QUANTITATIVE BONE HEALING PROCESS MONITORING ULTRASOUND SYSTEM

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Abstract: In this work, a model that simulates the interaction of ultrasound waves with a human body is utilized to monitor fractured Tibia healing process. The effect of frequency of operation on the measurements is investigated. The Tibia bone fracture is one of the most difficult fractures to treat because of the slow healing process which may take up to 5 months. The human body is modeled as a four layer medium: skin, fat, muscle and bone. Each of these layers is characterized by its acoustic impedance. These impedances influence the incident, transmitted and reflected ultrasound pressure waves. A mathematical model which describes these interactions is used to compute the reflection coefficient at the transducer side. This coefficient will then be used to quantify the bone healing process.

Introduction

When a bone fractured, partial or complete break in the continuity of the bone occurs. This happens under mechanical stress that exceeds the limits of the bone's strength and stiffness [1]. A Bone fracture causes a sudden loss of bone continuity, stability and integrity. A fracture usually interrupts or cuts the blood vessels that result in the reduction of blood supply. One to two mm of bone dies on either side of the fracture due to the lack of blood supply. Treatment usually involves reduction or realignment of the bone, immobilization and restoring function through rehabilitation. The healing process starts when new bone forms and fills in the fractured area to restore the original continuity and solidity. It generally goes through the same series of stages for all fractures: inflammation, soft tissue formation, hard tissue formation, new bone and remodeling [2]. Healing is considered to be complete when the bone is almost identical to its original shape before injury and has regained its normal stiffness and strength.

A bone fracture may be diagnosed by many diagnostic imaging tests such as X – Rays, Computed Tomography (CT), Magnetic Resonance Imaging (MRI) or Vibratory devices [3-5]. Since the bone healing process may take several months in certain pathological cases, repetitive tests are required. X – Ray radiation is hazardous and is unsafe for such repetitive tests. Moreover, fractures are typically visible on primary radiographs. However, the healing process' Soft Tissue stage is difficult to visualize on radiographs. In other words, the fracture site is seen on an x-ray like a cloudy area. It becomes visible after three to six weeks after injury. Furthermore, X-Rays can not be used for pregnant women or patients who had a barium contrast media or any medication containing bismuth.

MRI tests are long and of high cost and can not be repetitive. The MRI machine itself is prohibitively expensive for small hospitals and therefore, it is not available everywhere. Additionally, people who are claustrophobic, nervous, or disturbed by the loud noise caused by MRI machines must be given some ant-anxiety medication before the examination.

Vibratory devices are mechanical sine wave vibrators that are applied to the fracture site [5]. An electromagnetic shaker is pushed against the skin near the distal end of the bone. A miniature accelerometer captures the response of the bone-prosthesis system to the applied vibrations. If there is no fracture, the bone system behaves linearly and only the excitation frequency will be detected. However, if there is fracture, the system will behave in a non-linear mode and harmonics will be detected by the accelerometer. Experiments have shown that an excitation frequency between 100 Hz to 200 Hz is well suited to detect fractures. Unfortunately however, vibratory devices suffer from the increased signal damping and locating the resonance frequency especially with obese subjects.

Ultrasound is the second most popular imaging modality after the X-rays. Ultrasound has a major advantage over X-rays that it is being non-ionizing radiation [6-8]. This is in addition to being a low cost and high resolution imaging modality. Ultrasound frequency is a very crucial factor in the image display or resolution. The wave propagation speed is also an important characteristic of the ultrasound wave. The speed varies depending on the frequency of operation and the acoustic impedance of the medium. Ultrasound imagers transmit sound waves (with frequencies in the MHz range) to the body using a piezoelectric transducer. Ultrasound waves interact with different materials such as skin, tissue, and bone. Each material has its own distinct acoustic impedance; therefore some of the sound waves energy will be reflected back to the transducer

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while the rest propagates through the medium. The reflected waves will be picked up by the same transducer which converts sound waves to pure electrical pulses. This work is devoted to model and simulate the interaction of ultrasound waves with a human body. Results of simulation will be utilized to develop a system that can be used to quantitatively monitor the fracture healing process.

