Abstract—In this article, a method of using ultrasound to image a residual limb is presented. The method employs a compound scanning technique to reconstruct a cross-sectional image (a slice) of the limb in a transverse plane. By scanning the limb in many transverse planes, a three-dimensional (3-D) volumetric image can be obtained from which either a transverse slice, a longitudinal cross section, or a 3-D surface of the limb can be displayed. The compound process circumvents the problems associated with the large attenuation of bones and enables reconstruction of a complete image of bones and adjoining tissues. In addition, the compound process improves the lateral resolution and reduces the speckle noise. Results obtained from a pair of thin wires, a contrast-resolution phantom, and a human limb demonstrate the beneficial effects of the compound process. To maximize the benefits, however, an accurate pixel registration in image reconstruction is essential. Sources of pixel misregistration and the potential means of minimizing misregistration are discussed.

Key words: lower limb prosthesis, prosthetic socket design, 3-D imaging, ultrasound.

INTRODUCTION

The main challenge in lower-limb prosthetic socket design is to obtain an appropriate loading pattern, or pressure distribution, at the limb-socket interface. To achieve this goal, prosthetists need to address two key questions: Where should pressure be applied? and How much pressure should be applied? In regard to the first question, prosthetists generally agree that the locations of weight-bearing and pressure-relief areas are closely associated with the bony structure of the residual limb. For example, the patellar notch and the medial and lateral flares of the tibia form the major weight-bearing areas, while the tibial head, distal tibia, and distal fibula define some of the major pressure-relief areas (1,2). Since current socket design is implemented by modifying the shape of a limb model, which may be either a positive plaster model or a digitized computer model, the areas for shape modification are identified on the surface of the limb model. To accomplish this, the prosthetist first palpates the limb to identify special bony structures and draws marks on its surface using a special pencil. These marks are then copied to the inner surface of a plaster cast, and finally transferred to the surface of a plaster model or to a computer model during shape digitizing. The current method of landmarking is subject to positioning errors, due to the difficulty in projecting deep bony structure to limb surface and the possible shift of landmarks during casting and shape digitizing. If the internal bony structure of a
residual limb can be recorded and displayed on the computer model, the landmarks may be generated directly on the model, and the above sources of positioning errors can be eliminated.

The question of how much pressure to apply is more difficult to answer. First, there are no universal rules that prescribe an optimal, quantitative pressure distribution for a given residual limb. Second, even if such an optimal pressure distribution is prescribed, there are no ready methods to translate the desired pressure distribution to a required shape rectification. In fact, current prosthetic socket design is a highly empirical, largely trial-and-error process. As a result, the quality of socket fitting often shows a great variation (3). A promising method for quantitative investigations of limb-prosthesis interaction is finite element (FE) analysis. In such an approach, an FE model of the limb-prosthesis system is first established, based on the measured or assumed geometry, material properties, and boundary conditions of the limb-prosthesis system. The model can then be used to estimate the interface stress distribution and to evaluate prosthetic fit (4–6). Unfortunately, due to the difficulties in determining the tissue properties and boundary conditions, the accuracy of current FE models in estimating prosthetic interface stresses has not been satisfactory (4). As an alternative approach, a visual display of the internal structures of the limb may assist the prosthetist to better estimate the applied pressure during socket design by showing the type (e.g., fat, muscle) and thickness of the soft tissue.

From the above discussion, it is expected that three-dimensional (3-D) visualization of the external shape and internal structures of the limb can help the prosthetist both in locating the critical areas for shape modification and in determining the required quantity of shape rectification. Modern technology provides a number of modalities for imaging the internal tissue structure of a limb. For example, X-ray computerized tomography (CT) and magnetic resonance imaging (MRI) techniques have been used to acquire the geometry of a limb-socket system in FE modeling studies (5,6). More recently, spiral CT has been utilized to perform volumetric imaging of residual limbs (7). As a low-cost alternative, ultrasound can also be used to image the internal tissue structure.

