

Ultrasound Imaging of Trabecular Bone: Simulation and Experimental Results

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Abstract— A new ultrasound surgical guidance tool for accurately inserting screws during spinal fusion surgery is being developed. The technique requires generating B-mode images from within a small bore hole in trabecular bone. The purpose is to determine the viability and safety of the hole placement for subsequent insertion of a screw. This paper endeavors to develop an improved understanding of ultrasound propagation in trabecular bone and to understand the factors that govern B-mode image quality. Specifically, the results of numerical simulations on the effects of frequency and bone volume on image quality are presented along with demonstrating B-mode image formation obtained in-vitro on a human pedicle using a 3.2 MHz probe.

Index Terms—Imaging, Trabecular Bone, Spinal Surgery.

I. INTRODUCTION

Ultrasound imaging of trabecular bone remains difficult due to the high scattering and attenuation of bone. While many studies have been conducted using through-transmission ultrasound to measure properties of bone [1], few attempts have been made to perform ultrasound imaging within bone. As part of a larger project to develop an ultrasound guidance system for spinal fusion surgery [2], this paper reports our investigations on the ultrasound image quality that can be obtained in trabecular bone, and the effects of the percentage of bone volume and transducer frequency. For the purposes of the intended application, image quality was defined as the visibility of the trabecular and cortical bone interfaces. Simulation as well as experimental work was conducted.

II. METHODS AND MATERIALS

All human vertebral bone specimens were obtained through the anatomy department at the University of Toronto after receiving ethics approval from the research ethics board at St. Michael's Hospital (REB 06-223C) and the Coroner.

A. Simulations

Simulations were conducted using Finite-Difference-Time-Domain simulation program *Wave3000* (Cyberlogic), using the geometry imported from micro-CT scans. A human

vertebra (F, age 60, L4 lumbar) was cleaned of all soft tissue and bone marrow using a scalpel and water jet, and the vertebral body was isolated. The vertebral body was then scanned with micro-CT to yield three 3-D segments of trabecular bone. For logistical reasons, the first segment was obtained with a SkyScan 1172 micro-CT scanning at a resolution of 57.3 voxels/mm, while the second and third segments were obtained with a SCANCO Medical micro-CT 40 at a resolution of 66.7 voxels/mm. Each segment of bone was selected from a different anatomical region of the same vertebral body. The method of trabecular dilation/erosion was used to create three different bone volume % models from each segment (low, medium and high) as listed in Table 1. Each model was then divided into three slices, each slice being 3.0×4.9×3.0 mm in size.

The trabecular geometry from each slice was used in the simulation program to generate A-mode images for both 1.0 and 2.0 MHz plane-wave transducers transmitting high bandwidth pulses. Details of the simulation setup and material properties are shown in Figure 1. The trabecular bone was aligned so that the ultrasound beam traveled along the direction of dominant trabecular alignment, which will cause the highest attenuation and allow the worst-case scenario to be investigated [3]. A 1 mm thick layer of cortical bone was added to the end of each slice.

The transducer was a flat 3×3 mm square, occupying an entire surface of the simulation space, at a distance of 2.1 mm from the trabecular bone. The boundary conditions were selected to create a plane wave; infinite boundary conditions were used at the ends to prevent reflections. Water was substituted for bone marrow in the pores of the trabecular bone. Bone volume % was calculated as the volume of bone divided by the total volume of the trabecular bone (including the water). Water was also used for coupling between the transducer and the trabecular bone.