Experimental System

The proposed system for the Monitoring of Bone Healing Process Using Ultrasound is composed of several stages as shown in figure 1. A pulse generator followed by a power amplifier and transducer form the transmitter circuit. The transducer, amplifier and a computer represent the receiver part of the system. A timing circuit is used to control the transducer.

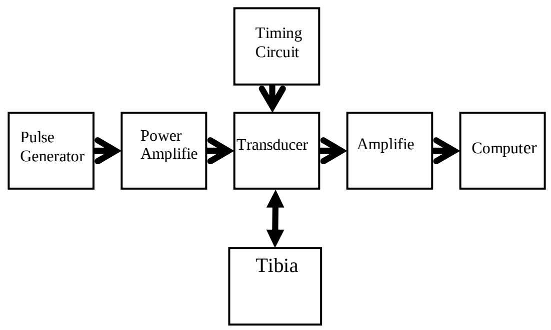

Figure 1: Block diagram of the ultrasound proposed bone healing process monitoring system.

Mathematical Model and Sample Description

Ultrasound testing techniques are applied in a monostatic or a bistatic fashion. In addition, these measurements may be conducted from only one side (echo/reflection measurements). Generally, resolution is dictated by the footprint of the ultrasound sensor which is usually small. However, covering a wide area takes a lot of scanning time. Additionally, in the application at hand, we only care about monitoring the healing process at a given location.

In an ultrasound regime the human tibia is modeled as a three layer medium: skin, fat, and bone. Each of these layers is characterized by its acoustic impedance. These impedances influence the incident, transmitted and reflected ultrasound pressure waves. The presence of a fracture or a crack can be modeled as an additional layer of callus on top of the bone. When an acoustic wave is incident on a structure, similar to the human body, part of it will be reflected at the boundary between any two layers and the rest will travel through. Waves reflect whenever they travel from one medium (layer) to another medium. The reflection coefficients between the layers are calculated until the wave reaches an infinite half space. By comparing the reflected wave from a layer to the incident wave on that layer, the reflection coefficient is obtained at that layer. The reflection coefficient at the first layer (effective reflection coefficient) can be calculated and measured as well. A mathematical model which describes these interactions is developed and used to compute the reflection coefficient at the transducer site. This coefficient will then be used to quantify the bone healing progress. The details of the mathematical model are given in [9]. The mathematical model is based on the transmission line technique to model the interaction of a pressure wave with a multilayered structure backed by an infinite half space of bone. The formulation was expanded for any number of layers by cascading reflection and transmission formulae shown below, respectively.

$$X_k = \frac{\frac{Z_k}{Z_{k-1}} Y_k - 1}{\frac{Z_k}{Z_{k-1}} Y_k + 1} \qquad \text{and} \qquad Y_{k-1} = \frac{1 + X_k . e^{-2j\gamma_{k-1} d_{k-1}}}{1 - X_k . e^{-2j\gamma_{k-1} d_{k-1}}}$$

Where, k is the layer number. Calculation starts from the last layer where Y_n =1 for the first iteration if the last layer is an infinite half-space (n = number of layers). By solving for all X_k and Y_k, X_1 will be the effective reflection coefficient.

Figure 2 shows a structure made of four layers, where the last layer is an infinite half space of bone. A code was developed to calculate the effective reflection coefficient due to the interaction of an ultrasound wave with any layered structure. The inputs to the code are the number of layers, their acoustic impedances and their thicknesses as well as the frequency of operation. The output of the code is the reflection coefficient. The presence of a crack is associated with a layer of callus that forms around the crack location. As the healing process progresses, this callus layer begins to vanish till it totally disappears. The developed model was applied to a stratified structure consisting of four planner layers as described below:

Layer 1: Skin, Z = $1.69\times10^6\, kgm^{-2}s^{-1}$, thickness: 1.5 mm,

Layer 2: Fasciae & Fat, Z = $1.38\times10^6\, kgm^{-2}s^{-1}$, thickness: 1.0 mm,