Medical ultrasound has mainly been used for examining soft tissues such as the fetus, liver, heart, breast, and prostate. Bony structure, on the other hand, cannot be imaged satisfactorily using conventional B-scan imaging techniques. Due to strong impedance mismatch at the bone-soft tissue interface and the large attenuation of bony material upon ultrasound, little echo signal can be received beyond the front surface of a bone. To obtain a complete image of the limb, compound scanning can be used. Using this technique, the limb is scanned from different directions and the individual B-scan images are combined to produce a compound image. In addition to circumventing the problems associated with the large attenuation of bones, the compounding process also improves the image quality by reducing the speckle noise. Ultrasound speckle refers to the granular appearance of a B-scan image which gives a false impression of the actual structure of the target (8). Using compound scanning, Sehgal et al. (9) obtained cross-sectional images of turkey and dog limbs, showing bones and adjoining soft tissues. To study the effects of intermittent pneumatic compression treatment on a swollen upper arm, Fares et al. (10) developed an ultrasound 3-D compound imaging system to measure the changes in compartment volume of an upper arm.

In this article, preliminary results using ultrasound to image human lower limbs are presented. The goal of this study was to explore the feasibility of developing an ultrasound-based prosthetic socket design system that can assist prosthetists in socket design and fitting, as well as being used as an alternative tool in basic prosthetic research using FE analysis.

METHODS

Scanning and Data Acquisition System
The structural details of the system have been reported previously (11). In brief, the system consists of a water tank, a seating and lifting device, a portable ultrasound scanner, and a personal computer (PC), as shown in Figure 1. The water tank is made of Plexiglas and measures 50 cm in diameter and 76 cm in height. Two stepping motors located underneath the tank provide the rotational movement and vertical translation, respectively, of an ultrasound transducer. The chair is mounted on the top of a telescope-type actuator through a rotating coupler. To perform the ultrasound scan, the residual limb is submerged into the water. Scanning normally begins at the distal end of the limb. At each level (in a transverse plane) of the limb, the ultrasound transducer rotates 360° around the limb and a B-scan image is acquired every 10°. The transducer is then translated proximally 3 or 5 mm and scanning is repeated at the new level. The scanning dura-
tion for each level is 12 s and the speed of vertical translation of the transducer is 15 mm/s. Depending on the length of the residual limb, the total scanning time is 12–15 min.

The ultrasound scanner used in this study is equipped with a mechanical sector transducer. Each B-scan image is composed of 128 scan lines (called A-lines), which radiate from the pivot point of a rocking transducer element and form a sector of 70°. Echo signals received by the transducer are first amplified by the portable ultrasound scanner and the detected echo envelope signals are then transmitted to the PC for digitization and image reconstruction. For each A-line, 1024 samples are digitized at a sampling frequency of 5 MHz.

Determining the Number of B-scan Images to be Acquired at Each Level

This section provides calculations for determining the minimum number ($N_i$) and the maximum number ($N_X$) of B-scan images to be acquired at each level. Based on these two (lower and upper) bounds, a particular number is selected for this study.

The minimum number, $N_i$, of B-scan images to be acquired at each level can be determined with the help of Figure 2. The figure shows the relative positions of the transducer, the center of rotation (point O) and a segment of limb surface (centered around point B) closest to the trajectory along which the transducer is rotating during compound scan. Theta ($\theta$) is the angle of B-scan sector and $\phi$ is the angle between transducer position I and posi-

$$\phi \equiv 2 \tan^{-1} \left[ \frac{BC}{BO} \right] \approx 2 \tan^{-1} \left[ \frac{AB \tan(\theta/2)}{AO - AB} \right] \quad [1]$$

The minimum number, $N_i$, of B-scan images to be acquired at each level is equal to $360^\circ/\phi$. In this study, $AO=112$ mm, $AB=30$ mm, and $\theta=35^\circ$ (only 64 of the 128 A-lines are actually used for compounding, see DISCUSSION), the maximum $\phi$ calculated from equation 1 is $13.2^\circ$; and $N_i$ is then equal to 27.

