The Integration of Soft Tissue Data Into a 3D Model of the Human Head

by

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ABSTRACT

This thesis describes the acquisition of maxillo-facial soft tissue thickness data and its integration into a Computer Aided Design based solid model of the human head. The final outcome was a significant element in a novel approach to the creation of a 3D modeller for detailed study of the design of protective equipment at the interface with the skin tissues.

Anthropometric surface data of 300 head 3D profiles were acquired using a laser scanning system consisting of a dual mirror configuration and a CCD video camera. Approximately 70,000 data points are scanned in less than ten seconds. The resulting surface model has a resolution of ±0.5mm circumferentially and ±0.2mm radially.

Soft tissue thickness values have been measured using a A-scanning ultrasound system with a pulsed, 12.5MHz focused ultrasound probe with an axial resolution of 0.31mm and a lateral resolution of 2.7mm. The accuracy of the measurements has been estimated as ±0.25mm. A novel representation of these data has been suggested in which quasi iso-thickness zones have been identified. These zones, where the thickness values are often consistent to within as little as ±0.5mm, have been shown to be consistent with key anatomical regions. Colour spectral plots visualise the tissue zones.

The thickness data are referenced to the laser data using a magnetic spacing system, POLHEMUS by 3DSpace®. The resulting 3D soft tissue model has important uses in Finite Element Analysis methods for design of protective equipment.
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1. **The Relevance of Soft Tissue In Anthropometry**

1.1 **Overview and Aim**

This work has been completed as part of a project commissioned by the Defence Research Agency (DRA) in 1990. The purpose of the DRA project was to do a modelling study of human head anthropometry for use as a database for the design of a new generation of flying helmet, oxygen mask and ancillary equipment. The objective of the research described in this thesis was to investigate the development of an anthropometric solid model of the human head using modern data acquisition methods and the eventual integration of the acquired data into a computer aided design system. In particular, an objective was to attempt to incorporate underlying soft tissue thickness into the computer model to provide a soft tissue shell.

This thesis describes the acquisition of the topographical and thickness data, and a new approach to the union of this data into a model. The concept of anthropometry and its applications, as well as a brief anatomical description of the relevant soft tissues are introduced in sections 1.2 and 1.3. Applications for use of this research, including the above mentioned DRA exercise, range from maxillo-facial surgery to forensic reconstruction and are outlined later in this chapter.

1.2 **Background of Anthropometry**

Quetelet (1796-1874) introduced the term "Anthropometry" in the early 1800's. It is a derivation of two Greek words, *Anthropos* meaning human, and *metrikos* pertaining to measurement. Thus anthropometry, as derived, became associated with a technical use of the measurement of human beings. As early as Greek and Roman times, scientists studied body dimensions and proportions and the resulting implications of man's interaction with his environment. For example, the Roman architect Vitruvius in the 1st century BC published quantified sketches of human body proportions and highlighted their importance in architectural design.
Leonardo da Vinci later created his famous drawing of a man based on Vitruvius' theories of proportions as seen in figure 1.1.

In the 13th century, systematic investigations into human body size and shape began with Marco Polo's explorations. Marco Polo recorded and compared large varieties of body size and build in the different races around the world. In the early 1700's scientist undertook to explain these differences based on geographic and temperate conditions. This became known as Comparative Racial Anthropology, through which anthropometry became its own discipline.

The ability to represent human dimension with mathematics and statistics was developed through the scientific method and applied to mass markets during the Industrial Revolution. Mass production of clothes in workhouses was dictated by the statistical tables and charts becoming available. Blumenback (1752-1840), was the first to compile and report on all the anthropometric data at his time.

Up to the early 1900's, anthropometrists were concerned mainly with acquiring human data for clothing and racial comparisons. Just prior to World War II (WWII), an interest developed regarding anthropometric data and its potential role in improving man-environment interaction. The problems were emphasised by aircraft cockpits which were too small for the pilots with controls difficult to reach. Although the military had shown interest in anthropometry as early as the Civil War, it was WWII that highlighted for them the importance good anthropometric data would have in the future of aircraft and protective gear design. Since the 1940's many anthropometric data tables have been published as a result of surveys done at military institutions. A selection of the most recent surveys include U. S. Air Force laboratories at Wright-Patterson Air Force Base, the U. S. Army Quartermaster Research and Development Command and the Royal Air Force research centres in the UK. These surveys include anthropometric data of the head that was taken using classical anthropometric techniques. Only the report from Wright Patterson lists data taken by laser scanning, a modern surface acquisition method (introduced in Chapter 2).

The measurement of detailed human shape has been limited by the capabilities of traditional anthropometric tools. In the past, human shape has been defined by the available linear dimensions, i.e. the length of a leg or the circumference of a head. As shape definition has progressed from linearity's to nonlinearity's, i.e. surface definition, with the advent of graphics and computer aided design packages, scientists have developed new approaches to the types of data collected.
It is only as recently as the 1950's that science has developed the tools to define, practically, non-linear surfaces with three-dimensional co-ordinates and spatial relationships. By the late 1980's this data could be acquired and processed quickly and efficiently. These modern methods, discussed in Chapter Two, have involved the work of many scientists not trained in the field of anthropometry. Roebuck, Kroemer and Thomson have defined the specialised field of engineering anthropometry which encompasses the different sciences used in modern anthropometry and exemplifies the approach used for this research, viz.

"Engineering anthropometry is the application of scientific physical measurement methods to human subjects for the development of engineering design standards and specific requirements and for evaluation of engineering drawings, mock-ups, and manufactured products for the purpose of assuring suitability of these products for the intended user population."

Engineering anthropometry is of interest to a large group of industrial and commercial sectors. Car manufacturers use the data to improve the ergonomics of the vehicles. With the knowledge of the population's reach spans and body sizes, they are able to improve the efficiency of human interaction with the car controls.

Documented variations in body dimensions also give the clothing manufacturers statistical size groups around which to design clothing. Marks & Spencer plc and Truelife in Dublin have expressed interest in using the latest technology of anthropometric data acquisition.

Sporting manufacturers are interested in the data for the prediction of improved sporting performance with changes in design of sporting accessories and protective equipment.

The unique feature of the work described here is the use of two measurement techniques, laser scanning and ultrasound, to acquire topographical and soft tissue dimensional data of the human head, and the combination of the data with that of the CAD environment to produce solid models. The data are referenced together with a third system, a magnetic positioning sensing device before being modelled in a computer aided design system, resulting in an accurate design tool. In addition, the data collected for the model development, form a new three-dimensional anthropometric database, from which statistical information can be derived.
1.3 Soft tissue and its relevance to the study

A general description of what is meant by the soft tissue layer is necessary for clarification. The soft tissue of interest to this study comprises three main layers. The skin (figure 1.2, layer 1) is the outermost of the three layers of tissue incorporated into our soft tissue work, the other two being the subcutaneous tissue (layer 2) that contains fat cells, and the deep facia or muscle layer. Two sub-layers, epidermis (layer 1a) and dermis (layer 1b), constitute the skin. The nature and function of these layers are described below.
1.3.1 Epidermis

The epidermis is the outermost layer of the skin. It is typically less than 1mm thick and is further composed of many cell layers. The cells in the deeper layers of the epidermis are living because of diffusion of nutrients from the dermis. The living cells are passed to the surface of the epidermis, and gradually die due to the lack of a vascular supply. As the cells move towards the surface, the nutrients passed on from the dermis no longer reach them. They become cornified, or tough, because of the presence of a protein called Keratin. In this form these cells act as a barrier against water and protection from microorganisms and minor mechanical damage. The thickness of the dead cell layer, also known as the "stratum corneum", is dependent on the location of the body. In protected areas such as the dorsal aspects of the body, it's thinner than in exposed ventral areas. An exception is on the hand and feet where the dorsal aspect is actually thinner than the ventral aspect. Thickness of the epidermis is also determined genetically to some degree.

1.3.2 Dermis

The dermis is the lower and thicker layer of the skin. It is comprised predominantly of dense connective tissue and contains the blood vessels, tactile nerves, hair follicles, small muscles that raise hair, and glands. The dermis itself has two layers, the papillary and reticular layers. The papillary layer contains the dermal papillae which projects into the epidermis, forming a connecting bond. The dermal papillae also contain blood vessels which diffuse nutrients to the epidermis. The reticular layer of the dermis contains collagen and elastin (connective tissue fibres.) These fibres give the skin its mechanical strength and elasticity. The dermis is able to repair deeper wounds and resist infection to the skin and regulate body temperature with its vascular supply. Its varying thickness largely determines the overall thickness of the skin.

1.3.3 Subcutaneous Layer

A primary role of the subcutaneous layer is the attachment of the skin to deeper structures in the body. The subcutaneous layer is loose connective tissue which stores fat unequally distributed and in abundant amounts in some regions of the body. Fat is absent, however, in the eyelids, penis, scrotum and nipple. The subcutaneous layer contains blood and lymph vessels, hair roots and the secretory portions of sweat glands.
1.3.4 Muscle

The muscle layer on the face controls facial expressions and the mastication movements of the jaw. In the regions of the masseter muscle and temporal muscle, figure 1.3, the muscle contributes to a large percentage of the soft tissue thickness. The muscle consists of fibres, or long cells, which are bundled together. They are embedded in a deep connective tissue layer. This connective tissue does not contain any fat and holds the muscles together, allowing for free movement.

Figure 1.3: Temporal and Masseter Muscles.
1.3.5 Relevance of soft tissue thickness measurements

A soft tissue model is of interest to a number of disciplines. The principle historical interest is that in medical diagnostics for the detection of tumours or other irregularities. Information about soft tissue thickness and their dependence on the underlying skeletal structure has major application in the areas of plastic and reconstructive surgery, forensic facial reconstruction, facial animation, and for military research agencies. Some information is available tabulating tissue thickness about the face. This existing data includes only about twenty positions on the face and can be identified through external or surface landmarks. Some scientists and surgeons in the above mentioned fields have stated that there is not enough thickness information available to support the potential work they could be doing. The following sections give a review of the way in which comprehensive data on soft tissue thickness would contribute to these research areas.

1.3.5.1 Plastic and Re-constructive Surgery.

A surgeon's concern in reconstructive surgery is that, as a final result, his patient should appear "normal" and have a high level of function. What constitutes normal facial form is an area that is still ambiguous. Facial form is dependent on both the skeletal structure and the soft tissue layer. But to what extent each defines facial form, and the possible dependence of one on the other is not yet fully specified.

In areas such as orthodontic and corrective surgery, it has been observed that the soft tissue facial profile changes with any skeletal alterations. The conditions and amount of change are still actively being researched. Nonetheless, this evidence exists to support the fact that facial form and tissue thickness is dependent on the underlying skeletal structure.

Re-constructive scientists rely on the works of anthropometrists like Leslie Farkas, who has established facial proportions of contemporary adults as in figure 1.4. This has become the basis of a surgeon's decision as to what is normal. The orthodontic and reconstructive surgeon can readily identify soft tissue characteristics that deviate from the normal proportions of the average face. In order to anticipate correctly how surgical changes to deformities affect facial form, there is a need for surgeons to have a clearer understanding of soft tissue thickness and the interaction with the skeletal structure.
1.3.5.2 Forensic Facial Reconstruction

The forensic field of facial reconstruction has been a main stimulus for measuring the limited data available, to date, on tissue thickness, out of the need to reproduce more accurately the identification of a person's facial form from skeletal remains.

An original technique of facial reconstruction was to build up manually muscle anatomy with clay on to a skull. The sculptor then added a covering or clay layer representing the fatty soft tissue. This was a highly subjective method only possible because of the artist's great skill and an anthropologist's input of the possible age, sex and race of the skull.

The use of tissue thickness data became very important in adding more quantitative information to the above procedure. This information was first available in the late 1800's when scientists produced tissue depth tables from cadavers. Rhine and Moore produced a table of thickness values by inserting pins into fresh cadavers. The depth of penetration against a bone surface was measured and published. However, these thickness values taken after death are not applicable to live models because of the changes in thickness data due to fluid dispersal or the release of interstitial tension. Most recently, B-scan ultrasound has been used to determine tissue thickness at a select number of points in American Caucasoid children. The data available is limited to thicknesses at bony landmarks at the same sites from which...
Kollman and Buchly and Rhine took their data, but the results were not directly comparable to the available data because the subjects were children. These sites only number twenty in total, and both Hodson and Rhine have stated that a larger sample of sites would be desirable.

A procedure for the forensic scientist is to attach pegs the length of the tissue thickness to its corresponding bony landmark. An example is shown in figure 1.5. Clay is then used to build up the soft tissue on the skull to the thickness at the bony landmarks. Although this method produces good similarities $^{38, 37, 41, 42}$, there is a large proportion of guess work in moulding the shape of the face between bony landmarks.

Figure 1.5: Forensic head reconstruction using pegs $^{37}$.

The latest generation of reconstruction scientists are using computers to "sculpt" analytically, allowing more people to take the role of the sculptor, without the artistic expertise. In this approach, a skull has pegs mounted at the bony landmarks sites, which as stated previously, only amounts to approximately twenty sites. The pegs have lengths comparable to the published thickness data available at these sites. The
skull and pegs are digitised into a computer by use of a video. The face reconstructor can, through a mathematical algorithm and with the aid of the computer, interpolate a tissue layer to the thickness pegs. In addition, the operator has the ability to add facial features from a database. This method is still very much dependent on the location and number of tissue thickness pegs, as with the manual method, the tissue growing algorithm is also very theoretical. Another technique used by Vanezis et. al has been to laser scan the skull and again "grow" the tissue thickness onto the skull. The resulting image is only the fundamental facial shape with no facial features. Vanezis has written that this method would be able to define the facial surface very accurately if it wasn't for lack of tissue thickness information, although he gives no indication on what his algorithms for growing the tissue are based. All of these techniques build up from skulls using unknown variables and for that reason are different from the methods used for this research.

1.3.5.3 Facial Animation

Work is being carried out in which a 3D representation of the face and its features are animated to create facial expression and speech. Within this field, one of the main areas of research is the generation of 3D facial models. Waters and Terzopoulos base their 3D face on the histology of facial tissues and the anatomy of facial muscle structure. The tissue model they use is tri-layer based on the cutaneous, subcutaneous and muscle layers. Each layer is represented as springs with different lengths and stiffness properties, assigned based on mechanical properties of the skin. The tissue thickness is not based on physical data collected from real models. Accurate and more comprehensive thickness data, as provided by this research, could only enhance the work being done in this area.
1.3.5.4 Defence Research Agency (DRA) Project

The previously mentioned DRA project is titled "A Database and Modelling Study for a New Generation of Flying Helmet, Oxygen Mask and Ancillary Equipment". The objective of that work was to create an interactive database 3D solid model design system based on a library of at least 300 head scans. Each head was to be able to be "interrogated" for the dimensions of selected points, sections and segments, areas and volumes. The individual head data could be used as a basis for a comparative anthropometric study. Aggregation of the heads could be done for the development of statistical head shapes. The techniques for aggregation are addressed in Chapter 5.

The subject of this thesis, i.e. the investigation and measurement of soft tissue geometry, has been a major and parallel part of the study to obtain data suitable for incorporation of a soft tissue layer in the database model.

A further soft tissue study, the investigation into the mechanical properties of the soft tissue layer as a whole, and of individual layers, has also been a major component of the DRA work at the University of Surrey. This was being accomplished using indentation and extension testing to record load-displacement data. Finite element work was to be developed using the soft tissue model and the mechanical properties information gained through this project.

1.4 Overview

In the preceding sections, it has been shown that anthropometry has become an integral part of engineering in the design of any tool, machine or clothing with which Man interacts. Soft tissue anatomy has been introduced and the hypothesis, that soft tissue thickness measurements can be combined with anthropometric data for the development of a sophisticated soft tissue model, has been suggested. It has been argued that a soft tissue model would be relevant to the fields of maxillofacial surgery, forensic reconstruction, animation, and this particular DRA project.

Chapter 2 discusses other published attempts to generate soft tissue models and provides a review of the anthropometric and anatomical data acquisition systems available. Chapter 3 describes the laser scanning system used for scanning volunteers' head shapes. Chapter 4, describes an ultrasound system, including the modification and validation work completed prior to acquiring the surface and thickness data. Chapter 5 discusses the method used to combine the surface and thickness data.
Chapter 6 presents the detailed acquisition protocol. Interpretation of the ultrasound trace and the extraction of soft tissue thickness data are discussed in Chapter 7. In Chapter 8, the thickness data is presented with an interpretation of the thickness data based on a proposed zonal approach. The validity of this zonal approach is considered as a method for improved soft tissue representation for the range of applications described. Soft tissue models generated on the CAD are also presented. Finally, Chapter 9 presents further ideas for continuation of this work and summarises its conclusions.
2.0 Review of Anthropometric and Anatomical Data Acquisition Methods

There is a restricted amount of literature describing the development of high resolution anthropometric head models containing anatomical information such as tissue thickness. Considerable work exists describing independently the progress of high resolution anatomical imaging tools in the field of medical diagnostics, and some work has been published on potential anthropometric use.

The following sections review the technology described in the literature, such as photogrammetry and laser scanning, used to acquire surface data. Methods for acquiring internal anatomical data are presented and an explanation of why A-scanning ultrasound was thought to be most appropriate for this work, is given.

2.1 Anatomical Models With Tissue Thickness Data

2.1.1 Invasive Methods

Dissection of the human body began in the Renaissance period. While it is an excellent means of examining anatomy, preservation and death can alter it, making dissection an inaccurate method as a basis for diagnostic information and in vivo databases. Dissection was shown to have been a very good tool for education purposes in many medical schools. However, anatomical education can now be had through human body visualisation software. For example, the ability to store large volumes of data has enabled documentation of the permanent imaging of an entire dissected human body through video and Computerised Tomography. At the University of California, Los Angeles a frozen cadaver was sectioned and videoed every 200μm. The 1024 x 1024 pixel camera images were digitised and reconstructed into an anatomically complete image that allows viewing of regional anatomy of complex structures. Although this technique results in accurate information, it is not an available method for modelling the soft tissue information of live subjects. 46.
2.1.2 Computer Aided Tomography

Currently, one popular method of imaging anatomical information is through the use of computer aided tomography, or CT scanning. It is used generally as a diagnostic tool for the detection of tumour and other anatomical disorders. It is used as a surgical aide for maxillo-facial reconstructive surgery and for the construction of educational anatomical atlases. CT scanning can also produce low level images of soft tissue.

CT scans are generated by the transmission of a beam consisting of a fan-shaped planar array of X-rays through a section of the body. The amount of X-ray that penetrates the section, and hence the attenuation, is measured by an array of detectors mounted opposite to the radiators. Each type of tissue allows varying amounts of x-ray penetration. The resulting attenuation is derived from an "attenuation coefficient" assigned to each tissue. The overall attenuation of the X-ray beam is equal to the summation of the attenuation of each component of the array. In order to acquire a 2D picture, in figure 2.1, the apparatus is rotated axially around the body. For one scan, several hundred views could be taken during 1-15 seconds. High level computation is needed to solve the many transmission equations and thus to derive radiological density distribution through each section.

It is important for a CT scanning system to have certain characteristics for successful imaging of the internal anatomical information. To acquired enough data for an adequate image of the head - approximate one hundred slices at 2mm intervals - the procedure often takes thirty minutes of radiation exposure to the subject. The obtainable section-section resolution with CT scans is 0.3mm. However, at a 0.3mm increment, it would take 1000 slices to image the head, and as patient comfort and movement must also be considered (patient movement degrades the computed image significantly), slices are usually 2-4mm apart. This separation has been shown to be sufficient in imaging the skull for maxillo-facial surgery.

The image reconstruction software uses the scanned data to produce an image from each section. The image is divided into an array of elements where each element is given a numerical value mathematically calculated from all of the data taken from the individual rotated views. This numerical value is called the CT number and is related to the apparent radiological density of the tissue. To convert the image into a video display, digital to analogue conversion is used, where each CT number corresponds to different shades of grey, the whitest areas representing bone.
The essential problem in reconstruction the slices into an image is segmenting out various anatomical details. Each slice is a grey-scale picture and tissue boundaries between each anatomical feature are not easily identified. This is a result of low contrast between tissue densities. Ideally, the boundaries between bone and tissue or bone and air could be easily identified. However, there are systematic problems such as artefacts and scatter which can cause ambiguity at the boundaries. Entire research teams devote their energies into developing the best method for automatically detecting these boundaries so that anatomical information can be more distinctly identified.

While CT scanning is an excellent imaging tool, it is primarily useful for establishing the bone structure because the high attenuation levels result in better quality images. The reverse is true for soft tissue, and therefore, CT scanning is not the best imaging tool for looking at structure and possible layer detail.

### 2.1.3 Magnetic Resonance Imaging

Magnetic Resonance Imaging (MRI) is a procedure which can be used to image soft tissue distribution because of the presence of hydrogen. A MRI system transmits a strong magnetic field, 1-2 Tesla (1 Tesla = 104 gauss) through a subject. Figure 2.2 shows a typical MRI scanner and a schematic of the process. Within the magnetic field, hydrogen atoms in the tissue contain protons which are constantly spinning about their axis, aligned in the direction of the magnetic field. By transmitting Radio Frequency currents from a coil within the magnet, at the resonant frequency of the
spinning proton, the protons can be driven into an excited state where their axis flip away from the direction of the magnetic field. In an excited state, the proton's energy increases. This energy is eventually transferred away from the nucleus so that the proton can realign itself with the magnetic field direction. The time it takes for the proton to transfer its energy and realign itself is known as relaxation, and is one of the parameters measured. The second parameter measured by MRI is the radio frequency at which the protons are resonating. The quantity of radio frequency received is indicative of the tissue type because of the amount of mobile water, or hydrogen content in the tissue. Spatial information of the proton is known by applying field strength gradients across the magnetic field to alternatively excited specific areas.
The resulting MRI image in figure 2.3 is a typical slice through the area of a head being imaged. To reconstruct this image, the slice was originally divided into rows and columns during the imaging, to form a matrix of tissue "voxels" or volume elements. The information stored in each voxel consisted of the frequency and phase data from the radio frequency recorded at that voxel. Fourier transformations are used to sort the radio frequency and phase signals into signal intensity and spatial data. When reconstructed into figure 2.3, the signal intensity and spatial data are displayed as an array of pixels and brightness. A signal of higher intensity appears brighter on the image, resulting in a grey-scale picture similar to CT scans. However, the highlighted anatomical features are the soft tissues, i.e. the brain or soft tissue layers of the skin.

Many of the same problems exist in MRI scanning that exist in CT scanning. As discussed in section 2.1.2, edge detection is ambiguous in segmenting anatomical features. Additionally, the time periods for imaging enough data for successful reconstruction are similar, and although there are not the worries of radiation exposure with this technique, patient movement can significantly affect the results. The presence of metal in tissue causes artefacts that result in the loss of data on the final image.

Figure 2.3: Reconstructed MRI scan.
The design of new surface gradients and coils in the late 1980's has resulted in the increasing resolution of MRI systems. This has not eliminated the boundary detection problems, but has increased the detail the MRI systems can 'see'. The French team of Querleut et. al. from the cosmetic laboratories of L'OREAL have used MRI to characterise the skin layers of the dermis and epidermis by establishing water content in relation to ageing. 62, 63.

Hohne et. al. (1992) have recognised the merits in combining the results from CT and MRI scans to produce a detailed soft tissue and bone anatomical image. They have produced an anatomical atlas of the head for teaching purposes. 64, 51 An anatomic atlas has also been done at the University of Colorado at Denver under the sponsorship of the National Library of Medicine. 65. Their anatomical reconstruction was based on a series of MRI and CT scans at 1-4mm intervals. It is available on the internet and is titled "Visible Human Project". This type of work has been termed "multimodality registration", and is another area of research within the medical imaging field 49, 66. One of the challenge of multimodality registration is determining the best technique for referencing different data sets together 67-69. The ability to reference data sets together is critical also to the work done for this thesis and will be discussed further in Chapter 5.

2.1.4 Ultrasound

Ultrasound is used in medical diagnostics for determining internal body structure. Ultrasound is also used to determine the existence of disease through detailed qualitative assessment of the data produced by the interaction of the ultrasound with the soft tissue. Over the past twenty years the work of R. Chivers, J. Gore and C. Hill has been instrumental in developing "tissue characterisation" through ultrasonics. At the outset, it was of interest to this work to distinguish tissue layers with the principles of tissue characterisation that had been developed. It became apparent that to apply the work developed by Chivers, Gore and Hill, among others, was an immense project in its own and beyond the limitations of the equipment that had been purchased. This is explained further in Chapter 4. It is not considered relevant to this work to critically review the research done in the area of tissue characterisation through ultrasound. Further reading can be found, in general, in the journals Ultrasound in Medicine and Biology, Physics in Medicine and Biology, Ultrasonics, and the Journal of Clinical Ultrasonics, and the following specific references:

- Echo Structure in Medical Ultrasonic Pulse-Echo Scanning 70.
The use of ultrasound, specifically A-scanning ultrasound, is a key feature in the work described in this thesis and the relevant theory will be discussed in detail further in Chapter 4. However, a brief introduction to ultrasound follows, together with a review of its application in anatomical imaging of soft tissue.

Ultrasound is a sound wave with a frequency above 18kHz. When this sound wave is propagated through soft tissue, usually through a gel or water bath medium at the skin, it can be reflected at the boundaries of tissue layers. The amount of sound reflected is dependent on the difference in the tissue density and the related speeds of sound through the tissues at any particular boundary. In A-scanning ultrasound, the reflected boundary information is displayed as a series of echoes in a signal on an oscilloscope as demonstrated in figure 2.4. Extracting anatomical information from an A-scan trace, in the form of distances between tissue layers, is done by examination of the signal to locate the echoes corresponding to tissue boundary layers and determining the time between them. Tissue thicknesses can be calculated by taking the product of the transit time and the average speed of sound through the relevant soft tissue.

In B-scanning, the reflected signals, or echoes, are recorded relative to a reference geometry either via hardware displacement or via fixed arrays, giving each echo a one dimensional reference co-ordinate. These are displayed on a cathode ray tube as a series of dots as in figure 2.5. The intensity of each dot is proportional to the amplitude of the returning echo signal. B-scanning ultrasound can result in a grey-scaled image superficially similar to CT and MRI images. The key difference is in what the data the grey-scaling represents.
For instance in B-scanning, as previously described, the grey-scaling is representative of the voltage amplitudes of the echoes which occur at reflecting surfaces. The amplitude is a function of the changes in tissue density at the boundary, and attenuation of the signal. The grey-scaling in CT scans represents the CT number which is a function of the radiological density of the tissue and is not boundary dependent. For an MRI scan, the grey-scaling represents the intensity of the radiological signal which is a function of the frequency and phase data of the radio frequency signal received from the tissue. This is indicative of the amount of hydrogen in the tissue.

Figure 2.4: A-scan trace

Figure 2.5: B-scan image
"Electronic callipers" are used to extract anatomical information from the image. A calliper is a cursor which can be moved around the screen and fixed at a screen pixel. It can be difficult to fix the calliper correctly at a tissue boundary because of the pixel size of the B-scan image, in that larger pixels may contain more information than just the boundary position. Also, signals with high amplitudes resulting from strong reflections at tissue boundaries can appear as very bright dots on the CRT screen, and these may hide information about the surrounding signal. Again, this would impede the correct placement of the cursor. Depending on the resolution of the image, these problems can result in significant errors.

Both A-scanning and B-scanning have been used successfully since the mid 1950's to image the superficial soft tissue layers. B-scanning has become more popular with the improvements in equipment capabilities. In this review, A-scanning and B-scanning, which have both been used to determine soft tissue structure and characterisation, are discussed separately for easier comparison. It is recommended the reader review Section 4.1 first if the concepts and terminology of ultrasound are not familiar.

2.1.4.1 A-scanning

Most literature credits the work of Alexander and Miller as a landmark for the use of A-scanning ultrasound to examine skin thickness in vivo. However, Alexander & Miller cite the work of Daly & Wheeler and their use of A-scanning to measure oral soft tissue thickness, as the predecessor to their own work. Investigation of the literature further back in time shows that the first reported use of A-scanning on biological tissue seems to be the work of Wild and Reid in 1952, who used the ultrasound in vitro on a dog's bowel to explore the technique as a viable tool to measure soft tissue thickness. Their set-up was to use a 15MHz transducer coupled with water contained in a rubber membrane to the dog's bowel tissue. They did not publish any thickness value, but showed they could see tissue boundaries and encourage further investigation into explaining why reflections arose at the boundaries.

The mechanics of tissue interaction with ultrasound were understood further by 1971 when Strakova and Markova used A-scanning to determine subcutaneous fat thickness. With a 4MHz transducer, they were able to distinguish the skin/subcutaneous and subcutaneous/muscle boundaries on 400 children. Comparing their results to calliper readings at the seven sites on the body showed that 60% were not significantly different, and that agreement between the two results was better at the
sites which had a thinner subcutaneous layer. This was because at very fatty sites, the skinfold layer can not be lifted, making the callipers difficult to use.

Alexander and Miller used a 15MHz transducer (of the order of 1000 times higher than the audible range) with a reflecting surface resolution in soft tissue of approximately 0.05mm. The signals were displayed on a CRT and the transit time was determined. The speed of sound value used was said to be taken from Daly and Wheeler as 1580m/s. Consulting the Daly and Wheeler paper shows that they had actually used a speed of sound value of 1518m/s. Either the value of 1580m/s was a typographical error in the Alexander and Miller paper or they used the wrong value in their calculations. Alexander and Miller compared their results to X-ray measurements taken at the same site. The results correlated at a 95% confidence level, with seven of the ten ultrasound values lower by 8%. If Alexander and Miller had used the accepted higher value of 1640m/s for skin, the results would have been comparable to the X-rays results. The remaining difference was more likely to be due to the difficulties in extracting exact thickness boundaries from the grey-scaled X-ray.

The above work had shown that it was possible to accurately use ultrasound to distinguish soft tissue boundaries. The work of Tan et. al. (1982) showed that ultrasound measurements were repeatable and valid as a measure of true skin thickness. The ultrasound equipment they used was similar to Alexander and Miller as described above. To test reproducibility, two technicians independently measured skin thickness at one site on the flexor side of the forearm on both arms, using pulsed ultrasound. They took five readings from each arm of twenty subjects. The mean of the five readings were plotted for each technician. A best fit line was drawn through the points and the results correlated to r=0.88. The coefficients of variations of the five readings were found to be 6.9% for one technician and 5.4% for the other. The thickness measurements taken on any one subject by each technician were not statistically significant at 1.0-1.7%. They were able to conclude that ultrasound is "highly reproducible" and therefore, the technique was used in confidence.

Tan et. al. wanted to determine that ultrasound was measuring the true thickness of skin, and was a better tool than X-rays or punch biopsies. They measured the thickness in vivo on the flexor side of the forearm of twelve people using ultrasound. After the ultrasound, punch biopsies were taken at the same sites and prepared for measurement with a microscope. The results showed that the in vitro thickness measurements were 76-88% higher than the in vivo ultrasound measurements. They repeated a slightly
different test on seven individuals at the following sites: four from the forearm, two from the upper arm and one from the leg. First they took in vivo measurements using ultrasound and X-rays. They then took punch biopsies of 4mm diameter of the sites and measured them in vitro with ultrasound. The in vitro measurements were 46% larger than the in vivo measurements taken by ultrasound and 28% larger than the X-ray measurements. The large difference between the biopsy and the ultrasound and X-ray was due to the loss of resting dermal tension when the skin was excised. Also, the dermis is distorted through the punch biopsy procedure. Hence, biopsy thickness measurements are not useful when measuring thickness values in vivo.