B. Experiments

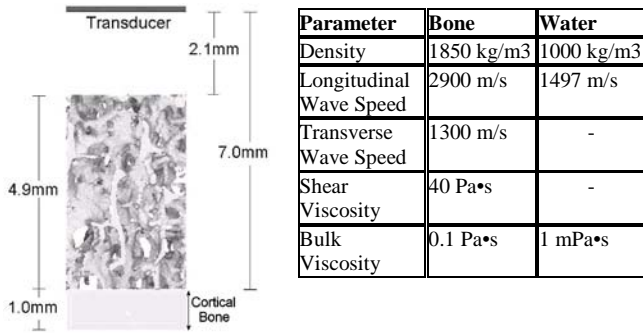
To conduct experiments, another human L4 vertebra (M, age 83) was obtained, and cleaned of all soft tissue and bone marrow. The specimen was cleaned with methanol and stored

in a freezer. The pedicle was isolated from the vertebral body and a hole was drilled along the long axis of the pedicle. This mirrors the procedure followed during spinal fusion surgery.

TABLE 1- BONE VOLUMES FOR MODELS AND INDIVIDUAL SLICES

Segment Size, Resolution	Bone volume %, each model	Bone volume %, individual slices	Slice #
3.3×18.3×3.5mm 57.3 voxels/mm		5.4%	1
		6.7% (low)	2
		9.0%	3
		9.5%	4
		11.4% (mid)	5
		15.0%	6
		12.8%	7
		16.9% (high)	8
		19.2%	9
3.0×17.0×3.0mm 66.7 voxels/mm		5.2%	10
		6.9% (low)	11
		10.5% (low)	12
		6.0%	13
		9.0%	14
		11.4% (mid)	15
		11.1%	16
		11.9%	17
		21.5% (high)	18
3.0×15.0×3.0mm 66.7 voxels/mm		4.7%	19
		6.8% (low)	20
		8.3% (low)	21
		7.1%	22
		9.2%	23
		11.7% (mid)	24
		12.3%	25
		12.4%	26
		17.3% (high)	27
	16.4%		

FIGURE 1- SIMULATION SETUP FOR A TYPICAL SLICE (LEFT) AND MATERIAL PROPERTIES (RIGHT)



An ultrasound probe with a 3.2 MHz centre frequency was then inserted into the pedicle. The probe has a diameter of 3.5 mm, with the transducer being 2 mm×4 mm in size. Both the probe and the bone were immersed into a water tank, which provides coupling. A Panametrics Model 5800 transmitter/receiver was used to transmit at a 200 Hz repetition frequency. The received signal was displayed on a Tektronix TDS3012B oscilloscope and captured with a custom software module. Since the ultrasound probe can obtain only one-dimensional A-mode images, many images are required to construct a two-dimensional image from inside the bone.

Sixty-seven images obtained at 5.4 degree intervals were used to create each 360 degree image. Low pass interpolation was used to smooth the images. Images were obtained at different regions along the length of the pedicle. The pedicle was then scanned with the SCANCO micro-CT at a resolution of 55.6 voxels/mm, allowing comparisons to be made between the ultrasound and micro-CT images of the bone.

III. RESULTS

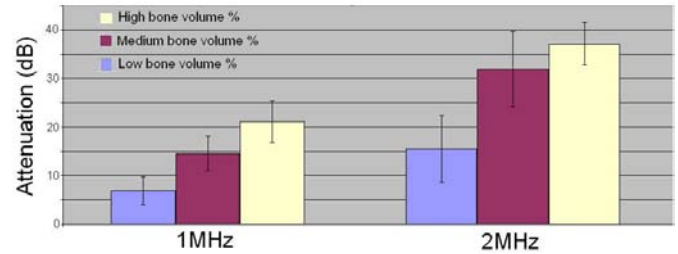
A. Simulations

Each simulation results in a one-dimensional A-mode image. The reflection from the cortical bone was isolated and its frequency spectrum obtained using FFT. The attenuation was calculated as

$$\alpha(f) = 20 \log\left(\frac{A_{sig}(f)}{A_{ref}(f)}\right), \quad (1)$$

where f is the centre frequency of the transducer, A_{sig} is the Fourier transform of the reflection and A_{ref} is the Fourier transform of the transmitted high bandwidth pulse. The elapsed time between the first peak of the transmitted pulse and the first peak of the cortical reflection was defined as the time-of-flight.

