inside the body, we can express the *X(x, y, z)* coordinate (Figure 3-15):

X(x,y,z)= f(P(x,y,z), T(x,y,z), L)

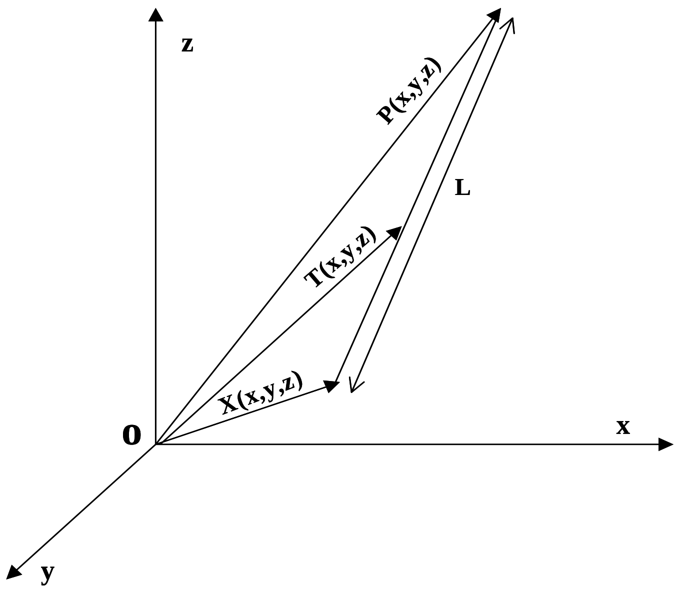

Figure 3-15: *3D representation of position of instrument*

An array of ultrasound receivers with a geometrical distribution in a horizontal plane situated above the surgical table will detect the position of the sensors (markers). Data is read at every corner of a square with the dimensions of 2x2m in this plane. In each corner a mini array of 4 receivers is placed to 3-dimensionally locate the transmitter positions. The 3D position is calculated directly from standard geometrical equations. The fourth receiver in each array will produce redundant information that can be used to check the validity of the position calculation [3.6] [3.7][3.8][3.9][3.10].

Another important aspect of the instrument localisation it might concerns the determination of the angular rotation. One transmitter placed on the laparoscop cannot provide this information. The solution to the problem involves placement of two transmitters close to each other. Figure 3-16.

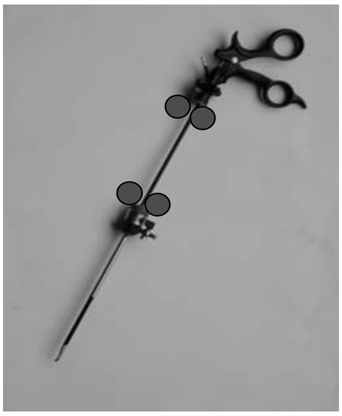

Figure 3-16: *Position of ultrasound markers on laparoscopic instrument*

The angular variation α of the median axis of the closed markers represents the change of orientation of the laparoscop.

The calculation of α derives from the 3D position of sensors and the known distance between the sensors.

The Figure 3-17 suggests the algorithm calculation for a change of orientation of the laparoscop for two successive positions.

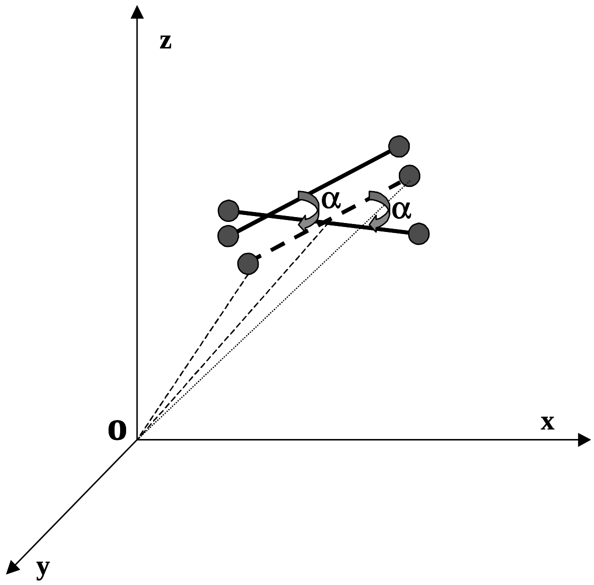

Figure 3-17: *3D representation of the orientation of an instrument*

The position of the tip of the instrument will be calculated considering the next equation.

$$\vec{X} = \vec{P} + (L(\vec{T} - \vec{P})) / \left|\vec{T} - \vec{P}\right| \tag{3-1}$$

The Figure 3-18 shows the schematic placement for the receivers and transmitters.

3.5 Concluding remarks

Chapter II and chapter III described the general requirements which should be respected in the process of developing a system used to localise the 3D position and the orientation of laparoscopic

instruments. They also discuss systems on the market and under development and the advantages and drawbacks specific to each.
In conclusion, it was shown that only a few devices could respect the requirements for laparoscopic intervention and a completely new system should be designed. Subchapter 3.4 gives a general view of the way to implement our solution to develop the system. A detection system based on the ultrasound markers for calculation of laparoscopic instrument positions was described further.

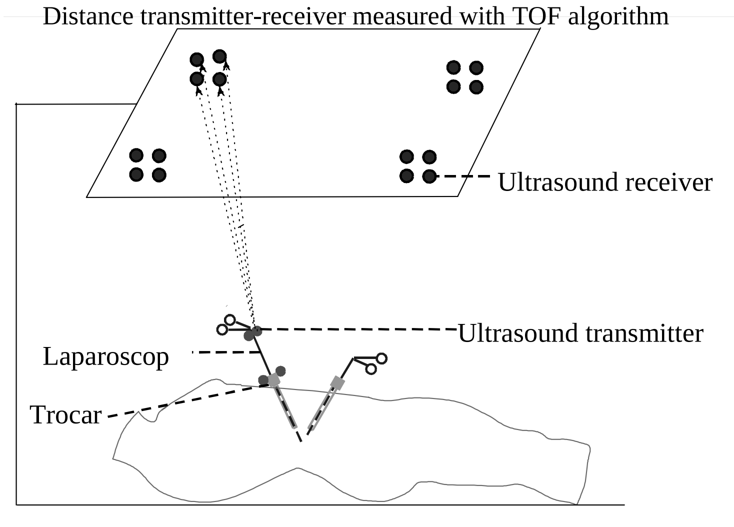

Figure 3-18: *The positions of ultrasound transmitters on surgical instruments and the receivers in the horizontal plane situated above of surgical table*

The next chapters will go into more detail on the detection methods used for the markers' position. Several approaches were studied and the measurements based on these methods will indicate the advantages and limitations of each.

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4 Distance detection using ultrasound

4.1 Introduction

In the previous chapter of this thesis the ultrasound system for detection of position and orientation of laparoscopic instruments for minimally invasive surgery was presented and the general set-up on which the instrument localisation is based was introduced. It was shown that the system should function in accordance with the requirements imposed by medical staff. However, besides those requirements, other aspects need to be investigated. To start with, this chapter presents a detailed analysis of the ultrasound transmitters and receivers used and the methods and ultrasound techniques used for distance detection and illustrates a series of measurements and results.

The ultrasound transmitters consist of piezoelectric transponders working at 40kHz. Before this transmitter was chosen, a number of possible ultrasound transmitters were studied.

4.2 Ultrasound transmitter-receiver

The first important criterion which determines the optimal choice of ultrasound transducer is the stability of the sensor related to variations in climatic environment.

The distance perceived by the acoustic sensor is directly proportional to the sound velocity. Investigating the propagation of sound through the air it was found that certain climatic factors exert an important influence on the speed of sound. Humidity and temperature represent parameters which have significant influence on distance estimation.

For the ultrasound waves, the amplitude of the sound pressure is reduced due to friction losses in the transmission medium. Knowing the value of this absorption loss, or attenuation, is crucial in determining the maximum range of a sensor. The attenuation of sound in air increases with the frequency and at any given frequency the attenuation varies as a function of humidity. The attenuation of sound as a function of frequency and humidity is illustrated in Figure 4-1[4.1]. The study of this function reveals that for humidity variations between 45-65%RH, the 40KHz frequency present minimal variations in speed of sound. Furthermore the humidity in an operating suite is usually well controlled and hence a humidity sensor is not required when an ultrasound sensor of 40kHz is used.

To estimate the influence of temperature on sound velocity the following formula was considered [4.3].

$$V = 20.056\sqrt{T} \qquad (4\text{-}1)$$

where V is the sound velocity and T is the absolute temperature.

Therefore, when the temperature is modified in the range of 0-40 °C the sound velocity changes in the range from 330-360 m/s. Such a large effect must be taken into account for the determination of the distance. In operating theatres the temperature is well controlled with a deviation of $\pm 0.5°C$. This implies speed of sound variations of approximations 0.5 m/s. For a distance of 2m, the resulting uncertainty in distance then becomes about 2.5mm.

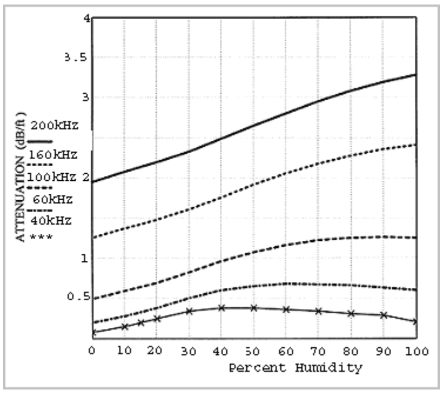

Figure 4-1: *The attenuation of sound as a function of frequency and humidity*

It was concluded that the optimal acoustic transmission frequency is around 40KHz. The attenuation of ultrasound waves around this frequency is minimally influenced by humidity variations. The influence of temperature variations on speed values, however, is too large to be ignored. Therefore compensation based on Eq. (4-1) should be applied. This means that real-time temperature measurement is required.

Based on the above considerations the air ultrasonic ceramic transducer developed by Quantelec with a resonating frequency of 40KHz was choose.

Characterisation of the ultrasound ceramic transducers

Ultrasonic Ceramic Transducers transfer acoustical energy to mechanical energy or vice versa. The Standard Open Type Transducer is built to incorporate the fundamental structure of a piezoelectric ceramic element of the monomorph type with a conical metal resonator. This combination provides high sensitivity (over 65DB/V/μBar), wider bandwidth, optimal temperature and humidity durability, stable electrical and mechanical characteristics and small size.

The general transducer design features a piezo ceramic disc bender that is resonant to a nominal frequency of 20-60kHz and radiates or receives ultrasonic energy. They produce sound waves above 20kHz that are inaudible to humans and the ultrasonic energy is radiated or received in a relatively narrow beam.

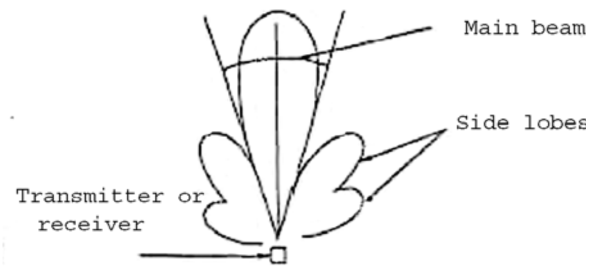

Figure 4-2: *The ultrasonic beam pattern for an ultrasonic ceramic transponder*

The “open “type ultrasonic transducers design exposes the piezo bender bonded with a metal conical cone behind a protective screen. The “enclosed” type transducer design has the piezo bender mounted directly on the underside of the top of the case which is then machined to resonate at the desired frequency. When compared to the enclosed transducer the open type will develop more electrical output at a given sound pressure level (high sensitivity) and exhibits less reduction in output as the operating frequency deviates from the normal resonant frequency (greater bandwidth). The open type transmitter will produce more output for a specific drive level (more efficient). The enclosed type transducer is more suited to dusty or outdoor application and the surgical environment is not characterised by those factors.

In the light of this information an open type transducer was chosen. The alternative of a close type transducer may appeal because it is easy to sterilise. However, the literature indicates that for the open type suitable sterilisation techniques also exist.

More reasons supported the decision to choose the resonating frequency of the transponder as 40kHz. During the process of selecting the proper transducer it is important to be aware of the principle of sound propagation. Since sound is a wave phenomenon, its propagation and directivity are related to its wavelength (λ). Due to the reciprocity of the transmission and the reception, the graph (Figure 4-2) portrays both power radiated along a given direction (in the case of wave production), and the sensitivity along a given direction (in the case of wave reception).

The angle which represents the half-width of the main beam could be expressed by the following[4.9]:

$$\sin\left(\frac{\alpha}{2}\right) = \frac{\lambda}{D} = \frac{V}{DF} \tag{4-2}$$

Where "D" is the effective diameter of the flexure diaphragm, "V" is the velocity of sound and "F" is the operating frequency. The above relationship applies if $\lambda<D$. For $\lambda>D$, the power patterns tend to become spherical in form. Thus, narrow beams and high directivity are achieved by selecting a large "D" in relation to λ. In the case of open type transducers, other factors should be considered. The beam shapes depends on the angle and diameter of the conical cone attached to the bender inside of the housing and it cannot simply be calculated by the diameter of the housing. In general it can be concluded that a higher resonance frequency in the 20kHz-60kHz spectrum will help to miniaturise the sensor.

Otherwise, for any wave propagation the intensity of the acoustic power is decreased with the distance to the sound source. The decreased intensity is a combination of two effects. The first is the inverse square law or spherical divergence in which the intensity drops 6dB per distance doubled. This is common to all wave phenomena regardless of the frequency. The second effect that causes the intensity to decrease is the absorption of wave energy by the air. Absorption effects vary with humidity and dust content of the air and, most importantly, they vary with the frequency of the wave. For lower frequencies the absorption is lower and the propagation is better for a long range.

Therefore an operating frequency of 40kHz seems to be an optimal choice also from the perspective of the transponder size and sound.

Considering all the information presented the 40KHz ultrasound transmitter (receiver) SQ-40-T(R) was chosen, see Figure 4-3:

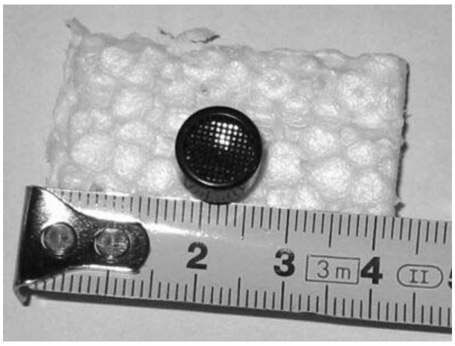

Figure 4-3: *The ultrasound transducer used in set-up measurement*

The table describes the parameters of the transmitter - receiver used[4.9]:

Table 4-1. *The transmitter and receiver characteristic*

ITEM	SQ-40-T	SQ-40-R
Transmitting sensitivity	110db	-
Receiving sensitivity	-	-70
Resonant Freq. Trans	40±1 KHz	-
Resonant Freq.Rec.	-	40±1 KHz
Directional Angle	Aprox 60^o	Aprox 60^o
Max.Input Volts Vrms	10	-
Impedance	700	30K
Capacitance	2000+20% pf	2000+20% pf

Table 4-1. *The transmitter and receiver characteristic*

Pulse rise time	0.7 ms	-
Temp range	-20+60 C	
Trans Selectivity	20 min	-
Rec Selectivity	-	25 min

Figure 4-4 represents the design for the shape of the transducer. The dimensions are listed below.

A = 9.8 (.39) mm (inch)
B= 7.7 (.30)
C= 6.6 (.26)
D= 8.5 (.33)
E= 4.8 (.19)
F= 0.5 (.02)

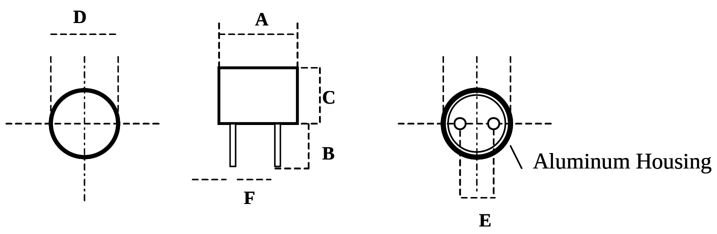

Figure 4-4: *Design for ultrasound transponders*

Speed of sound variation as an effect of temperature fluctuation

In the last section the influence of temperature variations on the speed of sound estimation was discussed. The error of 2.5mm for a detection range of 2m requires investigation of methods for reducing this effect.

Feynman demonstrated that sound velocity could be expressed as follows:

$$c^2 = \gamma R \frac{T}{\mu} \tag{4-3}$$

where c is the speed of sound, R the molar gas constant, T absolute temperature, γ the specific heats of the air and μ the molecular mass[4.11].

The formula explains the relation between the temperature and the speed of sound.

In literature the Feynman formula for the speed of sound in air is approximated as:

$$V = 20.056\sqrt{T} \tag{4-4}$$

For preliminary measurements we investigated the stability of the ultrasound transponder related to the compensation formula Eq. (4-4). The transmitter and the receiver (40kHz) were placed 30cm apart. An ultrasound signal was transmitted and received. The phase shift between transmitted and received signal was measured and related to the temperature variations[4.5].

For a temperature of around 23°C (standard temperature for an operating suite) we found a relation between phase shift and temperature (Figure 4-5).

$$\phi = \alpha T + \beta \tag{4-5}$$

Φ¸ phase shift, T¸ temperature expressed in °C

α = - 42.666, β= 1098.1

As presented, the main influence on speed of sound variation is temperature change. The humidity has small influence which can be neglected. For a fixed distance the reason for a change in phase shift is the variation of the speed of sound. Figure 4-6. compares the relative changes of sound velocity derived from Eq. (4-4), and the relative change in sound velocity found by Eq. (4-5)) plotted in time.