The ultrasound measurements of Tan et. al. were lower than the X-ray measurements by 18%. This discrepancy was caused partially by use of the wrong speed of sound value for skin as the set up was identical to Alexander and Miller.

Others have validated the use of ultrasound against skinfold measurements of subcutaneous adipose tissue. Borkan et. al. 146 claimed there existed a significant difference between the ultrasound and skinfold measurements. They admitted they couldn't be sure which was more accurate. They did, however, prefer the skinfold technique because it was not necessary to have such an in-depth knowledge of the underlying anatomy. Because they were experienced in using the skinfold technique, they found it to be more repeatable. They admitted the disadvantages for using skinfold techniques were that the adipose tissue could be compressed resulting in thinner measurements. Also it was only possible to use the technique at certain sites, and not at all on the obese. This author has some doubt about the ultrasound device Borkan et. al. used, called the Body Composition Meter, comprised of a 3.5MHz transducer. While the frequency used was low enough for adequate penetration of the ultrasound beam to the subcutaneous layers, the form of readout, i.e. the thickness display, consisted of LED lights which were not stable. Borkan et. al. questioned the reliability of the readout themselves.

Certainly the more accurate approach would have been to extract the thickness values from the A-scan trace just as Jones et. al. 147 did later in their attempts to validate ultrasound against their in-house designed skinfold callipers. They used a 5MHz transducer and ensured the ultrasound probe did not compress the tissue by enclosing the transducer in a 5cm surround, which more evenly distributed any applied pressure. Ultrasound measurements were validated by taking measurements on a cadaver and then dissecting the site to obtain a depth gage measurement. The results were satisfactory. Comparing the further ultrasound measurements to skinfold
measurements resulted in a correlation of $r=0.99$. They advised the ultrasound was a valid technique to use and was more comfortable to the patient.

Even today, it is difficult to validate perfectly any tool which measures skin or soft tissue thickness in vivo. Tan et. al. have shown that ultrasound is repeatable and consistent. Jones et. al. showed ultrasound was valid against skinfold techniques. An ultrasound trace shows reflections at the boundaries from which thickness measurements are then calculated, whereas, in X-ray images the grey-level represents radiological properties of the tissue. It is practical to conclude that to measure thickness of tissue layers, ultrasound should be used. The parameter that determines its accuracy is the speed of sound value. Ultrasound has become widespread in applications for which the knowledge of skin or soft tissue thickness is relevant, such as the diagnosis of skin disease.

Assessment of the nature of small skin tumours and disease has been attempted using A-scan ultrasound. It is possible in principle to detect abnormalities in the skin because the echoes arising from tissue boundaries are a ratio of the product of the density and sound in the tissue. This is described more in Chapter 4. Changes in the nature of tissue affects the characteristic echoes of the signal. Edwards et. al. used A-scan to monitor changes in composition and structure of tumorous tissue through changes in echo amplitude and spacing in the signal. The apparatus comprised of the Dermal Monitor 70 by Cutech which has been also used in the work of this project. The transducer centre frequency was 18MHz with a reflecting surface spatial resolution of approximately 50µm. Higher frequencies with shorter pulse widths allow better resolution when examining the superficial soft tissue layers. It is not uncommon to see centre frequencies around 100MHz being used for investigation of the skin. However, this increased resolution is compromised by a shorter scan depth. In Edwards' work, the transducer was coupled to the skin using a water bath. A
perspex cone of known geometry was used as the probe tip to ensure accurate location of the ultrasound beam just under the skin surface. Using the same configuration on the work for this thesis, it was found that the configuration of our ultrasound equipment wasn't accurately placing the focus of the sound beam within the tissue because the focal length was shorter than the length of the perspex cone. This was thought to be unique to our system. As a result, new probes were designed for this project. Section 4.3 discusses these modifications. Nonetheless, the manufacturer's configuration was appropriate for Edwards et. al. to conclude with the Cutech probe that echo amplitudes are a significant indicator of the types of lesions that can be found in skin.

Another skin diagnostic application was the work done by Mourad and Marks (1990). They used Cutech system and an identical configuration of the work of Edwards 87 to predict the severity of gravitational syndrome, by the measurement of skin thickness. A symptom of Gravitational syndrome is the thickening of the skin. By measuring skin thickness, they were able to successfully monitor the different stages of the disease.

A-scanning has been used in research to determine how different creams and coricosteroids affect the changes in skin thickness over time. Serup 88, 89 used a 15MHz centre frequency probe and a speed of sound value of 1518m/s to determine that skin thickness decreased by 21% on the forearm and 11% on the back with the application of coricosteroids. However, his results are questionable due to the low speed of sound value used in his calculations. A higher value would alter his findings by about 8%.

Wells and Mountford 90, 91 have used A-scanning for another application different from measuring skin thickness. In the mid 1970's they were able to establish that A-scanning could be used to determine the level of cirrhosis in the liver by the characteristics of the echoes in the signal. Ultrasound penetration in the body is inversely proportional to the centre frequency used. So to examine structure deeper within the body, such as the liver, lower frequencies must used. Wells and Mountford used a centre frequency of 1-5MHz. They concluded that echo amplitudes increased with the level of cirrhosis.

A-scanning is not a complicated technique to use and therefore, there isn't much one can be critical about when examining the various applications, especially of those determining skin thickness. The two critical factors dictate A-scan performance. The first is the centre frequency of the probe used. This affects the pulse width which

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determines the resolution achievable. So higher frequency ultrasound is capable of better resolution. The other debatable factor is the speed of sound used. The speed of sound is a function of the elasticity and density of the tissue through which it travels. Table 2.1 indicates the accepted values for the speed of sound in the different components of soft tissue \(^{92,70}\).

<table>
<thead>
<tr>
<th>Material</th>
<th>Speed of Sound (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skin</td>
<td>1640</td>
</tr>
<tr>
<td>Fat</td>
<td>1480</td>
</tr>
<tr>
<td>Muscle</td>
<td>1580</td>
</tr>
<tr>
<td>Blood</td>
<td>1570</td>
</tr>
<tr>
<td>Average</td>
<td>1540</td>
</tr>
</tbody>
</table>

Table 2.1: Values for speed of sound through soft tissues.

The speed of sound values that have been used in past work in determining skin or soft tissue thickness values have varies around 1520-1580m/s. It seems that much of the work that progressed from Daly (1971) and his use of A-scan on oral soft tissue has used the speed of sound quoted by him of 1518 m/s. Alexander and Miller (1979) cite Daly's figure as the value they use, but then used the contradictory value of 1580 m/s, which is more appropriate for skin. Because of this error, further published work has also cited 1580 m/s as the correct average value \(^{85}\). In 1989 and 1990 Marks was co-author on papers determining thickness in skin for soft tissue tumour assessment and in each year the values of 1580 m/s and 1518 m/s respectively, were cited. The confusion seems to be that some research, when calculating thickness for soft tissue, have used a value more appropriate for skin, and vice versa.

Gore has stated that by using the average value of 1540m/s, an error results of 1-2% in the thickness value for soft tissue. Around the head, this is an error of the soft tissue thickness of 0.035-.25mm. This work has used the value of 1540m/s as a standard value. Perhaps, changing this value to be weighted according to the tissue structure of any area would produce more accurate results. However, determining for each site what the value should be, is a major undertaking and not part of the scope of this immediate work, although could be part of further work on this project.

None of the work mentioned above used A-scanning to look at the entire soft tissue layer, including the muscle, to determine thickness. This is not true for B-scanning. It
has become more popular because of its imaging ability and therefore, has been used for more applications, including determining soft tissue thickness. The next section reviews some applications of B-scanning.

2.1.4.2 B-scanning

In the late 1970's B-scanning was also being used to image the soft tissue layer, specifically the skin histology, to attempt to predict disease severity by skin thickness that change as the diseases progress. The literature cites Rukavina and Mohar (1979) the first study to demonstrate that skin lesions and disease could be located using B-scan ultrasound. They used a 2MHz transducer which results in an approximate resolution of 3mm. They were able to distinguish tumours of sizes 2-9cm on different patients. There is one discrepancy in their work in that when discussing the existence of the tumours, they identify the superficial side of the tumour as being located in the skin layer. However with a resolution of 3mm, it is probable that in some regions of the body, they couldn't image the skin layers which are typically less than 3mm in thickness.

Payne (1981) realised that better resolution was needed of the ultrasonic transducer to be able to accurately image the skin and it's individual layers. He introduced the use of a new plastic film transducer, and with the help of Alexander and Miller, was the first to possibly image the epidermis and most certainly the dermis and subcutaneous layers.

In the same year, Cole et. al. used a 10MHz transducer with 0.5mm axial resolution to image the epidermis and dermis layers of the skin on the upper back. Their interpretation of the B-scanning image was that the epidermis appeared as a strongly reflective layer, the dermis was a "homogeneous medium grey tone", the fat was echoluscent and the muscle, coarsely echogenic. They published thickness values of 1-3mm for the epidermis and 3.3 ± 0.6mm for the dermis. From the example B-scanned image in their published work, it is difficult to see exactly how they identified the boundaries between the tissue layers. The epidermis is indeed very bright indicating a strong reflective layer which could be "hiding" the epidermis/dermis boundary. This can be a problem with B-scanned images as indicated earlier in section 2.1.4. The other boundaries layers in the example given, are very difficult to determine because of the grey level changes.
Hodson et al. (1985) reported work that is more directly relevant to the work of the project. They determined soft tissue thickness on Caucasoid children at 20 sites. Using a 7.5MHz probe with a 0.6mm wavelength, they scanned the children while they were lying on their backs with their heads stabilised. 22. The B-scans were stored and the films were produced from which they extracted the thickness using callipers that were accurate to 0.01mm. The emphasis of their work was to determine if measurements on boys were significantly different than girls. The only significant difference was the location just under the nose, point 5 in figure 2.6. They did not quote any errors, nor did they show examples of the B-scan images so it is difficult to ascertain with what accuracy they were able to extract the thickness. This author's experience with extracting thickness from MRI films which are similar to the B-scan grey scale picture, is that the varying levels of grey make identification of boundaries extremely ambiguous, even the soft tissue/bone interface. Additionally, it is possible that the protocol they used by having the child in a prone positioned, introduced thickness changes from an erect position around the face. It might be expected that areas around points 15, 16, 19, and 20 would appear thicker and points 5-10 thinner.

As in A-scanning, there are many examples of using B-scanning to determine skin thickness; more recently to determine the affects creams and cosmetics have on the skin 98. It is debatable whether B-scanning is the better tool, over A-scanning, to examine skin thickness. B-scanning definitely aids in visualisation, but the problem remains of accurately determining the boundary layer because of the grey tones are brightness of signals.

Figure 2.6: Diagram of thickness points taken by Hodson (1985) 22.
The above mentioned methods of CT, MRI and B-scanning for acquisition of anatomical data are well-developed, reliable techniques. Nonetheless, their use in this research was eliminated for a variety of reasons. The goal of this project was to not only develop a soft tissue shell, but acquire significant data on soft tissue thickness and possible tissue layer information around the head for a general applications as mentioned in Chapter 1. The discussed anatomical acquisition methods result in grey-scaled images. Soft tissue thickness, and especially tissue layer thickness, are difficult to process out because of indiscreet boundaries as mentioned previously.

CT scanning is an invasive technique which in large doses, can be dangerous to subjects. MRI is not invasive, but the acquisition times can be long depending on the quantity of data needed. The costs of using these devices and the software needed to process the data are high. While large companies and government bodies would afford these costs, it was a consideration of this work to develop a procedure involving tools which smaller companies and research groups could afford.

For these reasons, A-scan ultrasound was chosen as the tool for this research. With A-scan, the approach is different from the other techniques in that raw tissue soft tissue thickness data is taken and built up into a soft tissue shell, not processed out of an existing image. The extraction of the thickness is quick and reliable with better resolution than achievable by other techniques. Most importantly, it is a more accurate tool for distinguishing tissue layers, because boundary reflections are often clearly seen in the reflected ultrasound signal. Chapter 4 introduces the concept of A-scan and the system used for this work.

2.2 Anthropometric Data Acquisition Methods

Following the introduction to the methods available for the acquisition of anatomical data, it is now appropriate to discuss the methods available for acquiring anthropometric data

2.2.1 Classical Anthropometric Techniques

Classical or direct measurement techniques were the first means of acquiring anthropometric data on the human body and head. The tools developed by research
groups, used to measure straight line distances between anatomical features and estimated circumferences were basic. They were standardised in 1967 when the terminology, approach and technique to be used when acquiring anthropomometric data for health and engineering applications were defined. Using these standards, tools were developed in the 1960's and 1970's which became common to most anthropometric surveys. These tools, still in use today, are:

**Anthropometer:** This instrument, 'A' in figure 2.7, is similar to an engineering Vernier Callipers, only much larger, comprising a sectional rod with one fixed and one sliding edge, extendible over the length of the rod. This instrument is usually calibrated in millimetres and is used to measure distances up to two metres.

**Sliding Compass:** Instrument 'B' in figure 2.7 is a smaller version of the Anthropometer. It is more convenient to use to measure distances up to 300mm,

**Spreading Calliper:** In some regions of the body, especially around the head, protruding features make it difficult to align the straight edges of the standard anthropometers to measure distances. A spreading calliper, instrument 'C' in figure 2.7 has bowed arms which do not interfere with the obstacles mentioned. It can also be used to obtain measurements in re-entrant areas such as the ears.

**Tape:** Cloth tape measures are useful for circumferences about the head and body.

Figure 2.7: Classical anthropometric tools.
The DRA reports in their 500 Head Survey that these instruments produce accurate results to 1mm. Roebuck adds that for the anthropometer, the precision is usually only 5mm over larger measured distances. Although the instruments are easy to use, they require a skilled technician to produce repeatable and accurate lengths of the anatomical distance. The technician requires extensive knowledge of anatomy and the position of bony landmarks. Currently, when the anthropometric tools described later in this section can obtain repeatable precision to 0.5mm with little operator training, the use of these tools is diminishing.

The above anthropometric tools limited the scientist's ability to gather and quantify information about curvatures and shape. In attempts to define shape, various ingenious methods were used and for a while, two available methods dominated the field.

1. The multiple probe method, illustrated in figure 2.8, provided direct measures of body contour. The apparatus consisted of three, approximate 1.5 metre high walls connected at right angles to each other. Each wall has a grid of holes even spaced over the range of distances equivalent to body length and breadth. Graduated rods to one millimetre were placed in the holes, parallel to each other. The subject would sit in the box while the operator pushed the rods gently until they touched the person's body. The operator would record the rod number and the protruded amount of the rod to contact with the subject. Each rod number had a known X and Z position and the protrusion gave the Y so that a three dimension reconstruction could be done. This method was extremely time-consuming as each measurement had to be carefully done and immediately recorded. Sources of error came mainly from wrongly positioned rods and more significantly, reading errors off from grids of more than 20 rods. Subject movement was another source of error. This instrument was mainly used on the torso, so that respiratory breathing, which can cause the chest to move by 2-3 mm, would change the positions of the rods.
2. The method of casting is recognised as being an accurate permanent record of the body contours by using plaster of paris to physically record body surfaces. Since the plaster of paris moulds directly to the subject's skin, the resulting hardened surface is a very close replica of the surface moulded and is suitable for further examinations and measurement. This method can be useful if the objective is to produce clothing or protective gear which can be directly moulded from the caste. It is not a good method from which to take measurements later on because of the absence of landmarks and numerical data. Applying casting materials to the subject is also messy and uncomfortable, and the time requirements are high. Temperature changes, both heating and cooling, may affect the dimensional accuracy of the cast.

Generally, the problems are numerous when acquiring anthropometric data by direct methods. The data is usually taken at one sitting by one operator to maximise repeatability \(^{101}\). However, during these sittings, the subjects may suffer from fatigue, tremors and changes in muscle tone. These occurrences, along with breathing, can significantly alter the precision of the results.

Direct anthropometric methods, with the exception of casting, do not result in a permanent record of the subjects' sizes and shapes. If questions arise about the quality of the data, it is not a trivial task to re-check a measurement or take another measurement that was missed the first time.
Logically, scientists have developed anthropometric techniques involving photography to enable retention of permanent records. These techniques have been evolving since the advent of photography. They gradually began to replace direct techniques in the late 1960's and early 1970's. Although direct techniques are still used in some instances, modern photographic and video techniques have become the modern-day anthropometric acquisition tool. The next section reviews the evolution of photographic-based techniques and introduces laser scanning, the latest anthropometric acquisition tool available.

### 2.2.2 Photogrammetry

Photography had been used to attain records of the qualitative characteristics of people since its invention in the mid 1800's. In 1940, Sheldon published his work on somotyping which included photographs to support his classification of body sizes. In 1946, Tanner and Weiner adopted the photographic technique Sheldon used to determine body dimensions, standardised it and showed that photography could be used to the same accuracy as classical anthropometry methods. This was the start of widespread use of the camera as an anthropometric tool.

The original photogrammetric methods involved the inclusion of a grid in the photograph with the subject. This grid, placed either in front or behind the subject, was used to extract simple dimensions like the head breadth or nose width. This technique enhances the optical problem of parallax as demonstrated in figure 2.9.

![Figure 2.9: Parallax in photogrammetry.](image-url)
In the above configuration, \( x \) is the distance of the maximum diameter of the circle. This is the true dimension that an anthropometrist would want to measure. However, the camera only sees the distance \( x' \) because of the curvature of the head. When measuring the apparent head diameter from the grid in the photograph, the distance actually begin measured is \( a \). The parallax error \( P \) is defined as \( a-x' \). Most of the error comes from the apparent displacement of the head on the grid and not the error caused by curvatures. To eliminate the major component, the grid can be placed in the same plane as the point being measured, or superimposed on the head with light projectors.

In the work by Lovesey et al. \(^{100}\), they managed to eliminate both of the factors contributing to parallax. Referring again to figure 2.9, they increased the distance \( d \) using a telephoto lens, so that the curvature of the head no longer affected the camera's ability to see the point of maximum diameter. They also projected a fine colour grid of five stripes of 2mm width onto the face. A grid distorts when projected onto the face because of the curvatures. Using the standard black and white grids made it difficult to distinguish between grid marks in areas where the lines would converge, i.e. around the eye. Using colour allowed them to see the individual grid lines in areas of gross convergence. This technique resulted in rms\(^1\) errors of 0.66mm. The procedure was limited to use on the face and was not used for a significant anthropometric survey of any sort, making it difficult to surmise acquisition and interpretation times.

In 1989, Abeysekera and Shahnavaz attempted to model the anthropometric data obtained from the human head of Buddhist Monks using photogrammetry. \(^{103}\). Their procedure was to take four photographs of a subject, e.g. front, side, back and top of head. They took some measurements from a points on the head to determine the scaling factors between the actual and photographic measurements. From the photographs, they took 80 measurements to determine the minimums and maximums and then drew the profiles of each of four views. The profiles were used by a sculpture to model a head. From the model they determined that 80% of the measurements were accurate to \( \pm 2.5 \)mm and the error was more on the other 20%. Within the report they make no mention of the problems of parallax as discussed previously. Their scaling method would not correct for parallax consistently for every point on the head as the position of each point will result in a variable parallax error.

\(^{1}\)Root Mean Square (rms) value denotes the measured value as the arithmatic mean of the squares of the accuracy in each of the orthogonal directions, ie. \( \sqrt{\frac{x^2+y^2+z^2}{3}} \)
Another identifiable large source of error would probably come from the lack of a referencing method for the four views. The photographs from the four views were not all taken simultaneously. From experience on this project and others as reviewed in Chapter 5, the error in repeating the same head position can be significant. It doesn't appear that they had a reliable method for doing this between photos. Therefore, developing a 3D head model from the four photographs cannot be valid.

Acquisition of 3D data of the entire head from photogrammetry is not usually easy to achieve. This technique provided permanent records of the head (usually just the face), but the problems of parallax and the inability to extract accurate data did not make advantageous over the classical techniques. The development of stereophotogrammetry eliminated these problems, and has replaced standard photogrammetry as an anthropometric acquisition tool.

2.2.3 Stereophotogrammetry

The concept of stereophotogrammetry was developed before photography was ever invented. In the late 1830's, Sir Thomas Wheatstone developed the stereoscope which allowed viewing of two pictures, drawn from slightly different perspectives, to be viewed simultaneously, resulting in what seemed like a 3D image. He wrote to the Royal Society in 1838, confirming his work. Shortly after photography was invented in 1841, Wheatstone commissioned a photographer to take two photographs, that could be viewed in the stereoscope. This was successful, however photography was in it's infant stages so emulsions were slow and costs were high. The technique lay dormant for years until the start of the 1900's when "modern" stereophotogrammetry came into use, mainly to analyse the contours of terrain.\textsuperscript{104}

Stereophotogrammetry is based on human binocular vision as is demonstrated in figure 2.10.
Photographs are taken with a pair of stereo cameras. Light rays from the object e.g. points A and B, pass through the lens and intersect the emulsion plate where the image is recorded. To reconstruct a three dimensional image from the photographs, a stereoplotter is used. The overlapping photographs are placed in the stereoplotter and light rays project through the photographs to form a three dimensional image in space. A visual mark was then positioned on the virtual three dimensional image to extract the co-ordinate data. In the more recent systems, the photographs have been replaced by CCD cameras and reconstruction of the image involves a different technique which is discussed later in this section. Specific systems will be reviewed after a more detailed description of the stereoplotter is given to show that whereas the initial photogrammetric systems were always attempting to eliminate the parallax error for accurate results, the stereoplotter utilised the parallax error to calculate the depth dimension for the generation of the three dimensional image.

Referring to figure 2.10, in a stereophotogrammetric set up, the following geometric information is recorded to enable the extraction of the three dimensional co-ordinates.
1. The orientation of the central axis, \( \mathbf{OP} \) and \( \mathbf{O'P'} \), of the camera.
2. The length of the base the cameras are mounted on, \( B \), which is also the distance between the central axis of the cameras.
3. The position of the principle point on the emulsion plate, i.e. \( \mathbf{P} \) and \( \mathbf{P'} \). This is defined as the point where a light beam coincident with the central axis of the camera hits the emulsion plate.
4. The principle distance of the camera, i.e. the distance \( \mathbf{OP} \) and \( \mathbf{O'P'} \).
5. The position of control or reference points which are included in the photographs.

The above parameters are used to align the photographs in the stereoplotter. To measure the position of any two points, e.g. \( \mathbf{A} \) and \( \mathbf{B} \), the visual marker is first used to measure the parallax between the two photographs, of each point. In figure 2.10:

\[
\Delta W = W_2 - W_1
\]

Using equivalent triangles, the depth, \( \Delta D \) can be calculated through the following equations:

\[
\Delta D = \frac{\Delta W (D - \Delta D)D}{B(OP)}
\]

This can be rewritten to solve for \( \Delta D \) as:

\[
\Delta D = D \left( \frac{1}{\frac{B(OP)}{\Delta W* D} + 1} \right)
\]

When stereophotogrammetry was being established as a reliable anthropometric tool during the 1960's, the special stereometric equipment that was needed for the task was high in cost and demand. Burke and Beard (1967) attempted to combine the cameras and stereoplotter into one device. The intention was to cut down on expense and eliminate the problem of setting up the stereoplotting equipment to the same geometry of the cameras, which had been identified as a source of error. 102. The cameras were mounted over a subject who was lying down. Removable adapters allowed the
equipment to take the photos and then reconstruct the image by passing light rays through in the opposite direction. Their system geometry was restricted so that only part of the face could be seen. Errors were in the order of 2-4 millimetres in the rapidly receding parts of the face, e.g. from the zygomatic bone to the ear. The system was not used for a mass anthropometric study because of the large errors. It was an attempt, however, to push forward the field of stereophotogrammetry which appeared to have a limited future because of the high equipment costs.

In the 1970's progress in stereophotogrammetry seemed to slow down by the absence of much literature. Most systems were being evolved into hybrids of the old photogrammetry methods and stereophotogrammetric concepts, with the introduction of video equipment for the data acquisition. However, in 1991 stereophotogrammetry in its old form, appeared again as a technique to model the head and facial soft tissues.

Rasse et. al. (1991) used a stereophotogrammetric system to examine soft tissue changes pre and post-operative reconstructive surgery. Their system was similar to the one described above and they used landmarks on the face and control points. Their published work made no mention of the errors involved in this particular system, but one can infer that they were capable of getting accuracy's of ±0.5mm as it is similar to the systems developed by Burke (1967) and Herron (1972) which were both capable of this degree of accuracy. Their results were presented as contours of the face, of the areas where both cameras had the same view. This method is very accurate, but does not contain enough information to develop a soft tissue model.

The French team of Coblentz et. al. (1991) also chose to use a stereophotogrammetric system to develop a three dimensional model of the face shape of French adults to promote its use in the design of protective equipment. They maintain that as of 1991, this was the most accurate method for acquiring three dimensional data. With their system, again similar to that described above, they were able to obtain accuracy's of ±0.3mm. However they point out that the data processing times was high which limited the amount of data they were able to take. Introducing lighted structured back into the system, such as Loveday (1974), would make the problem of data processing easier.

The later generations of data acquisition systems have incorporated video equipment and structure light to overcome some of the limitations of the standard stereophotogrammic systems.
2.2.4 Hybrid Systems

In the past fifteen years, numerous systems based on the principles of stereophotogrammetry have emerged. A full review of all the various combinations is not necessary, but this section will briefly describe a few systems to give examples of the innovative works that have occurred.

In 1983, Dowideit et. al. introduced the use of video equipment to replace stereometric cameras in a stereophotogrammetry configuration. The real-time analogue systems were frozen and stored. An analogue to digital converter was then used to digitise the data. They used an in-house developed digital image correlation algorithm to extract the three dimensional data from the video images. This method was not described, however it must have been very rudimentary as the processing time for one image was 24 hours. The work was published when the system was still under development. The accuracy achievable with this system was limited to the pixel size of the video camera which was 1mm. The digital image correlation software is difficult to comment on without knowing any detail of the algorithms. This work was one of the first to use video equipment and develop the methods for processing data. With experience, it is probable that the algorithms would have become more sophisticated, reducing the processing time significantly.

It seems to be true for Deacon et. al. (1991) that eight years later, with the availability of better video which captures the data in a format suited for digital manipulation in more powerful computing equipment, that data acquisition and processing times were efficient and accurate. The equipment comprised of two CCD cameras with a pixel array size of 699 x 580 connected to a framestore which was capable of capturing the images from the two cameras simultaneously every 1/25 of a second. This was necessary to ensure there was no movement of the subject. A reference object was used to determine the cameras' geometry's which as mentioned before, is necessary to know in a stereophotogrammic set up. The subject was positioned and a dot pattern projected on the face. To process the data, landmarks were designated with a mouse and a "sheet growing" matching algorithm was used to extract the co-ordinate data. The processing was done on a Sun 3/160 with a speed of four million instructions per second (MIPs). This system was accurate to between 0.5-1mm which is consistent with other stereophotogrammic systems. However, the processing time to extract the co-ordinate data was approximately forty minutes. Upgrading their equipment to a Sun SPARCstation™ increased the processing time to under ten minutes. This system was limited to imaging the face only, and specifically regions which appeared in
and right image. From the published pictures, it appears that this restriction results in an image of ±45° about the mid-line of the face. With better computing power and another set of cameras, this system could be very successful as a 3D imaging tool.

A CCD-based stereo system similar to the one developed by Deacon et al. was used in the research of Gregory et al. 109. The emphasis of this work seems to be the software, e.g. matching algorithms, used to extract the three dimensional co-ordinate data. The specification for the cameras was a 600 x 500 array, giving a spatial resolution of about 1mm per pixel at the object surface. The objective of the image processing was to develop a method enhancing the contrast of the grid shown on the subject's face so that the processing would become simpler. They used a technique called line thinning which identified the grid lines from the image which could consist of a number of grey-levels and be more than a pixel in width, and reduced the data to a thin line of one pixel in width. They suggest that the accuracy of their system is equivalent to other stereophotogrammic systems because extraction of data was based on triangulation, as it was with the others. They did not do a validation exercise to determine if this was true.

A system developed by a Japanese group, 110 used standard cameras to obtain photographs of a subject simultaneously from three views. The subject was illuminated with a grid of 800 light points. Reference points in the form of luminous diodes were positioned in the photograph for later reference. An original feature was that once the photographs were developed, they were transmitted to a computer via a image scanner with a resolution of 200 dots per inch. Using the reference points and the geometric configuration of the cameras, the co-ordinates were calculated. The errors involved in this procedure were 0.2-0.3mm. Although these errors are lower than the previously mentioned system by 0.2-0.7mm, improvements could be made by increasing the resolution of the scanner, thus increasing the pixel array. This system was used only to image the face. Like the other systems dependent on the light arrays, in would probably be inadequate to image the entire head as the light distortions caused by hair would introduce errors. It is not known if the matching algorithms would cope with more than the two or three images used in most cases.

Stereophotogrammetric systems have been shown to be accurate to approximately ±0.5mm but their ability to acquire data 360° around the head has yet to be demonstrated. Extraction of 3D data involves expensive equipment or more recently with the incorporation of CCD cameras into the systems, complicated matching algorithms. These difficulties have lead to the development of laser scanning systems.
in the mid 1980's, which acquire the data in a format from which Cartesian or polar coordinates can be easily calculated. The next section discusses the fundamentals of laser scanning and reviews some current commercial and research systems.

2.2.2.4 Laser Scanners

This section introduces laser scanning systems and reviews most of the commercial and research systems which are reputable in this area of research. The problems associated with laser scanning are characteristic of this technique and as far as this researcher can tell through personal communications, are encountered by all users of laser scan methods. These problems will be described along with a complete discussion on the particular laser scanner used on this project is in Chapter 3.

The feature that distinguishes laser scanning systems in the use of a line projector to illuminate a profile on the subject. The profile is viewed by digital camera(s) either directly, or indirectly through mirrors geometrically aligned. Most systems have introduced rotation of either the camera(s) and laser rig or of the subject. When each profile is read by a CCD camera, a digital encoder determines the exact rotational position. The three-dimensional information is collected in the form of a height (or $H$), a radial distance away from the centre-of-rotation ($R$), and the encoder value ($\theta$) as in figure 2.11. These parameters are stored and later reconstructed using calibration algorithms that correct for any camera distortions.

Figure 2.11: Basic principles of laser scanning.
The accuracy levels of a laser scanning system are dictated by the resolution of the CD camera, the amount of data storage capabilities and the reconstruction algorithms.

Operation of these systems is inherently simple. The operator's task is to position the subject and start the system via a keyboard. The entire acquisition event is computer controlled, allowing easy routine operation by a trainee individual. When data acquisition is completed, the operator processes the data through menu driven instructions. Data output files consists of either the raw data, Cartesian co-ordinates or polar co-ordinates. Data analysis and display is done in a matter of minutes or less with an accuracy and precision of ±0.1-1mm, depending on which laser system is being used. Operator error is rarely a factor, and the system error is low, depending on the quality of the equipment and the care taken when calibrating.