The maximum number, $N_X$, of B-scan images to be acquired at each level is determined, based on the following requirement: to achieve maximum speckle reduction through compounding, the speckle patterns of the component B-scan images must be mutually independent. In other words, if transducer position II is too close to position I, the speckle patterns of the two B-scan images will be correlated with each other and the maximum speckle reduction will not be achieved. To produce mutually independent speckle patterns, the transducer must be translated by at least half its diameter between two scans.
If the diameter of the transducer is "a," \( N_x \) can be determined as:

\[
N_x = \frac{2\pi A0}{a/2}
\]  

[2]

For the transducer used in this study, a=20 mm. For \( A0=112 \) mm, \( N_x=70 \), corresponding to a minimum angle, \( \phi=5.1^\circ \).

From the lower bound (\( N_x=27 \)) and upper bound (\( N_x=70 \)) for the number of B-scan images to be acquired at each level, a value of 36 (corresponding to an angle \( \phi=10^\circ \)) was selected to be used in this study. One of the reasons of selecting this particular number is because 36 can be divided by many numbers (e.g., 9, 6, 4, and 2) so that the effects of compounding can be studied by combining 4, 6, 9, and 18 images taken symmetrically around the object.

**Image Reconstruction**

From the 36 B-scan images acquired at each scan level, a compound image is reconstructed that displays a transverse cross section (a slice) of the limb at that level.

There are three major steps in image reconstruction: sample position mapping, pixel value interpolation, and compounding. The process of sample position mapping is to determine the precise location of each sample in the final compound image. In our reconstruction scheme, the final compound image contains 256X256 pixels, representing a fixed image area of 180X180 mm. The center of this area is coincident with the physical center of transducer rotation during compound scan. This point, called the center of rotation (COR), is the reference point in all position calculations. Prior to image reconstruction, system calibration is performed by hanging a thin nylon wire near the COR, rotating the transducer around the wire, and acquiring 36 B-scan images of the wire. By averaging the positions of the wire samples in each sector, the location of COR, in terms of A-line number and sample number along the A-line, is determined. In this study, the speed of sound is assumed to be a constant: 1,500 m/s. For a sampling frequency of 5 MHz, the spatial distance between two consecutive samples is calculated as 0.15 mm. This provides a way to determine the spatial position of each sample in an A-line. By assuming that all A-lines spread evenly in the B-scan sector (a good approximation for A-lines in the central region of the sector), the spatial position of each sample within the B-scan sector can then be determined. Finally, knowing the direction of each individual B-scan (10°, 20°, 30°... from the home position), the position mapping from every sample in every B-scan sector to the final compound image can be established.

The above mapping process usually causes a sample to fall into the gap between pixels in the compound image (the pixels are located at discrete positions, 0.7 mm apart in our example). Bilinear interpolation is then used to distribute the value of the mapped sample to the neighboring 4 pixels. Finally, the contributions from different B-scan images to the same pixel are averaged to produce the final compound image. Using a 80486-based PC (50 MHz), the total time for image reconstruction is approximately 30 min.

**Image Display**

The above image reconstruction results in a stack of slices. Three forms are used to display the limb. The first form shows the original slice, a transverse cross section of the limb. The second form shows a longitudinal cross section of the limb either in a sagittal plane or in a frontal plane. The third form displays a 3-D surface of the limb.

The image displayed using either the first or second form is an amplitude-coded gray scale image in which the brightness of each pixel is proportional to the average magnitude of the echo reflected back by a certain tissue element. For example, since water is echo-free, it provides a black background for the limb image. Strong echoes reflected by the water-skin interface produce a clear skin contour. The high contrast at water-skin interface makes it easy to detect skin contour automatically. It also helps to maintain high accuracy in reconstructing the external shape of the limb. On the other hand, the bone boundaries usually are not very sharp. As the echoes obtained from inside the bone are extremely weak, bones are generally shown as black holes.