The comparison shows that the maximum uncertainty of speed of sound when Eq. (4-4) is used is about 0.5m/s. For the usual value of 344.5m/s at 23°C the uncertainty in speed after temperature compensation corresponds to an uncertainty in distance of 0.4mm for 2m range.

This remaining uncertainty is ascribed to secondary effects like possible changes of air flow (convection).

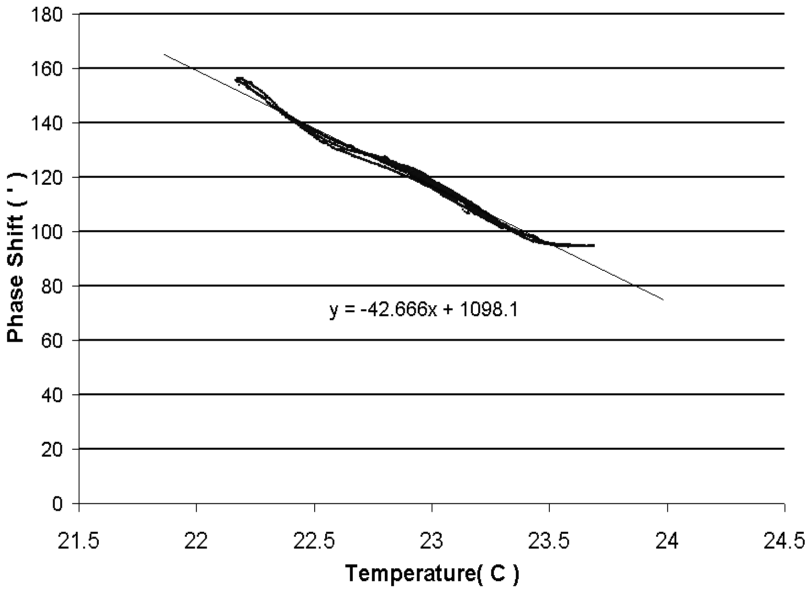

Figure 4-5: *Phase shift dependence on temperature at 23°C*

Speed of sound variation as an effect of humidity variations

For the same set-up the values of humidity were registered. For the humidity variations the phase shift did not show any significant change. The conclusion is that humidity has a minimal influence on speed of sound propagation.

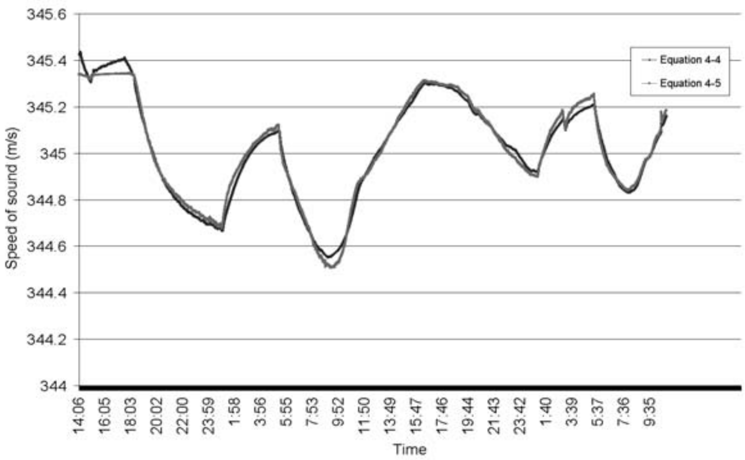

Figure 4-6: *The relative variations of sound velocity (over 46 hours) according to equations(4-4) and (4-5)*

Conclusions on ultrasound transponders

Up to this point the work concentrated on the determination of the characteristics of ultrasound transponders and sound propagation.

Different factors, effects and phenomena were considered during the identification process of sensor requirements.

Considering ultrasound wave propagation, the absorption factor, the "beam pattern" related to the frequency of the ultrasound transmitter and the climatic factor which affect the estimation of the speed of sound, an appropriate ultrasound transmitter/receiver operating at 40kHz was chosen.

The influence of the temperature and humidity on speed of sound estimation was investigated and also the possibility of compensating for these factors.

The results show that humidity has very little influence on distance measurement, while temperature has a significant effect. However the temperature variations can be measured and therefore compensated. For a temperature of around 23°C the results obtained using the phase shift method show that the uncertainty in distance measurement can be as low as 0.4mm for a 2m range.

The next subchapter will present the most commonly used ultrasound methods for determining distance. The advantages and disadvantages of each one will be discussed, along with the possibilities for improving the methods in original ways.

4.2.1 The phase shift method

One of the most representative principles for distance measurement is the **phase shift** (or phase detection) method which involves continuous wave transmission[4.4]. A beam of amplitude-modulated acoustic energy is directed towards the target. A small portion of the wave (potentially up to six orders of magnitude less in amplitude) is reflected by the object's surface back to the detector along a direct path. The returned energy is compared to a simultaneously generated reference that has been split off from the original signal, and the relative phase shift between the two is measured to ascertain the round-trip distance the wave has travelled. For square wave modulation at the relatively low frequencies typical for ultrasonic systems (20 to 200kHz), the phase difference between incoming and outgoing waveforms can be measured with a simple linear circuit. The output of the exclusive-or gate goes high whenever its inputs are at opposite logic levels, generating a voltage across capacitor C that is proportional to the phase shift. For example, when the two signals are in phase (i.e., $\Phi = 0$), the gate output stays low and V is zero; maximum output voltage occurs when phase Φ reaches 180°.

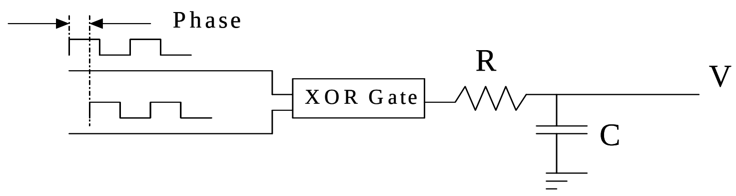

Figure 4-7: *Phase detection device used for ultrasound transmission*

While easy to implement, this simplistic approach is limited to low frequencies, and may require frequent calibration to compensate for drifts and offsets due to component aging or changes in ambient conditions.

For a transmitter-receiver setup, where the reflector is replaced with a receiver, the phase shift principle is applied in a similar way.

Figure 4-8 describes the principle of phase detection measurement.

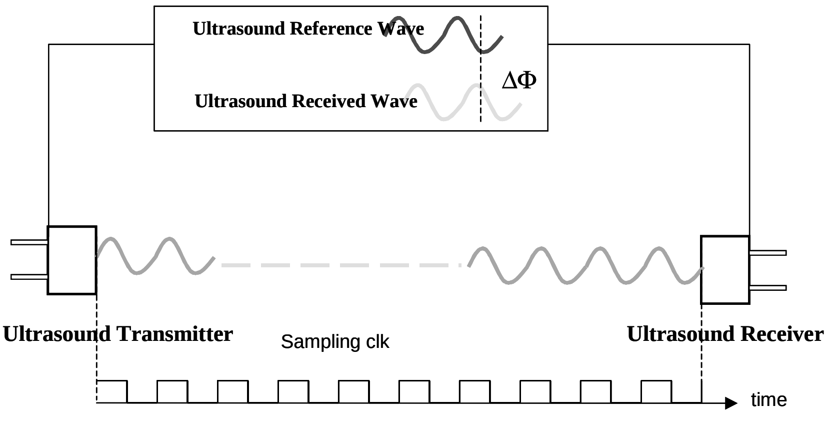

Figure 4-8: *Phase shift setup for measuring the distance between a pair of ultrasound transmitter-receivers*

Figure 4-9 describes the blocks used for the phase detection method. The transmitted sequence could be described by the following equation:

$$s(t) = A\sin(2\pi ft) \tag{4-6}$$

The received sequence will arrive after a time delay which could be written as follows:

$$t_{delay} = t_{\sigma} + \frac{\Phi}{2\pi f} \tag{4-7}$$

where t_σ = n* sampling clock periods, Φ is the phase shift

The received signal has the next form:

$$r(t) = B\sin\{2\pi f(t - t_\sigma) + \Phi\} \tag{4-8}$$

The s(t) is shifted by 90^o and the signal becomes:

$$s'(t) = A\cos(2\pi ft) \tag{4-9}$$

The multiplexer block produces:

$$s'(t)\times r(t) = \frac{AB}{2}\sin(-2\pi ft_\sigma + \Phi) + \frac{AB}{2}\sin(4\pi ft - 2\pi ft_\sigma + \Phi) \tag{4-10}$$

For the next step the signal is filtered. The DC value obtained is directly proportional to the phase shift Φ and the t_σ .

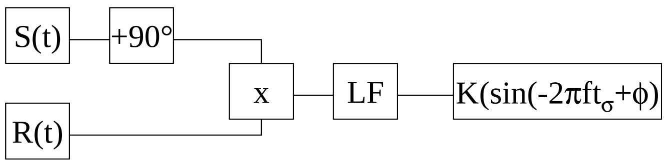

Figure 4-9: *Schematic block diagram of phase shift method.*

The method should find a way to count the t_σ, otherwise it is difficult to make an estimation of the distance. A possible way to solve this is to use recognition time markers on the transmitted signal and received signal. The time markers could be phase switches or frequency switches introduced in the transmitted signal at certain moments in time. Then the number of periods of the sampling clock are counted and the time delay determined.

The transmitter-receiver distance is obtained using the following formula:

$$d = t_{delay} \times V \tag{4-11}$$

where d is the transmitter-receiver distance and V represents the estimation of the sound velocity.

4.2.2 The time of flight method

This principle is based on the measurement of time of flight (TOF) which represents the time necessary for an ultrasonic signal to propagate on air from the transmitter to the receiver either in a direct manner or after being reflected from a target (Figure 4-10).

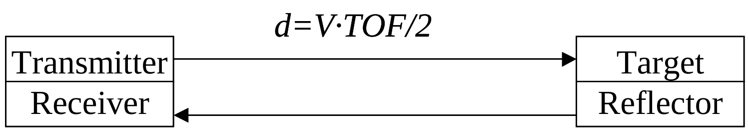

Figure 4-10: *The time of flight principle used for a reflected wave*

With reflection the transducers-target distance is

$$d = \frac{V \cdot TOF}{2} \tag{4-12}$$

where V is the speed of sound.
When the receiver is on the target (Figure 4-11) the distance d is expressed by the formula:

$$d = V \cdot TOF \tag{4-13}$$

Figure 4-11: *The time of flight principle used for a direct received wave*

Different types of ultrasonic techniques can be used for TOF calculation. The most common is the pulse-echo technique. More complex methods involving the modulation of pulse waves have been developed. [4.13] [4.14] [4.15].

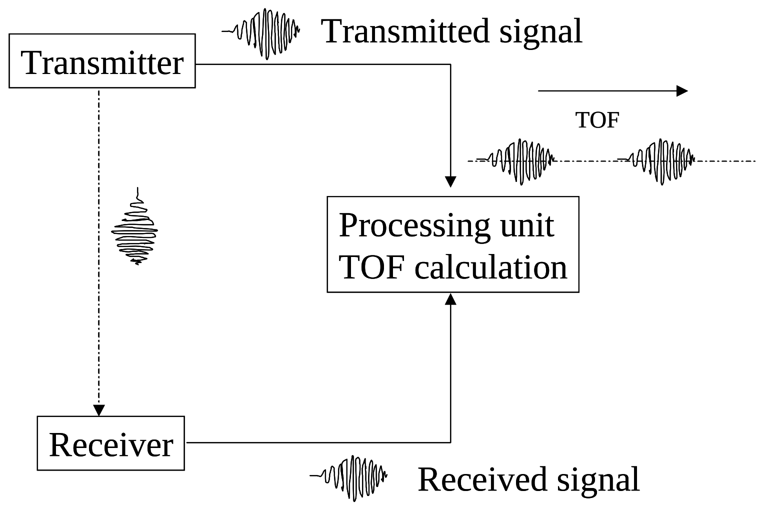

Figure 4-12: *The TOF setup or measuring distance between a pair of ultrasound transmitter-receiver*

Pulse techniques are often used in commercial systems for industrial applications and also for medical applications (as described in chapter III).

Figure 4-12 shows the general setup used to determine the TOF for a pulse wave between a pair of ultrasound transmitter-receiver With this method, a short train of waves is generated and subsequently detected by the receiver. The simplest and most common way to detect the ultrasound signal is the threshold technique, where detection occurs when the signal crosses a given amplitude level. The method is simple and inexpensive, but suffers from a low resolution, especially when the pulse has been attenuated. An improvement involves the adoption of an adjustable amplitude threshold, and in fact detection no longer depends on the magnitude but only on the signal[4.16].

In the presence of noise due to air turbulence or mechanical vibrations in the close proximity of the ultrasound transducer both methods are known to achieve a repetition of some tenths of the ultrasonic wavelength. The effects correspond to error distance measurement proportional to the transducer operating frequency and possible variation in sound velocity.

The following equations describe the model of the ultrasound signal transmitted and received:

$$x_T(nT) = s(nT) + v(nT) \tag{4-14}$$

$$x_R(nT) = \alpha s(nT - TOF) + n(nT)$$

where T is the sampling interval, while *v(nT)* and *n(nT)* take into account the discrepancies from the ideal model and can be considered as zero mean not correlated random processes.

s(t) is the signal generated by the ultrasonic transducer. It forms a short train of acoustic waves. The mathematical model for the *s(t)* signal is:

$$s(t) = a(t)\sin(2\pi f_0 + \Phi_0) \tag{4-15}$$

where f_0 is the resonant transducer frequency and the pulse *a(t)* represents the signal envelope and has finite duration, α is the factor of signal attenuation.

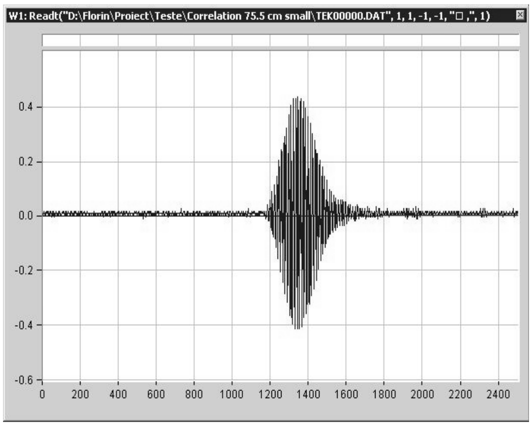

Figure 4-13: *A burst generated by an ultrasound transducer of 40kHz frequency*

Figure 4-13 represents the signal generated by an ultrasound transducer with a resonant frequency of 40 kHz.

The next section of this chapter compares the disturbing factors and properties of the TOF and phase shift methods.

4.2.3 Disturbance factors and advantages specific to TOF and phase shift estimation methods.

This section describes the specific advantages and disadvantages for the methods discussed above.

Both methods have disturbing factors resulting from variations in temperature and humidity.

While humidity has only very little influence on the distance measurement, the temperature has a significant effect. The temperature effect can be compensated with a simple equation that correlates variations in temperature in an operating theatre with variations in sound velocity.

Another important factor is the multipath transmission. Reflected and direct waves are composite and the detection of the signal becomes difficult.

The multipath has a strong effect in the **phase shift** measurement. The continuous direct wave interferes with reflected waves. Hence the detection of phase shift becomes very difficult.

For the **TOF principle** the multipath effect is minimal. The reflected bursts will always arrive later then the direct burst. The first burst gives the direct transmitter-receiver path.

Another important aspect is the complexity of hardware required to implement the principles. For **phase shift,** the technique is more expensive due to the often complex hardware needed to measure the phase and due to the difficulty in determining the number of integer wavelengths.

For **TOF** the simplest and most economical technique offers poor resolution and only produces accurate estimations with the aid of complex and expensive hardware. Solutions for improving TOF measurement with a clear increase in accuracy and considerable reduction in cost lie in the form of digital techniques [4.10].

To summarise the conclusions presented in this section, the phase shift method is not an optimal technique to be implemented in the discussed system. The TOF method, however, has less disadvantages compared with phase shift solution. Depending on the accuracy required, a specific digital technique will be chosen.

4.3 Distance calculation using TOF algorithm

4.3.1 TOF detection using the threshold method

In section 4.2.2 (Figure 4-12) a simple TOF detection technique was presented.

An ultrasound pulse wave signal described by Eq. (4-15) is sent and subsequently received by the ultrasound receiver. Depending on the method used the accuracy of the detection may be in the range of millimetres to centimetres.

The simplest method is the threshold technique. The ultrasound burst is detected when a specific voltage level is exceeded. However the presence of noise in the received signal and distance dependent attenuations induce uncertainty in the peak detection, within the burst. A wrong peak detection generates errors of multiple periods of

25µs in time (at a signal frequency of 40 kHz), which means multiples of 9mm in distance.

Measurement set-up

Our measurements were made using the commercial 40kHz piezoelectric resonant transducers presented, which generate ultrasonic pulses. The frequency spectrum of the received signal is presented in Figure 4-14.

For this set of measurements the distance between the receiver and the transmitter was about 60 cm. The position of the receiver was varied 5 times in the range 0-5cm above its null position (~60 cm). At every 0.5cm, the time of flight was measured and the results are plotted in Figure 4-15.