Laser systems have become popular over stereophotogrammetric systems because of the following characteristics:

- The attainable accuracy of laser systems is as good as stereophotogrammetric systems at ±0.5mm and with the latest models, resolutions of ±0.1mm have been achieved.

- Operation and calibration of laser systems are simple and fast once the operator has had some basic training.

- Data acquisition is on the order of 15-30 seconds. This is also true for stereophotogrammetric systems. However, data reconstruction for laser systems is approximately 30 seconds to 2 minutes which is faster than the reconstruction times, which can be hours, for the stereophotogrammetric system.

- The equipment costs for laser scanners can be less expensive.

To summarise, laser scanning systems are replacing stereophotogrammetric systems because laser systems are easier to use, have faster acquisition and data reconstruction times and can achieve levels of precision better than before.

Currently, the most commonly used, commercially available laser scanning system is Cyberware based in California. The first systems capable of head scanning were mainly located in the USA. In 1988, Cutting et. al. reported on their use of the
Cyberware system for monitoring changes from maxillofacial surgery. The Cyberware system uses a line projector and camera set in fixed positions on a platform that is rotated around the subject. At that time the Cyberware data acquisition consisted of 512 radial or profiles and 256 horizontal points from the CCD camera. The system was also initially only able to achieve a resolution of ±1mm. New features incorporate the use of a 32-bit colour camera which can be used to add colour to the resulting image, for increased realism. This is the only system to do this. The Cyberware system now also includes a CCD camera with increased resolution so that 512 horizontal points for each radial profile can be collected. Following data acquisition, the data is transferred to powerful workstations like the Silicon Graphics Iris 2400 to be used in software packages created specifically by the user or Cyberware Corporation.

Cyberware have also introduced a full-body scanner which uses linear scans, as well as the cylindrical scan. The laser/camera configuration translate across the body to be able to image areas a cylindrical scanner can miss. The whole body scanner can achieve a resolution of ±0.4mm on large objects and ±0.1mm on small objects. The Cyberware has been used by Wright Patterson in their anthropometric data acquisition, and the Defence and Civil Institute of Environmental Medicine in Ontario and is also becoming established in the UK at the National Engineering Laboratory and at SCRDE Colchester.

Another laser system, established in the UK and used on this project, is the system developed by University College London. The UCL system features a single laser and CCD camera which are mounted stationery while the subject rotates 360°. The system takes 245 radial scans with 320 horizontal points per profile. This system was selected for our work and thus full details of this scanner are in Chapter 3.

In 1993, Cole and Cross compared the UCL system against the Cyberware system and found very little difference in their performance. He found the Cyberware to be slightly longer to acquire the data, but the time to create the models and effort to acquire data were found to be the same.

The Human Measurement Anthropometric and Growth Research Group of Loughborough University under Professor Peter Jones have developed the Loughborough Anthropometric Scanner, otherwise known as LASS. It is a full body scanner comprised of two banks of TV cameras positioned opposite to each other. Each bank of cameras views profiles from two, of four, banks of light projectors. A
subject stands on a turntable and rotates 360° and while beams of light, passing through the centre-of-rotation, illuminating profiles on the subject. Raw data is recorded by the cameras and fed into a computer which produces a model through dedicated software. It takes seconds for the scanner to accurately measure size and shape. This author is not aware that LASS is commercially available. However, it has been used for the design of breast prostheses and in assessing treatment of liver disease.145

Rioux et. al. have designed a laser scanner that provides linear scanning in the X and Y planes. A six sided mirror captures the illuminated profiles and averages them together to eliminate noise or spurious data. In 15 seconds it samples a volume of 250mm x 250mm x 150mm at a resolution of 1mm in the X and Y directions and 0.25mm in the Z direction.

The above described scanners are all renown among human imaging research groups. The simplest configurations, involving a single camera and light projection, are less expensive, however, there is a compromise between the simplicity of the system and the amount of surface information that can be obtained. Hidden surfaces and swept volumes caused by limbs, can be imaged better by LASS or the linear scanners. With slight modifications, which are discussed in Chapter 3, the UCL scanner has met the needs of this project.

2.3 3D Display and Modelling Methods

In the previous sections, the general methods of data acquisition were described. The display and modelling of the data is the key to it being useful information. The following sections mention and describe current means of display involving the reviewed data acquisition techniques.

2.3.1 From CT and MRI Data

The images produced from software programs such as ANALYZE™118, which can produce 3D displays by amalgamating MRI and CT scans are visually very appealing which satisfies the clinicians interest in:

- Exploring the data to determine anatomical structures or the existence of malformities in the body.
• Analysing the image, i.e. to examine the extent of a tumour.
• Manipulating the data to aide in the planning of outcomes of surgical procedures such reconstructive surgery.
• Using the images as a navigational aide in surgical procedures.

Ultimately, the metric accuracy is determined by the number of scans taken and in most situations, because of the radiation exposure involved with CT scans and the times requirements for data acquisition for both methods, there is not an adequate amount of data to get good anthropometric information and the images can not be used as design tools. However, the software is continually being improved. A review of different methods for amalgamating the data and the resulting images is not relevant to this work. Interested readers should consult the SPIE Visualization in Biomedical Computing series and the following references as an introduction to this subject matter: 55, 61, 28, 29, 116

2.3.2 From B-scanning Ultrasound Data

There have been attempts since the 1970's to display the 2D image data from B-scanning in a 3D context. Coleman et. al. 119 attempted to develop other methods of representing the amplitude of echoes on the B-scan image. The amplitude of the signal is traditionally represented by the intensity of the dots on the screen. Coleman realised that some information was lost with high amplitude echoes "hiding" lower amplitude echoes. To preserve amplitude characteristics, he geometrically superimposed the A-scan signal on the B-scanned image. While this technique may have assisted technicians in examining tissue characteristics, it seemed to be displaying redundant information, as the B-scanned image already contained the A-scan trace data. It is this author's opinion that little extra information could be seen. This technique did not produce data in a format useful for building a 3D model.

In 1990, King et. al. developed a 3D tool which would help the technician interpret the B-scan image better by supplying information about the position and orientation of the scan with reference to a base scan. 120. The spatial data was taken with an ultrasound positioning system which could detect the position and orientation of the B-scan probe in space. To ensure the spacing system would work at every position of the probe, an elaborate framework was built with four ultrasound generators positioned above the subject. A further four ultrasonic receivers were attached to the B-scan probe. The resulting apparatus looked difficult to use. The resulting images consisted of a B-scan
with a line overlaid. The images are difficult to understand unless experienced in reading B-scan images and are not in a format for proper modelling.

Attempts have also been made to produce 3D models from the 2D B-scan images. Stiller et al. used a PC based software package called "BLOCK" to stack 2D B-scan images. With a 35-50MHz probe of axial resolution 60µm, data was acquired in real time. Thirty parallel B-scans were taken at 100 or 500µm intervals and the resulting block image could be rotated, tilt or sliced. From sectioning they could look at the tissue structure and determine tissue thickness. However it's use as a model is limited as it avoids the surface rendering techniques that are necessary to generate a full solid model of the head. Their technique only generated models from tissue sections of 3mm width.

A good attempt to model B-scan images was done by Ganapathy et. al. in 1992. They used an approach similar to that of King et. al. and of this project. Using an electromagnetic positioning sensor system (Isotrak 3SPACE), they measured the six degrees of freedom of the B-scan ultrasound probe. Ultrasound data was taken in incremental slices of 2.5mm. Reconstruction of the data consisted of "injecting" the data from the slices into the reconstructed volume from the spacing data. Because the data was taken arbitrarily, a powerful interpolation algorithm was used to fill in the holes. The resulting three dimensional models were noisy and have a spatial resolution of 2.5mm.

The work done by Ganapathy et. al. is similar to this project only in as far as an electromagnetic position sensing device is used in conjunction with ultrasound. They acquired an ultrasound data set and using dedicated software, were able to visualised the data using the position data. It was not in a format suitable for use in a CAD package. The resulting image has a large surface resolution which would not be suitable for acquiring accurate anthropometric data as required by our sponsors. Their purpose was to be able to three dimensionally visualise the ultrasound data. They did not attempt to reference the ultrasound data to another mode of data acquisition for enhanced results.

2.3.3 From Laser and A-scanning Data

To the knowledge of this researcher, no party has attempted to combine A-scanned ultrasound data with laser scanning data onto a computer aided design system for the generation of a solid model. Additionally, the novel technique used to achieve this
includes conclusions about soft tissue thickness around the head of which no prior information has been found.

2.4 Summary

Most of the current anatomical data acquisition systems were deemed inappropriate in 1990 when this research began because of their invasive natures and high cost, i.e. CT and MRI. The information about soft tissue, i.e. tissue thickness was difficult to extract from CT, MRI and B-scanning. A-scanning was chosen because it was easy to use and determine soft tissue thickness with high resolution, inexpensive and non invasive with possible additional information about thickness of individual layers.

Of the anthropometric data acquisition systems available, laser scanning is an accurate and fast system. Data is acquired in three-dimensional format that involves very little further processing.
3.0 Laser Scanning

3.1 Description of the UCL Laser Scanner

The University of Surrey's BioMedical Engineering Group commissioned the Medical Physics Group at University College, London for an optical laser scanner with a specification that would allow acquisition of 360° scans of heads and output these files in a Cartesian data format. The equipment delivered consisted of:

1) One semi-conducting, infra-red laser (700nm wavelength)
2) One CCD (charge couple device) Sony Avacam Video Camera with resolution 256 x 320 pixels
3) Four mirrors: 2 no. 25cm x 25cm; 2 no. 10cm x 10cm
4) One workstation comprised of: 486DX33 105MByte computer with a 20MHz T800 transputer, 1 Avery SVGA monitor, 1 EIZO Flex Scans 9060S video monitor, 1 Hitachi camera monitor, 1 custom designed power supply by Medical Graphics and Imaging at UCL.
5) One stationary platform and chair rotated by a stepper motor with a shaft encoder for feedback of angular position data.

Fig. 3.1: A schematic plan view of the laser scanner.
The laser, camera and mirrors rig was delivered mounted on an aluminium 'V' -shaped frame that was supported on temporary scaffolding owned by UCL. A new frame was designed to support the laser scanning mount according to the specification that it be sturdy enough to support the weight of the mount of 6-7kg, have the facility to adjust the height of the mount, and be aesthetically pleasing. Figure 3.2 is a photographic of the laser scanner mounted on the permanent frame.

Figure 3.2: Laser scanner

The layout can be seen with the aid of Figure 3.1. The chair platform and centre-of-rotation lie coplanar to the laser line. When a subject is seated in the chair, the laser illuminates a profile of the subject's face. The profile is reflected in Mirrors A and B and then to the Camera Mirrors. The CCD camera views the two profiles and interprets the data to deduce the spatial information in Polar or Cartesian format. The subject is rotated 360° in the darkness to collect a full 3D data set. The entire scanning process taking approximately 10-15 seconds. The details of calibration and acquisition are discussed in the next section.
3.2 Calibration of Scanner

3.2.1 Procedure

With the calibration procedure, parameters are defined which contribute to the overall definition of the location of a point in space. These parameters are:

1. The mirror angle.
2. The camera tilt.
3. The distance from the centre-of-rotation of the chair to the laser.

This section describes the physical protocol of calibration and section 3.2.2 discusses the analytical calculations done to determine the spatial data. Section 3.4 discusses a validation exercise and presents the level of accuracy achievable with this system.

Calibration must be done routinely, especially if the system has been physically moved causing possible changes in any one of the parameters. An attachment that supports a 15cm graduated rod is mounted over the chair perpendicular to the centre-of-rotation. Figure 3.3 is a schematic illustration of the set up. The zero position on the graduated rod corresponds to the centre-of-rotation and is positioned using a standard plumb bob. At every centimetre on the rod there is a groove cut for the purpose of hanging the calibration stick, a white flat plate size 25mm by 190mm, onto which the laser line is projected.

![Diagram of Calibration Set Up]

Fig 3.3: Calibration set up.

The first step in the calibration procedure is to adjust the side mirrors (mirrors A and B in figure 3.1) so that they are in the position to image the subject being scanned and reflect this image onto the back camera mirrors, from where it is recorded. This
procedure commences by suspending the calibration plate over the '0cm' slot. The software then prompts for adjustments to the mirrors' horizontal and vertical tilts based on the position of the projected laser light on the calibration plate. Continual adjustments are made until the mirrors are oriented in the correct position, which is indicated by the software.

A dual mirror system results in two mirrored images viewed by the camera. Ideally, these are symmetrical about the vertical axis (the centre of rotation), Figure 3.4a. If the camera has tilted around its horizontal axis, an asymmetrical image occurs, figure 3.4b, the software is unable to continue correctly its calculations. Corrections to the camera tilt are made until the profiles are symmetrical as in figure 3.4a.

![Figure 3.4a: No camera tilt](image1)

![Figure 3.4b: Camera tilt](image2)

Figure 3.4a, b: Examples of profiles on video screen.

Originally, the CCD camera came from UCL mounted on an aluminium plate. Rectification of the camera tilt was done by the turning of a positioning screw. The amount of twist could not be gauged and as a result much time was wasted attempting to straighten the camera. A modification was made by mounting the camera in a Merlot Precision Rotator with resolution of 3 arc seconds. With the precision rotator, correcting for camera tilt is easily achieved.

Once the mirrors and camera are positioned correctly, the final calibration step, which sets known radial distances from the centre-of-rotation, is executed. This is completed by suspending the calibration stick at each of the centimetre intervals where the illuminated profile is viewed by the camera and digitised.
3.2.2 Theory

The method by which the calibration distances are digitised is as follows. A 1mW laser is dispersed through a small cylindrical disk into a thin vertical line with a width of 0.7mm, the centre line of which is computed by the digitisation software for use as the reference profile representing the scanned object surface. The camera views and samples a profile of width 0.7mm and length 190mm. During sampling, the output from the camera is passed through a digital comparator that knocks out all signals less than 50% peak intensity of the previous signal. This eliminates most noise from spurious reflections in the room. The camera gives a raster scan of 287 horizontal lines; each line sampled 590 times.

For each horizontal raster scan \((H_m)\), where \(m=1-287\), the camera sees the image as illustrated in figure 3.5b. In the image, there are two profiles that represent the left and right mirror reflections. With an internal camera timing device, the values for \(A_m\) and \(B_m\) are determined as 'clock counts'. The distance of the profile away from the centre of rotation \((L_m)\) is simply:
where for each profile, \( L_m \) is calculated for \( m = 1-287 \). The entire procedure is repeated as \( x \) increases from 1-14 in incremental steps of 1. This method for extracting the distance from the centre-of-rotation to the profile is also used when acquiring full laser scans. The profile becomes a curve rather than the straight line illustrated above.

The calibration data is stored in a calibration file in the system. This calibration data is repeatedly used for the final processing of the output file from future scans.

Figure 3.6: Data output from laser scanner.

The final output file is of the format \((r_x, z)\) where, \( r_x \) is the radial distance from the centre of rotation to each point at every vertical \( z \) in figure 3.6. To calculate the values of \((r_x)\) and \( z \), algorithms are used which account for any possible errors due to parallax in the camera geometry or errors in readings of the measured data. These algorithms have been supplied to us by UCL. The inverse relation graph in figure 3.7 illustrates why errors can exist in the laser scanning method. In an ideal world, the laser scanner would yield measured data exactly comparable to the calibration data with a slope with the value of one (curve A). However, due to optical distortions, referred to as parallax, introduced by the closeness of the subject to the camera, the measured curve is irregular such as curve B. Using the Least Squares method, a line can be drawn
through curve B which represents the measured data. The slope of the measured data usually does not match the calibration curve because of electrical and mechanical changes. The offset of the best fit line from the datum represents the parallax error.

Fig 3.7: Correlation between calibrated and measured distances.

The calculation of \( r_z \) is as follows.

\[
 r_z = \frac{1}{\left( \frac{slope}{centre - prof\_pt} \right) + par}
\]

Eq. 3.1

The parameters are defined as:

\[
\begin{align*}
slope &= \text{the slope of the best fit line of the measured data.} \\
centre &= \text{the 'clock count' of the centre of rotation.} \\
prof\_pt &= \text{the 'clock count' of the profile being measured.} \\
par &= \text{the parallax error.}
\end{align*}
\]

'Z' is calculated by:

\[
 z = \left[ val11 + \frac{(val11 - val5) \times (r - 11.0)}{6.0} \right] \times \left[ pfirst + (\text{profile\_pos} - 159) \right]
\]

Eq. 3.2

The parameters' definition used in the calculation of \( z \) are:

\[
\begin{align*}
val11 &= \text{the 'clock count' of the 11th calibration value.}
\end{align*}
\]
val5 = the 'clock count' of the 5th calibration value.
pfirst + profile_pos-159 = the horizontal scan number (m) of the profile.

The first part of equation 3.2 is a scaling parameter for each scan line, and is derived from the calibration data. The second part of the equation represents the scan line number which is either positive or negative from the designated 'O' scan line in the vertical centre of the screen, or scan number 159 in figure 3.5b.

The data for each profile consists of 287 co-ordinate points of \((r_p,z)\). Around a 360° scan, 245 profiles are taken so that a full head scan yields approximately 70,315 data points. Image reconstruction within the laser scanner software is done using Gouraud interpolation. Originally, profiles were taken at even increments of approximately 1.5° around the head. Since 1994, we have been able to control the number of profiles taken at different intervals through the 360° so that a greater concentration of profiles can be taken on the face, where there is more detail, and less around the back of the head. An example of the current interval set-up is given in section 6.1.1 where the scanning protocol is discussed.

### 3.3 Scanning Difficulties

At the onset of this project, a long time was dedicated to determining the limitations of the scanner. In addition to volunteers, two standard objects (figure 3.8), a rugby ball and a funnel, were used to study the scanner performance because of the similarities in size and shape to the human head.

We were aware that the UCL laser scanner had been primarily designed for maxillofacial imaging and could not, therefore, adequately cope with a full 360° scan because the issue of hair artefacts in the images had not been dealt with.

By scanning individuals with varying hair length, we were able to determine that hair longer than 5mm and facial hair resulted in spurious data which would cause the scanner software to "hang up" before displaying a graphical output. Examining the raw data files showed that the data from the hair was marked as "bad data" and not stored in the format \((r_p,z)\).
The only options seemed to be to either have each volunteer shave his or her head, or to cover the hair with a smooth surface which the laser scanner could image. The later option was determined to be more feasible and the search was begun for a skull cap that would work well.

Figure 3.8: Rugby ball and funnel used for validation purposes.

The initially selected skull cap was specially moulded by the DRA. Figure 3.9 is an example of the image achievable at that stage. Although the fit was not ideal, this approach enabled the first successful 360° scan.

An additional problem was highlighted by this scan. The top of the head was "missing" from the scan. This was also true of the first images produced of the rugby ball in figure 3.10.
Figure 3.9: First successful 360° scan.

Figure 3.10: First scans of rugby ball.
The problem was that there was not a high enough intersect of reflected beam in the line of sight of the camera path. This problem had been seen to occur in the work of the other research groups. To the present, this seems to present a problem in use of scanners at re-entrant surfaces unless great care is taken.

The first attempt at a resolution was to acquire two scans for each subject - one with the subject bent over to scan the "top" of the head, and the other with the subject sitting in the normal position to capture the data of the face and neck. Three reference markers were placed on the head and had to be visible in both scans for the data to be merged together in a geometrically and dimensionally accurate way. The combination of the data was done on the CAD system. Figure 3.11 is an example of the results. Figure 3.11a shows the separately scanned volumes, indicating their different orientations and translated position. Figure 3.11b show the combined head model after manipulation of the internal reference axis.

Figure 3.11a: Attempting to combine to seperately scanned models into one.
Figure 3.11b: Aggregation of two head scans to capture the top of the head.

This technique proved to be very time consuming both in the acquisition and modelling of the data. It could be visually seen that when the subject leaned forward for the dome scan, the pull of gravity changed the tissue topography, making the face appear fuller in the cheek area. We felt it also defeated the philosophy of possessing a state-of-the-art scanning system allowing us to acquire all of the necessary data on the subject in seconds without manual intervention.

The second approach, the one we have adopted, uses a combination of an enhanced reflective surface over the dome of the head and a known reference point positioned at the top of the head to aid in the software interpolation. The quality of the image in the receding surface regions was studied using a standard anthropometric dummy, and it was found to be very sensitive to surface quality and colour. Reflective tape was put on the skull cap and by ensuring light, highly reflected surfaces in the difficult regions, an almost complete scan was achievable as a result of reduction in reflective signal losses.

To aid in a more accurate interpolation of the dome area, it was decided to suspend a rod with a small disk over the centre of rotation. The notion was that the laser scanner would image the disk and provide a surface flush with the top of the head that the laser scanning software could interpolate to. It would also provide a reference point for
later attempts to aggregate all of the scanned heads. Chapter 6 discussed this point further. Figure 3.12 is the result of the first successful attempt to image the dome of the head.

Figure 3.12: Laser scan including the dome of the head.

Fine tuning this approach involved replacing the DRA skull cap with a white nylon/lycra swimming cap. This increased the amount of reflection of the projected laser light. Additionally, a permanent rod was fixed over the centre of rotation. The resulting image of the highest achievable quality and reflection is shown in figure 3.13. The use of a skull cap creates a dimensional difference from the underlying skull surface. Use of the nylon swimming cap keeps the error small because it is tight fitting. However, to attempt to predict the level of error, a scan was taken of a volunteer with a shaved head and then wearing the skull cap. The images were aggregated together with the techniques discussed in Chapter 5. The errors were estimated to be from 1.5-5mm. The errors were dependent on the site from which they were taken. For instance, the seams of the skull cap gave a larger error. In other areas, where the skull cap was stretched, the error was smaller. These values include errors due to hair thickness for short hair. It was difficult to accurately assess the error from long hair thickness which was so variable between people.
3.4 Data transfer and modelling in CAD

Our objective of the research is the ability to model a soft-tissue shell onto a Computer Aided Design system, thereof it was important to ensure that the transfer of the laser scanned data to the CAD was possible and valid. Transfer of data from the scanner to the CAD software used, MEDUSA and VARIMETRIX, is achieved via an Ethernet link. The process involves three stages:

1. Scanning
2. Transfer of the raw data to CAD
3. Manipulation of the data within CAD.
Stage 1, the laser scanning data acquisition has been discussed and some results of models have been shown. A successful scan results in a binary file containing the data sets \((r, z)\) for each angle increment of the 248 profiles. Software has been written which converts the binary file to ASCII and calculates the Cartesian points as,

\[
\begin{align*}
    x &= r \cdot \sin \theta \\
    y &= r \cdot \cos \theta \\
    z &= \text{given in data file}
\end{align*}
\]

The angle \(\theta\) is the degree of rotation at which the profile was taken. The laser scanner has been validated by UCL \(^{115}\), but to ensure the models from the laser data are accurate and thus stage 2 valid, an object (the funnel) of known dimensions was scanned and modelled on the CAD. Dimensions were taken from the model (figure 3.14) and compared to the actual object. The results are shown in table 3.1.

<table>
<thead>
<tr>
<th>Model (mm)</th>
<th>Physical (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>153.87</td>
<td>154.00</td>
</tr>
<tr>
<td>29.76</td>
<td>30.00</td>
</tr>
<tr>
<td>84.74</td>
<td>85.00</td>
</tr>
</tbody>
</table>

Table 3.1: Measured dimensions versus modelled dimensions.

The above exercise showed that the transfer of data from the laser scanner to the CAD is valid and has an accuracy of 0.25 mm. These accuracy values can be expected after the system has been calibrated. If the apparatus is moved and the system is not re-calibrated, large errors of any magnitude can occur.

Manipulation of the data, stage 3, with the CAD is necessary for aggregation of the heads for statistical representation and is discussed further in section 6.1.2.
Figure 3.14: Model of funnel with indicated distances measured.
3.5 Resolution

Data is acquired with a 0.7mm width laser line at varying intervals. The data acquisition circumferential resolution of the laser scanner is a variable parameter. It is dependent on the interval between profile acquisition and the radius from the centre of rotation to the point of interest. Figure 3.15 shows the geometry.

\[ dr \sin \Delta \theta = dx \]

Figure 3.15: Circumferential resolution of the laser scanner.

Around the face the intervals are set to every 1°. This results in an approximate circumferential resolution in the location of the reference dots (about 50mm from the centre of rotation) of 0.17mm that is acceptable for imaging the dots. The interpolation between profiles produces a model of which it is possible to identify objects ±0.25mm circumferentially and ±0.2mm radially.

The laser scanner described in this chapter is a fast method for acquiring 3D data at an acceptable resolution and level of accuracy. The added attributes of the skull cap and the suspended rod enhances the quality and quantity of the data acquired. From the data, an image is modelled to the quality pictured in figure 3.13.

The next chapter presents A-scan ultrasound as the technique used to acquire anatomical information for integration with the laser scanned data.
4.0 Ultrasound System for Acquiring Soft Tissue Thickness

Ultrasound was chosen to determine soft tissue thickness about the head because of its low cost and its safe, non invasive nature. It has widespread acceptability as an effective mean of producing accurate dimensional measurements to fractions of a millimetre. "Where anatomy is (or considered to be) sufficiently simple that a 2-dimensional image is not necessary for guidance, the A-scan continues to have application and, in its r.f. form in particular, has value in obtaining the accurate dimensional measurements that are required.....". 79. Gore 70 has also stated that A-scan is a better tool with which to look at the tissue macro-structure. For the purpose of this research, we were not interested in the 2D images resulting from B-scan ultrasound because a "picture" of tissue anatomy was not needed. Instead, our interests lay in determining thickness dimensions with which Hill and Gore advocate using A-scan ultrasound.

The following sections discuss the theoretical background of A-scan ultrasound and present the ultrasound equipment used to measure the thickness of the soft tissues of the head.

4.1 A-scan Ultrasound

4.1.1 Ultrasound theory

Throughout the rest of the chapter, notable reference will be made to various ultrasound parameters. Therefore it is necessary to give a brief definition of these parameters.

Ultrasound follows the laws governing sound, only it has a frequency above 18kHz - the audible range being 20Hz to 18kHz. 123. Sound travels by the orderly sinusoidal longitudinal motion of particles in a medium. A source must generate the sound wave initially. For ultrasound applications, this source is most often a piezoelectric crystal. When a voltage is passed through the crystal, the whole crystal lattice moves, resulting in surface plane movement. Vibration of the crystal causes a net force on the atoms and molecules in the surrounding medium, resulting in mechanical vibrations that are transmitted through the medium at the same frequency of the source vibration. Figure 4.1 is a snapshot in time of a line of particles undergoing sound vibrations.
Figure 4.1: Snapshot of particles undergoing sound vibrations. Re-created from 124.

The distance along the sinusoidal wave where signal behaviour is the same, i.e. particles are producing the same motion, is termed the wavelength ($\lambda$), figure 4.2. One wavelength occurs over time ($T$), also known as the Period, resulting in a relationship between the speed of sound ($c$) of the particles and the wavelength.

\[ c = \frac{\lambda}{T} \]

Eq. 4.1

Figure 4.2: Sinusoidal wave showing wavelength ($\lambda$) and the period ($T$).

The frequency of a sine wave is calculated as the inverse of the period, $f = 1/T$ so that equation 4.1 can be written as:

\[ c = f\lambda \]

Eq. 4.2

The speed of sound through any medium is a function of the mass and spacing of the particles, or density ($\rho$), and the strengths of the forces of attraction between them,
known as the stiffness. Stiffer mediums propagate sound more quickly. Stiffness is described by the elastic modulus of the material (K), which is also defined as:

$$K = \frac{\text{Stress}}{\text{Strain}}$$  \hspace{1cm} \text{Eq. 4.3}

where for sound, the stress is defined as the applied pressure ($P_0$) on the surrounding particles and the strain is the fractional change in volume in the spacing of the particles. The compressibility of the medium is defined as:

$$\kappa = \frac{1}{K}$$  \hspace{1cm} \text{Eq. 4.4}

so that the speed of sound, defined as a function of density and compressibility is:

$$c = \sqrt{K/\rho}, \hspace{0.5cm} \text{or} \hspace{0.5cm} c = 1/\sqrt{\kappa \rho}$$  \hspace{1cm} \text{Eq. 4.5}

The parameter that dictates the amount of ultrasound signal that is reflected at tissue layers is the acoustic impedance ($Z$). The impedance is a ratio of the applied pressure on the surrounding particles, to the speed of sound of the particle.

$$Z = \frac{P_0}{c}$$  \hspace{1cm} \text{Eq. 4.6}

A unit derivation follows to show that the acoustic impedance can also be expressed as the product of the speed of sound and the density of the medium:

$$Z = \rho c$$  \hspace{1cm} \text{Eq. 4.7}

$$\frac{P_0}{c} = \frac{\text{Force}}{\text{Area}} = \frac{\text{kg} \cdot \text{m}}{\text{m}^2 \cdot \text{s}^2} = \frac{\text{kg}}{\text{m} \cdot \text{s}}$$  \hspace{1cm} \text{Eq. 4.8}

Multiply equation 4.8 by a factor of 1, or \( \frac{m}{m} \), results in the equation:

$$\frac{P_0 \cdot m}{c \cdot m} = \frac{\text{kg} \cdot \text{m}}{\text{m} \cdot \text{s}} = \rho c$$  \hspace{1cm} \text{Eq. 4.9}

which is also the product of the velocity of sound and the density of the medium.
As stated above electrical energy is converted to mechanical energy via the piezoelectric crystal. When the mechanical energy propagates through the tissue as a medium, the energy is absorbed by the relaxation and visco-elastic effects of tissue, leading to its heating. The rate of temperature increase is proportional to the intensity of the ultrasound and the absorption coefficient, and inversely to the tissue density. Therefore the intensity of the energy used in the ultrasound must be regulated for safety reasons, to protect structural damage to the tissue from heat.

A volume of tissue within an ultrasound field will reach an equilibrium temperature. The capacity to store heat is proportional to the volume of the tissue, so that for the same intensity, regions of tissue will have varying equilibrium temperatures. Additionally, the ability of tissue to lose heat is dependent on the surface area against the surrounding tissue.

A focused transducer affects a region of tissue smaller in volume and surface area so that the equilibrium temperature is higher. Pulse ultrasound has a thermal effect on tissue because the ultrasound is periodically switched on and off, and cooling of the tissue occurs in the off periods.