FIGURE 2 – ATTENUATION DEPENDENCY ON BONE VOLUME % AND ULTRASOUND FREQUENCY. N=9 FOR EACH BONE VOLUME %.



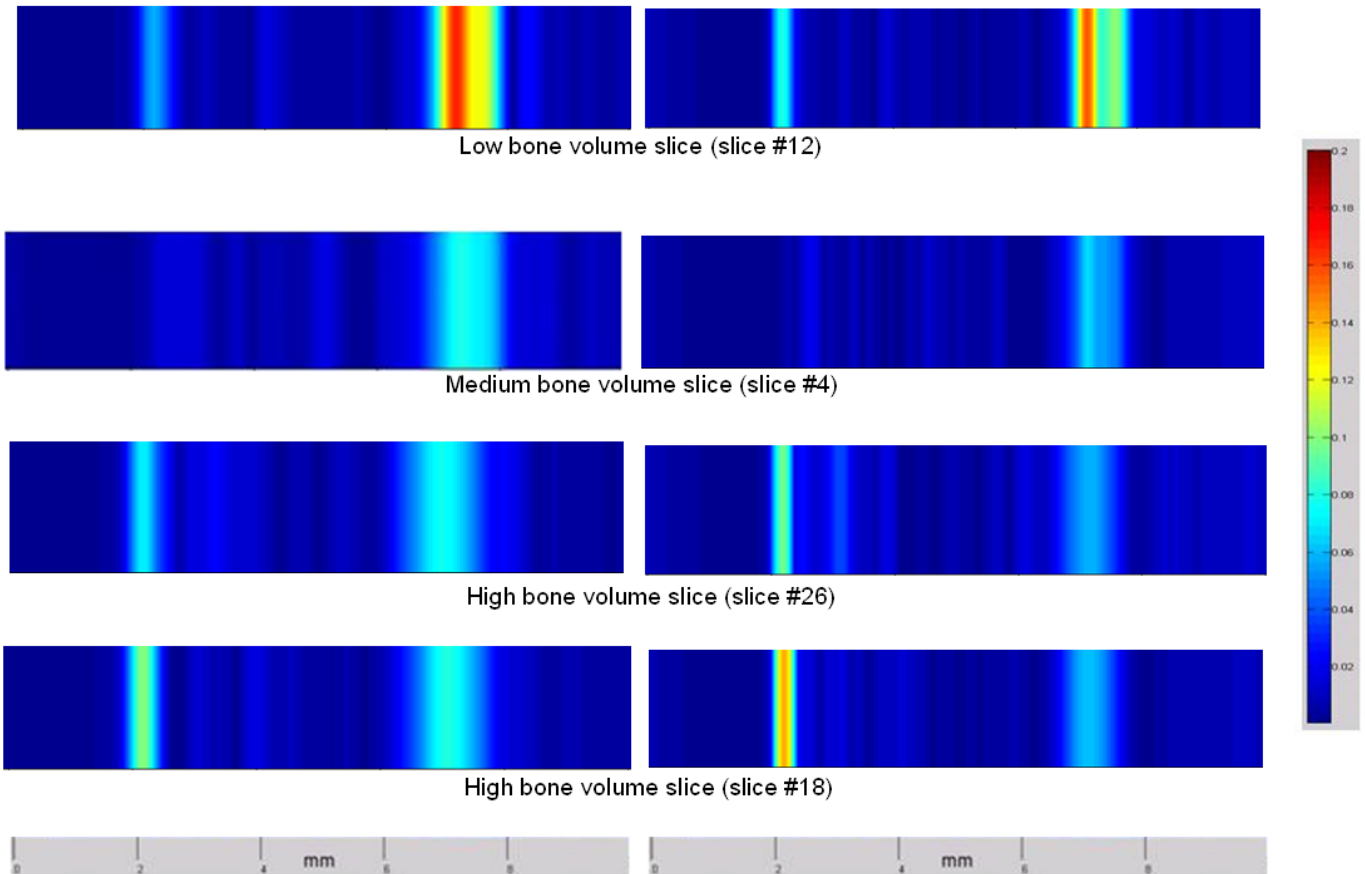
Simulation results shown in Figure 2 reaffirmed the expected correlation between bone volume and ultrasound attenuation. Specimens with high bone volume percentage exhibited the highest attenuation. At these frequencies, the main component of attenuation is most likely scattering rather than absorption [1]. A dependency between transducer frequency and attenuation can also be seen. This would suggest that images of high volume bone may have reduced visibility of the cortical bone.