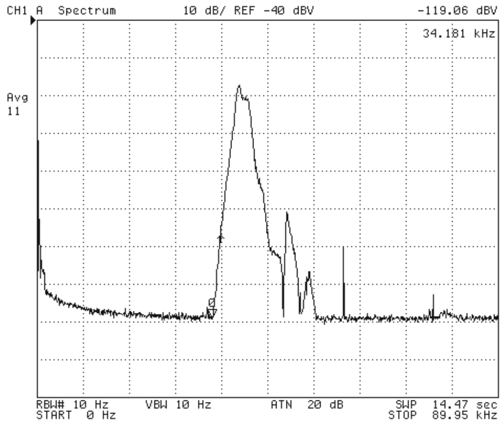

Figure 4-14: *Frequency spectrum for the received signal in a band of 0-90kHz*

Linear regression of the results presented in the graph gives the equation:

$$\Delta d = k \cdot TOF + \mu \tag{4-16}$$

Δd ¸ variation of distance between transmitter and receiver
κ = 0.0304 and μ = 0.002.

The variations of the time of flight have a good correlation with variations in distance. The systematic deviations in the 'tops' of the curve can be explained by hysteresis in the mechanical manipulator used to move the receiver.

The linear correlation between the time of flight and variations in the distance is visible in Figure 4-15.

But, as discussed before the presence of noise in the received signal induces some uncertainty in the detection to peaks which may lead to errors of multiples of 25μs in time, or multiples of 9mm in distance.

Figure 4-16 and Figure 4-17 show the different number of peaks in a burst and their positions detected for a variation of 1m in receiver-transmitter distance. It is clear that changes in distance may generate errors in detecting a certain peak within a burst.

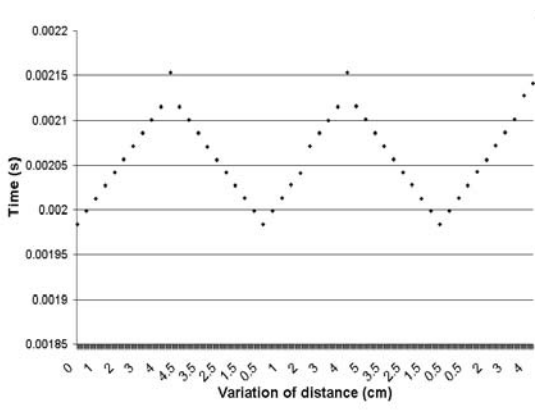

Figure 4-15: *TOF dependence on distance variation in the range 0-5cm*

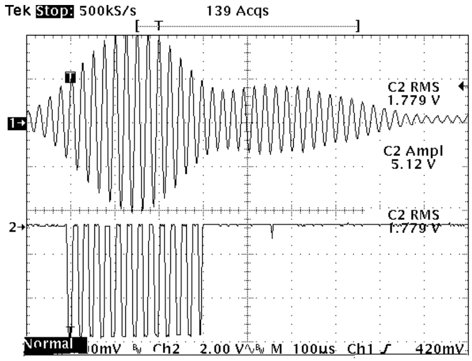

Figure 4-16: *Peaks detection for a transmitter-receiver distance at 60cm*

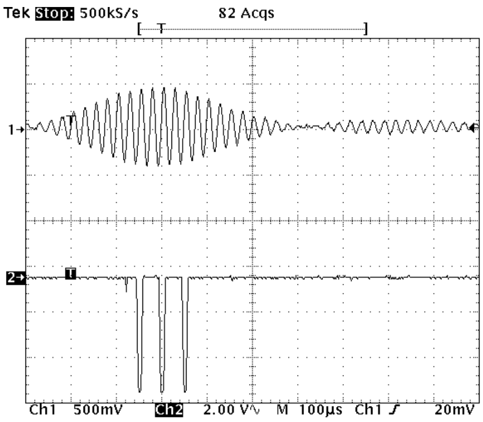

Figure 4-17: *Peaks detection for a transmitter-receiver distance at 160cm*

As a conclusion with the TOF threshold method, the variations of distance and time of flight have a satisfactorily linear correlation. However, the attenuation and the presence of noise in the signal induce errors in distance detection and make this method unsuitable for accurate location measurements.

4.3.2 TOF calculation using the cross-correlation digital technique

An option to improve the accuracy of TOF measurements is to use digital techniques. One of them is the cross-correlation algorithm[4.7].
The approach is described below.

For a well known signal the optimal processing is based on cross correlation, which searches for a facsimile of the known signal in the received signal with noise.

Assuming the signal s(t) and the noise given by n(t), the received signal may be expressed by:

$$r(t) = s(t-T^{'}) + n(t) \tag{4-17}$$

where $T^{'}$ is the delay time (i.e., the time of flight of the signal travelling from the source to the input of the detection system). Cross correlation amounts to generating and studying the function

$$F(T) = \int_{(-\infty)}^{\infty} s(t)r(t+T)dt \tag{4-18}$$

where T is an adjustable time delay.

Inserting the value for r(t),

$$F(T) = \int_{(-\infty)}^{\infty} s(t)s(t+T-T^{'})dt + \int_{(-\infty)}^{\infty} s(t)n(t+T)dt \tag{4-19}$$

The first integral will be maximised when T=T’. The second integral does not have an important contribution for any value of T since the integral is the product of independent oscillatory functions in time. So, the time of flight is found when F(T) peaks for T=T' [4.17]

One method of calculating F(T) is to consider the Fourier Transforms of the signal s(t) and r(t) named S(w) and R(w). Multiply them together, and find the inverse Fourier Transform of the results. The output will be the convolution of the two signals. Similarly finding the inverse transformation of the product of S(w) and $R^*(w)$ (the complex conjugate of R(w)) also yields the cross correlation of the s(t) and r(t)[4.12].

For the TOF measurements the system acquires the two digital sequences $x_T(nT)$ and $x_R(nT)$ representing the transmitted and received signals, respectively, which can then be written in the form

$$x_T(nT) = s(nT) + v(nT) \qquad (4\text{-}20)$$

$$x_R(nT) = \alpha s(nT - TOF) + n(nT)$$

where T is the sampling interval, while v(nT) and n(nT) take into account the discrepancies from the ideal model and can be considered as zero mean not correlated random processes.
s(t) is the signal generated by an ultrasonic transducer and is a short train of acoustic waves than can be represented in the form

$$s(t) = a(t)\sin(2\pi f_0 + \Phi_0) \qquad (4\text{-}21)$$

where f_0 is the resonant transducer frequency and the pulse a(t) represents the signal envelope and has finite duration and α represents the attenuation factor for the acoustic signal.
The cross-correlation of sequences is given by:

$$C(kT) = \sum_{k=-\infty}^{\infty} x_T(nT)x_R(nT + kT) \qquad (4\text{-}22)$$

The maximum values of C(kT) corresponds in time with the C (TOF). Thus the TOF is estimated by searching for the maximum of C(kT).

The innovative aspect of our idea is to replace the model for the transmitted signal with the last received ultrasound sequence digitised by a process unit. If the time slot between two consecutive transmitted ultrasonic signals is less than 100ms then the changes in received signal envelopes will be small, since the distortion at the envelope is expected to be almost constant over this small interval. Each sequence will be acquired and stored, after which it will be processed by a calculation unit (DSP or computer).

Figure 4-18 shows a last sequence stored as a reference trigger time and the next received one. The time delay between sequences represents the TOF for the transmitter-receiver distance. Figure 4-19 shows the cross-correlation function for those sequences.

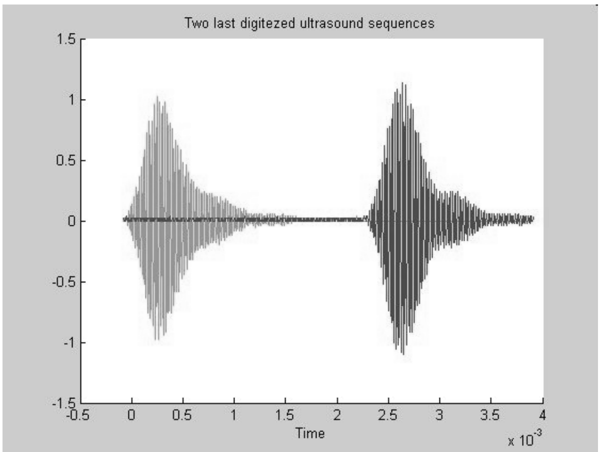

Figure 4-18: *Two consecutive received pulses will have almost identical envelopes when the time slot is small*

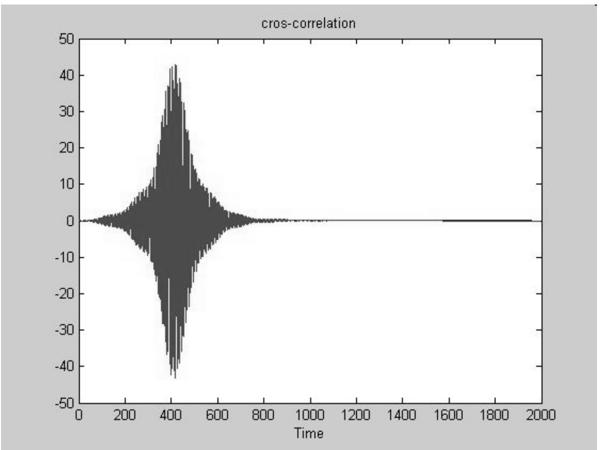

Figure 4-19: *The cross correlation function of two ultrasound signals processed by a DSP unit.*

Measurement setup

The first measurements for the ultrasonic signals were made for a transmitter-receiver distance of 1.69 m.
Two ultrasound transmitters react to two (different) RF trigger codes generated by the DSP. Figure 4-20 shows the ultrasound signals acquired from the two different ultrasound transmitters. The ultrasound transmitters are reacting exactly to their specific code.

The transmitter-receiver distance for one transmitter was changed 5 times in 5mm steps and the received signals were acquired and sampled with a 500 kHz sample rate. Figure 4-21 represents the two consecutive bursts acquired and the cross correlation function. The zero time for the TOF measurement is synchronised with a clock frequency of 100 Hz.

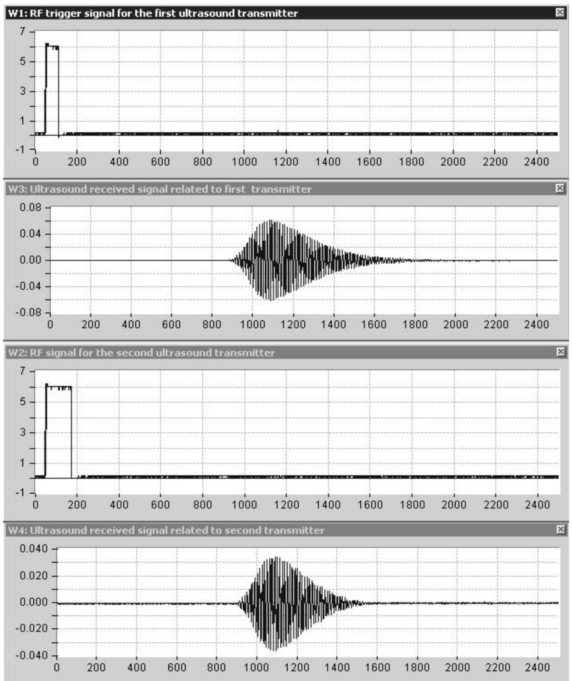

Figure 4-20: *Ultrasound bursts received from different transmitters, triggered with different RF codes*

The peak of the cross correlation function is correlated with 5mm deviation of the transmitter position.

The maximum of the cross correlation function does not change for a relative rotation of transmitter and receiver of ± 15^o. This shows that the distance between transmitter and receiver is accurately measured even for a large change in orientation.

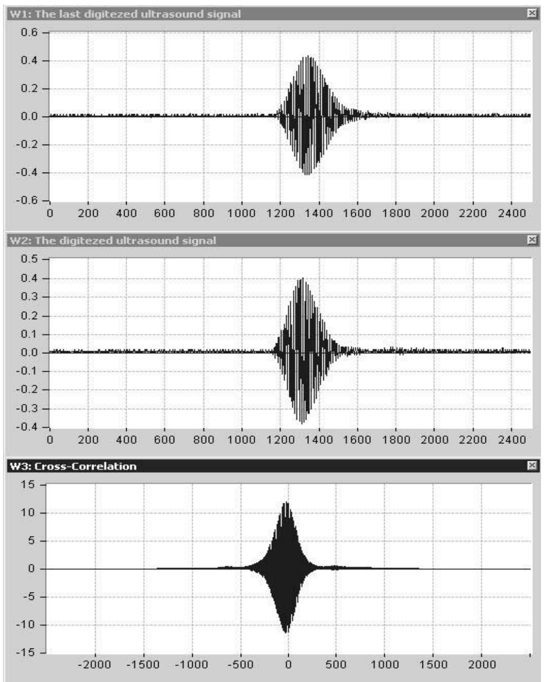

Figure 4-21: *The cross-correlation function of two signals received after a change of 5mm in transmitter position*

Analysing a set of consecutive cross-correlation sequences related to a set of 5mm standard deviations of the transmitter position we obtained a resolution of 0.7mm with a sampling rate of 500kHz. Increasing the sampling frequency will increase the resolution. The maximum absolute error of measurements was about 2.8mm for a distance of 1.69m.

The error of detection can be reduced in two ways: using DSP filter algorithms and reducing to 100ms time slot between two consecutive stored sequences

4.3.3 TOF calculation using the phase switch detection for an ultrasound burst

To reduce the error of the detection for the method presented above, we proposed a new approach: the phase switch detection method. The same frequency (40kHz) was used for the total burst, but half way through the burst a phase shift of 90° was introduced in the signal.

The phase shift transition moment was used as time reference for the detection of the ultrasound burst.

To improve the performance of phase switch detection the ultrasound received burst was cross-correlated with a reference signal of 40 kHz chapter [4.5].

The maximum peak of the cross correlation function marks the moment when the phase is switched.

$$s(t) = A_s \sin(2\pi f_0 t) \quad (-t_p < t < 0) \tag{4-23}$$

$$s(t) = A_s \sin(2\pi f_0 t + \varphi) \quad (0 < t < t_p) \tag{4-24}$$

Experimental set-up and measurements

The first measurements for the received ultrasonic signals were made for a transmitter-receiver distance of 1m.

A Lock-in Amplifier was used to lock the ultrasound received signal and the phase measurement was performed.

Figure 4-22a shows the 90-degree phase transition of the signal acquired from the ultrasound receiver. Figure 4-22b shows the stability of phase shift detected during 28 subsequent measurements in the first half of the ultrasound burst.

The measurements clearly indicate the transition of 90° in the received signal and a good stability of phase detection (around 1 degree). However, for this technique, the sampling frequency of the lock-in amplifier is not high enough to perform an accurate detection of phase transition moment versus time.

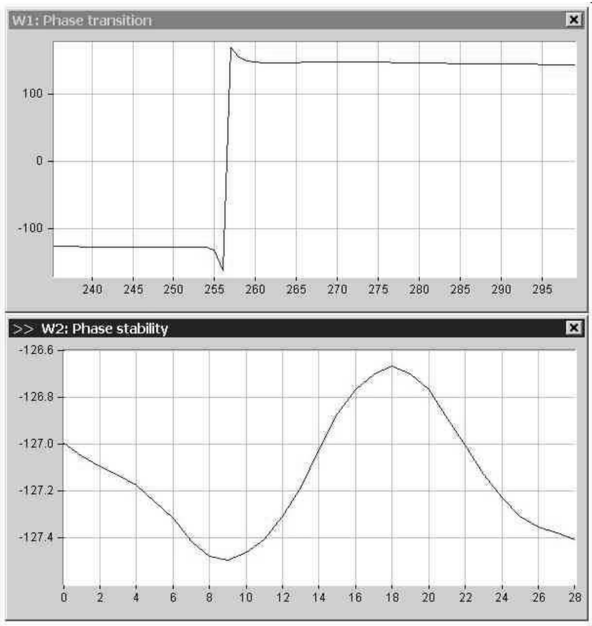

Figure 4-22: *a) The phase measurements close to 90-degree transition moment in an ultrasound burst. b) The set of 28 phase measurements*

Therefore the cross-correlation approach presented in the subsection 4.3.2 should be used.

For the second measurement setup an phase switched burst is acquired with a sampling frequency of 300Ksamples/sec. The sequence is cross-correlated with a reference signal of the same frequency.

The maximum peak of the cross-correlation function represents the phase transition moment versus time.

The method allows 3.3µs resolution in determining the time of flight measurement (TOF). The resolution is limited by the sampling frequency used (300 k sample/sec). The resolution of 3.3µs for TOF detection permits a calculation of transmitter position with a resolution of about 1mm. To achieve better accuracy the solution is to increase the sampling frequency.