The intensity is defined as "the rate of flow of energy through unit area oriented perpendicularly to the beam at that point". It can be calculated as proportional to the ratio of the applied pressure ($P_o$) to the acoustic impedance ($Z$).

$$I = \frac{1}{2} \frac{P_o^2}{Z}$$  
Eq. 4.10

An ultrasound beam with a high intensity and exposure time can be biologically hazardous. The American Institute for Ultrasound in Medicine have shown that for a spatial-peak-temporal-average (spta) intensity (defined as the power of the ultrasound divided by the cross sectional area of the beam) of 100mW/cm$^2$ with exposure times up to 500 seconds, there results no biological harm. The spta intensity of the equipment used for this work is 23mW/cm$^2$ and at no time was the exposure time longer than approximately 100 seconds. There was no danger of biological harm to our volunteers.
4.1.3 Propagation of ultrasound through tissue

Figure 4.3 is an example of how ultrasound reflects and propagates at tissue boundaries. A large proportion of the incident pulse is transmitted through the various interfaces of soft tissue until it reaches an air canal or bone structure, at which time, some of the remaining pulse is reflected back to the transducer (the pulse echo). The proportion of the signal that propagates through at tissue boundaries is variable depending on the acoustic impedance of the tissue. Also at boundaries, the sound can reflect normally, scatter, transmit and be absorbed so that much of the signal is lost. For example in this work, the value of T1 in figure 4.3 approximates 0.5mV, or only 2% of the original pulse. The value of T5 can range from approximately 0.25mV-1.5mV, or 0.8-5% of the original pulse. These phenomena are discussed further in sections 4.1.3.1-4.1.3.5

The time between the received pulses T1 and T5 from the gel/epidermis layer and the muscle/bone layer, is used to derive the distance travelled using the average velocity of sound through soft tissue as 1540m/s.

\[
D = \frac{(T5 - T1)\text{[sec]} \times 1540 \text{[m/sec]}}{2} 
\]

Eq. 4.11

Figure 4.3: Schematic of transmitted and reflected pulses.
Since the time \((T_5-T_1)\) measured is the duration of the pulse travelling to the reflecting boundary and back, the calculation of the distance travelled \((D)\) is divided by two.

Transmission of an ultrasound field through tissue results in a number of phenomena that significantly affect the resulting signal. An understanding of the phenomena is necessary to use A-scan as a diagnostic tool and to interpret the resulting signals satisfactorily.

4.1.3.1 Reflection

Reflection occurs at tissue layer interfaces where the acoustic impedance of materials before and after the interfaces are different. The ratio of the incident wave reflected at differential impedance interfaces is defined as:

\[
\text{Ratio of reflected/ incident pulse} = \left( \frac{\rho_1 c_1 - \rho_2 c_2}{\rho_1 c_1 + \rho_2 c_2} \right)^2, \quad \text{Eq. 4.12}
\]

where \(\rho_1 = \text{density of material 1}\) \(\rho_2 = \text{density of material 2}\)
\(c_1 = \text{speed of sound in material 1}\) \(c_2 = \text{speed of sound in material 2}\)

From equations 4.9 and 4.12, the proportion of sound reflected can be written as:

\[
\text{Proportion of sound reflected} = \left( \frac{Z_2 - Z_1}{Z_2 + Z_1} \right)^2 \quad \text{Eq. 4.13}
\]

Table 4.1 lists accepted values of acoustic materials in various soft tissues.

Reflections at the interface of the layers in soft tissue in similar impedance values are not expected to be significant. The densities of the epidermis, dermis and subcutaneous layer are nearly the same. The speed of sound is difficult to measure because we are not able to segregate and measure individual tissue layers \textit{in vivo}. Investigating the speed of sound value in an \textit{in vitro} sample of individual layers cannot be accurately used for \textit{in vivo} values because of variations in fluid contact and the changes in cellular membrane stiffness within the excised tissue.

To determine overall soft tissue thickness, it is predominantly the interface between soft tissue and bone or soft tissue and air that is of concern. We can calculate the expected ratio of the return echo to the incident pulse using the values in table 4.1:
Soft tissue/bone interface: 

\[
\left( \frac{1.7 \times 10^6 - 7.80 \times 10^6}{1.7 \times 10^6 + 7.80 \times 10^6} \right)^2 = 0.412
\]

Soft tissue/air interface:

\[
\left( \frac{1.7 \times 10^6 - 0.0004 \times 10^6}{1.7 \times 10^6 + 0.0004 \times 10^6} \right)^2 = 0.999
\]

When reflected by bone instead of tissue, 41% of the incident wave is reflected back and for sound emerging from tissue into air, 99.9%.

<table>
<thead>
<tr>
<th>Material</th>
<th>Acoustic Impedance [g/cm²s] (Z)</th>
<th>Velocity of Sound [m/s] (c)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air (N.T.P.)</td>
<td>0.00004 x 10⁵</td>
<td>1500</td>
</tr>
<tr>
<td>Blood</td>
<td>1.61 x 10⁵</td>
<td>1570</td>
</tr>
<tr>
<td>Bone</td>
<td>7.80 x 10⁵</td>
<td>3500</td>
</tr>
<tr>
<td>Fat</td>
<td>1.38 x 10⁵</td>
<td>1450</td>
</tr>
<tr>
<td>Muscle</td>
<td>1.70 x 10⁵</td>
<td>1580</td>
</tr>
<tr>
<td>Soft tissue (average)</td>
<td>1.63 x 10⁵</td>
<td>1540</td>
</tr>
<tr>
<td>Water</td>
<td>1.48 x 10⁵</td>
<td>1480</td>
</tr>
</tbody>
</table>

Table 4.1: Acoustic Impedance and Velocities of some materials. Condensed from McDicken (1981)

The remaining phenomena to be discussed have a large role in reducing the amplitude ratio of the reflecting signal.

To ensure most of the echo is detected by the transducer and not sent off "into space", it is necessary for the angle of incidence to be normal, or nearly normal, to the reflection surface. Small deviations from the normal can result in large reductions in the amplitude of the received signal. ⁹² This has been a challenging problem given that soft tissue at a depth (i.e. the reflecting surface) may not be parallel to the outer skin surface. Examination of various skulls and their angles of protrusions of the bones has proven to be very helpful to the operator in getting a "feel" for where the normal position of the probe should be in relation to the underlying surface.

### 4.1.3.2 Scattering

Scattering is said to occur when the ultrasound beam reflects in directions outside the defined beam or received direction (figure 4.4). This occurs when the reflecting surface is much smaller than the wavelength of the incident beam. The intensity of the
scattered wave echo is determined by the impedance of the reflecting surface. Scattering also occurs when the reflecting surface boundaries of the tissue have an uneven surface that is not normal to the incident pulse. As this is usually true in soft tissue, the resulting signal can contain echoes of variable amplitudes. However, when investigating the thickness of soft tissue as one entity versus examining individual layers, returning wave signals due to scattering do not greatly interfere with the analysis of the gel/epidermis and muscle/bone/air interface echoes because the reflected signals are a higher percentage of the amplitude of the incident pulse.

4.1.3.3 Refraction

Refraction occurs when the direction of propagation of a beam changes direction in transmission through an interface due to the two mediums having largely different velocities of sound. For soft tissue layers, refraction does not present a problem as the differences in the velocity of sound through the layers are not so varied. At the soft tissue/bone interface, 41% of the incident energy is reflected back, enough to result in a strong return echo. What is lost through refraction, after being transmitted through the bone is of little interest. At the frequencies used here, the bone becomes, effectively, a final block to further transmission and hence 'ends' the depth acquisition.

4.1.3.4 Absorption

As explained in section 4.1.1, the transmission of ultrasound through tissue is actually the vibrational energy of particles. The stresses that occur from the vibrations are relaxed through the absorption of heat into the tissue. This is where the safety aspects are of concern in diagnostic ultrasonics and have already been addressed in section 4.1.2.
4.1.3.5 Attenuation

The frequency, scattering and absorption play the largest role in attenuating the amplitude of an incident pulse as it transmits through the tissues, and similarly for the returning signal. The absorption is proportional to an increase in frequency. A higher frequency also provides more detail so that one chooses the right frequency to get a satisfactory balance of information to penetration depth, achieving the desired results. To determine soft tissue thickness, it is only necessary for enough of the energy of the pulse to survive the journey from the transducer to the reflection surface, and back, to give an echo of magnitude that can be detected by the operator. Up to 98% of the incident energy may be lost before the reflected echo is too small for satisfactory deduction of the tissue thickness in our system. The incident pulse we use is approximately 34mV when transmitted from the transducer. This researcher believes an accurate assessment of the return time of the pulse can not be made with an echo of amplitude less than 0.4mV after the 40dB amplification. The range of echo magnitudes in our signals are usually between 0.4mV-150mV after the amplification.

4.2 A-scan Acquisition System

4.2.1 System Hardware/Software Description

When choosing hardware for acquiring soft tissue thickness, both A-scan and B-scan ultrasound systems were considered. B-scan was eliminated because of its cost and the fact that the output is an image which would have needed further processing to extract the tissue thickness. At the time no readily available 3D analysis package was identified. With this in mind, the A-scan Dermal Monitor (DM70) manufactured by Cutech Ltd. was purchased. The hardware for sampling and digitising the waveform was chosen to be a 486DX2, 66MHz PC linked to a Hitachi digital storage oscilloscope. Section 4.2.2 discusses the ultrasound transducer and its operational specification in detail.

4.2.1.1 Dermal Monitor (DM70) by Cutech, Ltd.

According to the manual, the Dermal Monitor is capable of measuring depths of 0.05-50mm. We have determined that the transducers supplied with the equipment will only satisfactorily measure depths from 10-45mm. When the transducer is mounted in its probe casing, this results in a 0-35mm depth penetration within the tissue. An electrical pulse driver delivers a brief pulse, in the order of 250V, to the ultrasound transducer.
The piezoelectric transducer transmits and receives that pulse after interaction with the tissue. The received electric signal is directed toward either a radio frequency (rf) or video amplifier which increases the signals by 40dB and 60dB respectively. In our work the radio frequency signal was used. Although the video signal has a higher amplification, it is rectified unipolarly, from which is more difficult to identify zero-crossings, and thus, tissue boundaries. This is addressed further in section 4.2.2. Output from either one of the amplifiers is available for examination of the return echoes on an oscilloscope. There is also a trigger pulse output to synchronise the oscilloscope to the DM70. The Gain and Delay of the signal and trigger are controllable to alter the magnitude and to view particular parts of the signal. Figure 4.5 is a block diagram of the dermal monitor.

![Figure 4.5: Block diagram of features on DM70](image)

### 4.2.1.2 Hitachi Digital Storage Oscilloscope (VC-6025A)

It was recommended by Cutech Ltd. that an oscilloscope with a bandwidth not less than 50MHz was used to view the A-scan signal. Various oscilloscopes were considered and the Hitachi VC-6025A oscilloscope met the system requirements. The oscilloscope has an acquisition memory of 2,000 words but only 1,000 words of data can be transferred to the PC. The waveform data of the digitised signal is 8 bits. This corresponds to 256 vertical samples over the ten divisions of the cathode ray tube.
(CRT) screen. For the setting used in our data acquisition of 5mV/div, the vertical sample resolution is every 0.2mV. This vertical resolution is considered adequate since the emphasis of this work was not to consider signal characteristics for echo amplitude, for which a better resolution would be needed.

The sampling resolution of the time-base for a setting of 2 microseconds per division, or 1,000 words over typically 25 microseconds of signal, is 0.025 microseconds. This is acceptable to acquire the data of signals with a central cross-over frequency of 12.5MHz, or every 0.08 microseconds.

The data is transferred to the PC via a RS-232 serial line. The appropriate wiring diagrams can be found in the Hitachi manual. 126.

4.2.1.3 486DX2, 66MHz Personal Computer

A 486DX2, 66MHz personal computer was purchased to drive the acquisition system. Using a RS-232 connection to the oscilloscope, function commands written in QBasic and recognisable by the oscilloscope control the sampling, transfer of data to the PC, manipulation of file formats and saving, of the data. The data collected comprise 14μs duration of A-scan amplitude versus time display. These are saved for subsequent analysis to derive the tissue thickness.

Two pieces of software are used in this system for the acquisition and interpretation of the ultrasound signal. The acquisition software was written by this researcher while the interpretation software was supplied by Hitachi and is called Himes. Himes is a standard signal analysis program and its use, as well as the acquisition software, is described in section 4.2.1.5.

4.2.1.4 Acquisition Software

The acquisition software, entitled "Grabdata" (see Appendix B) was written in QBasic. The commands sent to the oscilloscope are pre-programmed by the manufacturer. The commands allow sampling of a signal, saving to the oscilloscope's memory, reading the data from the memory, transferring it to the PC and saving in a file. Figure 4.6 shows a logic diagram of the acquisition software.
Another program called FORM.BAS changes the file format as downloaded from the oscilloscope, to one that can be read by the Himes software. The format output from the oscilloscope first lists the number of data points in the signal. Next it lists consecutively the amplitude value of each sample on the signal. The format needed for file input to the Himes software includes a header where on each line a different parameter is set, such as the time base and amplitude per division, that was used. The data is then listed so each sample in on a separate line. Figure 4.7 is a logic diagram of the formatting software.
4.2.1.5 Hitachi Himes Software

This software has many capabilities common to other signal analysis software. However, for this work only certain functions were needed. Within this software the time the pulse takes to transmit through the tissue and reflect back to the transducer was identified. When loaded into the Himes, the signal was always miniaturised, but with the proper scale. No explanation could be found as to why the signal was not loaded as a "fit to page" graphic. It was necessary to enlarge the signal using a zoom function within the software, so that the signal detail could be seen. This operation retained the accurate scale. Cursors were placed on the appropriate parts of the signal for determination of the time base. Chapter 7 discusses interpretation of the signal to allow accurate placement of the cursors.

The next section describes the most critical front end piece of hardware for this research, the ultrasound transducer.
4.2.2 Ultrasound Probe Description

A parameter that is significant to the pulsed transducer is the frequency. A pulse of sound is the aggregation of several sinusoidal wave forms of differing frequencies. Therefore the frequency parameter quoted is actually the central, or "zero-crossing" frequency, which is approximately equivalent to an average of all the frequencies in the pulse. The zero-crossing frequency is defined as the frequency of a sine wave having the same number of crossings on the time axis per unit time, as the pulse. Figure 4.8.

![Zero crossing 17 times](image)

![Sinusoid wave with 17 zero crossings and frequency of f=1/T](image)

Figure 4.8: Graphical representation of zero-crossing frequency.

The transducers supplied with the Dermal Monitor were specified to have a zero-crossing frequency of 12MHz. To confirm this the transducers were put through a validation test described below. Other parameters that were investigated were the location of the focus and the focus width. The values for these parameters are important to enable the focus, or area of maximum intensity, to be accurately positioned at the tissue depth of interest.

The transducers were mounted, coupled with gel to a very thin polythene window on the side of a tank containing water. Sound waves, which were propagated from the transducer through the polythene into the tank of water, were measured by a Hydrophone with a polyvinylidene fluorine (pvdf) 100μm diameter transducer. Figure 4.9. A Hydrophone is a small transducer used to measure pressure strengths. It is made from a piezoelectric crystal approximately 1-2mm diameter and is independent of the angle of incidence of the pulse it is measuring. The pulse shape and peak-peak voltages were measured along the central axis of the transducer to find the maximum
intensity, or focus, of the transducer. The pulses are shown in figure 4.10. Each transducer was found to have a different zero-crossing frequency, listed in table 4.2. Plots of distance versus intensity are shown in figure 4.11. The distance at which the maximum intensity occurs, which is the location of the focus, can be identified.

Figure 4.9: Schematic of hydrophone measuring transducer properties.

Examination of the graphs for the 12.5MHz, 10.0MHz, and 13.2MHz for the in figure 4.11 reveals that the maximum voltage occurs at 25µs, 20µs and 25µs, respectively. These figures are converted to distances in millimetres by multiplying the times by the speed of sound through water, e.g.:

\[ 1480 \text{m/sec} \times 25 \mu \text{s} = 37 \text{mm} \]

Table 4.2 shows the location of the focus for each transducer. Apparently they are nearly the same.

<table>
<thead>
<tr>
<th>Transducer</th>
<th>Width of focus</th>
<th>Frequency (MHz)</th>
<th>Maximum Voltage</th>
<th>Distance focus-transducer</th>
</tr>
</thead>
<tbody>
<tr>
<td>88169</td>
<td>2.7mm</td>
<td>13.2</td>
<td>57mV</td>
<td>37mm</td>
</tr>
<tr>
<td>88163</td>
<td>1.1mm</td>
<td>10.0</td>
<td>57mV</td>
<td>29mm</td>
</tr>
<tr>
<td>88002</td>
<td>no focus</td>
<td>12.2</td>
<td>17mV</td>
<td>37mm</td>
</tr>
</tbody>
</table>

Table 4.2: Parameters of ultrasound transducers.
When the maximum intensity was found, the Hydrophone was then scanned perpendicular to the central axis to determine the width of the focus, which is the diameter of the ultrasonic beam at the focus. The results as seen from the graphs in figure 4.12 for each transducer are repeated in table 4.2.

There is a variation in intensity across the ultrasound field within the focus width. An ultrasound field can be detected for a width of 5mm, but the voltage intensity varies from 5mV minimum to 60-80mV maximum. The values in table 4.2 were taken from an average voltage value of 30-40mV. Notice that transducer 88002 does not have a focus width due to the fact that the maximum intensity level is hardly above the noise level, hence, it was difficult to determine the width.
Figure 4.10: transducer pulses
Figure 4.11: Plots of intensity versus distance.
Figure 4.12: Pulse widths.
It is possible to determine the theoretical position of the focus for these transducers with the specified wavelengths and their geometry's. The radius of curvature of the transducer \( R \) can be calculated using the depth \( h \) of the transducer (the depth of the transducer is a result of the curvature) and the transducer radius \( A \) as per figure 4.13.

\[ R = \frac{A^2 + h^2}{2h} \quad \text{Eq. 4.13} \]

The equation for determining the maximum theoretical voltage intensity was taken from O'Neil.

\[ z_{\text{max}} = R - \frac{12R}{k^2 h^2 + 12} \quad \text{Eq. 4.14} \]

where \((k)\) is determined from the wavelength,

\[ k = \frac{2\pi}{\text{wavelength}} \quad \text{Eq. 4.15} \]

The results are listed in table 4.3:
Table 4.3: Location of theoretical maximum intensities.

<table>
<thead>
<tr>
<th>Transducer #</th>
<th>Radius of Curvature (R)</th>
<th>Location of Max Intensity (Z_{max})</th>
</tr>
</thead>
<tbody>
<tr>
<td>88169</td>
<td>45.6mm</td>
<td>44.85mm</td>
</tr>
<tr>
<td>88163</td>
<td>43.8mm</td>
<td>42.66mm</td>
</tr>
<tr>
<td>88002</td>
<td>94.8mm</td>
<td>87.77mm</td>
</tr>
</tbody>
</table>

It was obvious by comparing the locations of the measured focus (actual measured intensity), the theoretical maximum intensity (Z_{max}), and the present designs of the ultrasound probes that the manufacturer of this equipment designed the existing probe tips with the intention of the focus occurring at Z_{max} in figure 4.14, i.e. the end of the probe tip. For unknown reasons, the focus actually occurs inside two of the probe tips and outside one.

The requirement, therefore, was to redesign probe tips so that the focus occurred at the expected soft tissue/bone or soft tissue/internal air boundaries, allowing a stronger echo signal for easier analysis. Soft tissue thickness on the face varies from approximately 3mm to 30mm, making it impossible to put the focus right at the boundaries for all maxillo-facial locations. Consulting figure 4.10 shows that over a distance of 11mm, the intensity remains at greater than 50% of its maximum value and at 30% over approximately 43mm. Therefore, according to the following design, the focus was placed at a position of approximately 22mm deep within the soft tissue.
With the actual specifications of the transducers known, new probe tips were designed and manufactured to allow for placement of the focus at an appropriate depth within the soft tissue. The new design consisted of two pieces - a type of push piece and housing arrangement - that screw into the existing casing. Between the two pieces, a thin polythene material that contained the water used as a transmission medium, inside.

From use on a preliminary test, it was obvious that the contact area of the probe tip was too large to reach some of the critical areas on the face, i.e. around the nose and eyes. It was decided to design an entirely new transducer housing which would optimise use of the beam shape, and provide minimum contact tip area.

From the graphs in figure 4.11, the focus width transducer 88169 is approximately 2.2mm for two of the transducers. This meant the beam shape transmitted from the transducer of 13.0mm diameter can be approximated as figure 4.15:

![Figure 4.15: Geometric representation of the focused ultrasound field.](image)

To minimise contact area, it was hoped that a contact surface of less than 1cm could be obtained. Appendix C contains engineering drawings of the final probe casing design.

The probe tip is the crucial element affecting successful transmission and accurate placement of the focus. Positioning the focus at approximately 20-22mm within the tissue required a probe tip of length no longer than 15mm, with the tip lying flush to the transducer. In figure 4.15, the beam converges at 81.7°. Over the length of 15mm, the probe tip converges and the resulting aperture at the contact surface is 8.6mm which ensures all of the beam is transmitted and not internally reflected. With a 2mm probe tip wall thickness, the contact area was 12.6mm, just over the desired 10mm.
With the view that ultrasound has been validated as a means of measuring thickness values, a quick validation exercise was done to ensure the ultrasound probe in this work would produce reliable results. The ultrasound probe was mounted in a water tank approximately 450mm wide by 300mm deep by 350mm high. In the bottom of the tank a piece of perspex was placed that had been measured with a Vernier calliper to be 3.5mm thick. The ultrasound beam was propagated through the perspex and from the resulting A-scan trace, the front and back surfaces of the perspex were found to be 1.3µs apart. With the speed of sound through perspex equal to 2680 m/s, the thickness of the perspex, measured by the ultrasound, was found to be 3.48mm. This result was satisfactory because the error of 0.02mm was considered to be due to the interpretation of the A-scan trace.

4.2.2.1 Resolution

The axial resolution (along the central axis of the transducer) of A-scan ultrasound is defined as the distance of the minimum separation of two surfaces that give rise to two identifiable echo signals. This distance should be at least one pulse width of the beam. If the pulse width is very long, reflections from surfaces too close together will merge into one echo. Therefore a shorter pulse width results in better resolution. The axial resolution for the transducers used on this project is as follows. For a 0.2µs pulse propagated through a soft tissue medium with a speed of sound of 1540m/s:

\[
0.2\mu s \times \frac{m}{s} \times \frac{1s}{10^6 \mu s} \times \frac{10^3 mm}{1m} = 0.308 mm
\]

This is considered acceptable for measuring soft tissue depths.

This chapter has presented the A-scan ultrasound equipment and the modifications necessary for successful incorporation into this project. Additionally, characteristic behaviour of ultrasound fields has been discussed to provide a basic background for the reader. The results of the ultrasound work are presented in Chapter 8.
5.0 Spatial Registration

Chapters 3 and 4 described acquisition of surface anthropometric data and soft tissue thickness of a human head. This information is valuable in itself; however, the combination of these data sets into one comprehensive data set, resulting in a solid soft tissue model, is unique and more informative as a design tool. The title of this chapter, "Referencing Methods" refers to the techniques used to identify in each individual data set a "world" reference frame, enabling aggregation of the data resulting in soft tissue detail related to anatomical location.

5.1 Objectives

The underlying rationale in building a soft tissue shell was that for every surface point, a known vector could be subtracted, in 3D space, to define the position of the inner surface of the soft tissue shell. The magnitude of the vector corresponded to the thickness measurement of the ultrasound. The orientation of the vector was dependent on the orientation of the ultrasound signal when propagated through tissue and reflected off the underlying bone surface or air pocket. As mentioned in Chapter 4, the highest amplitude echoes within the signals are recorded when the sound is propagated normal to the internal reflecting surface which is not necessarily normal to the outside surface of the head. Therefore to get an accurate inner shell shape (hopefully a shape similar to the skull), the orientation of the ultrasound probe against the outer surface was a critical piece of information to be incorporated into the model.

A type of "spacing system" was needed to record location and orientation data of the ultrasound probe tip at the surface of the skin, relative to the reference frame of the laser scanned data at a resolution of ±1mm and ±0.5°. A facility to acquire space data upon command by a PC was a requirement to be able to record the position of the ultrasound probe tip nearly simultaneous to the acquisition of the ultrasound signal.

The most common type of registration system used with ultrasonic scanner is a mechanical arm. A mechanical arm is usually comprised of potentiometers recording electric resistance, and therefore three degrees of movement. It has been incorporated into B-scan systems to give the two-dimensional spacing information that defines B-scan different to A-scans with its one-dimensional nature. Since B-scan has become a primary diagnostic tool, mechanical arm systems are very common and reliable within the limitations of the arm engineering. We initially acquired a second-hand mechanical arm to determine if this type of spacing system was appropriate to our
work. It was immediately apparent using this type of device was not going to work for the following reason. When using our ultrasound equipment, it is necessary for the operator to have a full six degrees of motion to get the signal of maximum amplitude. This particular mechanical arm was limited to three degrees of movement and although devices are available enabling six degrees of movement that could obtain the resolution of ±1mm, they are more expensive than other methods of acquiring spatial data, such as acoustic or magnetic tracking systems.

King et al. 120 attached an acoustic tracking system to a B-scanner to obtain three dimensional spatial information for each B-scan image. Three sound emitters were fixed to a triangular plate which was attached to the transducer. Overhead of the subject, four point microphones were positioned to detect the emitted sound waves and thus, calculate the position and orientation in space of the transducer. This type of system was not considered viable with our A-scanner because of the need to take data 360° around the head, instead of in planes from one side of the body. Hence for much of the time, the sound emitters would not be "visible" to the microphones and the spatial data would not be recorded.

As mentioned in Chapter 2, Ganathapy et al. 129 used, in a setup similar to King et al., a magnetic spacing system instead of the acoustic system. They attached a magnetic sensor to a B-scan transducer to acquire position and orientation data of each B-scan slice. The approach was similar to our work with the difference being they acquire three dimensional spatial information for an array of points rather than per point, as required for this project.

5.2 Equipment

A magnetic-based, spatial acquisition system was chosen because it met the requirements mentioned above and is sensitive to most positions and orientations within a defined volume. Magnetic spacing systems in general consist of either one or two magnetic field transmitters and one to four sensors (figure 5.2). The transmitter is secured to a stationery surface while the sensors are attached to the object whose position and orientation is desired. Inside the transmitter are three orthogonal coils which when sequentially electronically excited in each of the axis directions, produce a magnetic field in the respective axis plane (figure 5.1). Each sensor also contains three orthogonal coils which all detect the transmitted magnetic field from each of the x, y and z coils in the transmitter. The magnitude of the sensed voltage is determined by
magnitude and direction of the magnetic field, and thus, the position along each reference axis. The ratios of the voltages determine the orientation in space of the sensor. Thus six degrees of freedom: \( x, y, z, \alpha, \beta, \gamma \) are an output from the system's calculations.

![Figure 5.1: Magnetic fields produced with POLHEMUS transmitter.](image)

This work used two magnetic spacing systems. The first system, SPACYN, was donated by the DRA and was purpose-built for use in an aircraft cockpit to monitor pilot head motion in flight. Unfortunately, the control unit failed during the first measurement period in that the data acquisition was false, which was confirmed by its lack of repeatability. As it was an older model, it was not repaired. It is important to note this in conjunction with the author's comments on the ultrasound data collected with the SPACYN system in Chapter 8. However, because of the SPACYN system becoming obsolete, nothing of the validation work for it will be discussed.

It is with the second system, 3D Space\(^\circledR\) by POLHEMUS, that most of the data was acquired. The POLHEMUS system is similar to the SPACYN in operational theory only. Because it is commercially marketed, it is much more user-friendly. The system comprised of one programmable control unit, a power supply, one transmitter and two sensors, as seen in figure 5.2. There are numerous options in driving the control unit. For example, sending the command "u" via a serial link to the control unit changes the default units from inches to centimetres. Likewise, the command "P" instructs the control until to sample. Other commands enable the data output format to be altered, or change the sampling technique. For this work, only the commands "u", and "P" were used. The control unit is interfaced to a computer via an RS-232 serial link with the following settings made within the control program.
Figure 5.2: Photograph of POLHEMUS system.

The manufacturer's specification of the system is stated below:

Position Coverage: Best performance with receivers operating in a volume of ±101.6mm to ±711.2mm. However, the system will yield values up to ±1524mm.

Static Accuracy: Position - 2.54mm RMS
Angular - 0.75° RMS

Resolution: Position - 0.762mm
Angular - 0.1°

Output Update Rate: 30 times/second for 2 sensors

Power Supply: 120V, 60 Hz, 60 W
A series of validation exercises was done to ensure this specification. The raw data results are listed in Appendix A. Below is a brief description of the exercises and the results.

Static Accuracy:

Both receivers were mounted in receptacles (see section 5.3 for a description of the receptacles) at arbitrary positions, pictured in figure 5.3. The procedure was to allow the system to sample 30 times, and data was recorded at an approximate 2 second interval. The results are the standard deviations from the means and are shown in Table 5.1.

<table>
<thead>
<tr>
<th>Static Accuracy</th>
<th>Receiver 1</th>
<th>Receiver 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>X</td>
<td>0.66mm</td>
<td>0.19mm</td>
</tr>
<tr>
<td>Y</td>
<td>0.17mm</td>
<td>0.24mm</td>
</tr>
<tr>
<td>Z</td>
<td>0.13mm</td>
<td>0.12mm</td>
</tr>
<tr>
<td>α</td>
<td>0.04°</td>
<td>0.02°</td>
</tr>
<tr>
<td>β</td>
<td>0.03°</td>
<td>0.02°</td>
</tr>
<tr>
<td>γ</td>
<td>0.05°</td>
<td>0.03°</td>
</tr>
</tbody>
</table>

Table 5.1: Static Accuracy

The values result in an RMS value for receivers one and two of 0.69mm and 0.19mm, respectfully, for position and 0.43° and 0.026°, respectively, for orientation. These figures compare favourably to the manufacturer’s specification as listed above.

Resolution

The coils that register the magnetic fields are embedded in a casing of dimensions 10mm x 10mm x 15mm, which comprises the sensor. To set up a testing apparatus which could accurately measure the movement of the sensor in three dimensional space was thought to be time consuming. The manufacturer’s specified resolution of 0.762mm was considered acceptable. This author had no reason to doubt the manufacturers specification as changes in that order, in the position data, were observed.
Repeatability

After the problems encountered with the SPACYN system, a test was done to check the repeatability of the POLHEMUS. A receptacle for the sensor with a Velcro attachment was fixed firmly to a wooden board. The sensor was positioned in the receptacle and the data was taken. The sensor was removed and re-positioned in the receptacle 10 times, each time the data was recorded. The raw data can be reviewed in Appendix A and the results are listed in Table 5.2. The values represent standard deviations from an arbitrary mean.

<table>
<thead>
<tr>
<th>Repeatability</th>
<th>Receiver 1</th>
<th>Receiver 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>X</td>
<td>0.18mm</td>
<td>0.61mm</td>
</tr>
<tr>
<td>Y</td>
<td>0.22mm</td>
<td>0.30mm</td>
</tr>
<tr>
<td>Z</td>
<td>0.13mm</td>
<td>0.27mm</td>
</tr>
<tr>
<td>α</td>
<td>0.32°</td>
<td>0.10°</td>
</tr>
<tr>
<td>β</td>
<td>0.66°</td>
<td>0.07°</td>
</tr>
<tr>
<td>γ</td>
<td>0.49°</td>
<td>0.25°</td>
</tr>
</tbody>
</table>

Table 5.2: Repeatability

The rms values from the data listed above for position are R1: 0.18mm and R2: 0.69mm and for orientation, R1: 0.5° and R2: 0.16°. Theses values are within the manufacturer's specified Static Accuracy values and were considered adequate for this work.