Layer 3: this layer is a fat layer with a thickness of 0.2 mm except for a small circular area (1 cm^2) where the fat is replaced by callus (with properties like bone). The callus thickness is estimated to be 0.2 mm at the beginning and decays to 0.0 mm when the healing is complete. The thickness of the callus depends on the healing stage. Consequently, monitoring this callus layer reveals information about the healing stage. To calculate the acoustic impedance in the third layer, a mixing model based on the portion of area seen from fat and callus is used. Consequently, the acoustic impedance of the layer consisting callus varies between that of fat to that of callus depending on what portion of each is seen by the sensor,

Layer 4: bone, Z $=7.8\times10^{6}\,kgm^{-2}s^{-1}$, thickness: 0.3 mm.

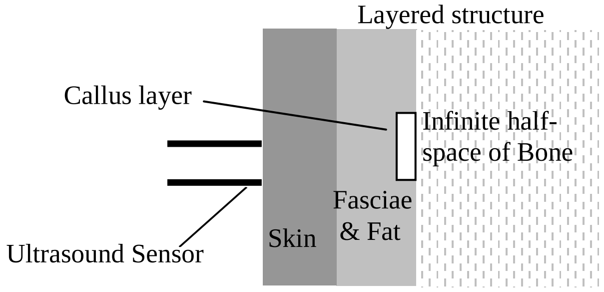

Figure 2: Fractured human tibia model at ultrasound frequencies.

Results

To study the influence of frequency of operation, frequencies in the range 5 MHz – 15 MHz were used. Simulation results were obtained for callus layer thicknesses of 0.2 mm, 0.15 mm, 0.10 mm and 0.05 mm. All graphs represent scans over the tibia starting just outside the callus layer and passing over the callus layer, i.e. the start and end points of each scan show the reflection coefficient when no callus is present. Figure 3 shows the reflection coefficient calculated as a function of scanning over a callus of thickness 0.20 mm at 5 different frequencies (6, 8, 10, 12 and 14 MHz). The change in the reflection coefficient begins to happen as the transducer (footprint = 1cm^2) partially begins to see the callus in the third layer. The variation ends as the callus is totally out of the footprint of the sensor. The intensity of the ultrasound wave is assumed to be uniform within the footprint. The results show change at all frequencies; however, the change in the reflection coefficient at 8 and 10 MHz is relatively large and no fluctuations around the no callus level are observed. This fact has a very important practical ramification since the fluctuations might mask the detection of the presence of callus in a measurement system. Figure 4 shows similar scans to the ones shown in figure 3 for a 0.15 mm thickness of the callus layer. Again, there are more fluctuations in the reflection coefficient level associated with this thickness of callus. However, these fluctuations are less at 10 and 12 MHz. Figures 5 and 6 show similar results for callus thicknesses of 0.10 mm and 0.05 mm, respectively. For callus thickness of 0.10 mm 6, 8 and 10 MHz show a large variation of the reflection coefficient while 12 and 14 MHz show less variation and more fluctuations, especially at 12 MHz. For the 0.05 mm thick callus all frequencies have a relatively large range of change in the reflection coefficient.

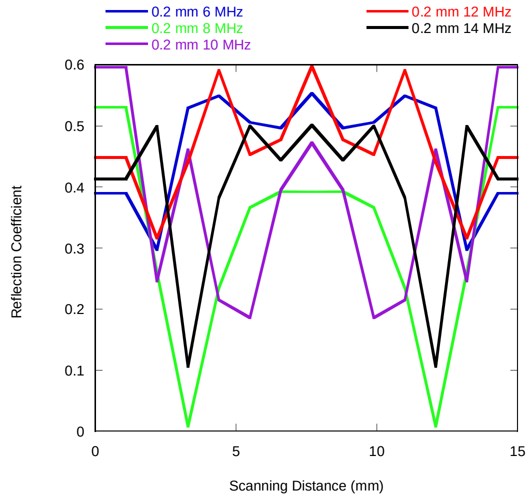

Figure 3: Reflection coefficient as a function of scanning over a callus layer of thickness 0.20 mm at 6, 8, 10, 12 and 14 MHz.