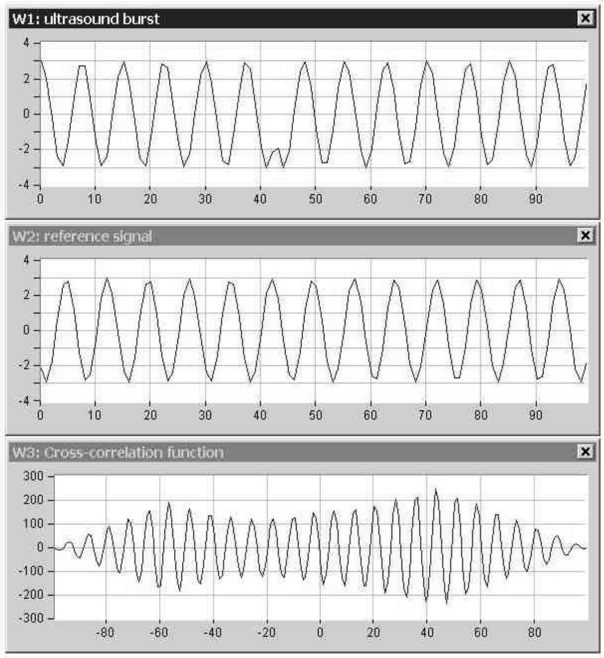

Figure 4-23: *The cross-correlation function of the reference signal and the ultrasound acquired signal*

4.4 Concluding remarks

The transponder used as part of the detection system generates and detects the waves. Some research work was necessary to find the proper sensor for this specific application. Depending on their mechanical-physical characteristics different types of transponders will react differently. In subsection 4.2 the considerations for choosing the optimal ultrasound transponder were presented. Different factors, effects and phenomena were considered during the process of sensor identification. Considering the ultrasound wave produced by an ultrasound transponder, the absorption air factor, the pattern beam, related to the frequency of ultrasound transmitter and climatic factors affecting the speed of sound, the choice was made for an ultrasound transmitter/receiver of 40kHz. The climatic factors were investigated and the stability of the sensor related to these factors was characterised.

The results show that on the one hand humidity has very little influence and on the other hand temperature has a dominant effect on speed of sound. The equation Eq. (4-4) was derived to compensate for the temperature effect. The study on the stability of the chosen transponder Eq. (4-4) estimates the values of the sound velocity which corresponds with an error of measured distance less then 0.4mm for 2m.

In subsections 4.2.1 and 4.2.2 the basic principles of measuring distance were presented. The Phase Shift method and the TOF method were described and the most common advantages and disadvantages specific to each one were presented. The adverse influence of multipath reflections and the complexity of the hardware necessary to build a system based on the phase shift method, led to the conclusion that the time-of-flight principle was to be investigated further.

The subsections 4.3.1-3 present the different techniques based on the time of flight principle used to measure the TOF for an ultrasound burst transmitted and received by the ultrasound transponder for different distances. The measurements for the simplest one, which calculates the TOF using the threshold of voltage detection technique, shows large errors as a result of the presence of noise in the spectrum of the received signal. The other two techniques are based on digital processing and the results achieved prognoses good detection of the time of flight of the ultrasound burst and so far for the transmitter-receiver distance. The best accuracy of determining the transmitter-receiver distance was obtained using the cross-correlation digital technique presented in subsection 4.3.2. With a sampling rate of 500kHz, we obtained a resolution of 0.7mm in determining the transmitter position. Increasing the sampling frequency will increase the resolution. The maximum absolute error of measurements was about 2.8mm for a distance of 1.69m without temperature compensation. Therefore, for the final setup using this technique we expect to obtain measurement results without significant errors in position and orientation estimation of laparoscopic tools.

In the next chapter, issues related to the final ultrasound sensor system will be presented.

4.5 References

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5 System for detection of laparoscopic instruments

5.1 Introduction

The developmental phase of designing the localisation system based on the ultrasound methods described in the previous chapter will be further presented. The electronic blocks used to trigger the ultrasound transmission and the hardware and software methods implemented to process the information derived from the acoustic sensors are described and further integrated in the complete set-up of the ultrasound wireless positioning system.

The measurements performed for the evaluation of the transmitter-receiver distance, the accuracy and the resolution achieved in testing the cross-correlation method will be discussed in the last part of the chapter. Furthermore the influence of temperature and air flow were investigated and the limitations derived from those factors will be presented.

Respecting the requirements presented in chapter II, the system gives the surgeon the location and orientation of the instruments inside the patient. The measuring system employs ultrasound markers placed on the instruments, outside of the human body. Using the cross-correlation method presented in subsection 4.3.2 and implemented for the final system set-up, the transmitter-receiver distance was measured with resolutions of 0.7mm and 1.2mm. The resolution values are dependent on the acquisition frequency (500k samples/sec and 300k samples/sec) used to sample the ultrasound signal. To solve the wireless requirement of the system, a radio wave (RF) unit was designed to perform the communication with the ultrasound markers.

5.2 The RF trigger block and the ultrasound generation modules

As presented in subsection 4.3.2 the ultrasound transmitters are triggered by the RF communication unit used as a time reference for the TOF calculation. The RF signal triggers the ultrasound transmitters. The assumption of the instantaneous transmission was based considering the high difference value between the speed of light (3×10^{8}m/s) and the sound velocity (344m/s). The RF unit generates 16 RF codes. Each ultrasound transmitter is hardware configured to react at a specific RF code. If the code does not correspond, the trigger operation for emitting the ultrasound burst is delayed until the right code arrives. Therefore, the conflict of the possible ultrasound multitransmission or confusion between the positions of different sensors is resolved.

The schematic view of the RF unit blocks, the recognition code and the ultrasonic generation modules are presented in Figure 5-1, Figure 5-2.

All those blocks induce an constant delay time of 186µs, time which will be considered in the TOF determination.

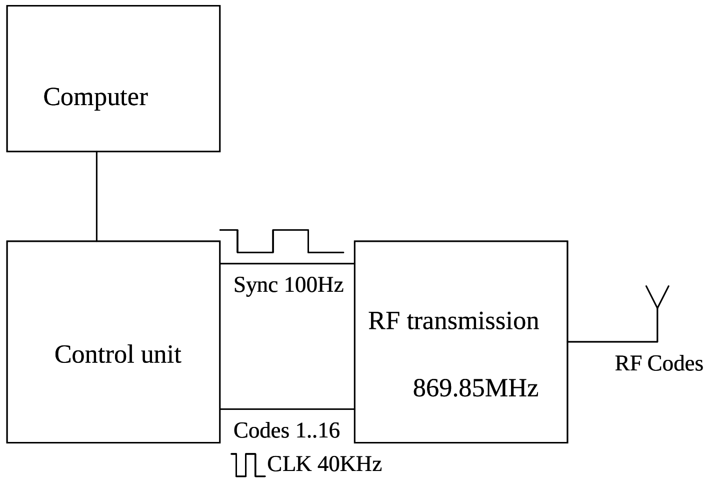

Figure 5-1: *RF transmission unit*

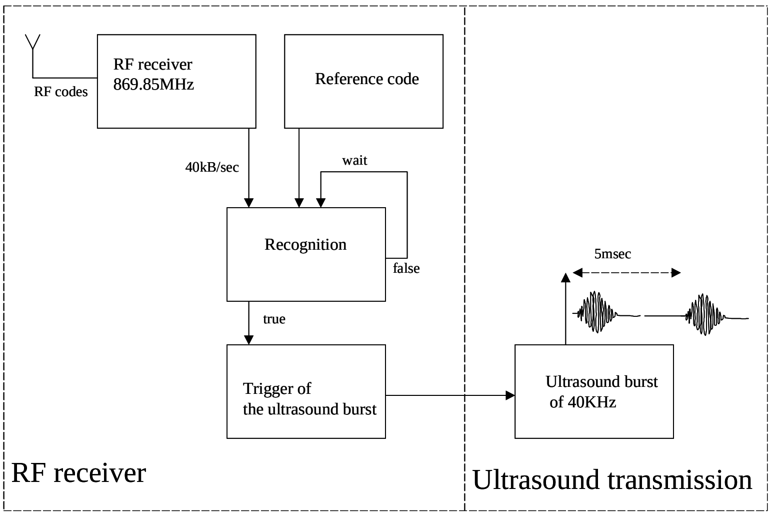

Figure 5-2: *RF receiver unit, code recognition block and ultrasound generation unit*

The RF transmitter and receiver modules are presented in Figure 5-3.

Figure 5-3: *The RF modules*

The ultrasound burst is received by the array of receivers placed above the surgical tools. Receiver set-up was described in section 3.4.1. To estimate the TOF, different hardware and software techniques are applied to the received ultrasound signal.

5.3 The hardware-software block used to acquire the ultrasound signal and to calculate the transmitter-receiver distance.

5.3.1 The electronic filter to improve the signal/noise ratio

Figure 4-14 in subchapter 4.3.1 indicates in fact that a filter operation needs to be applied to the 40kHz received signal in order to reduce the noise observed in the signal spectrum. Therefore the quality of the ultrasound received signal was increased using a high-pass band filter.

The description of the characteristics of the implemented hardware filter is presented below. Figure 5-4 shows the design of the filter.

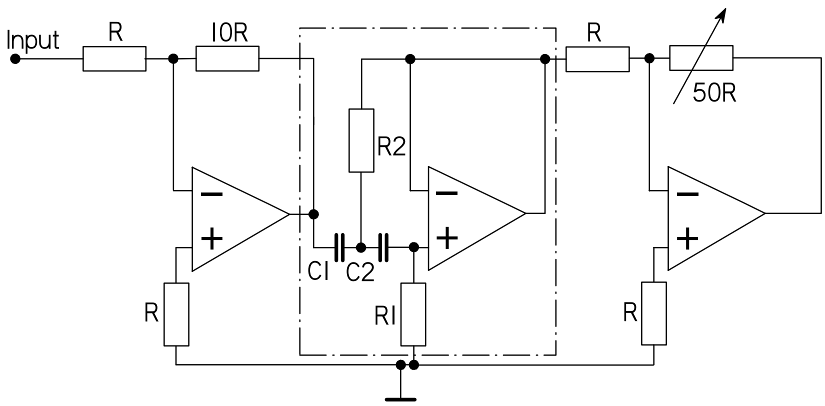

Figure 5-4: *The filter used for filtering the signal.*

5.3.2 Ultrasound signal acquisition.

The first measurements were performed with a Texas Instruments 6711 DSP Board with an acquisition rate of 1.2M samples/sec. The board offers the possibility to extend the sampling rate to 8M samples/sec. During the programming process for the DSP unit many difficulties were encountered regarding the configuration of the board for real-time acquisition of the signal.

Therefore a data acquisition card (National Instruments DAQ) was preferred for the final set-up.

The card is a multifunction analogue, digital and timing I/O device, with 8 differential analogue-input channels, 2 analogue-output channels, 8 digital input-output channels and 2 up/ down counter/timers. The analogue input and analogue output resolution is 16 bits, while the maximum sampling rate and the maximum update rate is 3.33×10^5 samples/s. To read data from the analogue filters and to send the trigger pulse to the RF unit, one analogue-output and one analogue-input channel were configured.

The trigger pulses used as input for the RF unit were generated by the DAQ analogue output with a sampling frequency of 10,000k sample/sec and a periodicity of 10mHz.

For the acquisition procedure the continuous read-out of data with the specific rate of 3.33×10^5 samples/s was implemented.

From each receiver a sequence of 2500 samples was acquired after the triggered pulse was generated.

The Block diagram of the acquisition system is illustrated in Figure 5-5.

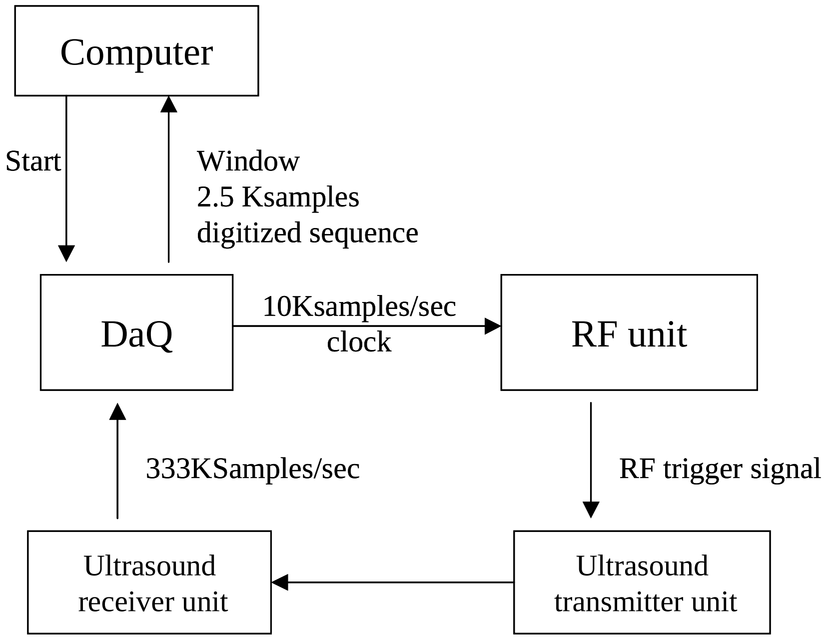

Figure 5-5: *Schematic view of the acquisition unit*

5.3.3 Data processing to determine transmitter-receiver distance

To determine the transmitter-receiver distance the TOF method presented in subchapter 4.3.2 was used.

The read-out digitised data from the ultrasound transponder were used as input for an software program, based on the cross correlation method to determine the TOF values. The program was built using the National Instruments LabVIEW 6.1 product.

The hierarchical representation of the program is illustrated in Figure 5-6.

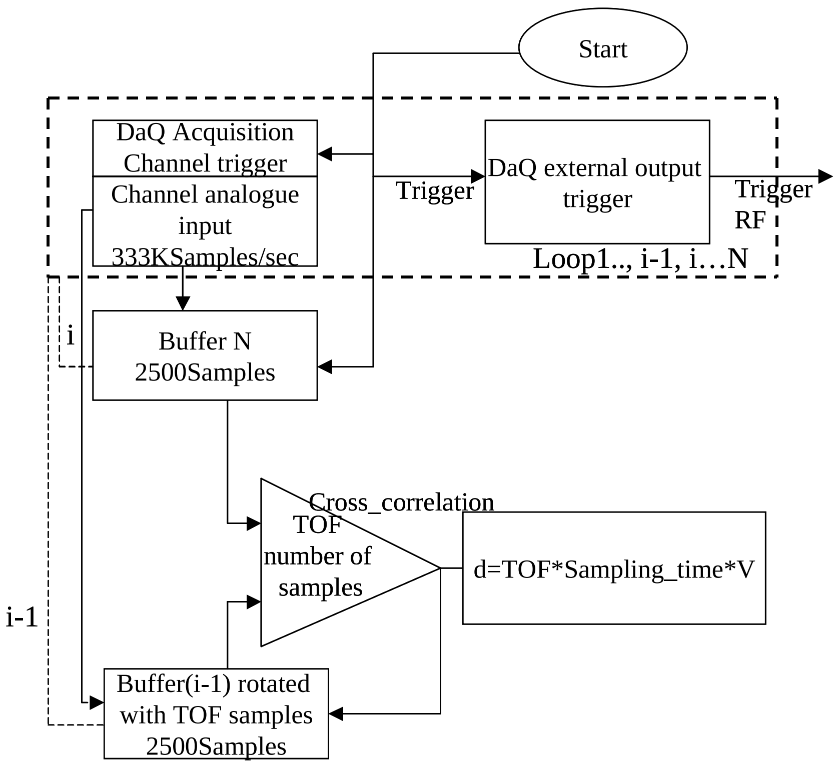

Figure 5-6: *Hierarchical representation of the software used for the TOF estimation*

5.4 Final setup and measurements

As presented in subsection 5.3.2 the ultrasound transmitters are triggered by radio wave signals. For this purpose the marker modules include simple RF receivers that react on the correct identification code only. These RF trigger codes are sent by a central RF transceiver unit and provided by a DAQ unit, which controls the whole process and derives position information from the ultrasound signals. The schematic of the whole system is described in Figure 5-7.

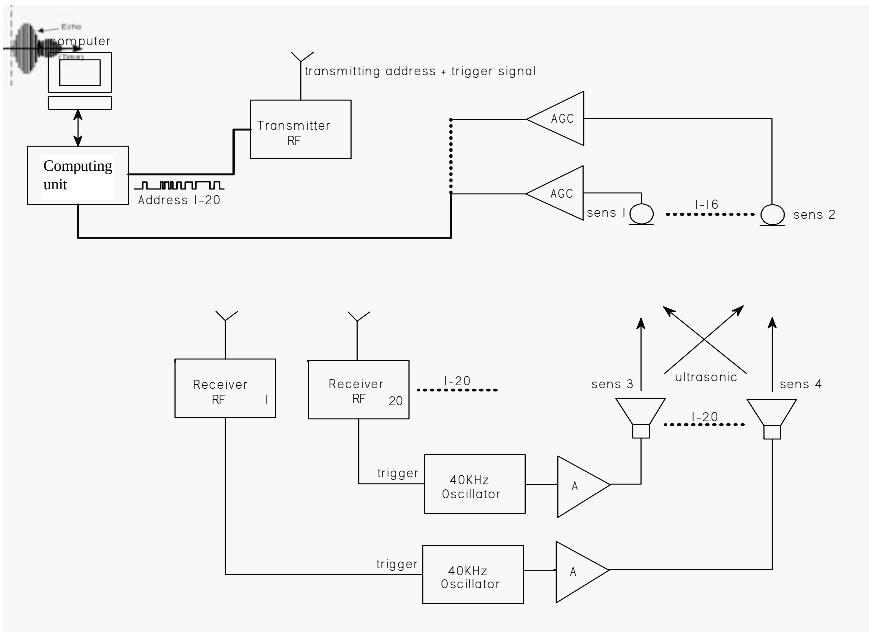

Figure 5-7: *Schematic view of the localisation system*

The measurements done using this system and the conclusions reached after the interpretation of data are presented in the next subchapters.

5.5 Results

Figure 5-8a. shows the time of flight dependence on distance measured for a constant transmitter-receiver position. The variation of measured distance is explained by the temperature variation. During the first 110 measurements the temperature was constant and after that, in the next 90 samples the time of flight was affected by temperature variation. That measurement concludes that the time of flight will be estimated accurately if the temperature compensation technique is used.