To obtain a longitudinal section of the limb along a particular cut line in a two-dimensional (2-D) slice, all pixels along that line are extracted from each slice, and the pixels from all slices then form a 2-D matrix. Since the pixel density is much higher in the transverse direction (256 pixels per 180 mm) than in the longitudinal direction (1 pixel per 3 mm or 5 mm), 2-D linear interpolation is performed in the longitudinal direction so that the final image has the same scale in both transverse and longitudinal directions.

To display a 3-D surface of the limb, the skin contour is first extracted from each slice using a morpholog-
ical outlining process (14). By stacking together the skin contours from all the slices, the limb surface can be displayed as either a shaded solid object or a wire-frame.

The programs for scanning control, data acquisition, and image reconstruction are written in Borland C++ (version 3.1) running on a 80486-based PC. Software for image display and 2-D interpolation is MATLAB (version 4.2) for the PC.

Experiments

Three different objects are scanned under various conditions. The images obtained from these experiments are used to evaluate the image resolution, measurement accuracy, and image quality of the system.

The first object is a pair of nylon wires vertically suspended in the water tank. The diameter of each wire is 0.14 mm and the distance between the two wires is 3 mm. The distance between the transducer and the center of the two wires is approximately 112 mm. Since the focal distance of the transducer is 70 mm, the target is placed in the far field. The wires are scanned using three methods and all scans are performed in the same transverse plane (no vertical translation of the transducer). In the first two experiments, simple B-scans are used (no compounding). In the first experiment, the two wires are aligned along the axial direction (the direction of ultrasound propagation), and in the second experiment, the two wires are aligned along the lateral direction (the direction of beam sweeping). In the third experiment, the wires are scanned from 36 different directions and a compound image is reconstructed. These images are used to determined the image resolution of the system.

The second scanning object is a contrast-resolution phantom manufactured by ATS Laboratories, Inc. (Bridgeport, CT). It has a cylindrical shape with a height of 15 cm and a diameter of 113 mm measured by a caliper. The phantom contains a concentric inner cylinder (contrast target), which has a diameter of 25 mm and a relative reflectivity of +12 dB with respect to the background material (both according to the manufacture’s specification). The phantom is held vertically in the water tank and the distance between the transducer and the central axis of the phantom is approximately 112 mm. Thirty-six B-scan images in the same transverse plane are obtained using the compound scanning technique. These images are used to estimate the measurement accuracy as well as to demonstrate the effects of compounding in speckle reduction.

The third object is the right leg of a healthy volunteer. Only a portion of the leg near the knee is scanned. The scanned segment has a length of 120 mm with the patellar notch located near the middle. A total of 41 levels are scanned and the vertical distance between two adjacent levels is 3 mm. At each level, 36 B-scan images are obtained and a compound image is reconstructed. The resultant 41 slices are used to display both transverse and longitudinal cross sections of the limb.

RESULTS

Image of the Wires

An image system is characterized by its point spread function (PSF), which is the 2-D distribution of image intensity of an hypothetical point target (15). In conventional B-scan imaging (noncompounding), the PSF is often replaced by the axial and lateral resolutions. Experimentally, these resolutions can be assessed by scanning a very thin wire and reconstructing an image of the wire. The axial resolution is the size of the wire image along the direction of ultrasound propagation and the lateral resolution is the size of the wire image along the direction perpendicular to the axial direction. In a compound image, the target is scanned from many different directions and the concept of axial and lateral resolutions no longer applies.

Figure 3 shows the images of two nylon wires under three different scanning conditions. The first two images, a and b, are simple, noncompounding B-scan images and the third, c, is obtained by compounding 36 B-scan images. In image a, the two wires are placed along the axial direction and the transducer (not shown in the image) is located at the top of the image. Knowing that

![Figure 3](image_url)
the actual space between the two wires is 3 mm, we use the distance between the centers of the two wire images in a as the reference to determine the size of each wire image in the axial and lateral directions. Using this method, the axial resolution is estimated as 1.5 mm and the lateral resolution as 4.5 mm. In b, the two wires are placed along the lateral direction. Since the lateral resolution is larger than the distance between the two wires (4.5 mm vs. 3 mm), the two images overlap, and the two wire targets are no longer distinguishable. The compound image of the two wires is displayed in c. The image of each wire target becomes a symmetric dot, and the two wire targets are clearly distinguishable. The resolution of the image, judged by the diameter of each dot using the same reference, is estimated as 1.5 mm.