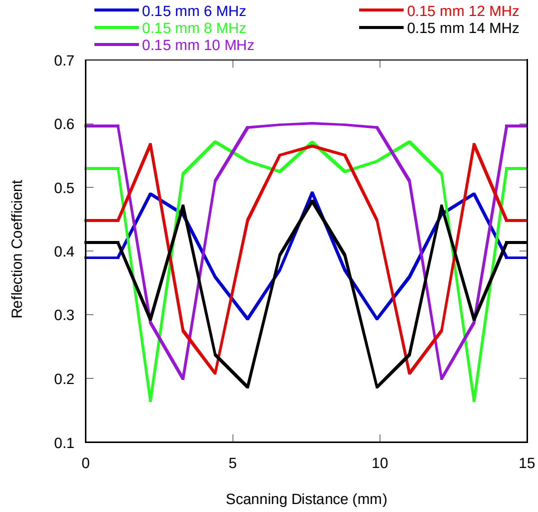

Figure 4: Reflection coefficient as a function of scanning over a callus layer of thickness 0.15 mm at 6, 8, 10, 12 and 14 MHz.

The results obtained thus far indicate that in order to achieve detection and estimation of the callus layer using the proposed ultrasound technique, multiple frequency measurements are needed. Figures 7, 8 and 9 show the results obtained at 8, 10 and 14 MHz for the different thicknesses of the callus layer. It is evident that the value of the reflection coefficient changes as the thickness of callus varies. This fact indicates that quantitative monitoring of bone healing may be accomplished. However, due to the fluctuations of the signal level, multiple frequencies should be utilized with a decision making system to avoid any uncertainty in

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the quantitative estimation of the callus layer thickness and hence the progress in the healing process.

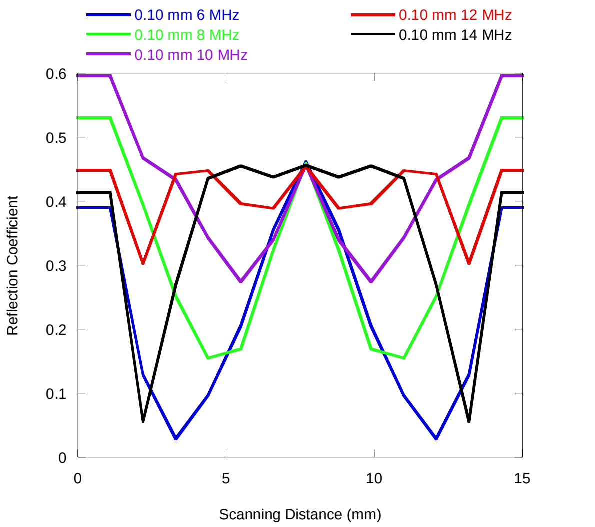

Figure 5: Reflection coefficient as a function of scanning over a callus layer of thickness 0.10 mm at 6, 8, 10, 12 and 14 MHz.

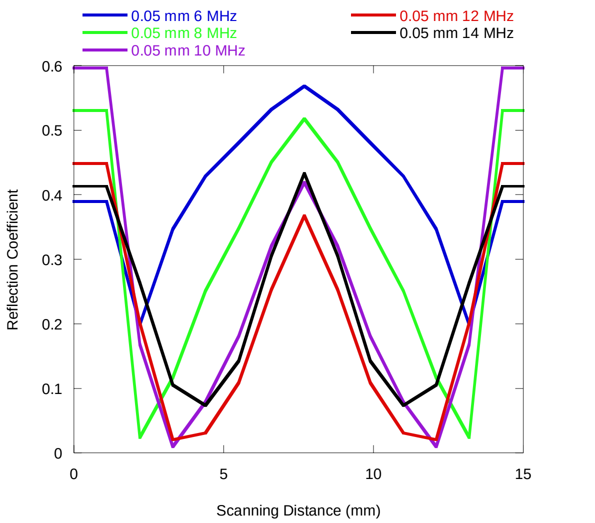

Figure 6: Reflection coefficient as a function of scanning over a callus layer of thickness 0.05 mm at 6, 8, 10, 12 and 14 MHz.