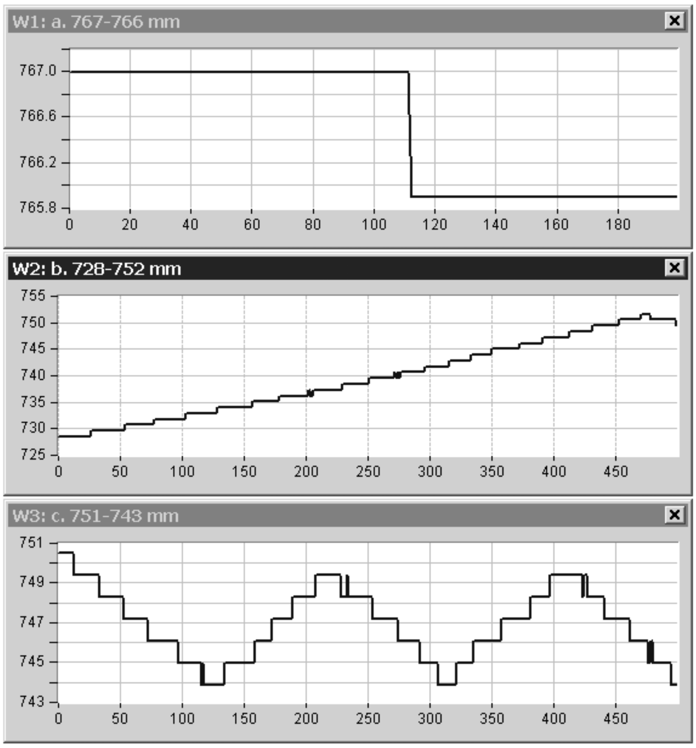

Figure 5-8: *The time of flight dependence on distance variations in the ranges a) 767-765.8mm, b) 728-752mm, c) 751-743mm*

Figure 5-8 b)c) and Figure 5-9 a)b) show the time of flight dependence on distance variations. The distances measured are in the ranges: Figure 5-8 b) 728-752mm, Figure 5-8 c) 751-743mm, Figure 5-9 a) 773-764mm, Figure 5-9 b) 773-782mm. The resolution achieved is approximated at 1.2mm. This relates directly to the uncertainty introduced by the limited sampling rate used for the acquisition of the signal.

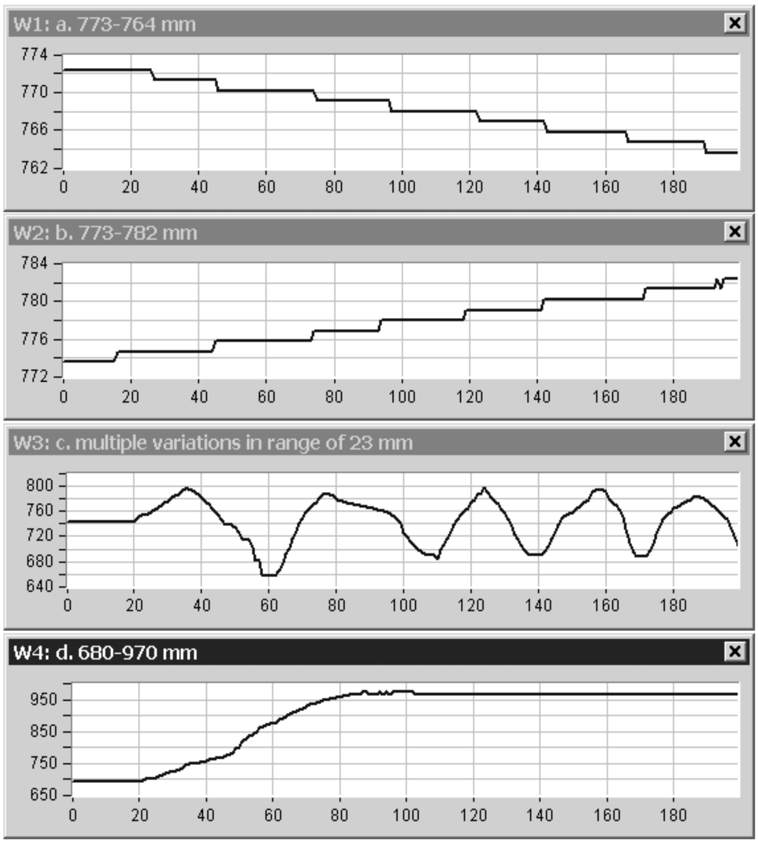

Figure 5-9: *The time of flight dependence on distance variations in the ranges a)773-764mm, b)773-782mm, c) multiple variations 650-800mm, d) 700-970mm*

Figure 5-9 c) and Figure 5-9 d) show the time of flight dependence of distance for the dynamic movement of sensor in ranges of Figure 5-9 c) 650-800mm, Figure 5-9 d) 700-1000mm. For a constant temperature the variation of distance is a continuous function depending on the TOF values. Therefore for a large variation of temperature over one day the TOF measurements for a fixed position transmitter-receiver were recorded. Figure 5-10 shows the temperature measurements, the TOF measurements and the TOF estimations compensated for the variations in temperature.

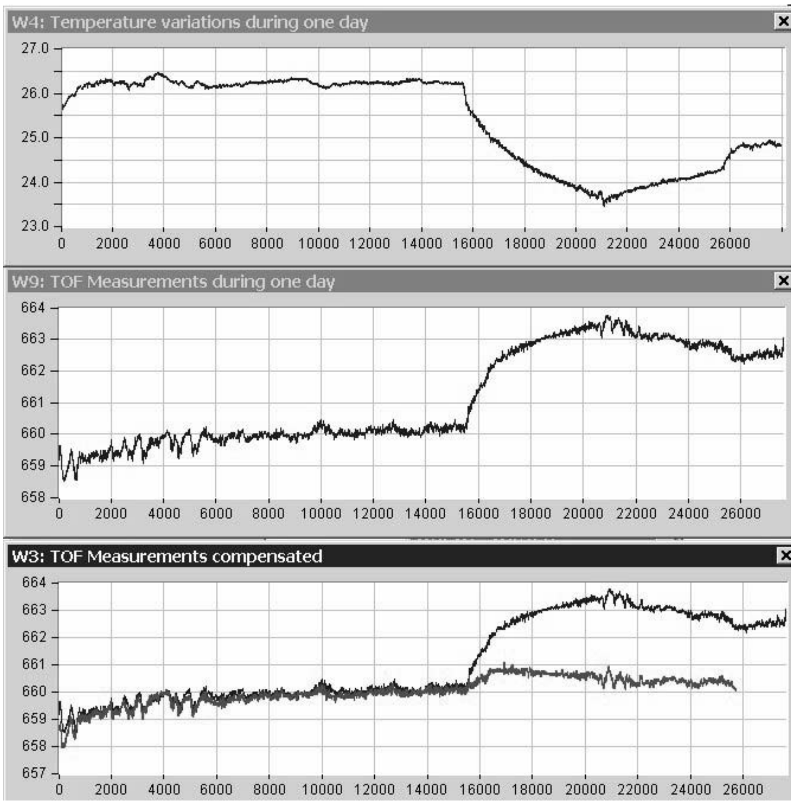

Figure 5-10: *a)Temperature variations over one day (x axis-hours), b) TOF measurements for a fixed distance transmitter receiver during one day, c) TOF measurement compensated for temperature variation influence*

For the next phase of measurements air flow disturbances were induced. For airflow around 1m/s the measurements registered errors about of 9-10mm (Figure 5-11). The consequences of airflow disturbances are reflected in two major phenomena which influence the cross correlation method and TOF estimation. The first is wrong estimation of the speed of burst due to the unpredictable air flow speed which is combined with the speed of sound. The result is the unknown variation of the burst speed and also incorrect calculation of the transmitter receiver distance. The second consequence is reflected in induced changes on the envelope of the ultrasound signal. The assumption that the two consecutive signals are almost identical is not longer respected and the cross correlation method causes errors.

Therefore the measurement shows a satisfactory resolution and might give good accuracy if temperature variations are compensated and a technique to reduce the air flow influence is found. Also, increasing the sampling frequency will improve the resolution. The maximum absolute error of measurements without air flow disturbances induced were about 2.8mm for a distance of 1.69m. In the presence of the air flow the error of distance measurement increased up to 9-10 mm. A simple solution to reduce the error is to use the DSP filter algorithms.

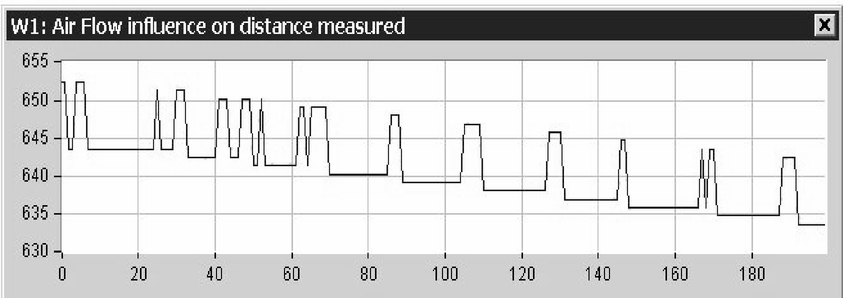

Figure 5-11: *The errors of distance measurement for a slow movement due to air flow disturbance (approx. 1m/s)*

To simulate the practical situation, we fabricated a patient model and used it for one dimensional measurements. The one-dimensional measurement tests were performed with one ultrasound sensor attached to a laparoscopic instrument. The endoscope was replaced with a video camera which was used as a positioning references for the sensor. The information regarding the sensor position superimposed on the monitor proved the accuracy of the sensor detection. Future research should continue with the optimisation of the total system, the miniaturisation of the sensors and tests regarding the packaging of the sensors in order to respect the sterilization procedure.

5.6 Conclusions

In this chapter a new system for position detection of laparoscopic instruments using ultrasound is discussed. The RF solution for triggering the ultrasound transmission is described. Hardware and software methods used for acquisition and filtering of the ultrasound signal are presented. For the estimation of the transmitter receiver distance the digital cross correlation technique is applied. This technique allows a resolution of 0.7mm (subsection 4.3.2) or 1.2mm (subsection 5.3) for a sampling frequency of 500k samples/sec respectively 300k samples/sec for determining the instrument position[5.1][5.2]. The influence of temperature and disturbances due to air flow presence were also estimated. Using the temperature compensation technique presented in the section 5.5 the accuracy of the system is about 2.8mm. The presence of the air flow (around 1m/s) induced errors of 9-10 mm. Therefore it is concluded that the resolution and the accuracy obtained after temperature compensation are sufficient for laparoscopic application. The presence of the air flow constitutes an important factor which demands for more research into techniques for reducing the errors due to air turbulence.
One of the methods tested “The 180° phase shift introduced after half way through the burst" was highly successful at reducing the air flow influence[5.3][5.4]. The method and the results will be described in the next chapter.

Until now we have presented a system that works with satisfactory accuracy and fulfil the requirements. The next chapters will present different methods of improving the system and its performance and eventually to extend its use to the medical field.

5.7 References

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6 Improving detection using the phase shift method

6.1 Introduction

Chapter V presented the ultrasound sensor system and the conclusions derived from the distance measurement. The system currently under development achieves 1mm resolution of distance measurements. Accuracy is limited by variations in temperature and air flow fluctuations. Although the temperature effect is easy to compensate for, air flow requires a more complex technique to reduce the uncertainty on the burst speed calculation due to air speed influence.

To overcome the disadvantages introduced by airflow turbulence or accidental multipath interference, the phase switch method is proposed as a solution. The same frequency (40kHz) is used for the total burst, but half way through the burst a phase shift of 180° is introduced into the signal. The effect will lead to a well defined envelope of the received signal. As a consequence the algorithm of the cross correlation function described in chapter V will lead to a better estimation of the time of flight evaluation. The consecutive signals emitted by the same sensor, it will conserve a standard shape for the its envelopes and will be less distorted by air flow influences or multipath interferences.

Mathematical model of the received signal

To study the model of the signal with the phase shift introduced at the half way point, we considered that the ultrasonic waves produced by the piezoelectric transducer's can be modelled by the empirical equation [6.5][6.6][6.7].

$$f(t) = At^n e^{-\alpha t}\sin(\omega t + \phi) \tag{6-1}$$

In this mathematical model n, α, ϕ, and ω are the transducers dependent parameters, ω being the undamped angular resonance frequency of the piezoelectric transducers. For the transducers used $n = 2$, and α is estimated to $10^4 s^{-1}$, and $\omega/2\pi = 40kHz$.

Considering the burst as a summation of the two waves

$$f_1(t) = At^n e^{-\alpha t}\sin(\omega t + \phi)u(t) \tag{6-2}$$

$$f_2(t) = A(t-t_r)^n e^{-\alpha(t-t_r)}\sin(\omega t - \omega t_r + \phi + \pi)u(t-t_r) \tag{6-3}$$

where t_r is the time when the phase shift occurs, and u(t) the unit step function.

The output of the transducers is:

$$f_1 + f_2 = Ae^{-\alpha t}[t^n\sin(\omega t+\phi)u(t) - (t-t_r)^n e^{\alpha t_r}\sin(\omega t - \omega t_r + \phi)u(t-t_r)] \tag{6-4}$$

After t_r the reflection of signal f_1 in the transducer will coexist with signal f_2 and the result is an interfered ultrasonic wave

Also considering that $t_r = 2k\pi/\omega$,

the equation of the transmitted wave becomes:

$$int(t) = At^n e^{-\alpha t}\left[1 - \left(1 - \frac{t_r}{t}\right)^n e^{\alpha t_r}\right]\sin(\omega t + \phi) \tag{6-5}$$

$(t \geq t_r)$

the envelope is zero for $t = t_0$, satisfying:

$$t_0 = \frac{t_r}{1 - e^{-\left(\frac{\alpha}{n}\right)t_r}} \tag{6-6}$$

for a switch of t_r is equal to 10 cycles of 40kHz frequency (0.25ms) the t_0 is a constant values.

Figure 6-1 shows the graphic representation of the simulations of waves f_1, f_2 and the interfered signal f_1+f_2. The zero level on the envelope signal has a good determination in time and divides the envelope into two lobes. The presumption is that the first lobe will be little influenced by the variation in air flow or the multipath reflection effect in comparison with the second one.

Also for the 90° phase shift a simulation was carried out, but the results are significantly different to those for the 180° simulation. Instead to achieve a zero crossing in the envelope of the ultrasound signal a small valley is obtained. Therefore the signal does not show much of a difference compared to the regular signal used for the measurement described in chapter V and the 90° solution does not give enough reasons to be implemented.

The next subchapter will present the electronics used for the hardware implementation to produce the 180° phase shift method.

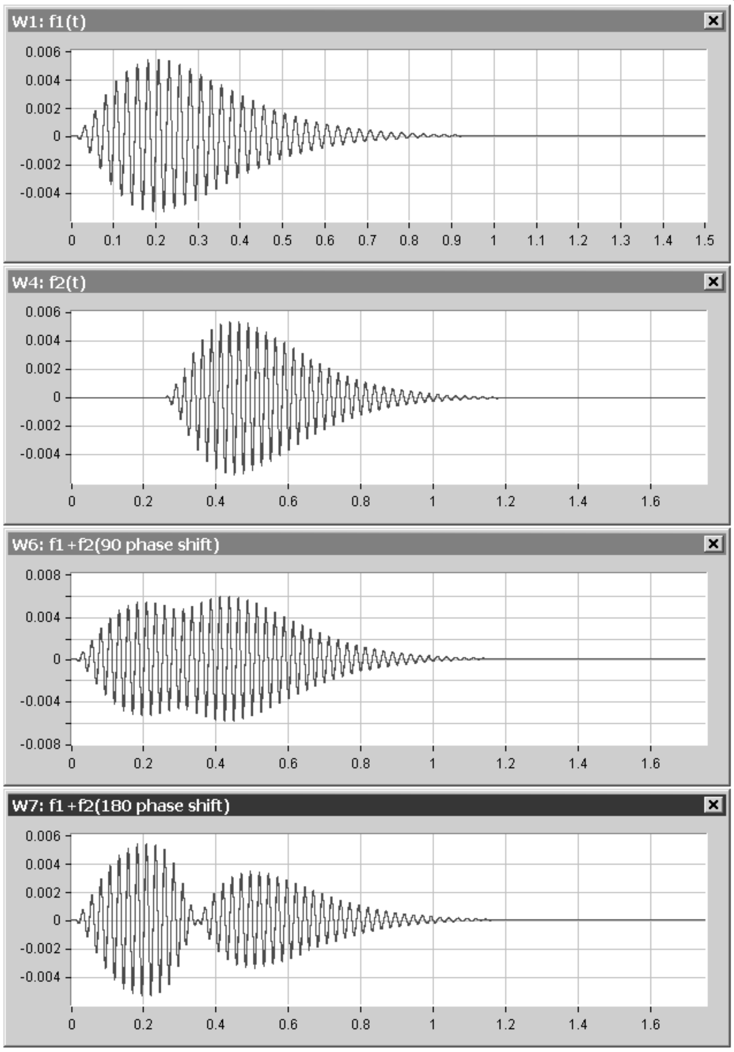

Figure 6-1: *Graphic simulation for the 90° and 180° phase shift*

6.2 Block diagram of the implementation of the phase switch method

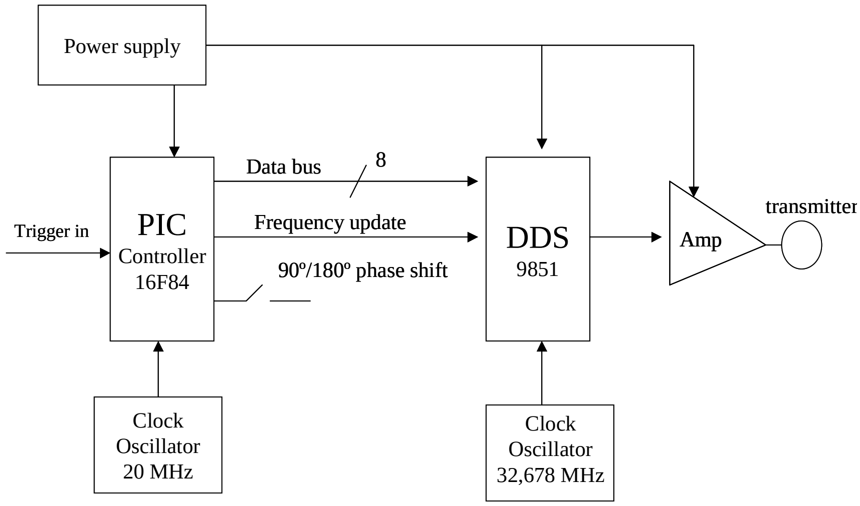

Figure 6-2: *Diagram blocks for the generation of the ultrasound burst*

To generate the special ultrasound burst (40kHz) with a phase shift of 180° introduced at the half way point, a DDS 9851 (Direct Digital Synthesis) was used. The PIC (Programmable Interrupt Controller) 16F84 triggers the DDS unit and sets values for the frequency and the phase shift required. The signal generated by the DDS unit is amplified and applied to the ultrasound transponder. The burst is formed by 10 cycles of 40kHz followed by 10 cycles shifted by 180°. The silent time between two bursts is programmed as zero frequency and zero phase signal.