Image of the Phantom

Figure 4 shows the cross-sectional images obtained from the contrast-resolution phantom. The image shown on the upper left is a simple B-scan image. To improve the image quality (see DISCUSSION), only the central 64 A-lines and only 800 samples in each A-line are used in image reconstruction. The granular appearance (ultrasound speckle) of the image is quite evident. Images on the upper right, lower left, and lower right are obtained by compounding 4, 9, and 36 B-scan images, respectively. As the number of component B-scan images increases, the appearance of the compound image becomes smoother.

To examine the measurement accuracy of the system, the image shown on the lower right is used to measure the external diameter of the phantom and the diameter of the contrast target in the image, and the results are compared with the actual values. The external diameter of the phantom in the image is measured along six directions (0, 30, 60, 90, 120, and 150° from the vertical) using the following procedure. Along each direction, a search routine is executed that examines the gray scale value of each image pixel. If the pixel is in the water region, its value is near zero. At the phantom boundary, the pixel value increases sharply (in our example, the gray scale values of the boundary pixels are normally more than 180). The first pixel having a gray scale value above a threshold (120, in our example) is identified as the boundary pixel. In this way, two boundary pixels, one on each side of the phantom, are identified along the chosen direction. The physical distance between these two pixels, which is defined as the measured diameter of the phantom along the chosen direction, is then calculated from their image coordinates (conversion factor: 180 mm per 256 pixels). The average value of the phantom diameter measured along the six directions using the above method is 114.5±0.5 mm. The diameter of the contrast target (central circle) is measured along four different directions (0, 45, 90, and 135° from the vertical) using the same search routine, and the average result is 23.3±0.4 mm. As compared with the actual diameter of the phantom (113 mm, measured using a caliper) and the diameter of the contrast target (25 mm, according to the manufacturer’s specification), the absolute errors in these two measurements are 1.5 mm for the external diameter and -1.7 mm for the central target, respectively.

Images of The Limb

Figure 5 shows a 3-D surface of the limb segment. The figure provides an anterior view of the limb with the medial side on the right. Figure 6 shows the transverse (top figures) and longitudinal (bottom figures) cross sections of the limb segment. In each transverse-slice image, the knee is facing up and the medial side is on the left. The level of the slice shown on upper left is approxi-
Figure 5.
Front view of a 3-D display of the surface of a segment of a right limb. The medial side is on the right. Total length of the segment shown in the image is 120 mm.

Figure 6.
Transverse and longitudinal cross sectional images of a segment of a right limb (total length = 120 mm). **Upper left:** A transverse slice of the limb sectioned at 35 mm distal to the patellar notch (white line on corresponding lower left image). Knee facing up, medial side on the left. **Upper right:** Transverse slice of limb sectioned just distal to patellar notch (white line on corresponding lower right image). **Lower left:** Longitudinal section of segment of limb in a frontal plane, posterior view. (Position of the cut = horizontal white line on corresponding upper left image). **Lower right:** Longitudinal section of limb segment in a sagittal plane, medial view. (Position of cut = vertical white line on corresponding upper right image).

DISCUSSION

One of the purposes of this article is to provoke an interest and discussion on the potential applications of ultrasound imaging techniques in lower limb prosthetic socket design. In addition to ultrasound, the internal structure of the limb can be imaged by CT or MRI, both of which provide superior images of bones and soft tissues. However, with CT imaging there is a concern of ionizing radiation associated with X-ray. In addition, both CT and MRI may be too expensive, in terms of equipment and operational costs, for the application of prosthetic socket design. Finally, in current CT or MRI scan, the subject must lie down in either a supine or prone position. The shape change of the limb caused by the gravitational force may introduce significant measurement uncertainties (4). In contrast, in ultrasound scan, the limb is in the natural vertical position. Although the water in which the limb is submerged applies a gradual hydraulic pressure to the limb (increase 1 mmHg for every 1.36 cm of depth in water), this pressure is applied symmetrically in the radial direction. As a result, the change in limb shape is less significant.