Validation

A validation exercise was completed which consisted of using the Polhemus sensor/ultrasound probe configuration to spatially outline the rugby ball used previously in the validation of the laser scanner. The validity of the transformation algorithm was also tested using this configuration.

The procedure was to mount one sensor at the top of the rugby ball to monitor its reference position which was stationary in this exercise. The probe/sensor was then positioned in vertical "stripes" down the contour of the ball at approximately 20mm intervals. Sixteen "stripes" were taken in total around 360°. A visual check of the data was done to confirm approximate dimensional accuracy. The data file was
transferred to the CAD system and modelled. The diameter and length of the model were accurate to ±0.25mm. The model was then referenced to a laser scanned model of the ball and the two models visually coincided to our satisfaction.

At certain points, small discrepancies were evident. Some of these errors were due to indentation of the ball from too much applied pressure on the ultrasound probe during the spatial registration. Another probable explanation was that for the laser scanned model, there were more data points from which the model was calculated, resulting in little error due to interpolation. The model from the Polhemus data was based on approximately 130 points, resulting in a larger interpolation error. The results and thus, equipment, provided results valid enough to proceed with the data acquisition.

5.3 Referencing Protocol

In this section the overall referencing protocol is discussed. The details of procedure, the acquisition software and the algorithms are discussed in full in Chapter 6.

To be able to reference the POLHEMUS data set to the laser scanning data, a common datum, identifiable in both data sets, had to be designated so that the two sets of surface data could be modelled and merged together. The first attempt was to use the sensor receptacle pictured in figure 5.3. The receptacles are made of nylon with a groove cut to the shape off the sensor, enabling it to lie firmly within. One receptacle has a flat underside to which Velcro was attached to hold the receptacle firmly in
The sensor receptacle was attached by Velcro to the swimming cap the subject wore while being scanned. After the laser scanning, the sensor was placed in the receptacle for the duration of the ultrasound measurement.

Two problems immediately arose using this method. Although the receptacle is easily identifiable in the laser scanned data, the actual datum lay within the sensor and thus, within the receptacle and was not easily identifiable. More significantly however, was the movement of the skull cap with respect to the head. The hair acting as a bearing, and the elasticity of the swimming cap material contributed to large movements in the position of the receptacle. An elastic rubber strip was placed circumferentially around the head in an attempt to increase the friction between the swimming cap and the head. This was not successful and no other way could be found to ensure the elimination of sliding movement.

The referencing method that was adopted into the data acquisition protocol was inherently more accurate. An overview of this technique is given next. The full data acquisition procedure is discussed in Chapter 6. During the laser scanning, three green light-absorbing dots, 3mm in diameter and with adhesive backing were positioned in a triangle on the subject's forehead. Due to the light absorption when the subject was
scanned, these three dots appeared as marked "bad" points in the file on the otherwise good data of the forehead. The centre of the dots was detected with an accuracy of ± 0.25mm which is the detectable accuracy of the laser data in the CAD package.

When the laser data was modelled on the CAD system, the three dots formed a triangular plane of a known size, position and orientation in space. By manoeuvring the plane in space, the shell of the head was translated and rotated to a desired position for later alignment with the POLHEMUS data.

The dots were left on the subject's forehead during the ultrasound and corresponding spatial acquisition. The receiver (R1) that monitor's the head movement was directly adhered to the forehead using double-sided sticky tape. A swab of alcohol rubbed across the forehead prior to attaching the procedure, ensured a good adhesion between the skin and the receiver. An arbitrary datum position of R1 was registered to which all of the subsequent data was later transformed. This was necessary because of the subject's movements.

Next, the spatial data was taken around the head using the second sensor (R2). Receiver 2 was attached to the ultrasound probe by a receptacle that screwed on to the end of the probe, figure 5.3. The position of the three dots was taken first and stored in the header of the data file of the ultrasound position and orientation. When the spatial position was registered, the distances away from the transmitter in the x, y and z direction were taken along with the roll (γ), elevation (β) and azimuth (α) of R2 (Figure 5.4). The central axis of R2 was positioned in line with the central axis of the ultrasound probe so that the angles of orientation were coincident for the receiver and the ultrasound probe, making further calculations simpler. Because the spatial data recorded was for R2 positioned at the end of the ultrasound probe, it was necessary to transform the co-ordinates to the tip of the probe which was in contact with the surface of the head. This was done using trigonometry and is described in full detail in section 6.4.1.
With the known surface co-ordinates the POLHEMUS data was reconstructed three-dimensionally on the CAD system, with the three-dots forming the same triangular plane as in the laser scanned data. Through a series of translations and rotations, the two triangular planes, and likewise the spatial data from the POLHEMUS and laser, were positioned coincident.

Details about spatial data transformations, changes in strategy for acquiring the spatial data, and the algorithms for "subtracting" the spatial data from the CAD model are given in the next chapter.
6.0 Data Acquisition Methods and Processing

The next sections discuss in detail the techniques used to acquire the laser, ultrasound and POLHEMUS data. Also included are descriptions of the software and data processing methods used to control the hardware and process the raw data.

6.1 Laser Data Acquisition and Modelling

The method used to derive anthropometric measurements in the past has been to aggregate individual measurements of, e.g. a length, or a volume, where the dimension of interest was one of several taken from a standard measurement protocol of discrete features. The use of electronic imaging allows greater flexibility in selecting such features, however, the method of aggregation tends to be that of identifying the value of the selected parameter, e.g. leg length, mouth width, breast volume, for each individual image data set and averaging that parameter over the population sample.

One objective of the programme of work for the sponsor was to attempt to develop a 3D database that would allow the user to display "typical" or "representative" heads on screen in 3D, and to examine statistically shaped facial features. There are serious obstacles to achieving this. The head is an irregular asymmetric shape with no geometric or proportionally consistent features or landmarks. This makes automatic registration, control and manipulation of the head model difficult as well as raising the question of which feature or landmark to use as the datum in the aggregation.

The sponsor requested a system that would allow the user to have freedom of choice when determining the datum criteria, depending on the region of the face under examination. Originally, the plan was to decide upon a fixed physical datum to enable development of the aggregation techniques. This datum was positioned apart from the head at the end of a rigid rod, hung over the centre of rotation, which could be seen in each laser scan. This is also the rod mentioned in section 3.3 (and is shown in figure 3.13) which aids in interpolation that enhances the dome of the head. The position of each head relative to the pole had to be consistent, so that when aggregated, the heads would all fall within a larger volume or envelope that would be statistically representative of the sample of head sizes.

Section 6.1.1 next discusses the protocol used in the laser scanning to acquire 3D data for each head and the methods of aggregation are reviewed in section 6.1.2.
6.1.1 Laser Scanner Data Acquisition Protocol

The specifications of volunteers for this study were that of being young healthy adults of both sexes from the normal student population between the ages of 18-30yrs. The total number of head scans in the data base numbers three hundred, while the number of volunteers for the tissue work was stipulated to be not less than thirty.

The volunteer was first told about the safety standards of the laser scanner. At its point source, the laser is a 1mW, Class III infrared laser. According to European standard EN60 825, for a 700nm wavelength laser, the maximum permissible exposure at the cornea is:

\[ I_{\text{mpa}} = 18\times t^{0.75}\cdot m^2, \quad \text{where} \quad 1.8\times 10^{-5} < t < 10^4 \]

Eq. 6.1

The entire laser scanning process takes approximately 15 seconds during which time each eye is exposed to the laser for less than 1 second. At the position of the subject, 1 metre from the laser source, the beam is diverged over 1 metre so that the intensity of the laser is 1\mu W/1000mm\(^2\). Substituting this into equation 6.1 results in a safe maximum exposure time of \( t = 2.1 \times 10^4 \) seconds.

The sponsor required that the eyes be open for the scan for the purpose of examining variations in inter-pupillary distance. However, scanning the eyes so that they appear parallel and looking forward proved to be a difficult problem. The most successful results were produced when the subject relaxed his/her eyes so that they were unfocused. This seems to reduce eye movement because the subject was not focusing from sight to sight as he rotated. From the large sample of scans taken it appeared that most pupils diverge approximately ±2mm from the parallel position.

The subject was then asked to don the skull cap to cover the hair and enhance reflection of the laser profile from the top of the dome, and sit in the laser scanner chair. Once the subject was seated comfortably, the alignment protocol was performed.

Investigation into the possible methods of consistently positioning heads was needed to give the aggregated results initial credibility, since at the early stages it was thought the CAD capabilities for manoeuvring the head model in space may require too much computer processing time and memory. If it was possible to align each head in
position and orientation by a definable protocol, it would ease further processing of the data. This approach was valuable for interactive manipulation of groups of head data during the development stage.

The literature has shown that most attempts to standardise head position come mainly from the fields of anthropometry and orthodontistry. The two most common reference positions are the Frankfurt Plane (or, Frankfurt Horizontal) and the Natural Head Position (or, Resting Head Position). The Frankfurt Plane is defined as the plane formed between the tragus (point F) and the lower orbital rim in a horizontal position as in figure 6.1. Because of anatomical variations and asymmetries, the resulting head position can be significantly different between people. This position does not necessarily have a relationship to an individual's head posture based on comfort or normal physiological positions. The use of the Frankfurt Plane as a standard seems to be more commonly used when the clinicians' interests lie in acquiring anthropometric dimensions of the head without any information on posture and the effect on function.130-132.

![Figure 6.1: Illustration of the Frankfort plane. Taken from Croney (1980) 133.](image-url)
The Natural Head Position (NHP) is the head posture a person most naturally holds and is dictated by musculo-skeletal conditions, growth trends and respiratory needs. It is indefinable in terms of cranial facial landmarks but has been said to be more representative of actual physical appearance. Cooke (1990) showed in a five-year natural head position reproducibility study that showed when taking cephalograms of children, after five years the change in NHP was not significant. The amount of data to quantify the NHP is limited because uniform methods for measuring the NHP have not been established.

The purpose of aggregating the head scans is to derive statistical variations in normal head sizes and features. The original work of Downs (1956), Bjerin(1957), and Moorees and Kean(1958) showed that the NHP varies from the Frankfort Horizontal by approximately 5°. The NHP is an individual's most comfortable and repeatable position and although not necessarily consistent between people, it was chosen as the standard in the protocol for the laser scanning. It was considered useful to our sponsor to have the added information of postural differences in head position.

Movement of the head is possible through six degrees of freedom. The NHP ensures a returnable position of the head relative to the person's body and also sets the elevation (tilting about the frontal plane). To set the azimuth (rotation in the horizontal plane about the vertical axis) and pitch (lateral tilting or rotation about a sagittal axis), as well as the vertical, lateral and forward position of the head relative to the reference rod (figure 6.2), the laser beam from the scanner was used for further alignment, according to the following procedure.

1. The volunteer was seated in the laser scanner chair and asked to look in a mirror at the image of his or her own eyes. According to Lundstrom (1991), Luthy, in 1912, defined this procedure as the best way of obtaining the NHP. This position sets the elevation so the volunteer is asked to hold still, only moving upon this researcher's request.

2. Next the chair was raised until the dome of the head is just in contact with the reference rod.

3. The subject was then rotated 90° where the chair was adjusted fore and aft to align the auditory meatus in the plane of the laser beam. The subject was rotated a further 180° to ensure the other auditory meatus lay in the plane of the laser as well. Since the auditory meatus are not located symmetrically and are 180° apart.
on the head, a discrepancy occurs. When aligning the laser beam, the chair was adjusted by approximately half of the distance of the discrepancy so that a mean position was taken.

4. The subject was rotated so that the laser beam illuminated the approximate midline profile of the face. It was sometimes necessary to make fine adjustments when establishing the lateral position, azimuth and pitch of the head. These adjustments did not affect the already established vertical, lateral and elevated positions.

![Diagram of head with six degrees of motion](image)

Figure 6.2: Six degrees of motion of the head.

When the subject was positioned, the scanning procedure began. The software that drives the scanner has a menu with six options:

<p>| | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Calibration</td>
<td>4. Registration</td>
</tr>
<tr>
<td>2. Acquisition</td>
<td>5. Acquisition Setting</td>
</tr>
<tr>
<td>3. Display</td>
<td>6. Quit to Dos</td>
</tr>
</tbody>
</table>

Option number 2, "Acquisition", requires that the scanning intervals discussed in Chapter 3 be set. The best set-up for acquiring good detail around the head, yet better resolution of the face was:
Table 6.1: Laser Scanner Acquisition Set-up.

<table>
<thead>
<tr>
<th>Interval</th>
<th>Increment Degree</th>
<th># of Profiles</th>
<th>Resolution (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>-180 to -135</td>
<td>4</td>
<td>12</td>
<td>0.7</td>
</tr>
<tr>
<td>-135 to -45</td>
<td>1.5</td>
<td>60</td>
<td>0.26</td>
</tr>
<tr>
<td>-45 to +45</td>
<td>1</td>
<td>90</td>
<td>0.17</td>
</tr>
<tr>
<td>+45 to +135</td>
<td>1.5</td>
<td>60</td>
<td>0.26</td>
</tr>
<tr>
<td>+135 to +180</td>
<td>4</td>
<td>11</td>
<td>0.7</td>
</tr>
</tbody>
</table>

The resolution was calculated for a radial distance of 100 millimetres.

There is the capability of saving a default set-up so that each time a scan is taken, the operator only has to answer "yes" or "no" to changing the default.

The next step was to register where the 0° profile is. The implications of doing this were twofold. Firstly, there often existed intra scan head movement, which was seen as a discontinuity as in figure 6.3. This was "shown" normally in the posterior head region so that the data of the face were continuous. Secondly, when the image was displayed, the 0° profile was positioned in the middle of the screen. If the operator was interested in looking at a deformity on the back of the skull, the 0° profile would be registered at that position and every time the image was displayed, that was the view that would first come up. This reduced the need for image manipulation, to see the point of interest. For this work, the 0° was positioned on the mid-line of the face so that a front view of the face was seen when the image was displayed.

The subject was then rotated through -180° to start the scan. By pressing the spacebar on the keyboard, the scan was activated. Throughout the scan, the operator was looking for movement of the subject with respect to the reference pole by viewing a video monitor that displayed what the camera was reading. Movement did not happen frequently because the pole stabilised the head if the subject maintained contact with it. If the subject slouched in the chair, or changed his head position during the scan, this contact was lost and resulted in a step in the image. When this happened, the scan was usually retaken.
Figure 6.3: Step in image resulting from movement of subject during scan.
Subject undergoing laser scan with referencing dots on forehead for referencing.

Ultrasound probe casing and transducer
The operator was also looking for spurious reflections that showed up as random white spots on the video monitor. These occurred when the camera read specular reflections off dust particles in the air or from intense light sources in the room. This problem was eliminated by reducing the signal amplification (gain) control on the power unit.

Upon successful completion of a scan, the data was saved under the following format.

```
##ABNNN.lsm
```

where ## is the year, i.e. 95
A is the sex of the subject, i.e. M or F
B indicates ethnic background, i.e. C=caucasion, AF=Afro-caribbean, etc.
NNN is the scan number, i.e. 1-300
.lsm is the suffix denoting a laser scanned file

The laser scanning protocol for the ultrasound volunteers was identical to the 300 sample with the exception of the placement of the referencing dots as described in Chapter 5. The positions of the dots on the forehead were placed arbitrarily in a triangle until it became clear that automatic location of these dots in the CAD software would be simpler if they were positioned consistently between subjects. An equilateral triangular template, with sides measuring approximately 30mm, was used on the remaining data. The first green dot were place just above the nasion and the position of the other two dots was dictated by the template.

With the exception of the thirty volunteers for the ultrasound testing, the subjects were free to leave after viewing the image of his/her head on the monitor to confirm the scan had been successful.

6.1.2 Protocol for Model Aggregation

The technique used to aggregate the heads evolved from different initial attempts. The first approach was to choose an arbitrary point around which to aggregate the heads. In most cases a bisected point on the interpupillary line was used as in figure 6.4. Any 3D aggregation procedure has to be based on three points to align six degrees of freedom. However, it was anticipated that the alignment protocol discussed in the preceding section would fix the head. Initial attempts at aggregating showed there was
still a large variation in head positions, figure 6.5, and highlighted the need to be able to manipulate the heads within the CAD software.

![Diagram of Aggregation Points](image)

**Figure 6.4: Aggregation points.**

At that stage in the research, software and hardware upgrades in the CAD system made it possible to rotate the head models almost in real-time, opening up a new set of options within the CAD system for aggregation. The head model is a solid model, and thus has a set of properties associated with it. One of the properties that can be extracted from the solid head model is the volumetric centroid. Choosing this point in conjunction with the pupils of the eyes defined a plane that could be oriented in space, figure 6.4. This technique was dependent on the volume defined from the head. The head model shows the neck which is variable in size. The neck influenced the model volume, and hence, the centroid position. To attempt to better control the neck's influence on the model, it was cut away from the head. However, the location of the cut-off was somewhat arbitrary due to the lack of surface anatomy or landmarks in this region. Presently, the cut is taken from just under the chin at a fixed angle. The production of amalgamated heads is shown in figure 6.6. It can be seen the volume
envelope of the aggregated heads decreased when using the centroid method. It was felt that this better represented the size differences between heads.

It has been a group effort to investigate possible methods for aggregation, however the modelling and writing of any code was done by Adrian Hugging. This work is currently on-going to entirely fulfil the requirements of the sponsors with an estimated completion date of the end of October 1995.

Figure 6.5: Amalgamation of heads using bisect on interpupillary line.
6.2 Ultrasound and Space Data Acquisition

In the flow of this research, three methods were used for the acquisition of the ultrasound and space data using the POLHEMUS. The changes were not in the method of taking the data, but in the philosophy of how the data should be taken as a whole. The software remained the same with only slight modifications. It is appropriate to describe the basics of the acquisition that were applicable to all three techniques. The actual changes in strategy are described in section 6.4.

Prior to testing a volunteer, certain preparation of the equipment was necessary to limit the testing times as much as possible. Figure 6.7 is a picture of the entire set-up. To begin, the computer, oscilloscope, ultrasound power unit and power supply for the POLHEMUS were powered up and the oscilloscope put in digital storage mode. The POLHEMUS must go through a bore sighting procedure, during which, the zeroed
orientation of the receivers are registered through a dedicated Microsoft QBASIC program (see Appendix B). When executing the program called "Boresight", the operator was asked to position consecutively each sensor and press any key on the keyboard, sending a POLHEMUS pre-programmed command from the computer to the POLHEMUS system. Each sensor was to be positioned in the same orientation for the transformation algorithms used later in processing the data, to work effectively. This was achieved by placing the flat side of the sensor on a fixed template. This action zeroed the orientation about the Z (α) and Y (β) axis, figure 6.8. The orientation about the X (δ) axis was set using a small spirit level designed for use with a camera and tripod. The error using this technique was demonstrated by this researcher to be ±0.5° Once bore sighting was completed, the POLHEMUS system was ready to use.

Figure 6.7: Acquisition equipment.
Preparation of the ultrasound probe involved filling it with a water medium within the probe tip. A piece of surgical rubber glove was stretched over the end of the probe tip and a ring was pushed over the sheath, holding it in place. This sheath contained the water and was replaced each time as the material weakened with age, allowing air bubbles into the water medium when holes developed. The entire probe was then submerged in distilled water and any air bubbles are released. The probe tip was screwed into the casing, trapping the air-free water inside. The probe is completely dried before being connected to any electrical equipment.

Once the equipment was ready for testing, the volunteer was asked to sit in a chair positioned approximately 150mm from the electro-magnetic transmitter of the POLHEMUS, and one of the sensors (R1) was attached to the volunteer's forehead as discussed in Chapter 5. The other sensor (R2) was placed in the receptacle attached to the back of the ultrasound probe.

The data acquisition was driven by the "Grabdata.bas" program discussed in section 4.2.1. When executed, the program prompted the user for a choice of acquiring both ultrasound and space data simultaneously, or either one individually. These options were made available as the acquisition protocols changed to accommodate the
The first task, prior to taking any ultrasound data, was to take the position of the sensor attached to the head (option 1 from the menu in figure 6.10). All subsequent data was to be transformed to this reference position. Next the position of the three dots was taken to reference the POLHEMUS data to the laser data. The procedure was that the computer repeatedly sampled the POLHEMUS system, until the operator hit "R" on the keyboard. The reference position of sensor R1 was completely arbitrary. The purpose was to monitor the movement of the head away from a known position (the reference position) so that any deviations away from the reference position (e.g. positions 1, 2 or 3 in figure 6.9) could be accounted for in the final ultrasound probe position. The position of the three dots was taken by placing the probe onto each dot consecutively and pressing "P" on the keyboard. The reference and dot positions were stored in a file "SabREF.txt" where "ab" represented the initials of the volunteer.

The next step was to choose option 4 from the menu. This enabled each ultrasound and corresponding spatial position to be recorded. The program first asked for the code (initials) and the anatomic region. The program then sampled the ultrasound trace on the oscilloscope at approximately 0.5 second intervals. When the operator saw the desired signal, pressing any key on the keyboard triggered the oscilloscope to acquire the ultrasound data, and the POLHEMUS, to acquire the spatial information.
It was difficult to know the exact delay between when the operator accepted an ultrasound signal and the POLHEMUS was sampled. This was dependent on the speed of the computer and the command travelling via the RS-232 link. However, it was faster than eye movement from the oscilloscope to the VDU of the computer and therefore, it was thought that errors due to hand movement during this delay were negligible.

**Figure 6.10: Ultrasound and Space acquisition software menu.**

For each sampled point, the corresponding ultrasound and space data were stored in separate files of approximately 96kbyte size. This was to aid in the identification of the data. Within each anatomical region the data was taken in horizontal rows and vertical columns at 5mm intervals, figure 6.11. This was particularly useful when interpreting the ultrasound signals because the spatial data corresponding to each ultrasound point was not modelled until after the ultrasound signal was interpreted. By coding the filename and consulting corresponding details taken when acquiring the data, the approximate position of each ultrasound point was known. The filename code for each point was:

```
Filename: Uabcd##.txt Sabck##.txt
Where: ab = first and last initial of the subject name.
       cd = code for anatomical area
       ## = row and column where the data was taken within a zone.
```

The program prompted the user to indicate if the sampled data was acceptable by answering "Y" or "N". If the data capture was successful, the data was downloaded to the computer under the appropriate filenames from the oscilloscope and POLHEMUS.
temporary storage memories. A "N" response started the sampling process over again. After each successful sample, the operator had the option of moving to the next row in an anatomical region, or quitting, by hitting "R" or "Q", respectively. To change the anatomical region, the operator could enter "N" which caused the computer to prompt for a new anatomic code.

Figure 6.11: Anatomical zones showing example positions of ultrasound data.
6.3 Data Processing

In its raw form, the data format for each spatial point collected is a set of Cartesian coordinates and the orientation of the sensor:

Sensor 1  \[X_1 \ Y_1 \ Z_1 \ \alpha_1 \ \beta_1 \ \gamma_1\]
Sensor 2  \[X_2\ Y_2 \ Z_2 \ \alpha_2 \ \beta_2 \ \gamma_2\]

Where the rotation around \(X_1 \rightarrow \gamma_1\), rotation around \(Y_1 \rightarrow \beta_1\) and the rotation around \(Z_1 \rightarrow \alpha_1\).

The format of the output file from the Hitachi scope is a series of numbers which represent the voltage amplitude at the sampling interval. Both sets of data have to be processed to result in an output file.

Processing the raw spatial data involves the use of another Microsoft QBasic program written by this author, called "Pol4.bas". The program first prompts the user to type in the initials of the subject. It then reads in all of the reference data from "SabREB.txt" and calculates the \(X_{Pp} \ Y_{Pp} \ Z_{Pp}\) of each of the referencing dots and places this information in the header of the final data file "OPOSab.usd". \(X_{Pp} \ Y_{Pp} \) and \(Z_{Pp}\) represent the distances from sensor 1 in its original reference position to the spatial point located on the head at the contact position of the ultrasound probe. The program prompts the user for the anatomical code and the number of rows and columns within each zone. It reads in the appropriate data files and calculates again the \(X_{Pp} \ Y_{Pp} \ Z_{Pp}\) of each data point and writes the results to the final data file. Figure 6.12 highlights the key functions of the programme that were applied to calculate the location of the referencing dots and the ultrasound signals.
Figure 6.12: Flow diagram of spatial transformation program.

With the aid of figure 6.13, the calculation of $X_{pf}$, $Y_{pf}$, $Z_{pf}$ can be demonstrated. The first step is shown in 'A' below - the transformation of the data read at sensor 2, to the probe tip end, P.
Sensor 2 was mounted so that its central axis coincided with that of the probe. Thus the orientation of the sensor was also the orientation of the probe. With $X_2$, $Y_2$ and $Z_2$ the co-ordinates of point 2 and $L_p$ the length of the probe, the position $X_p$, $Y_p$, and $Z_p$ was calculated by:

$$X_p = X_2 - (L_p \cdot \cos (\gamma_2) \cdot \cos (\alpha_2))$$  \hspace{1cm} \text{Eq. 6.1}
$$Y_p = Y_2 - (L_p \cdot \cos (\gamma_2) \cdot \sin (\alpha_2))$$  \hspace{1cm} \text{Eq. 6.2}
$$Z_p = Z_2 + (L_p \cdot \sin (\gamma_2))$$  \hspace{1cm} \text{Eq. 6.3}

The co-ordinates of point P were calculated for the orientation the head when the ultrasound was taken, which was usually different from the orientation of the head at the reference position. Diagram 6.13b illustrates this. The head position was monitored by sensor 1 so that the change in orientation of the head from the reference position was known. The change in head position was calculated by taking the difference between the angles of sensor 1 at the time the ultrasound data was taken and the head's reference position and it's position when the ultrasound was taken. The position of sensor 1 at the reference position was then derived by the standard algorithms for rotating a point in space through three angles $^{137, 138}$.

$$X_R = (\cos \alpha_{R-1} \cdot \cos \beta_{R-1})X_1 - (\sin \alpha_{R-1} \cdot \cos \beta_{R-1})Y_1 + (\sin \beta_{R-1})Z_1$$
\[ Y_R = ((\sin \gamma_{R-1} \cdot \sin \beta_{R-1} \cdot \cos \alpha_{R-1}) + (\sin \alpha_{R-1} \cdot \cos \gamma_{R-1})) X_1 \]
- \[ ((\sin \alpha_{R-1} \cdot \sin \beta_{R-1} \cdot \sin \gamma_{R-1}) - (\cos \alpha_{R-1} \cdot \cos \gamma_{R-1})) Y_1 \]
- \[ (\sin \gamma_{R-1} \cdot \cos \beta_{R-1}) Z_1 \]

\[ Z_R = ((\sin \alpha_{R-1} \cdot \sin \gamma_{R-1}) - (\sin \beta_{R-1} \cdot \cos \gamma_{R-1} \cdot \cos \alpha_{R-1})) X_1 \]
+ \[ ((\sin \beta_{R-1} \cdot \sin \alpha_{R-1} \cdot \cos \gamma_{R-1}) + (\cos \alpha_{R-1} \cdot \sin \gamma_{R-1})) Y_1 \]
+ \[ (\cos \gamma_{R-1} \cdot \cos \beta_{R-1}) Z_1 \]

These angle differentials, \( \alpha_{R-1}, \beta_{R-1}, \) and \( \gamma_{R-1} \) were also used to derive the co-ordinates of point \( P_f \) using the same equations.

\[ X_{pf} = (\cos \alpha_{R-1} \cdot \cos \beta_{R-1}) X_p - (\sin \alpha_{R-1} \cdot \cos \beta_{R-1}) Y_p + (\sin \beta_{R-1}) Z_p \]

\[ Y_{pf} = ((\sin \gamma_{R-1} \cdot \sin \beta_{R-1} \cdot \cos \alpha_{R-1}) + (\sin \alpha_{R-1} \cdot \cos \gamma_{R-1})) X_p \]
- \[ ((\sin \alpha_{R-1} \cdot \sin \beta_{R-1} \cdot \sin \gamma_{R-1}) - (\cos \alpha_{R-1} \cdot \cos \gamma_{R-1})) Y_p \]
- \[ (\sin \gamma_{R-1} \cdot \cos \beta_{R-1}) Z_p \]

\[ Z_{pf} = ((\sin \alpha_{R-1} \cdot \sin \gamma_{R-1}) - (\sin \beta_{R-1} \cdot \cos \gamma_{R-1} \cdot \cos \alpha_{R-1})) X_p \]
+ \[ ((\sin \beta_{R-1} \cdot \sin \alpha_{R-1} \cdot \cos \gamma_{R-1}) + (\cos \alpha_{R-1} \cdot \sin \gamma_{R-1})) Y_p \]
+ \[ (\cos \gamma_{R-1} \cdot \cos \beta_{R-1}) Z_p \]

The difference in the x, y, and z directions, between \((X_R, Y_R, Z_R)\) and \((X_{pf}, Y_{pf}, Z_{pf})\) was calculated as \((X_p, Y_p, Z_p)\). The process is repeated for all the data and each derived \((X_p, Y_p, Z_p)\) was written to file as described above.

Interpretation of the ultrasound signals and the extraction of the thickness data is the dedicated subject of Chapter 7.

6.4 Introduction of the Zonal Approach

The above acquisition option was used for the acquisition of data from 9 heads, when the strategy was to subtract a thickness vector from the POLHEMUS data. This strategy was accurate, but was not successful because of the low resolution of the POLHEMUS data compared to the laser data. In figure 6.14, the orientation of sensor
as propagated through the tissue. The thickness measurement (Th) was defined as the length from the probe tip to the reflecting surface underneath, giving the soft tissue layer thickness along the wave path. Since this thickness dimension was in the same orientation of the probe, the co-ordinates of the inner shell of the soft tissue thickness were found using the equations 6.1-6.3. By substituting the length of the probe (L) by the length of the probe plus the thickness measurement (L + Th), the resulting x, y and z co-ordinates of the inner surface of the soft tissue shell were calculated.

![Figure 6.14: Calculation of co-ordinates of thickness shell.]

The inner surface co-ordinates were modelled on the CAD system using the reference technique described in section 5.3. The soft tissue shell was formed by the combination of the outer layer of the POLHEMUS and the inner surface of the soft tissue layer.

In principle this technique worked. However, problems arose because of the low resolution in quantity of the POLHEMUS data. Since a spatial co-ordinate was only taken for every ultrasound point, which in total is approximately 80-100 points on one side of the face, the resulting model was disappointing in detail compared to the laser scanned model. Taking enough data to significantly improve the POLHEMUS model meant time commitments of hours for each subject, which could not be expected of the volunteers.

The strategy changed to taking multiple probe position points to one ultrasound signal to increase the spatial resolution of the outer POLHEMUS shell. The protocol changed to take a cluster of approximately 5-10 spatial data points and within that
cluster, to acquire on ultrasound signal representing the entire cluster. Each cluster was about 1cm². Option 4 from the menu of "Grabdata.bas" was used with a repetition loop added which sampled the POLHEMUS a set number of times for every sample of the ultrasound signal. The resulting spatial data sets increased to 800-1,000 points in number, however this resolution shell was still poor compared to the laser scan resolution. In the end the new protocol was confusing and it was difficult to take the ultrasound in an orderly format.