Figure 6-2 shows the schematic blocks diagram. Figure 6-3 presents the protocol used to programme the DDS unit through the PIC controller.

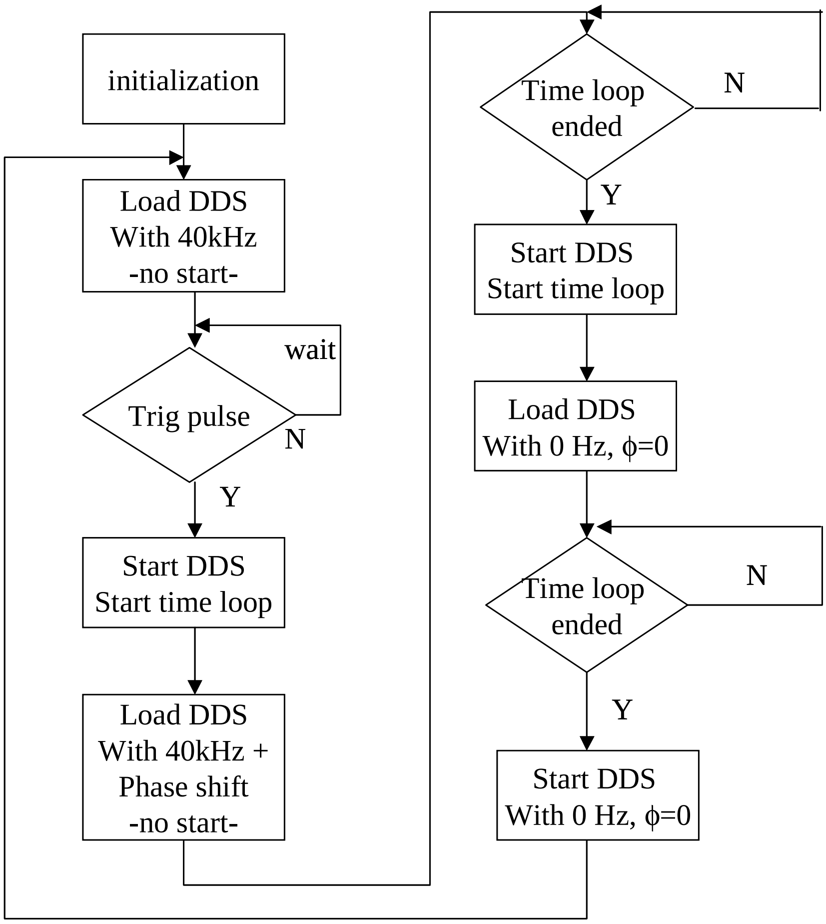

Figure 6-3: *Schematic of the DDS programming protocol*

6.3 Results

The measurements for the received ultrasonic signals were made for a transmitter-receiver distance of around 1m.

A NI card was used to acquire the ultrasound signal with an sampling rate of 300k sample /sec. The acquisition parameters were similar to the set-up described in Chapter V in order to compare the performance of the cross correlation method used with normal and phase shifted signals. The value of the frequency used for acquisition of the signal was chosen as the frequency corresponding to the much noise measurements performed in the previous set-up.

Cross-correlating the ultrasound sequence received with the previous one, we obtained the most accurate time of flight measurements.

Figure 6-4 shows the electrical signal used to excite the ultrasound transponder. The 180° phase was introduced after 10 cycles of 40 kHz.

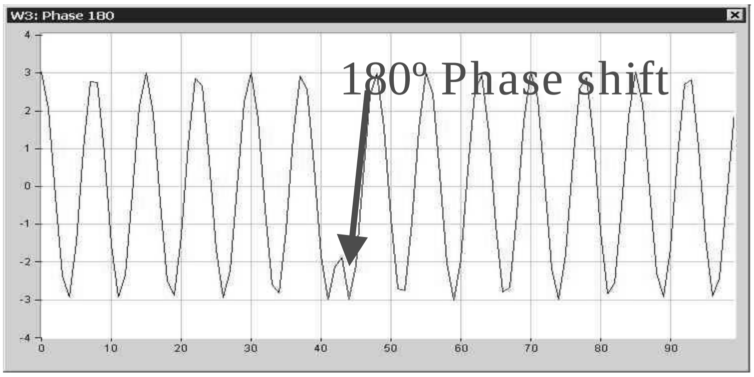

Figure 6-4: *The electrical signal(180 phase shift introduced half way) used to form the ultrasound burst*

Two consecutive sequences were acquired and the cross correlation function of these is represented in Figure 6-5. The signal received confirms the presumption that the envelope of the signal is well divided in two lobes characterised by a standard shape which does not show important change in the case of two consecutive signals influenced by variations in environmental conditions (air flow or multipath transmission generated by reflection from objects close to the trajectory of signals). To perform the measurements a ventilator was used to generate the air flow and different objects were placed close to the burst trajectory without blocking the signals.

The calculation of the transmitter-receiver distance for the different positions of the transmitter is shown in Figure 6-6. The random movements of the ultrasound transmitter are represented by the irregular variations of the distance values.

The first graph represents the stability of the distance measurements for a fixed transmitter position. The second graph shows the distance variations for a slow movement of the transmitter. The resolution of the distance measurement is about 1mm.

Graph 3 and graph 4 show the variation in distance for random movements of the ultrasound transmitter. The constant variation with +1, 0 and -1 mm indicate that the error of determining the transmitter receiver distance is lower then 1mm.

The method allows 3.3 µs resolution in determining the time of flight measurement (TOF). The resolution is limited by the sampling frequency used (300 ksample/sec). The resolution of 3.3µs for the TOF detection permits a calculation of the transmitter position with a resolution of about 1mm. The method presents a good stability and proves that the shape of the envelopes is less influenced by air flow and accidental multipath interferences. The only drawback of the method is incorrect estimation of the speed of the ultrasound burst which is a composition of the sound velocity and the air flow velocity. One solution is to measure continuously the air flow velocity and perform compensation for the burst velocity. A second solution consists in a double TOF measurement for both transmitter-receiver directions. In this case the effect of the airflow velocity component which influences the burst speed estimation will be eliminated. The last solution was tested and implemented through another TOF measurement method with an accuracy of 100µm in determining the transmitter-receive distance. The accuracy achieved was very high for laparoscopic application. This means new potential directions for additional medical applications where accuracy is required to the micrometre for the localisation of surgical instruments. The tests performed show that the influence of air flow may be compensated and will be a further subject of discussion in Chapter VII.

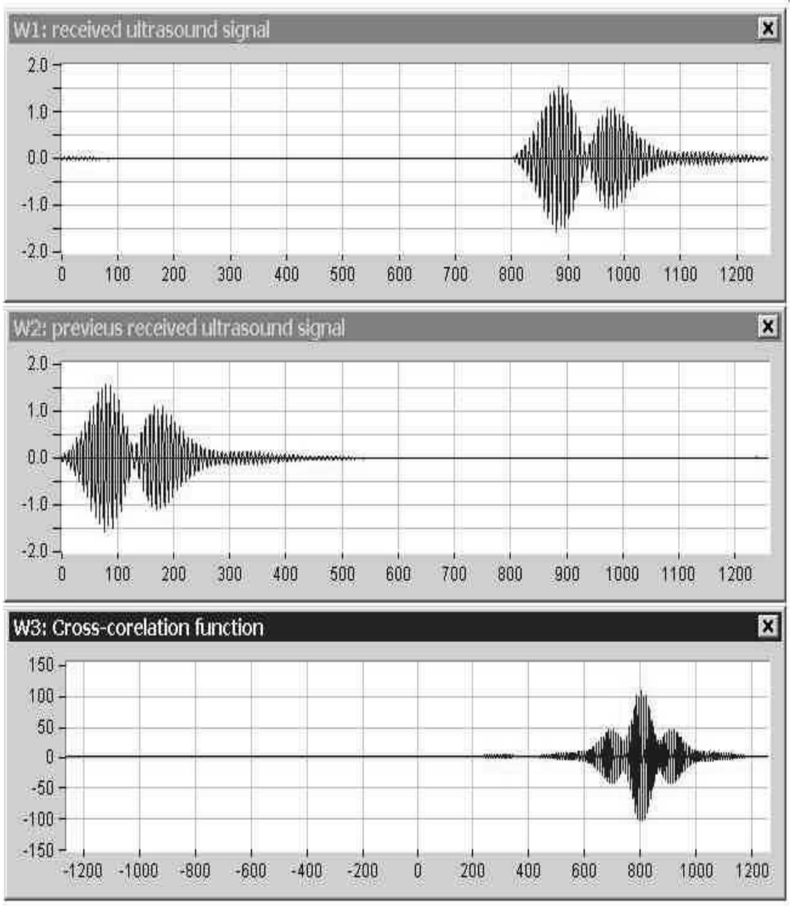

Figure 6-5: *The cross-correlation function for the reference signal with the ultrasound-acquired signal*

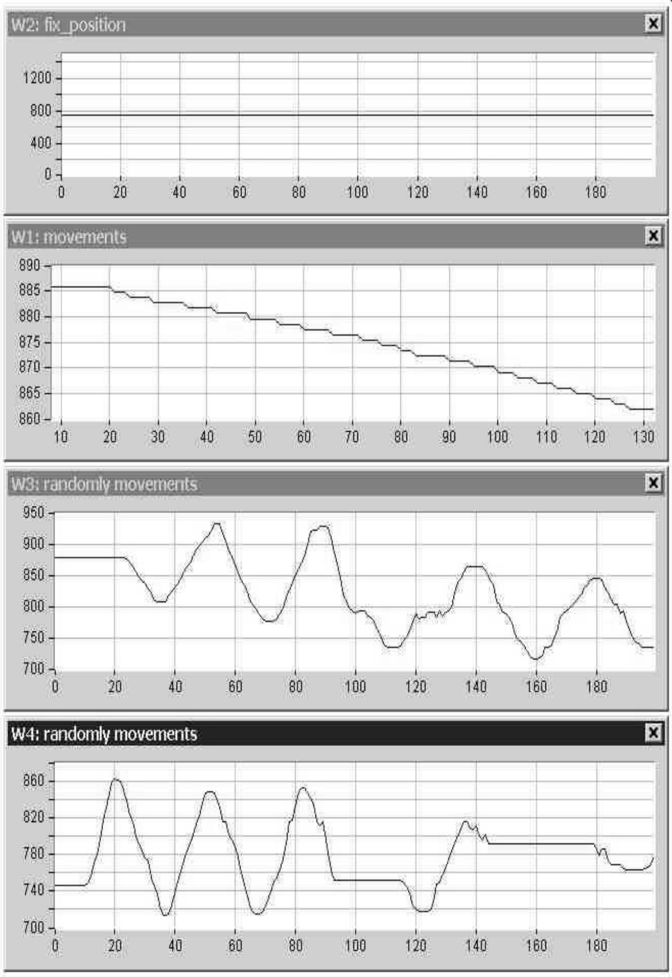

Figure 6-6: *The calculation of the transmitter-receiver distance (mm) for different positions of the ultrasound transmitter versus time*

6.4 Conclusions

The phase shift technique used to improve the TOF measurements was investigated and the results discussed. As was expected, the cross-correlation function of the signal sequences switched by a 180° phase half way through the burst yields a better estimation of the transmitter-receiver distance. Airflow turbulence or multipath interference were introduced and the method presents a good stability for the distortions introduced by those factors.

The only drawback of the method is the difficulty in measuring the airflow velocity, which leads in errors in burst velocity estimation and also in distance calculation.

Therefore chapter VII discusses solutions which may compensate entirely for airflow influences. The solutions were tested in a new method set-up which allows micrometre resolution for the distance calculations.

6.5 References

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7 Two frequency method

7.1 Introduction

Chapter V and Chapter VI presented the ultrasound sensor system and investigated solutions for carrying out distance measurements. These solutions, which are based on TOF measurement techniques, allowed a resolution in distance measurement of approximately 1mm and reduced the errors introduced by the variation of temperature and multipath reflections. Moreover, air flow turbulence shows less influence on distance estimations in the case of the phase switch method (chapter VI).

Whatever the method used, this does not solve the possible detection problem entirely due to the uncertainty in burst speed estimation. The burst velocity is a combination of the speed of sound and the speed of the air flow which exists along the burst trajectory. The difficulty to estimate and further to compensate the influence of the air flow velocity in the TOF calculation algorithm assessed for new research direction. Therefore, a two frequency method was investigated and tested. The new measurements were compared with the results of the methods/principles described before (chapters IV, V and VI) and performed better in distance detection. The air flow fluctuation phenomena identified as a drawback in previously studied methods and considered as an unsolved problem in chapter V, was also totally compensated. The technique used and the results achieved will be presented in this chapter. The micrometre accuracy obtained for the distance detection has new implications for microsurgery field.

7.2 Two frequency method

The mathematical model of the received signal

The innovative nature of this approach is related to the acoustic time of flight (TOF) measurements[7.1][7.2][7.3]. These are performed using an ultrasound burst which consists of the summation of two different frequency signals corresponding to the transducers resonance peaks, applied simultaneously to the ultrasound transmitter. A series of cycles of f_1 frequencies and a series of cycles of f_2 frequencies are applied to the transmitter. The combined signals generate an ultrasound burst characterised by a series of periodically zero-crossing points. The periodicity is directly related to the smallest common multiple of f_1 and f_2.

The signals are described by the following formulae [7.1]:

$$f_1(t) = A_1 t^n e^{-\alpha t} \sin(\omega_1 t + \phi_1) \tag{7-1}$$

$$f_2(t) = A_2 t^n e^{-\alpha t} \sin(\omega_2 t + \phi_2) \tag{7-2}$$

where n, α, φ are the transducers parameters and ω the undamped angular resonance frequency. Supposing that two such a waves travel simultaneously in space, the result is the interfered signal $f_{int}(t)$ described further. Considering:

$$A_1 = A_2 \tag{7-3}$$

the interfered signal $f_{int}(t)$ is described by the formula:

$$f_{int}(t) = f_1(t) + f_2(t) \tag{7-4}$$

$$f_{int}(t) = 2At^n e^{-\alpha t}\sin(\omega t + \phi)\cos(\Delta\omega t + \Delta\phi) \tag{7-5}$$

where,

$$\Delta\omega = \frac{\omega_2 - \omega_1}{2} \tag{7-6}$$

$$\omega = \frac{\omega_2 + \omega_1}{2} \tag{7-7}$$

$$\Delta\phi = \frac{\phi_2 - \phi_1}{2} \tag{7-8}$$

$$\phi = \frac{\phi_2 + \phi_1}{2} \tag{7-9}$$

Therefore the maxima for the envelope of f_{int} when occurs:

$$[\cos(\Delta\omega t + \Delta\phi)]' = 0 \tag{7-10}$$

The solution gives a series of time points characterised by the equation:

$$t_{max} = \frac{2\pi k - \Delta\phi}{\Delta\omega} \tag{7-11}$$

The major resonance peaks of the transducer used in the laparoscopy set-up are centred on 40kHz and 50kHz frequencies. Particularising the equation for f_1=40kHz and f_2=50kHz and assuming that the waves start in phase, then $\Delta\varphi = 0$, and $\Delta\omega = 2\pi 10$kHz. Solving the equation, the maxima serie of the envelope occurs with a periodicity of 100μs (Figure 7-3). The first zero crossing after the maximum of the envelope is considered as a time reference for the f_{int} signal. The detection of this point gives the possibility for a very accurate TOF estimation.

The uncertainty of zero crossing detection (3.3 µs) as a result of the sample frequency used (300k sample/s) was overcome using a linear interpolation technique (the last two samples before and the next two after the zero crossing were used for the interpolation algorithm).