Figure 3 shows a sample of 8 simulation images, obtained by using the Hilbert transform to calculate the envelopes of the echo signals and then converting the time axis into distance using an assumed speed of sound of 1545 m/s. In most images the water/trabecular interface (at 2.1 mm) and the trabecular/cortical interface (at 7.0 mm) create visible reflections. The magnitude of the reflections depends on the bone volume of the particular slice. Several trends were observed:

FIGURE 3 – SIMULATION A-MODE IMAGES SHOW DEPENDENCY ON BONE VOLUME % AND TRANSDUCER FREQUENCY.

1MHz images

2MHz images

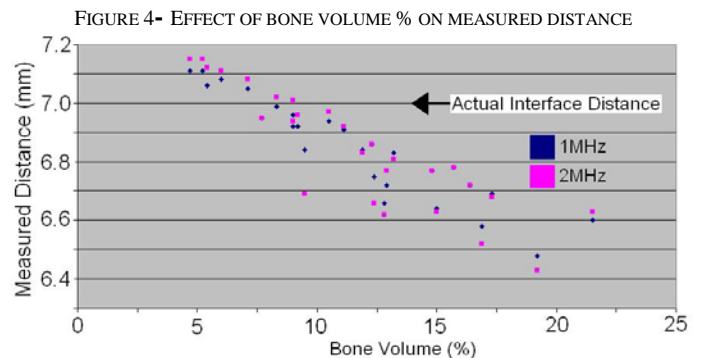


- In lower volume bone, reflections from the trabecular bone interface are generally weaker. This is most likely due to the acoustic impedance of low volume bone being close to that of water.
- Reflections from the cortical bone interface are generally stronger for low volume bone. This indicates that more acoustic energy passes the trabecular interface and is less attenuated in its passage through this medium.

The opposite trends were observed for high volume bone. Bone volume and transducer frequency therefore contribute to the differences observed between different slices. However, there is also considerable variation between slices from the same bone volume model. This is due to two factors: variation in bone volume between difference slices from the same model, and differences in the trabecular geometry between slices.

An additional effect can be seen as bone volume is increased: a shift in the distance measured for the cortical bone. Generally, in high volume bone, the cortical bone boundary appears closer to the transducer. This is most likely due to the material properties chosen for bone in these

simulations, in which bone has a longitudinal wave speed of 2900 m/s, approximately twice the assumed speed of sound used for distance calculations. This discrepancy becomes more pronounced in high volume bone. Another factor may be the greater predominance of fast waves in high volume bone compared to low volume bone [4]. With decreasing bone volume the speed of the fast waves decreases and approaches that of water. Figure 4 shows the effect of bone volume on the measured distance.



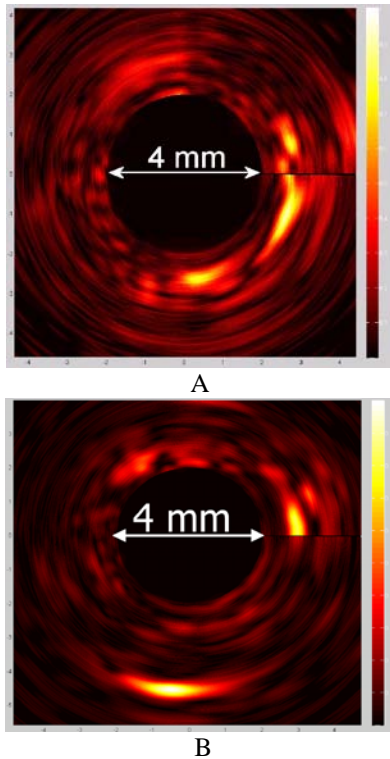
The simulation results showed that in most cases, the cortical bone is visible, and that the visibility of the cortical bone diminishes at higher bone volumes and higher transducer frequency. The accuracy of the measured distance was found to be within +0.2 mm and -0.6 mm of the actual distance, with a strong dependency on bone volume. Nonetheless, the accuracy is quite sufficient for the planned application in spinal fusion guidance.