The ultrasound data was suggesting that within certain areas the thickness measurements were less variable than had been expected. Enough data had been collected to justify another approach to incorporating the thickness data into the CAD model. A new protocol was established for the collection of the remaining data that involved zones of uniform thickness. These anatomical zones, where thickness variations were thought to be insignificant, were physically identified by palpation of muscles and bone landmarks. The border of the zone was then outlined using the POLHEMUS and these borders were modelled in the CAD system. These data were matched up with the laser scanned data using the reference dots, and the mean thickness vector for that zone was subtracted from every laser data point within the modelled POLHEMUS zone. This is described below.

The problem of insufficient data becomes less important with this approach, as the zonal definition implies non-variability within the zone and hence, every point within the zone can be assigned the same thickness. Effectively a thickness zone data set has the same size as the laser data, approximately 70,000 points.

The resulting file format that is exported to the CAD package is:

First line: Zone code (used to designate the anatomical region)
Lines 1-4: \( X_{PF} \quad Y_{PF} \quad Z_{PF} \) (4 co-ordinates that outline the boundary of the zone)
Final line: Thickness value (average thickness value for the Zone)

The function of the thickness algorithm was to subtract thickness values from the laser scanned surface shell. When the technique was originally to subtract a thickness
vector from each spatial data acquired with the POLHEMUS, both the magnitude and the orientation of the vector were needed. With the development of the zonal hypothesis, the orientation of the thickness vector at each point within the zone was normal to the skin's surface because it had the same thickness.

The POLHEMUS data set (which now contains the data points that outline the zones) and the laser data sets are combined using the method discussed in section 5.4. The data that lies within each zone is extracted and for each of these points, the soft tissue thickness is removed normal to both the horizontal and radial tangents. Figure 6.15 shows the geometry of these calculations. The radial scan line is the profile or vertical cross-section (figure 6.15a) of the head at the point of interest. The horizontal scan line (figure 6.15b) is the horizontal cross-section.

![Radial and horizontal profiles and the geometry for subtracting soft tissue thickness.](image)

In order to calculate the internal co-ordinates representing the soft tissue layer from each point of the laser scanned data, the slope of the surface at each point must be known to calculate \( \theta \) and \( \delta \) and therefore \( dx \), \( dy \), and \( dz \) from the equations:

\[
\begin{align*}
    dw &= t \times \sin \theta \\
    dz &= t \times \cos \theta \\
    dx &= dw \times \cos \delta \\
    dy &= dw \times \sin \delta
\end{align*}
\]
For the areas such as the eyes, nose and ears, because of the irregular shapes, the above algorithms were not applied. The thickness values were not constant so the zonal approach was not appropriate. To proceed with this work, the area were given a temporary, thin, constant thickness value of 1mm so that "holes" would not be present in the data when modelling. The result of this on the model was the appearance of features that should not be there in areas of the nose cavity and the eye orbits.

The resulting model is a skull-like surface as in figure 8.25. Comments about these results are found in Chapter 8. The algorithms in this section were developed by Adrian Huggins, a research assistant for the DRA contract.

This chapter has been an overview of the methods required to produce the results presented in Chapter 8. Further discussion on how the zones were define can also be found in Chapter 8.
7.0 Signal Interpretation

The task of interpreting an A-scan signal can vary in difficulty depending on the information being extracted. The main concern of this work was to extract soft tissue thickness values from the maxillo-facial region of the human head. To do this, an assumption was made that the strongest echo would occur at the tissue/bone or tissue/air boundaries. This assumption, although usually true, does not always occur because of attenuation in signal intensity. Large magnitude echoes are not necessary for accurate extraction of thickness data. An experienced operator can identify the reflecting surface echo, either through familiarity of the signal or monitoring the affects of changing the probe orientation. Additionally, since most of the signal is reflected at the bone or air boundary, no echoes appear after the soft tissue/bone or soft tissue/air boundary.

The phenomena discussed in Chapter 4, e.g. scattering, absorption, and irregular reflecting surfaces, reduce the amplitude of the pulse as it travels through the tissue, as does reflection due to impedance differences at the boundaries. Table 7.1 lists the percentage of the incident pulse which would be reflected back at each boundary layer due to the impedance differences between tissue layers. The reader can refer to figure 4.3 for a schematic representation of the boundaries.

<table>
<thead>
<tr>
<th></th>
<th>Skin/fat</th>
<th>Fat/Muscle</th>
<th>Muscle/air</th>
<th>Muscle/bone</th>
</tr>
</thead>
<tbody>
<tr>
<td>(Z_1) (kg m(^{-2}) s(^{-1}))</td>
<td>1.63 x 10^6</td>
<td>1.38 x 10^6</td>
<td>1.7 x 10^6</td>
<td>1.7 x 10^6</td>
</tr>
<tr>
<td>(Z_2) (kg m(^{-2}) s(^{-1}))</td>
<td>1.38 x 10^6</td>
<td>1.7 x 10^6</td>
<td>0.0004 x 10^6</td>
<td>7.8 x 10^6</td>
</tr>
<tr>
<td>% Reflected (= \frac{100}{\left(\frac{Z_2 - Z_1}{Z_1 + Z_2}\right)^2})</td>
<td>0.7</td>
<td>10.4</td>
<td>99.9</td>
<td>41.2</td>
</tr>
</tbody>
</table>

Table 7.1: Percentage of reflected signal at each soft tissue boundary layer.

The figures for this table were calculated using equation 4.13 in Chapter 4. Table 7.1 shows that at the reflecting boundary of muscle/air or muscle/bone, 99.9% and 41% respectively, of the incident pulse is reflected back. In an ideal tissue in which no scattering or absorption occurs, and very little of the signal is reflected at the soft tissue boundaries, this would result in an echo of 34.7mV or 14.4mV. In actuality, the return echo is of the order 10^-4mV. The echoes are amplified in the ultrasound.
control unit by 40dB so that the signal used for the extraction of the thickness has an amplitude of the order 10^4mV.

In the regions such as the forehead and chin where thin soft tissue reduces attenuation effects or it is easier to position the probe so that the angle of incidence is normal, echoes with amplitudes of magnitude between 100-150mV(after amplification) are possible. But in areas such as the cheek and temporal region where the tissue is thicker and hence attenuation increases, or where it is difficult to position the probe normal to the reflecting surface, the return echo can be difficult to detect.

In the next section example traces taken at similar locations on three subjects will be presented and discussed for each of nine regions of the face. It will be shown how thickness values from these traces were calculated and the results tabulated for every point on each subject. Finally the potential interpretation errors and resulting inaccuracies will be explained.

7.1 Example A-scan Signals

The signals shown in figures 7.3-7.10 were taken from the anatomical areas of the frontal bone, temporal region, maxilla, cheek, masseter muscle, bottom mandible, chin and the lips. These example traces have been chosen because they are typical of the signals taken at that point on the face. Discussion on these signals will be broken down into each area to highlight the challenges in interpreting signals from particular areas, and hence extracting thickness values.

Measurements on the signal can be taken with the methods described below, and the results do not depend on the technique used. Nonetheless consistency of approach should be taken on each signal.

The first technique is to locate a cursor on the leading edges in an unrectified pulses shown in figure 7.1. This point corresponds to the front face of the ultrasound transducer.
It was felt that this approach was not the best method to use because it is often difficult to differentiate the leading edge of the pulse from the noise. Point A in figure 7.1 is an example of this situation. A filter could have been applied to the signal to threshold out the noise, however this was not done because the next technique described was easier to apply and more dependable.

Measurements can also be taken from the cross-over point before or after the maximum amplitude echo. For example, points A and B in figure 7.2 are valid and consistently easy to locate.

Figure 7.2: Measurement of pulse placing cursor at the leading edge.

Figure 7.2: Measurement of pulse placing cursor at crossover points before or after the point of maximum intensity.
The convention used for this work was to take point B, or the crossover point after the maximum echo because this point has been shown to be the most reliable marker with the view that point A was sensitive to tissue type and density.

An original expectation of this work was to automatically extract the soft tissue thickness measurements and the thickness of the tissue layers. There have been attempts at computer analysis of A-scan ultrasound signals but with criteria that is difficult to apply to biological tissue, as explained below.

Taylor et. al. 139 have said that attempting to diagnose chronic spleen enlargement from an A-scan trace, they have had to set up signature libraries. The quantitative analysis of A-scan signals with large specular echoes was unlikely to be informative because the amplitude depends upon the random orientation of the reflecting surface with respect to the beam.

Trier et. al. 140 have said that in order to classify tissue by the echo features, it was necessary to know the characteristic quantities of the signal features and the upper and lower variances. Additionally, weighting factors would have to be established which compared the signal features to histology and clinical data. In their work, the A-scan traces used were taken from the human eye. For the comparisons to be valid they had to ensure the same angle of incidence and constant distances between the transducer and the tissue surface. They also stated they were only able ultimately to use the leading echo and return echo because true isolation of the internal echoes was nearly impossible because of the biological influences on the echo structure.

The success of the above work was dependent on very rigid testing criteria and the development of the signal libraries. This was possible to do because the signals they were acquiring were for a specified area with consistent anatomical structure. It was felt that for this thesis including the quantity and variability of data acquired, the above mentioned type of criteria could not be used because of time limitations. The construction of a signal library was thought to be another in-depth study and was not taken on as part of this work. As far as we can see, the only successful automated thickness acquisition was done on signals taken from a non-biological event where the signal characteristics and thickness values were predictable 141.

The signal interpretation discussed in the following sections was manually done and only the gel-soft tissue and soft tissue/bone or soft tissue/air boundaries were identified.
7.1.1 Frontal Bone

Signals taken in the frontal bone area are easy to acquire and to interpret. Figure 7.3 on page 131 shows two signals which are very similar taken at point FB11 in figure 6.11. Point A on both the top and bottom traces represents the gel/tissue interface. This marks the leading edge of the transit of the pulse through soft tissue. Point B on both traces is the reflection occurring at the soft tissue/bone interface. This echo is high in magnitude (approximately 200mV after 40dB amplification) because the soft tissue thickness in which the pulse has travelled is comparatively thin and the opportunities for absorption and scattering are low. The frontal bone is almost parallel to the skin's surface so that it is easy for the operator to position the pulse appropriately normal to the bone. It can be seen in many of the traces such as figures, 7.4, 7.7, 7.9 and 7.11 that the echo from the reflecting bone surface or air pocket decays to the noise level. The maximum reflection at the boundary is caused by the maximum amplitude of the transmitted pulse which has a width of 2μs. The decaying characteristic of the transmitted pulse also causes a decaying characteristic in the reflecting echo. This is more easily seen to occur at the soft tissue/air boundary when most of the signal is reflected back, or at the soft tissue/bone boundary in this soft tissue regions where there are small loses due to attenuation.

7.1.2 Temporal Area

Data from the temporal region in figure 6.11 was difficult to acquire and hence the results presented later in section 7.2 are variable. The difficulties lie in the shape of the underlying bone in this region. Figure 7.12 illustrates this point. It is a top cross-sectional view that shows the shape of the bone is irregularly concave against the skin surface. It is difficult to align the transducer normal to the bone without compressing the soft tissue. The transducer was positioned so that signals were reflected from surface A.
A further difficulty lies in the fact that the superficial temporal artery lies in this region extending from the ear to the scalp. The flowing blood within the arteries affects an ultrasound field by scattering the signal, sometimes to the extent that very little of the original pulse transmits to the reflecting bone surface. When this was found to occur in the temporal region, the ultrasound probe was moved to another point in the area.

Figure 7.4 illustrates two traces taken from this region. Point A is the skin surface echo and point B is the reflection echo from the bone. The reflecting echo in the top trace results in a thickness value smaller than in the bottom trace, which was expected because the points were taken at different sites.

### 7.1.3 Zygomatic Arch

Signals in figure 7.5 from the zygomatic arch are very easy to acquire due to the thinner soft tissue and the good, even bone surface. The signals in this region are always highly echogenic and similar from point to point and between people. This probably can be attributed to attachment areas of the zygomatic major. Point A is the skin surface reflection and point B is the bone reflecting echo. Along the zygomatic bone, at the "ball" of the bone, acquisition of the signals was troublesome. The results are also variable due to the large variations in size and shape of people's cheek bones. In some cases the bone is quite flat and then good data can be acquired. In other instances the cheekbone can be round or even pointed. This has the effect of scattering the ultrasound beam so that very little of the signal returns to the transducer.
7.1.4 Maxilla

Data from the maxilla was acquired with ease in most cases. However, closer to the teeth, the signal was difficult to get because of the ridges created by the roots of the teeth and the teeth themselves. These curved shapes scatter the signal. With practice and care, the probe could be successfully positioned to acquire a signal such as figure 7.6. Point A represents the skin surface and point B, the reflecting bone.

7.1.5 Cheek

The cheek was the most difficult region from which to acquire interpretable ultrasound echoes. Two characteristics of this regions contribute to the difficulty. First the soft tissue is can be thick (on the order of 20mm) which provides more opportunity for the signal to be absorbed and scattered. Secondly there is not a surface (bone) from which the ultrasound can be reflected. There does, however, exist a air pocket that theoretically is a stronger reflector than bone. When the subject has his mouth shut, there is not a large amount of air present. Inevitably the subject was asked to puff his or her cheeks lightly to exaggerate the air pocket without undue distension of soft tissue structure. As with the other regions, point A on figure 7.7 is the skin surface and point B is the soft tissue/air boundary.

7.1.6 Masseter Muscle

This region covers the masseter muscle which overlies the ramus of the body of mandible, thus providing a good reflecting surface. The ramus is smooth and relatively flat. Although the masseter muscle is approximately 14-22mm, enhancing absorption and scattering, it was usually easy to acquire signals because the ramus is a good reflector in that it is relatively smooth and even. Its shape is similar to the contours of the face so that it was easier to align the probe normal which resulted in a high amplitude signal. Figure 7.8 shows a consistently typical signal from this region. It can be seen that the top and bottom traces are very similar although they came from different people. The echo, point B, always lies at the end of the trace. In some instances, the thickness was above 22mm, resulting in a trace that was too long for our acquisition system to capture. This was a rare event and at these points, traces could not be digitised. However they were still visible on the oscilloscope so that the thickness values were taken manually.
7.1.7 Bottom Mandible

Good signals could be acquired on the bottom mandible. The soft tissue thickness is approximately 12-14mm which did not seem to attenuate the signal greatly. The bone surface is fairly even and lies nearly parallel to the skin surface. The signals in figure 7.9 show that the reflecting echo at B has a magnitude of approximately 55mV on the top trace and at least 80mV on the bottom. The signals within this area have different thickness as data is taken towards the jaw line. This is discussed more in chapter 8.

7.1.8 Chin

The chin was another relatively easy location from which to extract an interpretable ultrasound trace except in the extreme cases when the subject has quite a protruding mentus creating an acute angle shape of the bone. In this situation a signal can be acquire by indenting into the soft tissue but the resulting thickness is not reliably accurate. Figure 7.10 shows traces taken from the chins of two subjects. They show similar thickness values with point A being the skin surface and point B, the reflecting surface from the bone.

7.1.9 Lip

It was not difficult to acquire signals from the upper lip region because the reflecting surface is an air pocket between the soft tissue and maxilla or teeth. The reflecting echoes are usually high in magnitude. There is little variation in the thickness of the tissue in this region and this can be seen from the traces in figure 7.11. Again, point A is the skin surface and point B is the reflection from the soft tissue/air boundary.

7.2 Calculation of Thickness Values and the Results

Ideally, calculation of the tissue thickness should be broken down into the individual tissue layers because each layer has a slightly differing speed of sound. The calculation first presented in chapter 4, Eq. 4.11 is as follows:
Tissue thickness = Speed of sound * Time travelled/2

If the tissue thickness is the sum of individual layer thickness', the equation becomes:

\[ T = (D_1 \cdot t_1) + (D_2 \cdot (t_2 - t_1)) + (D_3 \cdot (t_3 - t_2 - t_1)) + (D_4 \cdot (t_4 - t_3 - t_2 - t_1)) \]

where \( t_{1-4} \) are the thickness values for the epidermis, dermis, fat and muscle layers and \( D_{1-4} \) are the corresponding speed of sound values. These values have been proven difficult to establish in vivo which is self-evident in that it is not possible to verify the values until post mortem. Additionally, at this stage in the project it has been shown to be difficult to extract thickness for individual layers. As discussed in section 2.1.4.1 the convention is to take a single value representing the average of the speed of sound through each layer. This value of 1540m/s changes the above equation to be:

\[ \text{Thickness} = \left[ 1540000 \text{ (mm/s)} \times t_{1-4} \text{ (\mu s)} \times 10^{-6} \right] / 2 \]

Table 7.2 lists the thickness values for the traces in figures 7.3-7.11.

<table>
<thead>
<tr>
<th>Region</th>
<th>Figure Number</th>
<th>Time travelled ((\mu)s)</th>
<th>Tissue thickness (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frontal Bone</td>
<td>7.3 top</td>
<td>5.71</td>
<td>4.39</td>
</tr>
<tr>
<td></td>
<td>bottom</td>
<td>6.053</td>
<td>4.66</td>
</tr>
<tr>
<td>Temporal</td>
<td>7.4 top</td>
<td>11.55</td>
<td>8.89</td>
</tr>
<tr>
<td></td>
<td>bottom</td>
<td>16.02</td>
<td>12.33</td>
</tr>
<tr>
<td>Zygomatic Arch</td>
<td>7.5 top</td>
<td>8.878</td>
<td>6.84</td>
</tr>
<tr>
<td></td>
<td>bottom</td>
<td>8.295</td>
<td>6.39</td>
</tr>
<tr>
<td>Maxilla</td>
<td>7.6 top</td>
<td>9.911</td>
<td>7.63</td>
</tr>
<tr>
<td></td>
<td>bottom</td>
<td>11.83</td>
<td>9.1</td>
</tr>
<tr>
<td>Cheek</td>
<td>7.7 top</td>
<td>19.3</td>
<td>14.86</td>
</tr>
<tr>
<td></td>
<td>bottom</td>
<td>15.1</td>
<td>11.63</td>
</tr>
<tr>
<td>Masseter Muscle</td>
<td>7.8 top</td>
<td>22.6</td>
<td>17.4</td>
</tr>
<tr>
<td></td>
<td>bottom</td>
<td>23.43</td>
<td>18.04</td>
</tr>
<tr>
<td>Body of Mandible</td>
<td>7.9 top</td>
<td>19</td>
<td>14.63</td>
</tr>
<tr>
<td></td>
<td>bottom</td>
<td>13.48</td>
<td>10.38</td>
</tr>
<tr>
<td>Chin</td>
<td>7.10 top</td>
<td>8.49</td>
<td>6.54</td>
</tr>
<tr>
<td></td>
<td>bottom</td>
<td>12</td>
<td>9.24</td>
</tr>
<tr>
<td>Lip</td>
<td>7.11 top</td>
<td>9.55</td>
<td>7.35</td>
</tr>
<tr>
<td></td>
<td>bottom</td>
<td>14.3</td>
<td>11.01</td>
</tr>
</tbody>
</table>
Appendix C lists tables showing tissue thickness values for every point taken on every subject of the sample. Discussion of these results occurs in chapter 8.

7.3 Errors and Accuracy

There are a number of factors that can affect the accuracy of the thickness values listed in Table 7.2 and Appendix C. These are:

1. Resolution of the ultrasound.
2. Resolution of the Himes signal analysis package.
3. Interpretation errors due to selection of correct echoes in the signal.
4. Errors in the correct use of the value for the speed of sound.
5. Indentation of the skin when taking measurements.

The resolution of the ultrasound has been stated to be approximately 0.308mm. The Himes package has a resolution of 0.04mm and therefore does not contribute significantly to an overall error. The largest error source must come from the signal interpretation and the following discussion will show why this occurs.

It is convenient to use figure 7.5 for this explanation. It highlights a problem that seems to occur as is illustrated at point B on the top trace. It can be seen that just prior to this point a number of echoes with large amplitudes occur in a cluster. Either these echoes are due to reflections from a surface prior to the bone, such a connective tissue, or they are actually part of the decaying signal. In the first scenario, point B is accurately placed. However if the situation is as the second scenario, a decaying signal, then the appropriate position of B is actually at B2. The wrong position of B can produce an error of up to 0.5mm. With experience the interpretation error in most cases is equivalent to the resolution of the Himes package.

The average speed of sound value for soft tissue, 1540m/s, used by most researchers, results in an approximate error of 0.01-0.024mm per millimetre, depending on the proportion of fat and muscle. For tissue thickness up to 20mm, the error could be as high as 0.48mm.

Finally, an error occurs if the skin is being indented with the probe. This can easily occur but was avoided by using the following technique. Once the signal was located, this author would pull the probe away from the skin until the signal disappeared. The
signal would then be re-established until the point where it was about to disappear again. It was at this point ensuring the least amount of pressure on the skin that the signal was sampled. It is not possible to predict accurately how much of an error occurs from indenting because it is variable depending on the amount of pressure being exerted on the soft tissue by the ultrasound and the properties of the tissue. However an estimate based on experience is and the above technique is 0.1mm.

In the best case scenario, where the interpretation error and indentation have been kept to a minimum, the resulting error would approximate ±0.24mm. If care is not taken to avoid indentation and someone inexperienced extracts the tissue thickness from the signal, the error could be as much as ±1.5mm.

The results are the data acquisition and interpretation are discussed next in Chapter 8.
Figure 7.3: Ultrasound trace from the frontal bone.
Figure 7.4: Ultrasound trace from the temporal region.
Figure 7.5: Ultrasound trace from the zygomatic arch.
Figure 7.6: Ultrasound trace from the round area of the zygomatic bone.
Figure 7.7: Ultrasound trace from the cheek.
Figure 7.8: Ultrasound trace from the masseter muscle.
Figure 7.9: Ultrasound trace from the bottom mandible.
Figure 7.10: Ultrasound trace from the chin.
Figure 7.11: Ultrasound trace from the lip.
8.0 Discussion on Thickness Data and Proposed Zonal Approach

8.1 Development of Zonal Theory

8.1.1 Results of SPACYN Thickness Data

The idea that anatomical zones can be defined in which there is relatively small variation in soft tissue thickness grew from the results of nine initial subjects, five of which are listed in tables 8.18-8.22 in Appendix D. When the data acquisition began, it was assumed that there would be large differences in the thickness around the face.

The ultrasound data in tables 8.18-8.22 was taken while the SPACYN system, briefly mentioned in chapter 5, was obtaining the spatial position of each data point. At that time it was not known that the SPACYN system was failing, so only rough notes were taken on the actual location of the points. Because the SPACYN spacing system malfunctioned, there was no permanent record of the location where the ultrasound was taken other than the rough notes. Further study of the notes and memory recall from this author to attempt to identify the position of the data suggested some possible thickness trends. The listing of data in tables 8.18-8.22 is an attempt to classify it into the present zonal divisions. Some of the data could not be confidently classified which is why many of the columns are missing. A quick study of the data shows the statistical means and standard deviations cover a wide spread. A detailed explanation of the data is not done because it is not considered to be able to be validated. However the data does demonstrates the basis from which the zonal concept grew.

To establish if there was a sound basis for the zonal theory, the first step was to study the anatomical detail of the face. The areas of the eye and nose were not investigated because examination of the anatomy indicated that the zonal theory would not apply to these features. The data listed on "OC" in the tables is data from around the eyes, which was originally thought to be needed for the soft tissue CAD model, but it has not been studied in depth in this research. This will be discussed further in section 8.3.

The next sections reveal the zonal definition based on an anatomical study of the face.
8.1.2 Anatomical Study

The purpose of this part of the study was to determine if there is support for the hypothesis that there is small tissue thickness variation within defined zones on the face. The first approach was to examine the anatomy of major bones of the skull and to subdivide them into regions of constant or flat reflecting surfaces. Once these areas were determined, a study of the muscle anatomy was done to establish how the muscle structures influenced soft tissue thickness. The subcutaneous connective tissue blends with deeper facia that surrounds the muscles. Within the subcutaneous layer lie the fat cells (adipose tissue), which are more evenly distributed on the face. Hence, it was thought that muscle would have a greater affect on thickness changes. Each area will be introduced, described and a discussion of the muscle structure will follow. The reader should refer to figure 8.1 which highlights each discussed area.

The first region is the frontal bone. It is a large area which is relatively flat with the exception of two bulges on the forehead called the frontal eminence. Looking at the surface anatomy, the frontal bone zone is considered to be the region extending from the hairline to the superior orbital ridge, e.g. the top of the eyes. The frontal bone is covered with the occipitofrontalis muscles which are comprised of flat bands over the forehead from the skin of the eyebrow to its superior attachment into the galea sponeurolicus muscle which covers the scalp.

On the lateral side of the orbit lies the temporal fossa. This is a depression in the temporal bone which is deep near the orbit and laterally shallows out above the ear. Figure 7.12 in chapter 7 illustrates a cross-section view. Within this depression lies embedded the temporalis muscle. This muscle is can be 17mm thick at the deepest part of the fossa and grows thinner as it out to cover the side of the head. This muscle is attached to the temporal bone over the entire surface.

Next is the portion of the zygomatic bone that runs from just above the auditory meatus to the medial edge of the nose. The zygomatic process can easily be palpated from the surface of the skin. The anterior surface is consistently flat. Very little muscle overlays this section of the bone. However there are muscle attachments for the zygomatic major which extends from the part of the bone located inferior to the lateral side of the orbit, down to the mouth where it merges with the orbicularis oris.
Figure 8.1: Zonal representation
The masseter muscle also connects on the inferior edge of the process and extends to the ramus where most of the muscle is attached.

The rest of the zygomatic bone, commonly known as the cheekbone, is a highly irregular surface. Two muscles are attached here, the zygomatic minor and the levator labii superioris. Both of these muscles extend and merge into the orbicularis oris.

The maxilla is a relatively flat reflecting surface except for the region along the upper jaw line where there are eminence as a result from the embedded tooth root. The area of the maxilla around the jaw slightly protrudes resulting in a concavity over the entire surface of the maxilla. There are many muscles that lie over and connect to the maxilla. The already mentioned Levator labii superioris, zygomatic major and minor and the orbicularis oris are present. Additionally, the levator anguli oris extends from below the orbit to the mouth where it blends partly into the skin of the mouth and partly into the orbicularis oris. The buccinator arises from the maxilla and mandible and extends into the orbicularis oris.

The final area initially defined was the bottom mandible. This appeared to be very flat and even. The surface of the face seems to follow the curves of this bone. The muscles that cover this region are the masseter muscle described above, the depressor labii inferius, depressor anguli occuli and the risorius. These muscles all have attachments into the orbicularis oris and project indifferent directions to attachments on the bottom mandible. The mantalis also arises from the mandible and inserts into the skin of the chin.

8.1.3 Zonal Definition

The hypothesis was that, based on sample tissue thickness data and the anatomic skeletal and muscular structure as described above, the following zones could be designated in (figure 8.1) The zonal boundaries were identified by palpation of the face. The question of the accuracy of the boundary locations is addressed when the results of the data are reviewed. A statement about the expected thickness trends before the data was collected is also given.

1. **Frontal Bone:** This zone extends from the superior boundary of the hair line to the inferior supraorbital ridge. It covers the entire forehead to the lateral frontal process of the zygomatic bone. Within this zone it was anticipated that the
thickness variations would be minimal due to the flat bone surface and the complete coverage of the occipitofrontalis muscle.

2. **Temporal**: The lateral side of the frontal process of the zygomatic bone marks the superior boundary and the superior edge of the zygomatic bone marks the inferior boundary. It extends medially to the orbital ridge and laterally to the hairline. This zone is simple to locate anatomically. However due to the variable depth of the temporal fossa and the embedded temporal muscle, it was anticipated that differential thickness of the soft tissue would occur, varying from 5mm to 17mm.

3. **Zygomatic Arch**: It was expected that an area along the process, which is highly visible in some people, would have constant thickness. There is little muscle that overlays the process and therefore no reason to suggest high variability in the soft tissue thickness coverage. This zone has been defined as having its boundaries at the superior and inferior edges of the process. The medially boundary has been chosen as the point where the process dips inferiorly which can be felt through palpation. It was felt that just medial to this point are muscle attachments which could affect the variability of the thickness of this zone and therefore, it seemed an appropriate boundary.

4. **Zygomatic bone**: This zone has been defined to run from the lateral border with the zygomatic process as just defined above. Following the curve of the bone to its junction with the maxilla defined the inferior boundary. On the medial side, the boundary is taken up to where the nasal bone and cartilage start to protrude. The superior boundary is the lower ridge of the orbit. Within this zone lay the muscles: the levater labii superioris alaequi nasi, the levator labii superioris, the zygomatic minor and the zygomatic major. It was thought that these muscles, although relatively thin, along with the variable shapes of cheekbones, would contribute to thickness differences within this zone.

5. **Lip**: The lip zone covers much of the maxilla. It extends from the superior boundary of the join with the zygomatic bone down to the inferior boundary of the mouth. Medially it begins at the mid-sagital plane and laterally ends at an imaginary line extending perpendicular to the most lateral edge of the lip. There are a number of muscles in this region and it was not known exactly how they would affect the overall thickness. These muscles are the zygomatic major and minor, levator anguli oris and the levator labii superius, depressor septa, nasalis and orbicularis oris. The muscles do not cross over and with the relatively flat
reflecting surface of the maxilla it was anticipated the thickness variability would be minimal.

6. **Cheek**: This zone was defined with a superior boundary as the inferior edge of the zygomatic process. The inferior boundary is the superior edge of the bottom mandible. It laterally ranges to the edge of the masseter muscle and medially, to the imaginary line perpendicular to the lip as defined above. The buccinator is the main muscle lying within this region. As it is a facial expression muscle it is considered to be relatively thin. The cheek is not attached to any bone surface and with the reflecting surface being an air pocket, it was through there would be small differences in the thickness of this region.

7. **Masseter Muscle**: This zone comprises solely the masseter muscle which is known to be thicker because it is used for mastication and is more powerful. It overlays the ramus so that the zone extends from the superior boundary of the inferior side of the zygomatic process to the inferior boundary of the angle of the bottom mandible. Laterally it begins at the lateral edge of the ramus. The medial boundary is the edge of the muscle itself which can be felt through palpation.

8. **Bottom Mandible**: This zone covers the bony surface of the mandible from the superior boundary of the tooth line to the inferior edge of the mandible. It extends laterally to the medial edge of the masseter muscle. The medial boundary is at the somewhat arbitrary imaginary line drawn perpendicular to the lateral point of the lip, as with the medial boundary of the cheek. The muscles, the depressor labii inferioris, depressor anguli oris and the playsma have attachments on the mandible in the medial side of this zone. Whereas there is no muscle coverage or attachment on the lateral side. Therefore it was expected that an increase in thickness would occur towards the medial boundary. A protrusion in the mandible towards the inferior edge was expected to cause a thinning of the soft tissue along this region.

9. **Chin**: As the name implies this zone covers from the superior boundary of the bottom lip to the inferior edge of the mandible. Medially, the boundary is the mid-sagittal plane. The lateral boundary is the imaginary line at the lateral edge of the lip. The mentalis, orbiculi oculi and depressor labii inferioris muscles cover this region and nearer to the superior boundary they cross so that a thickening in tissue was expected. As with the bottom mandible zone a thinning at the inferior boundary was anticipated.
10. **Brow**: The brow zone lies just above the supraorbital ridge in an indentation in the bone. Since this indentation is not visible from the surface anatomy, it was expected the tissue would be thicker than on the frontal bone. This zone is outlined by the supraorbital ridge on the inferior boundary. The superior boundary is the end of the depression in the bone. The medial and lateral boundaries occur at the corresponding ends of the supraorbital ridge.