To produce the interfered signal a PIC Controller was used to programme two DDS (Direct Digital Synthesizer) modules which generate the 40kHz and 50kHz signals. For the non-emitting "no signal" period between two bursts the DDS units were programmed to produce zero frequency signals with zero phase shift. Figure 7-1 shows a schematic of the electronics block used.

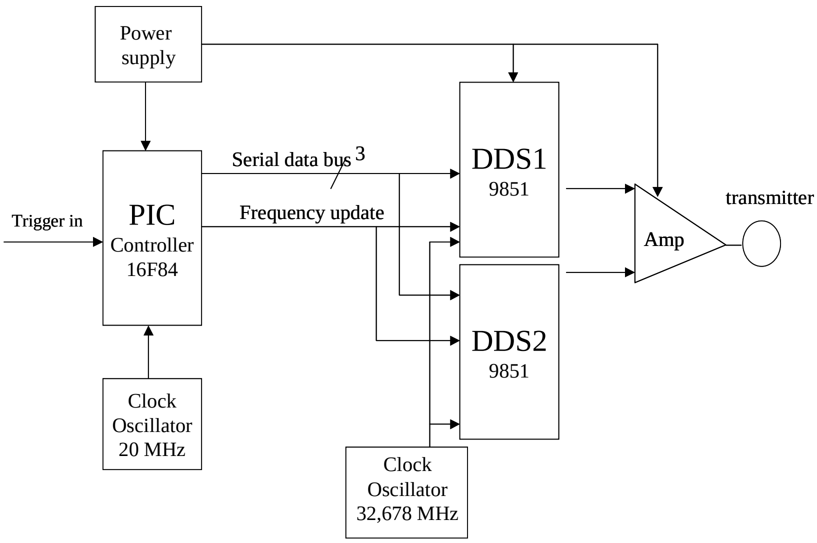

Figure 7-1: *The description of the electronics blocks used to generate the interfered signal of 40-50KHz*

Figure 7-2 describes the algorithm of the PIC controller.

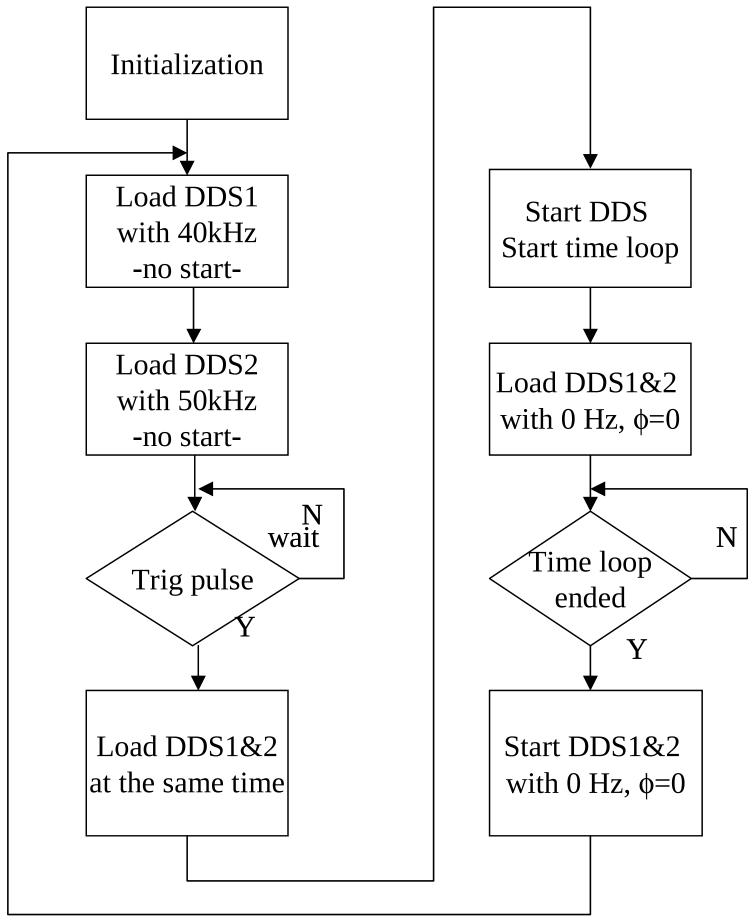

Figure 7-2: *The description of the protocol used to programme the PIC controller*

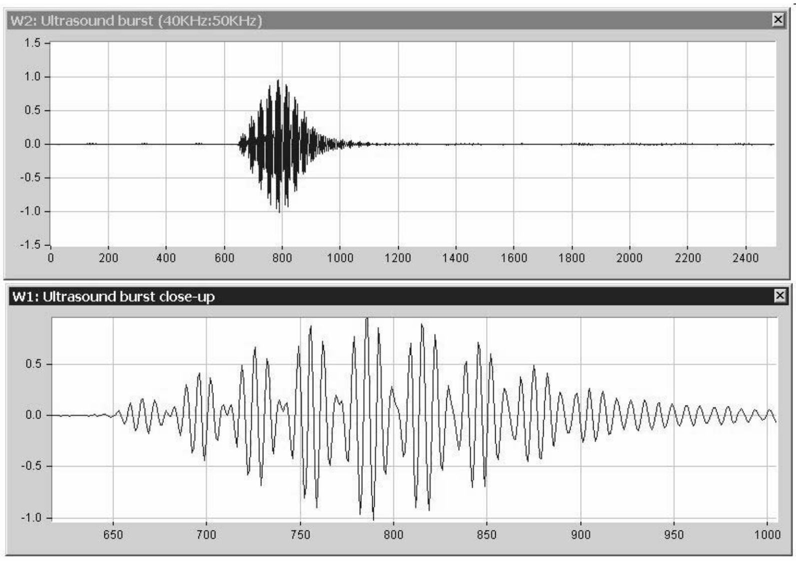

Figure 7-3: *The ultrasound burst obtained as a combination of 40 kHz and 50 kHz signals [Horizontal: sample number, Vertical: arbitrary units]*

Preliminary measurements based on two frequency method

The measurements were performed for a transmitter-receiver distance of about 1m. The environmental climatic conditions of temperature and humidity did not register large variations. The air flow speed in the proximity of the sensor was less then 0.3m/s. The ultrasound burst was acquired with a sampling frequency of 300ksample/sec and the TOF was determined. The distance calculations were based on TOF measurements performed with a repetition period of 50ms. Figure 7-4a) shows a better than 80µm accuracy and a resolution of about 5µm achieved in determining the fixed distance of transmitter-receiver at 714mm. The slow variations in the range of 80µm is explained by changes in environmental conditions (variations in temperature and/or airflow). Figure 7-4b) shows the tracking of the distance variations introduced by a micromanipulator.

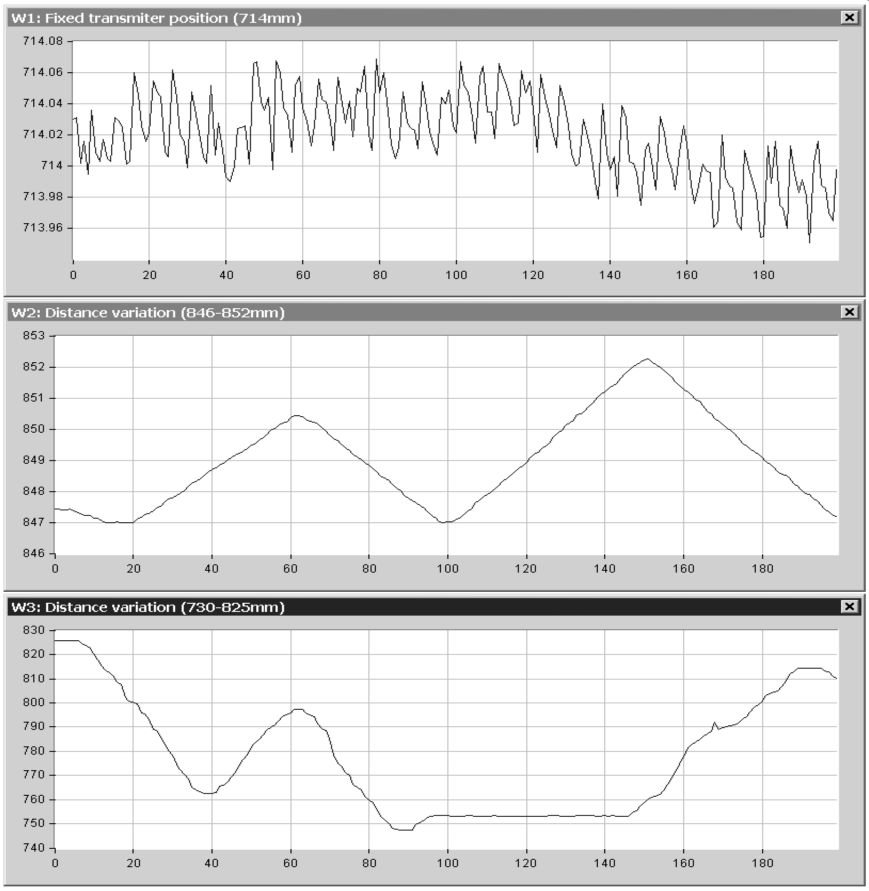

Figure 7-4: *a) The measured fixed transmitter position (714mm) shows a resolution of about 5µm and accuracy of less than 80µm in distance measurements. b) Measurements at distance variations in the range of 846-852mm (introduced by micromanipulator). c) Measurements at distance variations in the range of 750-825mm (the movement of the transmitter was introduced by hand manipulation with an average speed of 0.5m/sec)*

Figure 7-4c) shows the tracking of the distance variations introduced by manual movement. This type of movement is similar to the movement of the doctor's hand during laparoscopic intervention. The measurements show a high accuracy in distance measurements. Resolution and the accuracy achieved suggest that the method may also be used for the high precision location of instruments used in micro-surgical intervention. Before accepting the method as a viable solution the factors influencing distance calculation should be investigated.

The next equation explains the dependence between the distance calculation and factors as variation of the temperature or air flow fluctuation.

$$d = tof \times v_{burst}((\Delta T), \Delta(v_{airflow})) \tag{7-12}$$

v_{burst} is the burst speed which is sensitive to temperature variations ΔT and the air flow fluctuations $\Delta_{airflow}$. The measurements and the results presented in Figure 7-4 were not performed for large variations in these parameters.

Therefore the next investigations present the stability of the method considering large variation of those factors and solutions to compensate for them.

Influences of temperature variation

As was presented in subchapter temperature variation represents an important factor in sound velocity calculation. A proper determination of the sound velocity values implies a better estimation of the burst velocity. Therefore several distance measurements for a fixed position transmitter-receiver were taken and the temperature variation was registered.

Figure 7-5 shows the distance and the temperature measurements achieved over 10 hours with an acquisition rate of 10Hz.

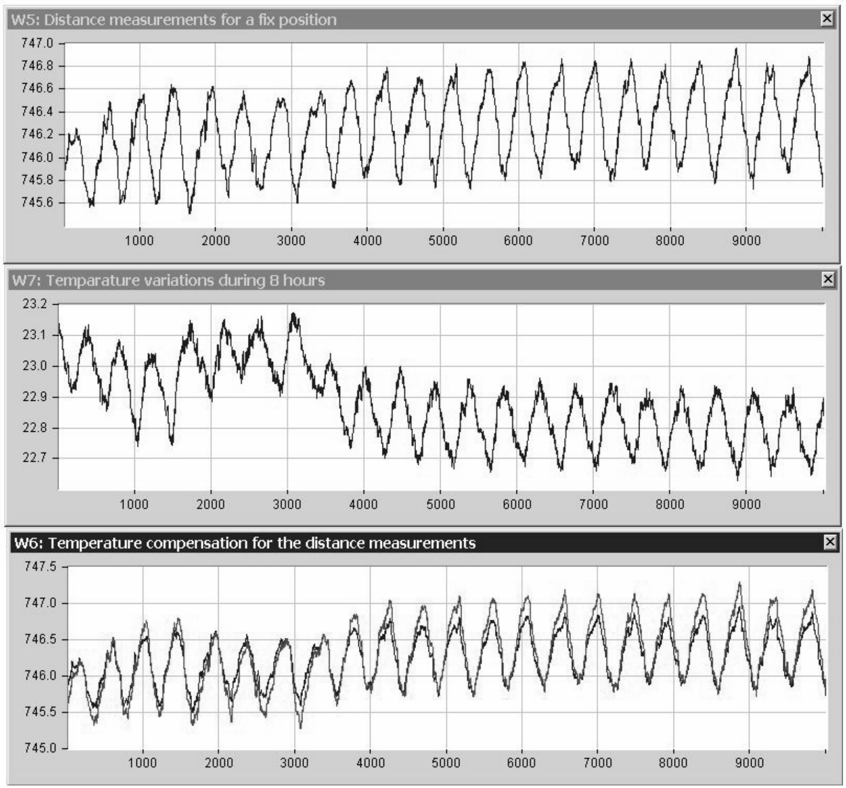

Figure 7-5: *a) Distance measurements for a fixed position-transmitter receiver around 746mm. b) Temperature variations during the distance measurements c) Distance measurements compensated for temperature variation influences.*

The distance measurements were compensated for temperature-related speed of sound variations. Figure 7-5 c) shows that after temperature compensation the distance may be estimated to an accuracy around of 200µm. For the temperature measurements a regular temperature sensor was used and the speed of sound estimation was based on Eq 4-4., chapter IV. The air flow influence was not investigated during the temperature measurements.

The next subchapter presents the results of the distance measurements in the presence of induced air flow and the accuracy achieved after compensation for this factor.

Influence of air flow fluctuation.

To investigate the significance of air flow variation on distance calculation, in the proximity of the sensor air flow with a speed of 0.6m/sec was induced. The results are presented in Figure 7-6 and show a large variation in distance compared to the accuracy of 80µm reported for small variations in airflow.

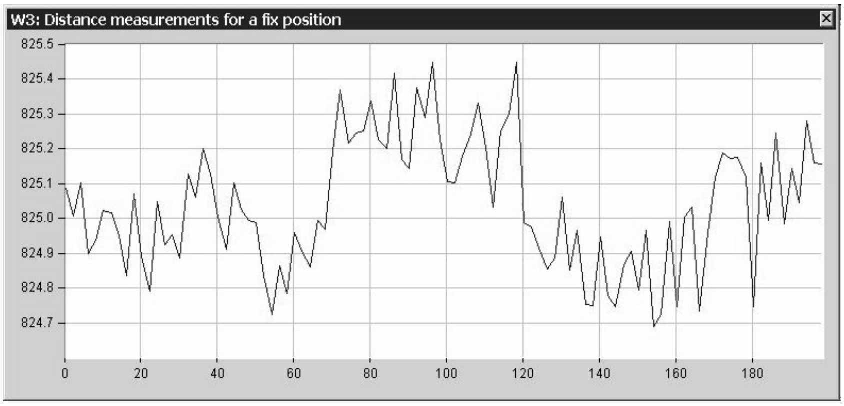

Figure 7-6: *The influence of air flow (0.6m/sec) disturbance on distance estimation for a fixed position [Horizontal: sample number, Vertical: mm]*

Therefore the influence of the airflow speed in burst velocity determination is explained further. Due to the vectors composition of the sound velocity and the air velocity the calculation of the speed of burst is represented by the equation:

$$\vec{v}_{burst} = \vec{v}_{sound} + \vec{v}_{air} \tag{7-13}$$

Eq. (7-13) shows that air flow has greater influence when the velocity vectors have parallel directions and has less influence when the vectors are perpendicular. Therefore a dual way measurement technique was considered in order to compensate for the airflow component, parallel with the velocity field. Figure 7-7 describes the set-up for the bidirectional TOF measurements.

After the reception of the burst, the transponders switch the reception-transmission mode status. The ultrasound burst is retransmitted and considering that the influence of airflow for the return path has an opposite effect for distance calculation, the disturbances due to airflow are compensated. tof_1 and tof_2 are the time of flight calculations for the same distance in opposite directions. The unknown distance d is extracted from the following equations

$$d = tof_1 \times (v_{sound} + v_{air}) \tag{7-14}$$

$$d = tof_2 \times (v_{sound} - v_{air}) \tag{7-15}$$

Reducing the v_{air} from these two equations the distance d is expressed as follows:

$$d = 2v_{sound} \times \frac{tof_1 \times tof_2}{tof_1 + tof_2} \tag{7-16}$$

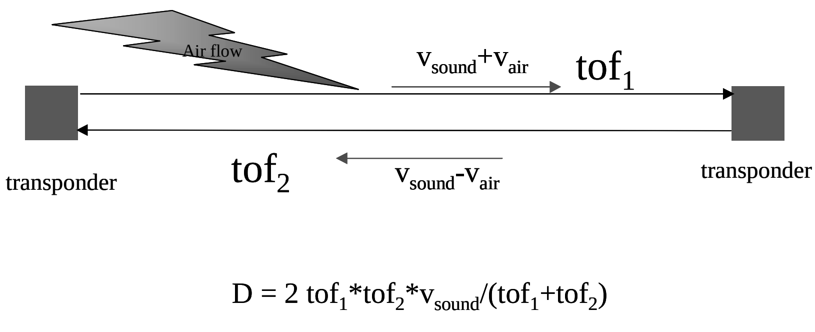

Figure 7-7: *Schematic of the air flow compensation*

Figure 7-8 shows the distance measured after the airflow is compensated. The air flow speed registered by a flow sensor varied at around 1m/s. The large signal represents the distance measured with airflow disturbances. The small signal is the distance after the compensation algorithm was applied. The accuracy achieved with this compensation technique remains in a convenient range of 150µm.