The main disadvantage of a conventional ultrasound B-scan image is its relatively low quality, which is characterized by the poor lateral resolution and the presence of ultrasound speckle. For a low-cost ultrasound scanner (less than $10,000), such as used in this study, the lateral resolution also changes significantly with scan depth. In imaging soft tissue, poor lateral resolution and the presence of ultrasound speckle blur the image and reduce the
ability to distinguish the details of tissue structure. In theory, the resolution of a fully compounded image in any direction can approach the axial resolution of the original B-scan image, which is normally in the range of 1–2 mm. However, any error in sample position mapping may cause pixel misregistration and worsen the resolution of the compound image. In practice, pixel misregistration may arise due to the error in determining the COR, inaccuracy in determining the direction of ultrasound propagation (A-line position), variation in speed of sound in different tissues, and refraction of the ultrasound beam. While the first two sources of error can be controlled through careful system calibration, the latter two are more difficult to control and may be inevitable. This is due to the fact that in ultrasound imaging using pulse-echo technique, all distances are calculated from the echo arriving time and the speed of sound. While the echo arriving time can be determined very accurately (within ±0.2 µs for a sampling frequency of 5 MHz), the speed of sound is medium dependent. For example, the speed of sound is 1480 m/s in water (at 20 °C), 1459 m/s in fat, and 1580 m/s in muscle (16). If the speed of sound used in distance calculation is incorrect, position errors occur. This can be demonstrated using the following example. Let us assume that a tissue reflector is located exactly 100 mm away from the transducer. Without knowing what type of tissue is between the transducer and the reflector, we may use a sound speed of 1500 m/s to perform all distance calculations. If the medium between the transducer and the reflector is all fat, an apparent distance of 102.81 mm (100 mm × 1500 m/s ÷ 1459 m/s) would be registered, which represents a position error of +2.81 mm. On the other hand, if the medium between the transducer and the reflector is all muscle, the apparent distance would become 94.94 mm (100 mm × 1500 m/s ÷ 1580 m/s), representing a position error of −5.06 mm. In practice, the ultrasound beam would encounter a mixture of media, composed of water, skin, fat, muscle, and so forth, along its route of transmission and reflection. If a speed of sound of 1500 m/s is used (a compromise between the speed of sound in water and soft tissues), the estimated achievable resolution of a compound image of a residual limb is 3 mm. Independent research is needed to determine whether this resolution is adequate for prosthetic socket design.

A more certain improvement in image quality is provided through speckle reduction by compounding. The theoretical signal-to-noise ratio (SNR) of an amplitude-coded B-scan image of diffuse scatterers is 1.91 (8). If N independent images of the same scatterers are averaged, the SNR increases by a factor of √N (8). Unlike the improvement in image resolution, the benefit of speckle reduction by compounding is less affected by the errors in pixel registration. However, image compounding increases the time for data acquisition as well as the time for image reconstruction. In addition, the increase in the number of component B-scan images for compounding also increases the likelihood of pixel misregistration, which will degrade image quality by decreasing the resolution.