The next section will discuss the data collected from each zone defined above. The discussion will include a presentation of the data in spectral plots which visually enhances the results of the data and in some cases, suggests alternatives for the boundary definition of particular zones.

### 8.2 Data Results

The data for each subject is tabulated per zone and listed in Appendix D. Discussion of the data is enhanced with the aid of the spectral plots shown in figures 8.3-8.16. The data in Appendix D was exported to a software package called Stanford Graphics, version 3 by Visual Numerics, Inc. One of the functions of the package was to create colour spectral plots where each colour represented, in this scope of this work, a tissue thickness value. For each zone, a separate colour spectral plot was generated showing the changes in thickness in that zone. The spectral plots for each zone were then fit into a template of the face so that the changes in thickness across the face were seen. The size of each zone was "proportion" to the actual zone on the head.

In the following paragraphs, suggestions are made that there are ambiguities in the data which, retrospectively, may result from using an acquisition resolution which was too low or improper definition of the zonal boundaries. It is suggested that dedicated studies be done to explain these ambiguities. These studies were not completed within this work because of the following reason. The hypothesis of the existence of zones of constant thickness developed from the data and later became part of the original thesis. Investigation into the ambiguities that the data shows is the next step in the progression of this work. It is felt that acquisition of a new set of data, at the suggested enhanced resolution would take at least three months and because of time limitations, was not completed as part of this work.

The data shows that for any one subject the spread of thickness data in the frontal bone zone is minimal. The consistent colour in each of the spectral plots
demonstrates this. The range of thickness mean of the sixteen subjects is from 3.5-5.65mm, while the standard deviations of the mean range from 0.33-0.92mm. The consistency of thickness data was expected for this zone and the results are acceptable. It is reasonable to state that the frontal bone can indeed be classified as a zone with acceptable boundaries.

Moving down the face to the brow, again there exists a consistency of thickness in this zone. The spread of statistical mean between subjects is from 4.16-6.83mm and the range of standard deviation of the mean is 0.09-0.94mm. This region is generally thicker than the frontal bone which defines it as being a zone of its own. There are three subjects, figures 8.7, 8.8 and 8.11, who have large thickness variations of up to 3.0mm. Figure 8.7 shows an increase of thickness at the centre of the brow, for which no anatomical explanation can be given. In figures 8.8 and 8.11 increases in the thickness at the lateral edge of the brow zone can be seen.

This characteristic of a change in thickness at the boundary occurs frequently in the data. At some of the boundaries between zones, abrupt changes in thickness occurs. This is sometimes justifiable based on the anatomy. In most cases it was expected there be an increase or decrease of thickness at the boundaries between zones. For instance, between the lateral side of the brow and the temporal fossa lies the lateral edge of the orbit. This region has not, as of yet, been defined as a separate zone but it seems to have thickness characteristics of its own. It suggests that another zone should be added. The appropriateness in the selection of boundaries will be discussed where relevant to the zones and again in chapter 9 under further work.

Within the temporal zone there exists the large variations in tissue thickness that were expected. It was thought that the variation would follow a trend where the thickest tissue would occur on the medial side and gradually laterally thin out following the slope of the temporal bone. This trend did occur in the data of figures 8.3, 8.4, 8.10, 8.11, 8.14 and 8.15. In the other figures, the changes in thickness seem to be random. The range in statistical mean is from 10.13-14.3mm and standard deviations of the mean, from 0.81-3.14mm. The variations are anatomically justified so that a valid anatomical means of breaking down this zone into further zones is not immediately apparent. It would be a necessary exercise to do a further, in-depth, study of the changes in thickness to determine if an algorithm could be developed that described the variations. Data spaced at an interval of 1mm or less would be needed to derive descriptive algorithms. For this work the data was taken at a 5mm interval to determine if thickness variations were present.
The zone defined over the zygomatic arch contains data which is non-variable for a given subject. One can see the consistency in colour when viewing the spectral plots. Only two, figures 8.13 and 8.15 show a noticeable change in colour within the zone. The range of statistical mean for the subjects is from 5.25-9.72mm while the standard deviations from the mean range from 0.39-1.03mm. In most of the spectral plots, the colour of this zone relative to its surrounding zones is quite distinct. This is expected for the superior temporal zone and the inferior masseter muscle. It is not necessarily true for the medial zygomatic bone zone. The boundary between the two is not as anatomically definite. Figure 8.3, 8.4, 8.8, 8.15 and 8.16 show a large colour change on the medial boundary of the zygomatic bone zone. This change of colour supports the boundary location. However, some of the other figures with a gradual colour change may indicate the zonal boundary was improperly chosen as in figures 8.6, 8.7 and 8.14. Perhaps, such as figures 8.10 and 8.11 suggest, there does not exist a change in thickness between the two zones and therefore they should be combined.

The acquired data has highlighted this boundary definition location as inconsistent between people. This author has the opinion that the data suggests two zones do exist and the question is where to locate the dividing boundary. Further data should be taken on a sample of at least 15 and with a higher acquisition resolution than what was specified for this thesis, with the emphasis on placing the exact location of the boundary.

A discussion of the zygomatic bone zone naturally follows. The data within this zone is variable. The range of statistical mean between the subjects is from 7.14-11.9mm and the standard deviations from the mean ranging from 1.4-2.7mm, are considered to be high. However examination of the spectral plots reveals a trend. Over the rounded part of the zygomatic bone the data seems to be constant. In some cases, such as figures 8.3, 8.5, 8.7, 8.10 and 8.16 the data thins out as it medially approaches the nose. The opposite occurs in figure 8.11. This subject had a very flat cheekbone with a thin soft tissue covering that could be felt through palpation.

In figures 8.3-8.6, 8.8, 8.10, and 8.13-8.15 a thickening in the medial bottom corner of the zone can be seen. This can probably be attributed to the acquisition technique. The data suggests that in most cases, data could have been taken from the maxilla area where the tissue is thicker. This author had difficulties when taking data from this zone because, although the zone may be correctly defined, it is practically difficult to take data in this region. The precise location of the reflecting surface is difficult to
know because the junction of these two regions is not parallel to the skin's surface. Where this author thought the data was being reflected from the cheekbone, the data suggest otherwise.

The data within the zygomatic bone zone taken from the maxilla region which was originally designated as part of the lip zone, has thickness values which are more similar to the data within the cheek zone. Figures 8.3-8.6, 8.10, 8.14-8.16 illustrate this. The data suggests the medial boundary needs to be relocated more towards the mid-sagittal plane. The data is more variable within the zone than what was predicted from the local anatomy. The range of statistical mean between subjects is 11.4-15.6mm and the standard deviations from the mean range from 0.67-3.52mm.

In almost all of the subjects there is a distinct thinning toward the medial side of the zone. In many cases, such as figures 8.5-8.8, 8.12, 8.13, 8.15 and 8.16 the thickness values on the lateral side of the zone are similar to the thickness values of the masseter muscle. In this instance it is inaccurate to suspect the boundary is misplaced or that data represented as the lateral edge of the cheek zone is actually the medial edge of the masseter muscle zone. This is because the edge of the masseter muscle is distinct and thus very easy to locate through palpation. Instead it is reasonable to conclude the buccinator muscle contributes more towards the thickness and there may be a decrease in the thickness of the fatty layer toward the medial side of the zone.

A peculiarity of the data is the abrupt change in colour on the medial boundary to the lip zone. It was expected this would be a gradual change but as figures 8.3, 8.4, 8.7, 8.11-8.16 demonstrate, this was not always the situation. Figures 8.6 and 8.8 do, however, support the observation. This author suspects the data acquisition did not approach the boundary closely enough and there is an actual gap in the data because of the acquisition resolution. It is suggested that this boundary be investigated at a higher resolution in future work.

In many of the spectral plots there is an abrupt change of colour on the superior boundary. This was expected because of the sudden protrusion of the zygomatic bone. On the inferior boundary, there is a gradual colour change as would also be expected.

The data in the lip zone does, in most subjects, remain constant. The statistical mean between subjects vary from 6.1-10.7mm and the standard deviations from the mean range from 0.56-3.2mm. The upper end of the range of standard deviations is due to three subjects with some artefacts in the data consisting of high thickness values.
range of standard deviations of the subjects, less the three with high thickness values, is from 0.56-1.2mm. Figures 8.8, 8.10 and 8.16 show a large variability. However 8.8 and 8.10 also show an increase in thickness on the lateral side of which values are similar to the medial side of the cheek zone, suggesting a cross-over of the boundary during the data acquisition. Figure 8.16 shows a thinning towards the lateral boundary. This is most probably an artefact rather than a characteristic of the tissue thickness because it is not seen in any other subject and the tissue thickness in that area should be higher due to the slope of the maxilla.

On the whole, the results of the data from the masseter muscle zone are satisfactory in that the thickness values are higher than any other point on the face, as was expected. However, there is more variability in the data. The statistical range of mean between subjects was 12.5-16.9mm. The range of standard deviations, from 0.67-2.9mm was considered high. It was expected that all of the spectral plots for the subjects would show constant thickness as in figures 8.6, 8.12 and 8.14. Instead the trend is a thinning of the data towards the inferior boundary, especially the lateral side of the zone as in figures 8.3, 8.4, 8.8, 8.9, 8.13, 8.15 and 8.16. This data suggest the masseter muscle is not of constant thickness and does not extend as far as the angle of the bottom mandible. The colour changes at the boundaries of the cheek and zygomatic arch zones have already been discussed in the relevant paragraphs.

The bottom mandible zone shows very high thickness variations in the statistical mean and standard deviations, 8.86-13.5mm and 0.79-3.32mm, receptively. The data is misleading and this can be seen in the thickness charts listed in Appendix D. The problem is due to the drastic thinning of the tissue towards the inferior boundary because of the underlying slope of the mandible. Data usually was taken in two rows. The top row show thickness values usually ranging from 10-13.5mm whereas the bottom row is usually from 8.86-10mm. Figures 8.3, 8.5, 8.7, 8.10, 8.11, and 8.14-8.16 show this decrease toward the inferior boundary. This division suggests that the zone should be redefined as two zones. There are no external landmarks which could be used for a zonal redefinition. There is a protrusion on the inferior of the mandible that can be felt through palpation. The suggestion for further data acquisition would be to use the superior edge of the protrusion as a boundary separating the bottom mandible zone. It is expected the data would be more consistent in thickness within these two zones supporting the existing data.

Finally, the chin zone has shown thickness variability over the mental protuberance. People with strong protuberances have a thinner soft tissue covering such as figure
8.12. The strong protuberance also produces a "concavity" in the bone just above it. The data in figures 8.6, 8.9, 8.11, 8.13, and 8.15 suggest that this concavity creates a bed for thicker soft tissue. The range of mean values within the chin zone was from 7.2-10.3mm and the range of standard deviations from 0.79-3.32mm. Figures 8.3-8.5, 8.8 and 8.16 show a consistency in thickness.

### 8.3 Comparable Data

The existing thickness data available are limited for comparison. The only published thickness data for adult Caucasian are the tables of Rhine and Moore recorded in Table 8.1, along with the data obtained from this study. The mean values listed for the University of Surrey are the average of all the values that fell within the particular zone for the 12 male and 4 female subjectes. The error is the Standard Error of Measurement derived as:

\[
S\text{.Meas.} = \frac{\text{Standard deviations of the measurements}}{\sqrt{2}}
\]

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<th>Anatomic Region</th>
<th>University of Surrey</th>
<th>Rhine and Moore</th>
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<tr>
<td>Brow</td>
<td>5.1 ± 0.6</td>
<td>8.25</td>
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Table 8.1: Comparison of existing thickness data.

The frontal bone, zygomatic arch, and lip thicknesses are comparable to the Rhine and Moore data. This was expected because, within these zones, the data are least variable so that the mean value is more representative of the thickness.

The zygomatic bone, cheek and masseter muscle zones thicknesses, as they have initially been defined, have a wider spread so that when comparing the mean values of, for example, the male data of 8.93mm, 17.70mm and 15.63mm, respectively, to the Rhine and Moore data of 9.75mm, 18.50mm and 17.75mm respectively, the Rhine and Moore data is thicker. This is explained by the fact that the mean values of the Rhine
and Moore data comprise of the mean of one point from many subjects. Whereas the mean value of the University of Surrey data comprise of the mean of all the data points within the zone from many subjects. It has already been shown that the thinning of tissue near the boundaries accounts for the lower mean values. The datum point from the Rhine and Moore set which has been compared to the zygomatic bone zone is actually located on the maxilla, proximal to the boundary for this zone. If a comparison is made between this point and actual data taken from that region, values ranging from 10-15mm have been taken. The Rhine and Moore data point taken on the cheek is near to the lateral boundary defined in the cheek zone, from which values range 10-19mm. Finally, the point on the masseter muscle taken from the Rhine and Moore data lies at the medial boundary of the muscle for which we have obtained values up to 17mm.

The data which show a significant discrepancy are for the chin and the brow zones. The Rhine and Moore data is significantly higher than what has been obtained through this work. However, a point to point comparison does not as easily explain the differences as with the other zones. The Rhine and Moore data, at 11mm, is higher than the data from this work which range around 9mm. There is doubt that the differences are due to the ultrasound technique since it has produced comparable results on the whole. It could be due to tissue fluid swelling in the cadaver. It is indicated on the Rhine and Moore data that there sample size was small for some measurements, although sample numbers are not given. It may be possible, with the sample size for this work being small, that this discrepancy between data is not statistically important.

With the ultrasound data established as being comparable to other existing data, it was felt valid to begin to incorporate them into the soft tissue model as discussed next.

8.4 Soft Tissue Modelling

The dedicated software of the laser scanner was developed as an imaging tool for maxillo-facial surgery planning and anatomical illustrations. It visualised three dimensionally to allow analysis of structures for better diagnosis. Since the purchase of the equipment in autumn, 1991, the dedicated software was modified to enable importation of grey-level data from other techniques such as MRI, CT and B-scan ultrasound. Whilst it is an excellent visualisation tool, the extraction of exact three
dimensional co-ordinates is not possible, inhibiting its value as a competent design tool.

An objective of this work was the exportation of the data into a Computer Aided Design tool to facilitate analysis of the precise geometry of a solid model. Dimensioning, sectioning and properties analysis are also possible as is the incorporation of the data from other techniques such as the soft tissue thickness data from the ultrasound. Within CAD packages it is possible to examine the intersection between two objects using BOOLEAN operations. This is directly applicable to the study of fit between head and aircrew equipment which is the underlying interest of the sponsors. The data in the CAD package is also in a format conducive for export through established links to Finite Element Analysis (FEA), Computer Aided Manufacturing (CAM) and Stereo Lithography (SLA). FEA has been tried within the scope of the overall project and will be discussed in section 8.4.3.

Recently, there has been a decision from the sponsors to move the work previously done in the CAD packages MEDUSA and CADDS5 to a different package VARIMETRIX. This has not caused any of the scan acquisition algorithms to be changed.

The next section displays and discusses the results to date of incorporating the zonal tissue data into the solid model within the CAD package VARIMETRIX. The modelling work is an on-going part of the overall project and the results presented are an important first phase.

8.4.1 Results

It was considered unnecessary to present the graphics at full resolution because of the problems of computing power versus time. The graphical results in this section are of a lower resolution of 1/25 and even at this resolution the shading time for each graphic was approximately 15 minutes.

The technique developed to "subtract" the thickness data from the laser scanned solid model using zones has proven to work. However, there exist some obvious problems that need to be corrected to improve this technique. Within figures 8.18-8.20 are the front and lateral views of the zones positioned over the laser scanned data. It can be seen in each figure how the boundary edges are approximately linear when the actual boundaries follow contours of the bone structure or muscle. For computational
simplicity in our initial modelling attempts, the zones have been outlined with the POLHEMUS spacing system with only four points, designating the corners of the zone. This had the affect of the boundaries not being correctly defined so that large gaps result between the zones. It this were the only problem, the corners of some of the boundaries should match up for all of the zones on the maxillas and bottom mandible. In most cases, it can be seen that this is not the case. Clearly, acquisition of the zonal points has to be done more carefully. This would occur quite naturally if the zones were defined by points located approximately every 5mm along the boundary. The data has been acquired in this manner. However, we have not yet seen the results because of the present inability to model the zones with more than a four-point-defined boundary.

With the exception of the frontal bone zone, the zones are considered to be of a reasonable size and location considering the four-point boundary definition approach. In the frontal bone zone, the upper left corner has been located too low and this is most certainly an operator error. As stated before, the frontal bone zone is superiorly defined by the hair line. In this case, the intersection between the hair line and the temporal line which laterally defines the zone, was mistaken.

Figures 8.18 and 8.19 highlight the difficulties with the scalp. It was intended that the scalp be considered one zone from which a uniform thickness should be taken. However, the algorithms use to match the zones to the laser scanned data could not cope with the scalp defined as one zone because the algorithms are written to identify the zones within a 180° hemisphere. The scalp zone falls into both of the hemispheres. Next, the data was taken as two zones comprising the dome of the head and the rest of the scalp. The results of this approach was that a zone could not be located on the dome of the head because of the reason stated above. Towards the end of the data collection, the scalp was divided into two zones, one each for the left and right hemispheres of the head. Since each zone was defined in the 180° partition, this approach did work and the zones were modelled. Figure 8.19 also show a triangular shape to this zone. This is because the point which would have defined the lower right corner was missing from the data set.

It would not reveal much to present the results of the soft tissue modelling as a soft tissue shell. The appearance would be that of a laser scanned solid model as in figure 6.13, only the inner skeletal surface would be missing. Instead it is felt to be more valuable to show the would be skeletal surface once the soft tissue shell has been removed from the laser data. Figures 8.21-8.25 show on the left, the low resolution
solid model from the laser scanned data. The picture on the right is the skull-like result of removing the soft tissue shell.

Figures 8.21-8.23 show the application of the techniques using the spatial data that came from the SPACYN system. One can clearly see the zones seem to be congregated along the mid-sagittal plane. The temporal zones are impinging on the orbits, the bottom mandible zones are located at the chin and the zone sizes appear to be smaller.

The results in figures 8.24 and 8.25 include the spatial data taken from the POLHEMUS. The location of the zones appears to be better. In figure 8.24, the temporal lines, orbits and cheekbones have some definition, remembering that holes in the skull are filled in. The data in figure 8.25 begins appear more skull-like. In the front view, the depth cannot be seen, so that the cheekbones are obscured. Figure 8.25 would begin to look even more realistic with an increase in resolution.

8.4.2 Validation

The individual tools used to acquire the data for the model have been validated by this research, the manufactures of the equipment, and in the case of ultrasound by many groups in the past. An acceptable method of validation may be to compare the resulting skull to that taken from a CAT scan. Alternatively, sections through a MRI model could be compared to sections of the CAD soft tissue model if the orientation of the MRI slices was known.

An attempt to reconstruct MRI data in the CAD package was done for this project. Data was extracted manually from each MRI slice, at a 2.5° increment for each of 90 slices. This data consisted of the angle and radial length to the outer skin surface and the inner soft tissue/bone boundary. Unfortunately, the attempt to model the data failed to the extent that viable comparisons could not be made.

It is felt that the key to the success of this technique is the boundary definition, which is dependent presently on algorithms being written for VARIMETRIX which will incorporate better zonal definition. This work should be completed prior to any future validation exercises. At that time, the MRI data should be exported to a dedicated visualisation package such as ANALYSE so that cross-sections can be taken for validation of our tissue model.
8.4.3 FEA Application

The data from a soft tissue model has been exported to an FEA package called PATRAN by other researchers employed by the sponsors. Preliminary work is being done which involves looking at tissue deformation under load. To date, three types of loads have been applied to the soft tissue model. The inner surface of the shell has been fix to approximate soft tissue attachments. A 1 Newton load was applied to the nose to simulate a punch in the nose. A 10-g downward load has been applied to the entire head. Finally, a contact load has been applied from the contact surface of an oxygen mask to the soft tissue.

The mechanical property information has come from testing done as part of the sponsored project. Through \textit{in vivo} tensile and indentation testing, values for Young's modulus and Poison's ratio have been extracted for the soft tissue as single viscoelastic material.

Computational time significantly increases with additional parameters and restraints so that a compromise of lower resolution occurs. This work is in its initial stages and continues to evolve as the quality of the soft tissue shell increases and the mechanical properties testing progresses.
Figure 8.2: Key to spectral plots.
Figure 8.11

Figure 8.12
9.0 Conclusion and Future Work

9.1 General Comments

The original goal of this work was to match soft tissue thickness values of the face to anthropometric surface data of the head and develop this data into a solid soft tissue model for use in a computer aided design system. The equipment chosen to acquire the data was a laser scanner developed by University College London and an A-scanning ultrasound kit by Cutech, Ltd. The techniques to reference spatially the data together had to be decided upon. It was additionally expected that the A-scan trace would reveal individual thickness values of the layers that comprise the soft tissue, e.g. epidermis, dermis, fat and muscle.

With modifications, the laser scanner acquires approximately 70,000 data points over a 360° revolution of the head with a ±0.5mm circumferential resolution and a ±0.2mm radial resolution. The A-scanner, comprising of a 13.2MHz pulsed and focused ultrasonic beam, has a 2.7mm lateral resolution and a 0.31mm axial resolution. From the A-scan trace, thickness values are extracted with an accuracy of ±0.25mm.

The spacing equipment finally settled upon to reference the laser scanned and ultrasound data is a magnetic spacing system called POLHEMUS by 3DSpace. With a resolution of 0.762mm linearly and 0.1° angularly it can acquire data with a position accuracy of 2.54mm rms and the orientation, 0.75° rms.

A new referencing method has been developed which "subtracts" zones of constant thickness values, spatially identified with the POLHEMUS, from the laser scanned model. The zones have been defined based on preliminary thickness data and the anatomic skeletal and muscular structure of the face.

The initial results have shown to produce a soft tissue model with a skull-like surface underneath. The soft tissue model in the CAD system Varimetrix is a valuable design tool for looking at the fit of air crew protective equipment. It has also been exported to an FEA package, PATRAN, where the affects of g-loads have been demonstrated.

The following sections give concluding remarks on the various aspects of the work and propose areas where further work could be done.

9.2 Testing Equipment
At the onset of this work in 1991 the equipment chosen was considered to be most appropriate for the type of data which was needed. In the last five years, experience and the introduction of new technology have not significantly altered these choices. The particulars are addressed in each of the subsections below.

9.2.1 Laser Scanner

Laser scanning is becoming more sophisticated and commonly used for imaging. Cyberware's advance since the early 1990's include linear scanning which overcomes such problems as swept volumes and enables imaging of odd geometric shapes. The user population of the UCL laser scanner is increasing. With the additional features of differential scanning and an increase in hardware memory, clear, quality images can be acquired with increased resolution. The performance of this scanner is satisfactory, especially with the established data links to the CAD. Other than hardware maintenance and possible software upgrades, further modifications to the surface data acquisition equipment are probably not necessary in the near future.

9.2.2 Ultrasound

In using the current ultrasound hardware, various aspects have been identified as needing modification. The first would be to include a hand held trigger to capture the data. The acquisition is controlled by a keyboard and this has proven to be physically inconvenient. Many times, reaching for the keyboard has caused the signal to be lost. A finger trigger held in the opposite hand to the probe, or even a trigger on the probe itself would positively affect the ability to acquire good signals.

Another, more significant change would include a modification to the sampling hardware. At the moment, the digital storage oscilloscope has a sampling storage memory of 1,000 words which is equivalent to 25μs of the signal, or 19.25mm. While reflecting boundaries lying deeper than this (up to 40mm) can be digitised on the oscilloscope, the data cannot be saved. All measurements deeper than 19.25mm in the presented data was hand recorded. Either a digital storage oscilloscope with larger storage capabilities or an input/out card in a PC which could sample at 50MHz would work. To change to a I/O card would call for a complete software change and might not be necessary with a better model of the Hitachi oscilloscope.

The question arises regarding the choice of A-scan, versus B-scan or MRI, as the choice method of data acquisition. Admittedly, experience has shown that extracting
layer information from the A-scan trace is at best, ambiguous. Additionally, as discussed in Chapter 7, the ability to process thickness data automatically from the A-scan trade is difficult, if at all possible, without setting up echo signature libraries and using sophisticated signal processing techniques. Whereas, some of the latest results using an ultrasound backscatter microscope are showing very distinctive layer information of the epidermis and dermis from a 100MHz transducer with 30µm axial and 94µm lateral resolutions.

The important link is the computability of grey level data with a CAD system. At the University of Chicago, Evenhouse et. al. used a CT scanner to input surface data from a skull of an entomb mummy to a CAD package. As far as this researcher can find, there have been no attempts to input B-scan or MRI data into a CAD system. The grey-levels represent amplitudes or densities which can not be modelled. The extraction of the 3D position of the tissue boundaries from this type of data is necessary for use in a CAD system. While A-scanning has provided accurate results for the overall tissue thickness, to advance the CAD model to include tissue information, a tissue acquisition system such as B-scan should incorporated in the future work of this project.

9.2.3 Referencing System

The POLHEMUS equipment has satisfactorily provided accurate spatial information for use with A-scan ultrasound. Since the zone definition has become separate to the ultrasound data acquisition, it is not necessary to mount the sensor on the back of the probe. Rather a "pen" for the purpose of digitising is offered as an alternative with the POLHEMUS equipment. Using a slender object with a small tip would ease the acquisition of the boundary data. Although a shift to another mode of tissue data acquisition, as discussed above, would include spatial information, the POLHEMUS system would still be needed to provide the orientation data of the slice information.

9.3 Data Collection Protocol

The acquisition protocol seems to work well. It is considered to be simple yet thorough. The necessary change is to acquire the boundary data with more than four points as has been done in this work. Careful attention should be given to this, taking the data no more than 5mm apart. One suggestion might be to divide the data
acquisition into two sessions. In the first session, the laser scan would occur and the boundaries would be defined. The second session could then be devoted entirely to the ultrasound acquisition. The main advantage would be a reduction in session time so that the operator would not be affected by fatigue during the acquisition of the ultrasound data.

9.4 Results

The thickness data results show that in some zones like the frontal bone and zygomatic arch, there is little thickness variation as was expected by the anatomical definition. Other zones as the cheek and bottom mandible are more difficult to define because of sloping or irregular bony surfaces. The data shows variability within these zones. It is recommended that for the bottom mandible, temporal and masseter muscle zones, a further study is done to determine if algorithms could better define the thickness trends within the zone. The collection of more data would support current zonal definitions and help define the necessary changes.

Once the zones had been confidently defined and validated through the collection of more data, the next step would be to investigate the nose, eyes and mouth to determine if patterns exist which can be incorporated into the CAD model. The variation of thickness data between the sample population could be calculated for different body size and weight to see what influence this has. It is suspected that for most people, the thickness in any zone are similar and it is the presence of thin to thick fatty layers which controls the population differences, coupled to different bone structure sizes.

Further work could be to scan a skull and build up the soft tissue with the zone layer as Evenhouse et al. 41, 42 and Vanezis et al. 24 have done. The final results could be compared, although their techniques have not been published.

It can be seen how once the zones are more firmly established and the soft tissue model validated, this work will be useful to the different disciplines mentioned in chapter 1, especially in conjunction with finite element work. To the plastic and reconstructive surgeon, the zone definitions would help to establish what is "normal" for soft tissue, enabling them to successfully plan surgical procedures. Zones of known thickness would aid the forensic facial reconstructor who attempting to add soft tissue to a skull either manually or through the rapidly developing computer algorithms.
9.5 Soft Tissue Model

The preliminary results presented in chapter 8 show promising results, but they are not yet complete, due partially to modifications in the zonal definitions that need to be made. Also, on the modelling end, the ability to incorporate full zonal definitions rather than the existing four points procedure needs to be developed. Until the algorithms can cope with the full boundary definitions, further validation of the soft tissue model against more data should not be done.

Once this has been successfully completed, the future work on the modelling end could be unlimited. Already as shown in chapter 8, the preliminary model has been successfully exported to the FEA package and the affects of g-forces and pressure forces are being demonstrated.

The development of the soft tissue model based on the established zones will greatly enhance the ability of the sponsors of the work and other interested research groups, such as the manufacturers of sporting equipment to provide more comfortable and safer protective equipment.
References


133 Croney, Anthropometry for Designers, 1980).


Appendix A: Validation data for Polhemus magnetic spacing system.
### Static Accuracy

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02 = Receiver 2

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Repeatability data:
01 = receiver R1
02 not tested (ignore data)

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02= receiver R2
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Appendix B: Dedicated software for borsighting of Polhemus sensors, ultrasound and spatial data acquisition, reformatting of ultrasound data and calculation and reformatting of 3D spatial coordinates.
***Boresight.bas is a program used to zero in space the Polhemus sensors.

OPEN "COM1:9600,n,8,1" FOR RANDOM AS #2
PRINT "PLEASE BORESIGHT SENSORS"
PRINT "SENSOR 1: PRESS RETURN WHEN READY"
INPUT S1$
PRINT #2, "B1"
PRINT "SENSOR 2: PRESS RETURN WHEN READY"
INPUT S2$
PRINT #2, "B2"
CLOSE #2
END
This software will acquire ultrasound and space data via an RS-232 link from the Hitachi VC-645 digital storage oscilloscope and the Polhmus spacing system.

```basic
DECLARE SUB COLLECTBOTH (NAME$)
DECLARE SUB COLLECTSPACE (NAME$)
DECLARE SUB COLLECTULT (NAME$)
CLS
DIM ULTRA(1 TO 10)
I = 1
J = 0
PRINT "SUBJECT'S INITIAL'S?"
INPUT NAME$
OPEN "COM1:9600,n,8,1" FOR RANDOM AS #2

'Choosing an option to take space data or ultrasound data or both

DO
CLS
PRINT "CHOOSE YOUR DATA COLLECTION OPTION."
PRINT ""
PRINT "1: TAKE POSITION OF REFERENCE POINTS."
PRINT "2: COLLECT SPATIAL DATA ONLY."
PRINT "3: COLLECT ULTRASOUND DATA ONLY."
PRINT "4: COLLECT BOTH SPATIAL AND ULTRASOUND DATA."
PRINT "5: QUIT."
INPUT OPT$
IF OPT$ = "1" THEN
PRINT "POSITION SUBJECT AND PRESS 'R' TO REGISTER REFERENCE POSITION OF SENSOR 1."
INPUT R$
REF$ = "C:\DOSSPACE\" + NAME$ + "RF.TXT"
OPEN "O", #5, REF$
PRINT #2, "u";

IF R$ = "R" THEN
   PRINT #2, "P";
   RSPA$ = INPUT$(94, 2)

   PRINT #5, "REF."; RSPA$
END IF

PRINT "REGISTER POSITION OF THREE REFERENCE POINTS BY PRESSING ANY KEY."
PRINT "POINT 1:" 
INPUT PS
PRINT #2, "P";
PP$ = INPUT$(94, 2)
PRINT #5, "PT1:"; PP$;
```
PRINT "POINT 1 REGISTERED!"
PRINT "POINT 2:";
INPUT PS
PRINT #2, "P";
PPS = INPUT$(94, 2)
PRINT #5, "PT2:"; PP$;

PRINT "POINT 2 REGISTERED!"