The measurements show a high accuracy in the distance measured. The resolution and the accuracy achieved suggest that the method also may be used for the high precision location of instruments used in micro-surgical intervention.

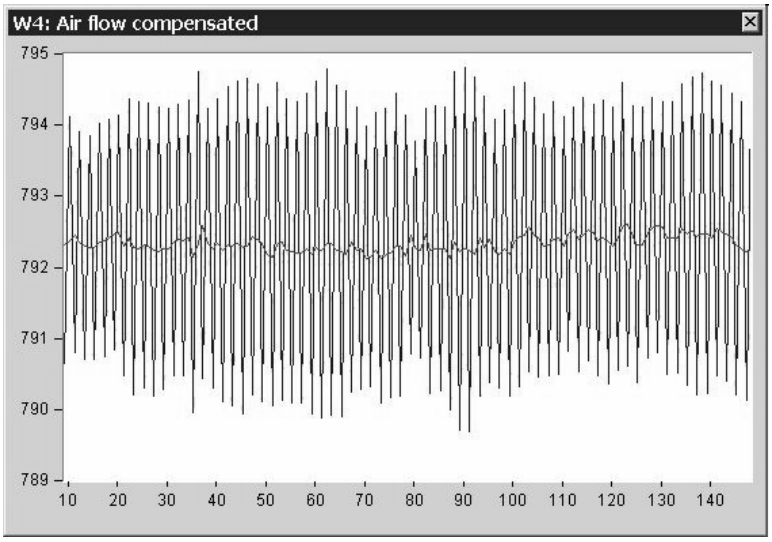

Figure 7-8: *The large variation of 5mm in distance measurement for a fix transmitter-receiver position due to airflow influence and uncertainty after compensation (150µm accuracy)[Horizontal: sample number, Vertical: mm]*

7.3 Conclusions

A new method was presented which opens new directions and possible applications in the micro-surgery field. Combining two frequency in the emitted signal, the received signal has very stable time reference points. Considering these references the TOF measurements were performed with micrometre accuracy. Because the signal envelope obtained as an effect of the signals summation, the multhipath transmission does not influence the TOF measurement. Temperature and air flow influences were investigated and methods were used to compensate their effects. After temperature compensation an accuracy of $200\mu m$ was obtained. Also, due to the vectors composition of the sound velocity and the air velocity "the double way" measurement setup was implemented. After the compensation of the airflow velocity component which is parallel with the ultrasound propagation direction, the calculation of the transmitter-receiver distance was performed with an accuracy of $150\mu m$. Probably a better estimation may be achieved if the temperature and airflow will be compensate in the same measurement setup. Also a possible explication of the distance variation in this range may be the result of the mechanical vibrations induced by the external factors present in the measurement room and which were not controlled or measured.

The final results which indicate that the one dimensional measurements with an accuracy better then $200\mu m$ may be achieved shows an stable and precise method to determine the informational position of different parts of surgical instruments. The compensation of the air flow disturbances represents a very interest approach which points that the most important drawback problem in the ultrasonic measurements was overcome. Finally a cheap ultrasonic system may provide a very high accuracy and desired stability for microsurgery applications.

7.4 References

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8 Conclusions and future research

Minimal invasive medical interventions are a modern way of treating disorders in many medical areas. Especially the laparoscopic procedures minimise patient trauma, mainly because of the small access site through which medical tools can be inserted in the body. Beside this important advantage the patient discomfort is reduced, the healing process is shortened and the risk of infection is minimised.

All these procedures need a way of monitoring the medical instruments once they are inserted into the body. An endoscope is used to show the surgeon what is happening inside the body, however this view is very limited and gives no information on instrument position outside the camera view. Even the use of 'standard' imaging techniques such as X-ray, ultrasound, MRI and CT scans before or during the operation do not solve this problem satisfactory. To be able to work, the surgeon builds a three dimensional in his head and tries to track the instruments inside. However, the position errors accumulate forcing surgeons to occasionally retract their instruments and /or the endoscope and ‘reset’ this mental model. Tracking the instruments becomes more difficult during the operation due to blood that obscures the few indicators which are used as guidance such as colour, texture and shape of the organs.

To alleviate this problem and because at present there is no suitable medical device that renders the information with respect to the position of the laparoscopic instruments, it was decided to develop a system that would add more information about the position and orientation of laparoscopic instruments and to reduce the number of retraction of instruments. The device should record the movement of the instruments during the surgical procedure, being especially useful to train less experienced young doctors. The research work and the development phase of the device, yielded important conclusions.

Because the medical environment and the specific requirements which should be respected in order to use this device in minimal invasive surgery, several types of sensors were investigated (magnetic, mechanical, optical, acoustic). Each one shows advantages and disadvantages which were compared. Finally, the research work leaded to an ultrasound wireless solution which was further implemented in our setup.

The sensor used to locate the laparoscopic tools consists of an ultrasound transmitter triggered by a RF receiver. Two pair of sensors placed outside of the human body, one pair on the tip of the laparoscope and the second on the trocar determine the unknown position and orientation of the instrument tip inside the body. To detect the position of sensors the Time of Flight principle is used. An array of ultrasound transmitters placed above the surgical table detects the signals emitted by the sensors.

To implement the wireless solution a RF unit produces codes which triggers and also identifies the ultrasonic transmitters.

As a first step towards the completion of the ultrasound sensor system several methods were used to estimate the distance between a pair of an ultrasound transmitter-receiver. To estimate the position of the ultrasound sensors relative to several ultrasound receivers (implicit the position and the orientation of the surgical instruments) the time of flight accurate estimation was crucial to determine the limits of the system.

The “cross-correlation” digital technique used in the final setup to calculate the time of flight show an accuracy better than 1mm in estimating the sensor-receiver distance.

Apart from the accuracy obtained choosing an optimal method to calculate the distance there are important disturbances factors (multhipath transmission, climatical variations) which should be identified and compensated for. Therefore the multhipath transmission phenomena, the influence of temperature and disturbances due to air flow were estimated. Using the temperature compensation technique the accuracy of the system remains in a range of 2.8mm which is sufficient for the laparoscopic procedure. The presence of the air flow (around 1m/s) induces an error of 9-10 mm in position detection. Therefore the presence of the air flow and its effect combined with the multhipath transmission constitutes an important factor which demands more research.

To simulate the practical situation, we fabricated a patient model and used it for one dimensional measurements. The tests were performed with one ultrasound sensor attached to a laparoscopic instrument. The endoscope was replaced with a video camera which was used as a positioning references for the sensor. The information regarding the sensor position superimposed on the monitor proved the accuracy of the sensor detection.

To overcome the disadvantages introduced by airflow turbulence or accidental multhipath interference, the phase switch method (Chapter VI) was used. The same frequency was used for the total burst, but half way through the burst a phase shift of 180° was introduced into the signal. The enhanced resolution of 3.3µs for the burst detection permits a calculation of the transmitter position with a resolution of about 1mm. The method shows a good stability and the shape of the envelopes is less influenced by air flow and accidental multhipath interferences. The only drawback of the method is the difficulty in measuring the airflow velocity, which leads to errors in burst velocity estimation and also in distance calculation.

Therefore, “two frequency method” was developed and tested. A series of cycles of frequency f_1 and a series of cycles of frequency f_2 are applied to the transmitter. The obtained signal presents a series of zero crossing which are a good reference in time for the ultrasonic signal. The ultrasonic signal detection measurements were compared with the results of the methods/principles described before (chapters IV, V and VI) and showed better performance in distance detection. The “two frequency method” which showed micrometers accuracy of sensor detection was tested in a special setup where “two way” measurements of transmitter receiver distance were performed. The two way measurement was implemented in order to compensate for air flow fluctuations. The air flow fluctuation phenomena identified as a drawback in previously studied methods and considered as an unsolved problem, was totally compensated. The accuracy achieved with this compensation technique remains in a convenient range of 200µm.

Finally, this thesis presents various aspects regarding the development of an ultrasound sensor systems for use in laparoscopic intervention. Considering the high accuracy (in the micrometers range), the flexibility and the ergonomics achieved for this system, the final device may be adapted very easily and used also in other microsurgery applications.

Future research

A number of various aspects can be investigated in a future research project, from which the most important would be the miniaturisation and integration of the RF receiver unit together with the ultrasound transmitter. To compensate totally air flow disturbances a microprocessor must be integrated in the sensor unit in order to process the double way measurements algorithm presented in Chapter VII.

The whole sensor unit needs to be packaged in order to allow sterilisation. For the measurement setup batteries were used to provide the power supply for the sensor unit. Therefore a wireless solution to recharge the batteries will be more suitable with respect to the sterilisation process.

From the performed measurements it was obvious that the data acquisition card played a significant role in the accuracy of the position detection. Especially for those applications where the accuracy should be less than 1mm, the information must be processed independently of the operating system running on the computer. A possible solution would be the use of a DSP unit to process all digital algorithms. The computer unit should become just a simple interface to represent 3D information to the surgeon.

The optimal placement position for the sensor must be defined depending on the size and the shape of the surgical instruments and 3D measurements performed. The errors due to mechanical placement of the sensors must be added to the final characterisation of the system.

Although in this research the location of the sensor was determined, it still has to be embedded in the medical reference image. This can be done by acquiring the medical image from the endoscope and superimposing the position and orientation of the sensor on it.

Summary

This thesis describes the development of a positioning system for use in minimal-invasive laparoscopic interventions. In such procedures, specially designed medical tools enable intervention to treat a large number of diseases with minimal damage to healthy tissues and with reduced trauma to the patient. An endoscope is used to show the surgeon what is happening inside the body. However this view is very limited because of the 2D nature of the images and in those situations when the instruments are outside of the camera view. This makes the operation procedures more difficult. To alleviate this problem and because at present there is no suitable medical device that renders the information with respect to the position of the laparoscopic instruments, an positioning system based on ultrasound was developed that would add more information about the position and orientation of laparoscopic instruments.

Discussions with an interventional laparoscopic surgeon at the Erasmus Medical Centre in Rotterdam, Prof. Jaap Bonjer and a representative of Olympus Multinational, producer of laparoscopic instruments, underlined the main concern of the medical profession regarding the possibilities to meet the requirements for such a system.

The sensor used to locate the laparoscopic tools consists of an ultrasound transmitter triggered by a RF receiver. Two pair of sensors placed outside of the human body, one pair on the laparoscope and the second pair on the trocar determine the unknown position and orientation of the instrument tip inside the body. To detect the position of sensors the time of flight principle is used. An array of ultrasound transmitters placed above the surgical table detects the signals emitted by the sensors and further a read-out circuitry was designed in order to transfers the information to a process unit were the time of flight was calculated.

Different factors, effects and phenomena were considered during the process of sensor identification. Considering the ultrasound wave produced by an ultrasound transponder, the absorption air factor, the beam pattern, related to the frequency of ultrasound transmitter and climatic factors affecting the speed of sound, the choice was made for an ultrasound transmitter/receiver of 40kHz. The climatic factors were investigated and the stability of the sensor related to these factors was characterised. Temperature variations, air flow or multihpath transmission were identified as the most important disturbing factors which may influence the accuracy of the system. For the final system setup several methods for time of flight calculation were investigated. The cross-correlation technique used allows a resolution of 0.7 mm of determining the transmitter position but the factors presented above may induces errors of 8-9 mm. To reduce the errors of the measurements, digital filters may be implemented and further the accuracy of detection increased.

To simulate the practical situation, we fabricated a patient model and used it for one dimensional measurements. The tests were performed with one ultrasound sensor attached to a laparoscopic instrument. The endoscope was replaced with a video camera which was used as a positioning references for the sensor. The information regarding the sensor position superimposed on the monitor proved the accuracy of the sensor detection. Future research should continue with the optimisation of the total system, the miniaturisation of the sensors and tests regarding the packaging of the sensors in order to respect the sterilization procedure.

In order to enlarge the applicability for such a system to the area of microsurgery applications, where an even higher accuracy is required the system setup based on cross-correlation technique was modified and two other methods were investigated.

One of the methods tested "The 180° phase shift introduced half-way through the burst" was highly successful at reducing the detection error. Even better results were achieved with the "two frequency method" which showed micrometers accuracy of sensor detection. The only drawback of these methods is the difficulty in dealing with the airflow velocity, which leads to errors in burst velocity estimation and also in distance calculation.

Therefore the "two frequency method" was introduced and tested in a special setup where bi-directional measurements of transmitter-receiver distance were performed. With this method it is simple to compensate for air flow fluctuations. After compensations for air flow disturbances and temperature variation the "two frequency method" was able to perform distance measurements with an accuracy of 200µm.

These last results proved that a cheap ultrasonic system may provide a sufficient high accuracy and stability for microsurgery applications.

Samenvatting

Dit proefschrift beschrijft de ontwikkeling van een ultrasonisch systeem voor het gebruik bij minimale invasieve laparoscopische interventie. Bij deze methode zijn speciale medische instrumenten ontwikkeld om een groot aantal ziekten te kunnen behandelen met een zo klein mogelijke schade aan gezond weefsel en met zo min mogelijk letsel voor de patiënt. Een endoscoop wordt gebruikt om de chirurg te laten zien wat er in het lichaam gebeurt. Echter het zicht is erg beperkt door de 2D eigenschappen van de camera beelden en in die situaties dat de instrumenten zich buiten het zichtveld van de camera bevinden. Dat bemoeilijkt de operatie handelingen. Om deze problemen te verminderen en omdat er op dit moment geen instrumenten zijn die positionele informatie van de laparoscopische instrumenten kunnen meten, is een ultrasonisch systeem ontwikkeld dat meer informatie geeft over de positie en oriëntatie van een laparoscopische instrument.

Na gesprekken met een chirurg van het Erasmus Medisch Centrum in Rotterdam, professor dr. Jaap Bonjer en een vertegenwoordiger van Olympus Multinational, een producent van laparoscopische instrumenten, werd duidelijk dat medici twyfels hebben over de haalbaarheid van zo’n systeem.

De sensor die gebruikt wordt voor het bepalen van de locatie van de laparoscopische instrumenten bestaat uit een ultrasonische zender die getriggerd wordt door een RF ontvanger. Twee paar sensoren die geplaatst zijn buiten het menselijk lichaam, één paar op de tip van de laparoscoop en één paar op de trocar bepalen de onbekende positie en orientatie van de tip van het instrument in het lichaam. Om de positie van de sensoren te bepalen is het "time of flight" principe gebruikt. Een groep ultrasound receivers die geplaatst zijn boven de operatietafel ontvangen de signalen, verzonden door de sensoren, verder bestaat het geheel uit een uitleeseenheid die ervoor zorg draagt dat de informatie verplaatst wordt naar een verwerkingseenheid waar de afstand berekend wordt.

Verschillende factoren, effecten en fenomenen zijn onderzocht gedurende het process van de sensor keuze. Bij het overwegen van het type ultrasound transponder, is gekeken naar de absorptie factor van lucht, het patroon van de akoestische bundel in relatie tot de frequentie van de ultrasonische zender/ontvanger en de klimatologische factoren die de eventuele snelheid van het geluid kunnen beïnvloeden, en is de keuze gevallen op een ultrasonische zender/ontvanger van 40KHz. De klimatologische factoren zijn onderzocht en de stabiliteit van de sensoren in relatie tot deze parameters gekarakteriseerd. Temperatuurvariaties, luchtstromingen en meervoudige geluidspaden zijn bepaald als de meest belangrijke verstoringen die de nauwkeurigheid van het systeem kunnen beïnvloeden.

Voor het uiteindelijke syteem zijn verschillende methoden voor het bepalen van de vlucht tijd onderzocht. Met de "kruis-correlatie" techniek kan er een resolutie van 0.7 mm voor het bepalen van de zenderpositie behaald worden maar hiervoor genoemde factoren kunnen nog steeds een fout van tussen de 8 en 9 mm veroorzaken. Om deze meetfouten te corrigeren en de nauwkeurigheid te vergroten kunnen digitale filters worden geïmplementeerd.

Om een practische situatie te kunnen simuleren hebben we een patiëntmodel gemaakt die we gebruikt hebben voor ruimtelijke metingen. De endoscoop werd vervangen door een videocamera die als positie referentie gebruikt is voor de sensor. De informatie betreffende de sensorpositie weergegeven bovenop het videobeeld heeft de nauwkeurigheid van de sensor detectie aangetoond. Toekomstig onderzoek zou zich moeten richten op het optimaliseren van het complete systeem, miniaturisering van de sensoren en het testen van de omhullingen wat betreft medische sterilisatie procedures.

Om het systeem ook geschikt te maken voor microchirurgische toepassingen, waar zeer hoge nauwkeurigheden nodig zijn, zijn nog twee andere methodes onderzocht.

Een van de methoden die erg goed werkte was de introductie van een fasestap van 180° halverwege het akoestische signaal. Hierdoor was een betere detectie van het verstuurde signaal mogelijk. Nog beter resultaten werden behaald met de "twee-frequentie methode" die micrometer resolutie mogelijk maakte. Het enige nadeel van alle voornoemde methoden is de moeilijkheid om de metingen te compenseren voor de luchtsnelheid.

Daarom is de "twee-frequentie methode" getest in een opstelling waarin bi-directionale metingen mogelijk zijn. Op deze wijze kon eenvoudig voor luchtstroomfluctuaties gecompenseerd worden. Na compensatie voor luchtstroom en temperatuur variaties werd een nauwkeurigheid van 200µm bereikt.

Deze laatste resultaten tonen aan dat een goedkoop ultrasoon system voldoende stabiel en nauwkeurig kan zijn voor microchirurgische applicaties.

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