B. Experiments

Experiments were conducted to determine whether the cortical bone will be visible. Some key differences between the experimental and simulation protocols include:

- The bone volume of the vertebral specimen was fixed and could not be adjusted as in the computer model.
- Transducer frequency was fixed at 3.2 MHz.
- The pedicle of the vertebra was imaged in experiments, whereas in simulations the models were created from the vertebral body.
- Due to the shape of the pedicle, the cortical bone was curved rather than flat as shown in the simulation model.
- Diffraction effects will be present in the experiments due to the finite size of the transducer.

FIGURE 5 – EXPERIMENTAL B-MODE IMAGES FROM A HUMAN PEDICLE. A) JUST INSIDE THE PEDICLE, AND B) 8 MM ANTERIOR.



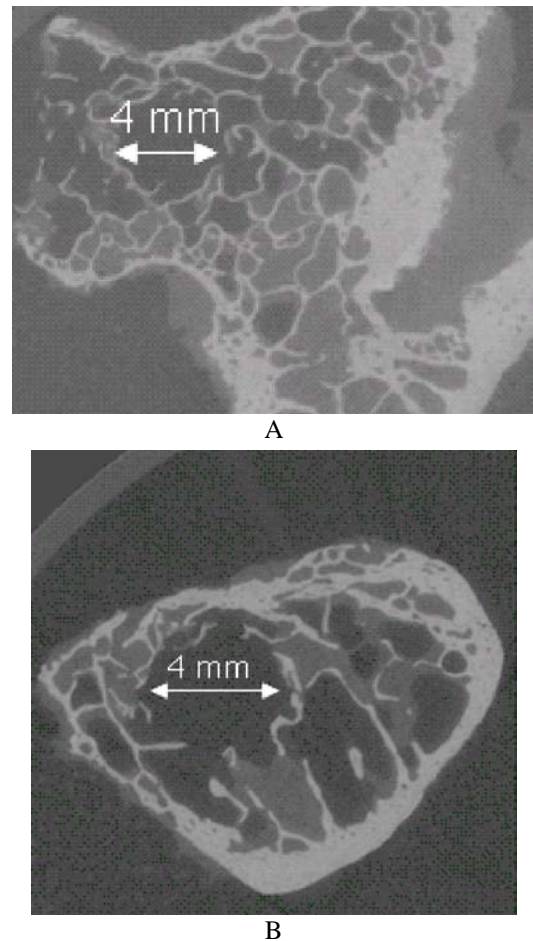
A speed of sound of 1545 m/s was assumed. The first image was obtained with the transducer placed just inside the pedicle. The second image was obtained 8 mm anterior. The ultrasound images are shown in Figure 5, with the medial direction to the right, and the lateral direction to the left. The

ultrasound probe occupies the empty space in the centre. Strong echoes are observed in the first image from the medial, superior, and inferior directions. These are most likely reflections from trabeculae rather than from the cortical bone. In the second image, the inferior cortical wall is the most visible. The lateral wall may be less visible because it is thinner than the medial wall and therefore may serve as a weaker reflector [5].