PRINT "POINT 3:";
INPUT PS
PRINT #2, "P";
PP$ = INPUT$(94, 2)
PRINT #5, "PT3:"; PP$;

PRINT "POINT 3 REGISTERED!"
CLOSE #2
CLOSE #5

ELSEIF OPT$ = "2" THEN CLOSE #2
COLLECTSPACE (NAME$)

ELSEIF OPT$ = "3" THEN COLLECTULT (NAME$)
ELSEIF OPT$ = "4" THEN COLLECTBOTH (NAME$)
ELSE
END IF
LOOP UNTIL OPT$ = "5"

END

'Subroutine Collectboth sequentially samples the DSO and the Polhemus

SUB COLLECTBOTH (NAME$)
CLOSE #2
DO
I = 1
J = 0

PRINT "WHAT ANATOMICAL REGION?"
INPUT REG$

DO
CLS
PRINT "OPENING COM PORTS"
OPEN "COM1:9600,N,8,1,CS,DS,CD,RB1000" FOR RANDOM AS #2
OPEN "COM2:9600,9,8,1,CS,DS,CD,RB1000" FOR RANDOM AS #1

DO
'IF J MOD 5 = 0 THEN
'PRINT J MOD 5
PRINT "SAMPLING DSO"

DO
PRINT #1, "S1"
LINE INPUT #1, RTN$ RTN = ASC(RTN$)
IF RTN <> &H41 THEN
   PRINT "ERROR STATUS =", HEX$(RTN)
ELSE
   END IF
   A$ = INKEY$
LOOP UNTIL A$ <> ""

PRINT #2, "u";
PRINT #2, "P";
PRINT "SAMPLING POSITIONING SYSTEM"
PRINT "LOADING SPATIAL DATA"
SPAS = INPUT$(94, 2)
PRINT SPAS

'LOADING OF ULTRASOUND DATA
PRINT "LOADING DSO DATA"
PRINT #1, "R1(1000,1000,A)"
RTN$ = INPUT$(4000, 1)

ELSE
PRINT #2, "P";
PRINT "SAMPLING POSITIONING SYSTEM"
PRINT "LOADING SPATIAL DATA"
'SPAS = INPUT$(94, 2)
PRINT SPAS

'END IF

PRINT "DATA LOADED!"
PRINT "DATA OK?"
INPUT Y$
LOOP UNTIL Y$ = "Y"
CLOSE #1
CLOSE #2
J = J + 1

'Writing to file
NEWS$ = "C:\DOS\DATA\U" + NAME$ + REG$ + LTRIM$(STR$(I)) + LTRIM$(STR$(J)) + ".DSO"
HPSS$ = "C:\DOS\SPACE\S" + NAME$ + REG$ + LTRIM$(STR$(I)) + LTRIM$(STR$(J)) + ".TXT"
PRINT HPS$

OPEN "O", #4, HPS$
'IF (J - 1) MOD 5 = 0 THEN
OPEN "O", #3, NEWS$
PRINT NEWS$
PRINT #3, RTNS$
CLOSE #3
'END IF
PRINT #4, SPA$
INPUT B$
IF B$ = "R" THEN
   I = I + 1
   J = 0
END IF

CLOSE #4

LOOP UNTIL B$ = "N"
PRINT "PRESS 'Q' TO QUIT"
INPUT C$
LOOP UNTIL C$ = "Q"
END SUB

'Subrouting Collectspace samples spatial data from the Polhemus.

SUB COLLECTSPACE (NAME$)
   DO
      I = 1
      J = 0
      PRINT "WHAT ANATOMICAL REGION?"
      INPUT REG$

      DO
         CLS
         PRINT "OPENING COM PORTS"
         OPEN "COM1:9600,n,8,1" FOR RANDOM AS #2
      DO
         PRINT #2, "u";
         PRINT #2, "P";
         PRINT "SAMPLING POSITIONING SYSTEM"

         'R1 COMM AND LOADING OF SPACE DATA
         PRINT "LOADING SPATIAL DATA"
         SPA$ = INPUT$(94, #2)

         PRINT SPA$
         PRINT "DATA OK?"
         INPUT Y$
LOOP UNTIL Y$ = "Y"
CLOSE #2

J = J + 1

HPSS$ = "C:\DOS\SPACE\$" + NAME$ + REG$ + LTRIM$(STR$(I)) + LTRIM$(STR$(J)) + ".TXT"
PRINT HPSS$
OPEN "O", #4, HPSS$
PRINT #4, SPA$
INPUT BS$
IF BS$ = "R" THEN
  I = I + 1
  J = 0
END IF
CLOSE #4

LOOP UNTIL B$ = "N"
PRINT "PRESS 'Q' TO QUIT"
INPUT CS
LOOP UNTIL CS = "Q"
END SUB

'Subroutine Collectult samples DSO

SUB COLLECTULT (NAME$)
DO
  I = 1
  J = 0
  PRINT "WHAT ANATOMICAL REGION?"
  INPUT REG$

  DO
    CLS
    PRINT "OPENING COM PORTS"
    OPEN "COM2:9600,N,8,1,CS,DS,CD,RB1000" FOR RANDOM AS #1

    PRINT "SAMPLING DSO"
    DO

    DO
      PRINT #1, "S1"
  DO

END SUB

vi
LINE INPUT #1, RTN$
RTN = ASC(RTN)$
IF RTN <> &H41 THEN
   PRINT "ERROR STATUS ="; HEX$(RTN)$
ELSE
END IF
A$ = INKEY$
LOOP UNTIL A$ <> '"'

'R1 COMM AND LOADING OF SPACE DATA
PRINT "LOADING DSO DATA"
PRINT #1, "R1(1000,1000,A)"
RTN$ = INPUT$(4000, 1)
PRINT "DATA LOADED!"
PRINT "DATA OK?"
INPUT Y$
LOOP UNTIL Y$ = "Y"
CLOSE #1

J = J + 1
NEW$ = "C:\DOS\DATA\U" + NAMES + REG$ + LTRIM$(STR$(I)) + LTRIM$(STR$(J)) + ".DSO"
PRINT NEW$
OPEN "O", #3, NEW$
PRINT #3, RTN$
INPUT B$
IF B$ = "R" THEN
   I = I + 1
   J = 0
END IF
CLOSE #3
LOOP UNTIL B$ = "N"
PRINT "PRESS 'Q' TO QUIT"
INPUT CS
LOOP UNTIL CS = "Q"
END SUB
CLS
PRINT "WHAT ARE THE INITIALS?"
INPUT NAMES$ 
PRINT "WHAT IS THE ANATOMICAL REGION?"
INPUT REGS 
PRINT "HOW MANY ROWS OF DATA IN THIS REGION?"
INPUT K 
FOR I = 1 TO K 
PRINT "" 
PRINT "HOW MANY COLUMNS OF DATA IN ROW"; I; "?" 
INPUT C 
FOR J = 1 TO C 
PRINT "ROW NUMBER? " 
INPUT I 
PRINT "PT NUMBER" 
PRINT J 
NEWS = "C:\DOS\DATA\U" + NAMES + REGS + LTRIM$(STR$(I)) + LTRIM$(STR$(J)) + ".DSO" 
PRINT NEWS 
OPEN "I", #1, NEWS$ 
NNEW$ = "C:\DOS\DATA\AIN" + NAMES + REGS + LTRIM$(STR$(I)) + LTRIM$(STR$(J)) + ".DSO" 
OPEN "O", #2, NNEW$ 
PRINT #2, "AZ" 
PRINT #2, CHR$(13) 
PRINT #2, "0.05E-0006" 
PRINT #2, "S" 
PRINT #2, "MV" 
PRINT #2, "2009" 
FOR L = 1 TO 2010 
K$ = INPUT$(1, 1) 
PRINT K$; 
IF L > 14 THEN 
PRINT K$; 
IF K$ = "," THEN 
PRINT #2, CHR$(13) 
ELSE 
PRINT #2, K$; 
END IF 
END IF 
NEXT L 
CLOSE #1 
CLOSE #2 
NEXT J 
NEXT I 
END
***Process.bas calculates 3D coordinates from the Polhemus raw data in a format for modelling in the CAD.

DECLARE FUNCTION NGAM1 (GAM1, RGAM1)
DECLARE FUNCTION FHZ2! (Z2!, BET2!, TH!)  
DECLARE FUNCTION FHX2! (X2!, BET2!, ALPH2!, TH!)  
DECLARE FUNCTION FYH2! (Y2!, BET2!, ALPH2!, TH!)  
DECLARE FUNCTION FY2! (Y2!, BET2!, ALPH2!)  
DECLARE FUNCTION FZ2! (Z2!, BET2!)  
DECLARE FUNCTION TALPH! (FY2!, FX2!)  
DECLARE FUNCTION TBET! (FZ2!, FX2!, FY2!)  
DECLARE FUNCTION NALPH! (ALPH!, RALPH!)  
DECLARE FUNCTION L! (X!, Y!)  
DECLARE FUNCTION NX! (X, Y, Z, ALPH, BET, GAM)  
DECLARE FUNCTION NY! (X, Y, Z, ALPH, BET, GAM)  
DECLARE FUNCTION NBET! (BET1!, RBET1!)  
DECLARE FUNCTION NZ! (X, Y, Z, ALPH, BET, GAM)  
DECLARE FUNCTION FX! (X1!, X2!)  
DECLARE FUNCTION FY! (Y1!, Y2!)  
DECLARE FUNCTION FZ! (Z1!, Z2!)  
DECLARE FUNCTION FX2! (X2!, BET2!, ALPH2!)  
CLS
PRINT "WHAT ARE THE INITIALS?"
INPUT NAMES
C = .017453292#

'******************************************************************************************************************
'DATUM DATA - senso 1 data and zeroed position and orientation.
'******************************************************************************************************************

RFS = "C:\DOS\SPACE\MARINA\S" + NAMES + "RF.TXT"
OPEN "+", #7, RFS
DT$ = INPUT$(394, 7)
CLOSE #7
PRINT DT$
DX = VAL(MIDS(DT$, 9, 7))
DY = VAL(MIDS(DT$, 16, 7))
DZ = VAL(MIDS(DT$, 23, 7))
DALPH = VAL(MIDS(DT$, 30, 7)) * C
DBET = VAL(MIDS(DT$, 37, 7)) * C
DGAM = VAL(MIDS(DT$, 44, 7)) * C
PRINT "REF"; DX; DY; DZ; DALPH; DBET; DGAM

'******************************************************************************************************************
'POSITION1 DATA - position of right dot or right eye
'******************************************************************************************************************

'RECEIVER 1 Raw Data - position and orientation of head
******************************************************************************************************************
X1P1 = VAL(MID$(DT$, 109, 7))
Y1P1 = VAL(MID$(DT$, 116, 7))
Z1P1 = VAL(MID$(DT$, 123, 7))
P1R1ALPH = VAL(MID$(DT$, 130, 7)) * C
P1R1BET = VAL(MID$(DT$, 137, 7)) * C
P1R1GAM = VAL(MID$(DT$, 144, 7)) * C
PRINT "R1P1"; X1P1; Y1P1; Z1P1; P1R1ALPH; P1R1BET; P1R1GAM

*XIP1 = VAL(MID$(DT$, 109, 7))
YIP1 = VAL(MID$(DT$, 116, 7))
ZIP1 = VAL(MID$(DT$, 123, 7))
P1R1ALPH = VAL(MID$(DT$, 130, 7)) * C
P1R1BET = VAL(MID$(DT$, 137, 7)) * C
P1R1GAM = VAL(MID$(DT$, 144, 7)) * C
PRINT "R1P1"; X1P1; Y1P1; Z1P1; P1R1ALPH; P1R1BET; P1R1GAM

`RECEIVER 2 Raw Data - position of probe back end

X2P1 = VAL(MID$(DT$, 156, 7))
Y2P1 = VAL(MID$(DT$, 163, 7))
Z2P1 = VAL(MID$(DT$, 170, 7))
P1R2ALPH = VAL(MID$(DT$, 177, 7)) * C
P1R2BET = VAL(MID$(DT$, 184, 7)) * C
P1R2GAM = VAL(MID$(DT$, 191, 7)) * C
PRINT "R2P1"; X2P1; Y2P1; Z2P1; P1R2ALPH; P1R2BET; P1R2GAM`

`PROBE TIP POSITION - TRANSFORMED FROM PROBE END`

TX2P1 = FX2(X2P1, P1R2BET, P1R2ALPH)
TY2P1 = FY2(Y2P1, P1R2BET, P1R2ALPH)
TZ2P1 = FZ2(Z2P1, P1R2BET)

`CALCULATION OF ANGLES AWAY FROM REF POSITION DUE TO HEAD MOVEMENT`

N1ALPHP1 = NALPH(P1R1ALPH, DALPH)
N1BETP1 = NBET(P1R1BET, DBET)
N1GAMP1 = NGAM(P1R1GAM, DGAM)

`TRANSFORMATION OF RECEIVER 1 THROUGH MOVEMENT ANGLES`

NX1 = NX(X1P1, Y1P1, Z1P1, N1ALPHP1, N1BETP1, N1GAMP1)
NY1 = NY(X1P1, Y1P1, Z1P1, N1ALPHP1, N1BETP1, N1GAMP1)
NZ1 = NZ(X1P1, Y1P1, Z1P1, N1ALPHP1, N1BETP1, N1GAMP1)

`TRANSFORMATION OF PROBE TIP POSITION THROUGH MOVEMENT ANGLES`

NX2 = NX(TX2P1, TY2P1, TZ2P1, N1ALPHP1, N1BETP1, N1GAMP1)
NY2 = NY(TX2P1, TY2P1, TZ2P1, N1ALPHP1, N1BETP1, N1GAMP1)
NZ2 = NZ(TX2P1, TY2P1, TZ2P1, N1ALPHP1, N1BETP1, N1GAMP1)
CALCULATION OF POSITION OF PROBE TIP FROM REFERENCE POSITION IN X,Y,Z

\[ \text{XP1} = \text{FX}(\text{NX1}, \text{NX2}) \]
\[ \text{YP1} = \text{FY}(\text{NY1}, \text{NY2}) \]
\[ \text{ZP1} = \text{FZ}(\text{NZ1}, \text{NZ2}) \]

'POSITION 2 DATA
'RECEIVER 1
\[ \text{X1P2} = \text{VAL}((\text{MID$(DT$, 208, 7)))} \]
\[ \text{Y1P2} = \text{VAL}((\text{MID$(DT$, 215, 7)))} \]
\[ \text{Z1P2} = \text{VAL}((\text{MID$(DT$, 221, 7)))} \]
\[ \text{P2R1ALPH} = \text{VAL}((\text{MID$(DT$, 228, 7)))} \]
\[ \text{P2R1BET} = \text{VAL}((\text{MID$(DT$, 235, 7)))} \]
\[ \text{P2R1GAM} = \text{VAL}((\text{MID$(DT$, 242, 7)))} \]
\[ \text{PRINT "R1P2", X1P2, Y1P2, Z1P2, P2R1ALPH, P2R1BET, P2R1GAM} \]

'RECEIVER 2 - POSITION OF MIDDLE DOT OR END OF NOSE

\[ \text{X2P2} = \text{VAL}((\text{MID$(DT$, 254, 7)))} \]
\[ \text{Y2P2} = \text{VAL}((\text{MID$(DT$, 261, 7)))} \]
\[ \text{Z2P2} = \text{VAL}((\text{MID$(DT$, 268, 7)))} \]
\[ \text{P2R2ALPH} = \text{VAL}((\text{MID$(DT$, 275, 7)))} \]
\[ \text{P2R2BET} = \text{VAL}((\text{MID$(DT$, 282, 7)))} \]
\[ \text{P2R2GAM} = \text{VAL}((\text{MID$(DT$, 289, 7)))} \]
\[ \text{PRINT "R2P2", X2P2, Y2P2, Z2P2, P2R2ALPH, P2R2BET, P2R2GAM} \]

'POSITION 3 DATA - POSITION OF LEFT DOT OR LEFT EYE

\[ \text{TX2P2} = \text{FX2}(\text{X2P2}, \text{P2R2BET}, \text{P2R2ALPH}) \]
\[ \text{TY2P2} = \text{FY2}(\text{Y2P2}, \text{P2R2BET}, \text{P2R2ALPH}) \]
\[ \text{TZ2P2} = \text{FZ}(\text{TX2P2}, \text{P2R2BET}) \]
\[ \text{N1ALPHP2} = \text{NLPH}(\text{P2R1ALPH}, \text{DALPH}) \]
\[ \text{N1BETP2} = \text{NBET}(\text{P2R1BET}, \text{DBET}) \]
\[ \text{N1GAMP2} = \text{NGAM}(\text{P2R1GAM}, \text{DGAM}) \]

\[ \text{N2ALPH2} = \text{P2R2ALPH} - \text{N1ALPHP2} \]
\[ \text{N2BETP2} = \text{P2R2BET} - \text{N1BETP2} \]
\[ \text{N2GAMP2} = \text{P2R2GAM} - \text{N1GAMP2} \]
\[ \text{NX1} = \text{NX}(\text{X1P2}, \text{Y1P2}, \text{Z1P2}, \text{N1ALPHP2}, \text{N1BETP2}, \text{N1GAMP2}) \]
\[ \text{NY1} = \text{NY}(\text{X1P2}, \text{Y1P2}, \text{Z1P2}, \text{N1ALPHP2}, \text{N1BETP2}, \text{N1GAMP2}) \]
\[ \text{NZ1} = \text{NZ}(\text{X1P2}, \text{Y1P2}, \text{Z1P2}, \text{N1ALPHP2}, \text{N1BETP2}, \text{N1GAMP2}) \]
\[ \text{NX2} = \text{NX}(\text{TX2P2}, \text{TY2P2}, \text{TZ2P2}, \text{N1ALPHP2}, \text{N1BETP2}, \text{N1GAMP2}) \]
\[ \text{NY2} = \text{NY}(\text{TX2P2}, \text{TY2P2}, \text{TZ2P2}, \text{N1ALPHP2}, \text{N1BETP2}, \text{N1GAMP2}) \]
\[ \text{NZ2} = \text{NZ}(\text{TX2P2}, \text{TY2P2}, \text{TZ2P2}, \text{N1ALPHP2}, \text{N1BETP2}, \text{N1GAMP2}) \]
\[ \text{XP2} = \text{FX}(\text{NX1}, \text{NX2}) \]
\[ \text{YP2} = \text{FY}(\text{NY1}, \text{NY2}) \]
\[ \text{ZP2} = \text{FZ}(\text{NZ1}, \text{NZ2}) \]
RECEIVER 1
X1P3 = VAL(MID$(DT$, 305, 7))
Y1P3 = VAL(MID$(DT$, 312, 7))
Z1P3 = VAL(MID$(DT$, 319, 7))
P3R1ALPH = VAL(MID$(DT$, 326, 7)) * C
P3R1BET = VAL(MID$(DT$, 333, 7)) * C
P3R1GAM = VAL(MID$(DT$, 340, 7)) * C
PRINT "R1P3": X1P3, Y1P3, Z1P3, P3R1ALPH, P3R1BET, P3R1GAM

RECEIVER 2
X2P3 = VAL(MID$(DT$, 352, 7))
Y2P3 = VAL(MID$(DT$, 359, 7))
Z2P3 = VAL(MID$(DT$, 366, 7))
P3R2ALPH = VAL(MID$(DT$, 373, 7)) * C
P3R2BET = VAL(MID$(DT$, 380, 7)) * C
P3R2GAM = VAL(MID$(DT$, 387, 7)) * C
PRINT "X2P3": X2P3, Y2P3, Z2P3, P3R2ALPH, P3R2BET, P3R2GAM

TX2P3 = FX2(X2P3, P3R2BET, P3R2ALPH)
TY2P3 = FY2(Y2P3, P3R2BET, P3R2ALPH)
TZ2P3 = FZ2(Z2P3, P3R2BET)

NALPHP3 = NALPH(P3R1ALPH, DALPH)
N1BETP3 = NBET(P3R1BET, DBET)
N1GAMP3 = NGAM(P3R1GAM, DGAM)

N2ALPHP3 = P3R2ALPH - N1ALPHP3
N2BETP3 = P3R2BET - N1BETP3
N2GAMP3 = P3R2GAM - N1GAMP3
NX1 = NX(X1P3, Y1P3, Z1P3, N1ALPHP3, N1BETP3, N1GAMP3)
NY1 = NY(X1P3, Y1P3, Z1P3, N1ALPHP3, N1BETP3, N1GAMP3)
NZ1 = NZ(X1P3, Y1P3, Z1P3, N1ALPHP3, N1BETP3, N1GAMP3)
NX2 = NX(TX2P3, TY2P3, TZ2P3, N1ALPHP3, N1BETP3, N1GAMP3)
NY2 = NY(TX2P3, TY2P3, TZ2P3, N1ALPHP3, N1BETP3, N1GAMP3)
NZ2 = NZ(TX2P3, TY2P3, TZ2P3, N1ALPHP3, N1BETP3, N1GAMP3)
XP3 = FX(NX1, NX2)
YP3 = FY(NY1, NY2)
ZP3 = FZ(NZ1, NZ2)

***********************************************************************
WRITING REFERENCE DATA TO FILE C:\DOS\SPACE\OPOS**.USD
***********************************************************************

FS = "C:\DOS\SPACE\OPOS" + NAMES$ + ".USD"
HS = "C:\DOS\SPACE\IPOS" + NAMES$ + ".TXT"
OPEN "O", #1, FS
OPEN "O", #2, HS
PRINT #1, ZP1, YP1, XP1
PRINT "PT1: "; XP1, YP1, ZP1
PRINT #1, ZP2, YP2, XP2
PRINT "PT2: "; XP2, YP2, ZP2
PRINT #1, ZP3, YP3, XP3
PRINT "PT3: "; XP3, YP3, ZP3
CLOSE #1
'CLOSE #2

'*****************************************************************************
'CALCULATION OF POSITION OF EACH DATA POINT
'*****************************************************************************

OPEN "A", #1, F$
'OPEN "O", #2, H$
PRINT "WHAT CODE IS DATA UNDER?"
INPUT NAME$
DO
PRINT "WHAT IS THE ANATOMICAL REGION?"
INPUT REG$
PRINT #1, REG$

PRINT "HOW MANY ROWS OF DATA IN THIS REGION?"
INPUT K
FOR I = 1 TO K
    PRINT 
    PRINT "HOW MANY COLUMNS OF DATA IN ROW", I; "?"
    INPUT V
    FOR J = 1 TO V
        GS = "C:\DOS\SPACE\MARINA\" + NAME$ + REG$ + LTRIM$(STR$(I)) + LTRIM$(STR$(J)) + 

    PRINT GS
    OPEN "I", #3, GS
    A$ = INPUT$(94, 3)
    PRINT A$
    CLOSE #3
    X1 = 0
    Y1 = 0
    Z1 = 0
    X2 = 0
    Y2 = 0
    Z2 = 0
    TX2 = 0
    TY2 = 0
    TZ2 = 0
    N1ALPH = 0
    N1BET = 0
    N1GAM = 0
    N2ALPH = 0
    N2BET = 0
    N2GAM = 0
    NX1 = 0
    NY1 = 0
NZ1 = 0
NX2 = 0
NY2 = 0
NZ2 = 0
OX = 0
OY = 0
OZ = 0

'CONVERSION FROM DEGREES TO RADIANS'

'VECTOR R1 PARAMETERS
X1 = VAL(MID$(A$, 4, 7))
PRINT "C"; C
Y1 = VAL(MID$(A$, 11, 7))
Z1 = VAL(MID$(A$, 18, 7))
PRINT "R1"; X1; Y1; Z1

'RECEIVER R1'S ORIENTATION
ALPH1 = VAL(MID$(A$, 25, 7)) * C
BET1 = VAL(MID$(A$, 32, 7)) * C
GAM1 = VAL(MID$(A$, 39, 7)) * C
PRINT X1; Y1; Z1; ALPH1; BET1; GAM1

'VECTOR R2 PARAMETERS
X2 = VAL(MID$(A$, 51, 7))
Y2 = VAL(MID$(A$, 58, 7))
Z2 = VAL(MID$(A$, 65, 7))
PRINT "R2"; X2; Y2; Z2

'RECEIVER R2'S ORIENTATION
ALPH2 = VAL(MID$(A$, 72, 7)) * C
BET2 = VAL(MID$(A$, 79, 7)) * C
GAM2 = VAL(MID$(A$, 86, 7)) * C
PRINT X2; Y2; Z2; ALPH2; BET2; GAM2

'PROBE TIP POSITION - TRANSFORMED FROM PROBE END'

TX2 = FX2(X2, BET2, ALPH2)
TY2 = FY2(Y2, BET2, ALPH2)
TZ2 = FZ2(Z2, BET2)

'CALCULATION OF ANGLES AWAY FROM REF POSITION DUE TO HEAD MOVEMENT'

N1ALPH = NALPH(ALPH1, DALPH)
N1BET = NBET(BET1, DBET)
N1GAM = NGAM(GAM1, DGAM)
TRANSFORMATION OF RECEIVER 1 THROUGH MOVEMENT ANGLES

NX1 = NX(X1, Y1, Z1, N1ALPH, N1BET, N1GAM)
NY1 = NY(X1, Y1, Z1, N1ALPH, N1BET, N1GAM)
NZ1 = NZ(X1, Y1, Z1, N1ALPH, N1BET, N1GAM)

TRANSFORMATION OF PROBE TIP POSITION THROUGH MOVEMENT ANGLES

NX2 = NX(TX2, TY2, TZ2, N1ALPH, N1BET, N1GAM)
NY2 = NY(TX2, TY2, TZ2, N1ALPH, N1BET, N1GAM)
NZ2 = NZ(TX2, TY2, TZ2, N1ALPH, N1BET, N1GAM)

CALCULATION OF POSITION OF PROBE TIP FROM REFERENCE POSITION IN X,Y,Z

OX = FX(NX1, NX2)
OY = FY(NY1, NY2)
OZ = FZ(NZ1, NZ2)

INCORPORATION OF THICKNESS DATA TO FORM INSIDE SHELL DATA

'PRINT "WHAT IS THE THICKNESS FOR POINT "; REG$, I, J; "?
INPUT TH
'THX2 = FHX2(X2, BET2, ALPH2, TH)
'THY2 = FHY2(Y2, BET2, ALPH2, TH)
'THZ2 = FHZ2(Z2, BET2, TH)
'N1ALPH = NALPH(ALPH1, DALPH)
'N1BET = NBET(BET1, DBET)
'N1GAM = NGAM(GAM1, DGAM)

'N2ALPH = NALPH(ALPH2, N1ALPH)
'N2BET = NBET(BET2, N1BET)
'N2GAM = NGAM(GAM2, N1GAM)
'NX1 = NX(X1, Y1, Z1, N1ALPH, N1BET, N1GAM)
'NY1 = NY(X1, Y1, Z1, N1ALPH, N1BET, N1GAM)
'NZ1 = NZ(X1, Y1, Z1, N1ALPH, N1BET, N1GAM)

'NX2 = NX(THX2, THY2, THZ2, N1ALPH, N1BET, N1GAM)
'NY2 = NY(THX2, THY2, THZ2, N1ALPH, N1BET, N1GAM)
'NZ2 = NZ(THX2, THY2, THZ2, N1ALPH, N1BET, N1GAM)

'IX = FX(NX1, NX2)
'FY = FY(NY1, NY2)
'IZ = FZ(NZ1, NZ2)

'PRINT "OK?"
'INPUT STOPS
'IF STOPS = "NO" THEN
'GOTO 10
'END IF

'PRINT #2, IZ, IY, IX

NEXT J

NEXT I

INPUT QUIT$
LOOP UNTIL QUIT$ = "QUIT"
CLOSE #1
CLOSE #2
END

FUNCTION FHX2 (X2, BET2, ALPH2, TH)
FHX2 = X2 - ((13.4 + TH) * COS(BET2) * COS(ALPH2))
END FUNCTION

FUNCTION FHY2 (Y2, BET2, ALPH2, TH)
FHY2 = Y2 - ((13.4 + TH) * COS(BET2) * SIN(ALPH2))
END FUNCTION

FUNCTION FHZ2 (Z2, BET2, TH)
FHZ2 = Z2 + ((13.4 + TH) * SIN(BET2))
END FUNCTION

FUNCTION FX (XI, X2)
FX = X2 - XI
END FUNCTION

FUNCTION FX2 (X2, BET2, ALPH2)
FX2 = X2 - (13.925 * COS(BET2) * COS(ALPH2))
END FUNCTION

FUNCTION FY (Y1, Y2)
FY = -Y2 + Y1
END FUNCTION

FUNCTION FY2 (Y2, BET2, ALPH2)
FY2 = Y2 - (13.925 * COS(BET2) * SIN(ALPH2))
END FUNCTION

FUNCTION FZ (Z1, Z2)
FZ = Z2 - Z1
END FUNCTION

FUNCTION FZ2 (Z2, BET2)
FZ2 = Z2 + (13.925 * SIN(BET2))
END FUNCTION

FUNCTION L (X, Y)
L = SQR(X * X + Y * Y)
END FUNCTION

FUNCTION NALPH (ALPH, RALPH)
NALPH = ALPH - RALPH
END FUNCTION

FUNCTION NBET (BET, RBET)
NBET = BET - RBET
END FUNCTION

FUNCTION NGAM (GAM, RGAM)
NGAM = RGAM - GAM
END FUNCTION

FUNCTION NX (X, Y, Z, ALPH, BET, GAM)
NX = (COS(ALPH) * COS(BET) * X) - (SIN(ALPH) * COS(BET) * Y) + (SIN(BET) * Z)
END FUNCTION

FUNCTION NY (X, Y, Z, ALPH, BET, GAM)
NY = ((SIN(GAM) * SIN(BET) * COS(ALPH) + SIN(ALPH) * COS(GAM)) * X) - ((SIN(ALPH) * SIN(GAM) * SIN(BET) - COS(ALPH) * COS(GAM)) * Y) - (SIN(GAM) * COS(BET) * Z)
END FUNCTION

FUNCTION NZ (X, Y, Z, ALPH, BET, GAM)
NZ = ((SIN(ALPH) * SIN(GAM) - SIN(BET) * COS(GAM)) * X) + (SIN(BET) * SIN(ALPH) * COS(GAM) + COS(ALPH) * SIN(GAM)) * Y) + (COS(ALPH) * COS(BET) * Z)
END FUNCTION
Appendix C: Engineering drawings of ultrasound probe casing and tip.
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<th>DRAWING NO.</th>
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**FOR INTERNAL USE ONLY**

DEPT. OF MECHANICAL ENG.

GUARDFORD G92 5XH

DATE 01-FEB-94

UNIVERSITY OF SURRY

1:1 SCALE

---

**GENERAL TOLERANCE**

SURFACE TEXTURE $N^7$

TREATMENT BREAK SHARP EDGES

**MATERIAL**

NYLON 66

**ALL DIMENSIONS IN MILLIMETRES**

$\varnothing 28.0$

$50.0$

$40.0$

$\varnothing 22.0$
Appendix D: Raw ultrasound thickness tissue data.
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