To maximize the overall benefits of compounding, we used a method called “partial compounding,” in which the size of the sampled area of each B-scan is reduced so that only B-scan images obtained from several neighboring angles are actually compounded. As shown in Figure 4, although the original B-scan image contains 128 A-lines and the maximum scan depth is 180 mm, only the central 64 A-lines and the first 800 samples (representing a scan depth of 120 mm, which is 8 mm beyond the COR) in each A-line are used for compounding. There are two reasons for shortening the depth of each component B-scan image to only slightly beyond the COR. First, for most A-lines, the ultrasound beam has encountered the bone before reaching the COR. Consequently, only the first several hundred samples contain useful echo signals. Second, shortening the depth prevents compounding over images obtained from the opposite sides of the limb where the tissue compositions, and the speed of sound, may be significantly different from each other. Compounding over neighboring B-scan images therefore reduces the amount of pixel misregistration. There are also two reasons for using only the central 64 A-lines for compounding. First, as the transducer element rocks (oscillates) back and forth to scan a sector, the sweeping speed of the ultrasound beam is more linear near the center of the sector. As a result, the sample position can be determined more accurately for the central A-lines. As the transducer element moves toward each side of the sector, deceleration of the transducer movement makes it more difficult to determine sample positions accurately. Second, the effects of wave refraction are more significant for the side A-lines, since the incident angle (the angle between the incident ultrasound beam and the line perpendicular to the skin surface, i.e., the angle β in Figure 2) increases as the transducer element moves away from the central position, as seen in Figure 2. As a result, the position mapping errors are greater for the samples of side A-lines.

It should be pointed out that the problems related to
the side A-lines are associated with a mechanical sector scanner only. These problems can be largely eliminated by using a linear array transducer. A linear array transducer uses electronics to steer the ultrasound beam and contains no moving parts. As a result, the A-line positions are more stable and can be determined more accurately, even for the side A-lines. Second, a linear array transducer produces parallel ultrasound beams. As a result, the incident angle of the ultrasound beam is small even for the side A-lines. Finally, a linear array transducer allows dynamic focusing that can achieve a better and uniform lateral resolution within the entire scan range. It is therefore expected that the image quality can be further improved by using a more advanced ultrasound scanner equipped with a linear array transducer. Previously, array transducers and dynamic focusing were only used by relatively expensive (more than $100,000) and large scanners. As the technology has become more mature, several relatively low-cost (less than $30,000) and portable scanners that provide these more advanced features are now available on the U.S. market.

In addition to the factors discussed above, the image quality can be significantly degraded if the limb moves during ultrasound scan. Currently, we use a pair of thigh clamps to secure the proximal limb and use a limb stabilizer to provide a light support to the distal end (see Figure 1). In a preliminary trial with 15 persons with amputations, we found that most could hold their limbs steady during the entire scanning period (12–15 min). However, we have also encountered persons who had difficulties maintaining the same position for more than 1 min. A very important area for future study is, therefore, to reduce the time needed for data acquisition, develop methods for detecting and quantifying limb movement during scan, and develop methods for removing the effects of limb movements and maintaining a high quality in 3-D reconstruction of the limb model.

CONCLUSION

This study demonstrates the feasibility of developing a low-cost, ultrasound-based system to perform 3-D imaging of a residual limb. By using the compound scanning technique, a complete cross section of the limb can be reconstructed, showing bones and adjoining soft tissues. In addition, by carefully calibrating the scan system and adequately choosing the reconstruction parameters, the compound process can improve the lateral resolution and reduce the speckle noise. The high contrast at water-skin interface allows the external shape of the limb to be reconstructed with a relatively high resolution and accuracy. The additional information about the internal tissue structure may help the prosthetist locate the critical areas for shape modification and estimate the required quantity of shape rectification.

Future works include improving image quality by using an ultrasound scanner equipped with a linear array transducer and the circuitry for dynamic focusing (we recently purchased an ultrasound scanner made by Hitachi, model EUB-405); improving the hardware for stabilizing the limb during scan; increasing the speed of data acquisition; investigating methods for detecting and quantifying limb movement during scan; and developing algorithms for removing motion artifacts in image reconstruction. In addition, new visualization tools will be developed to improve image display and enable the prosthetist to create landmarks directly on the limb image. Finally, clinical trials need to be conducted to test the hypothesis that a 3-D visualization of the limb can indeed improve the quality of socket design and fitting.

ACKNOWLEDGMENTS

The programs for data acquisition and image reconstruction were written by Mr. Qun Chen. The ultrasound scanning apparatus was constructed by the Instrument Shop at Wright State University under the direction of Mr. James Arehart. Their contributions to this study are acknowledged.

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Submitted for publication August 9, 1996. Accepted in revised form December 11, 1996.