Comparison between the experimental ultrasound images and the micro-CT images can help to interpret the ultrasound images as well as to verify their accuracy. Since each micro-CT image is a slice of thickness 0.018 mm, while the ultrasound transducer has a length of 4 mm, the ultrasound images actually encompass many consecutive micro-CT slices. Moreover the axis of transducer rotation may be slightly misaligned from the direction normal to the micro-CT slice. Both of these issues could cause difficulty in relating specific features of the two sets of images.

For the purposes of comparison, the micro-CT slice corresponding to the location of the centre point of the transducer was chosen. The second ultrasound image shows a strong reflection from the inferior direction, at a distance of 2.5-3.0 mm, close to that shown in micro-CT (Figure 6).

FIGURE 6 – CORRESPONDING MICRO-CT IMAGES FOR THE PEDICLE IN FIGURE 5 TAKEN FROM APPROXIMATELY THE SAME REGIONS.



IV. DISCUSSION

The simulation results suggest that for a wide range of bone volume, the cortical bone should remain visible, if greatly diminished due to the high attenuation. Methods such as time gain compensation may be utilized to compensate for the increased attenuation at high bone volume.

Distance measurement of the cortical bone was shown to be accurate to within 0.6 mm for the twenty-seven slices, which is sufficient for our purpose of developing a surgical guidance tool. The fact that an adjustment of the assumed speed of sound was not necessary to maintain good accuracy indicates that *a priori* knowledge of the bone volume may not be necessary to construct accurate images. It is also important to note that the accuracy of the distance measurement depends on the thickness of the trabecular bone and the amount of water in the coupling area between the transducer and the trabecular bone. In our simulations, these values were constant at 4.9 mm and 2.1 mm, respectively. These values vary in experiments from one region of the pedicle to another. More important than the absolute accuracy of the distance measurement is the relative distance of the different cortical boundaries (i.e. medial, lateral). In other words, it is more important for the ultrasound image to show the surgeon which cortical boundary is closest to the transducer.

One of the limitations of the simulations that have been conducted is the fact that the bone interfaces are flat and perpendicular to the direction of propagation. It was expected that the curved layer of cortical bone encountered in experiments would create less reflections. Also, due to the high frequency of the ultrasound transducer (3.2 MHz), our simulation results would indicate that the trabecular interface would reflect much of the incident energy. This may explain why the trabecular bone causes strong reflections in some of the experimental images. In other cases, the cortical wall is clearly visible while the trabecular bone does not cause any considerable reflection. This variation in experimental results was also seen in simulations, as different slices from the same region of bone can often exhibit significant variability in image clarity. This is most likely due to differences in bone volume and trabecular geometry even within the same specimen or simulation model. More experiments will be conducted in the near future on vertebrae from different individuals to determine the effect of inter-individual variability. A quantitative comparison between the micro-CT and ultrasound images will also be conducted.

V. CONCLUSION

The simulation results affirmed the feasibility of using ultrasound to detect the trabecular/cortical bone boundaries

with a sufficient degree of accuracy for surgical use. The simulations also allowed us to examine the effects of bone volume and transducer frequency. Importantly, the simulations provided insight on the level of variability to be expected when imaging a segment of trabecular bone. Due to the heterogeneity of trabecular bone, bone volume and trabecular geometry can vary from one region of bone to another, and this variability translates into differences in the reflections of the trabecular and cortical bone.

We expect our ultrasound probe will be more useful for surgical guidance in low bone volume patients. First, the lower bone volume translates into lower attenuation, more clearly distinguishable cortical bone, and higher accuracy. Second, spinal fusion surgery is more difficult in patients with low bone volume since it is more difficult for the surgeon to “feel” the cortical boundaries and ultrasound imaging would be more helpful in those cases.

Experiments showed clear visibility of some regions of the pedicle (medial, superior, and inferior) while the lateral region often returned only dim reflections. Since the spinal cord is located medial to the pedicle, and most cases of iatrogenic damage are caused by medial violations of the cortical bone, it is important for this region to be imaged clearly. Future experiments will be conducted to verify that these results are consistent over many specimens from different individuals.

VI. ACKNOWLEDGMENT

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