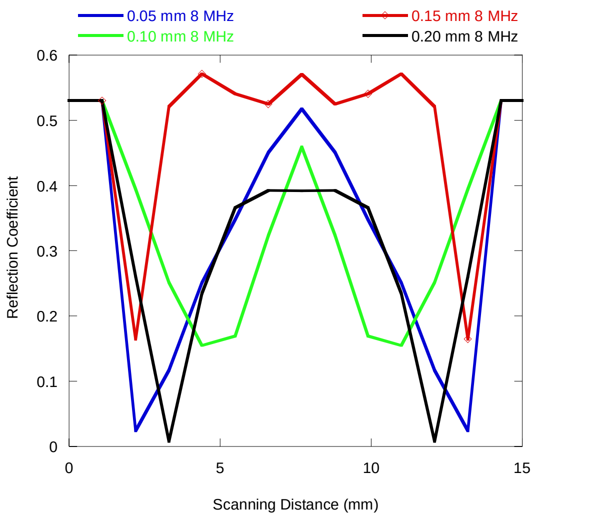

Figure 7: Reflection coefficient as a function of scanning over a callus layer of varying thickness at 8 MHz.

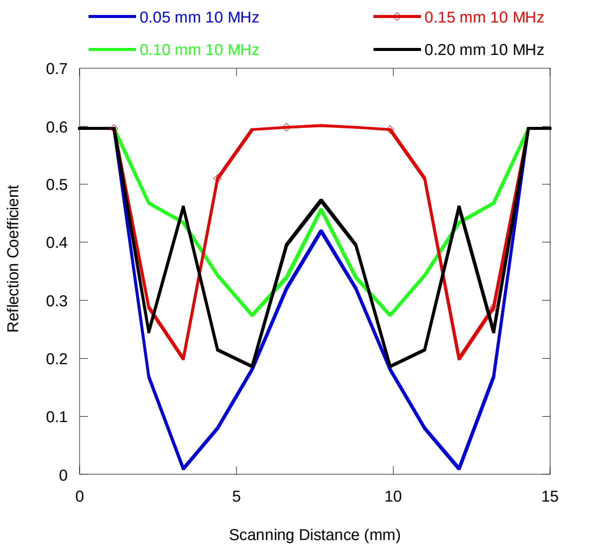

Figure 8: Reflection coefficient as a function of scanning over a callus layer of varying thickness at 10 MHz.

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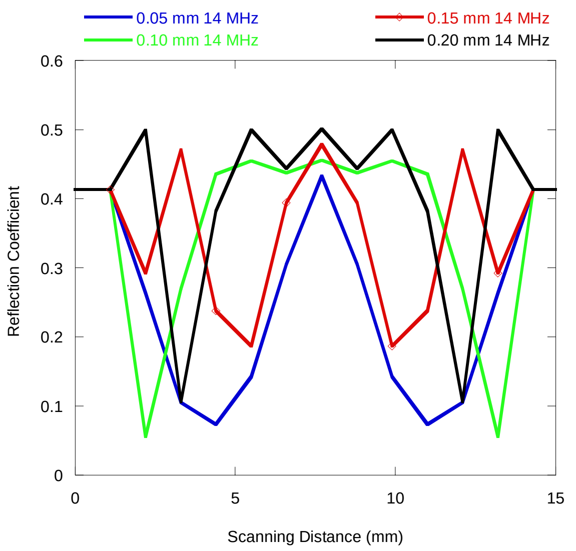

Figure 9: Reflection coefficient as a function of scanning over a callus layer of varying thickness at 14 MHz.

Conclusions

In this paper the fesability of using ultrasound quantitative monitoring of bone healing process was investigated. Simulation results indicate that monitoring the healing process of fractured tibia may be accomplished by monitoring the thickness of the callus layer. To arrive at a practical system, multi-frequency simulations were conducted. A decision making system may also be required to give an estimate of the callus layer thickness and resolve the uncertainty associated with the variation of the reflection coefficient observed. Once the measurement system is operational, the decision making system may be used to determine the shape of the callus layer and its extent.

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