Development of Ultrasound Based Techniques for Measuring Skeletal Muscle Motion

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Presentation Outline

- Introduction
- Thesis Objectives
- Mathematical Model and Principles
- Methods of Muscle Motion Measurement and Motion Artifact Removal
- Simulation Environment and Experiments
- In Vivo Experiments
- Conclusion
- Future Work

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Introduction

- The musculature of the body has a great deal to do with the overall health of an individual
  - Athletic injury
  - Muscular disorder
- Understanding the physical properties of muscles and how and why they move is of ongoing interest to many medical practitioners
  - Increasing the amount and type of information that can be measured could lead to a much better understanding of the musculature and its general effect on well-being

Thesis Objectives

- Design and implement ultrasonic techniques to study human skeletal muscle capable of measuring:
  - Internal tissue displacement and velocity
    - Remove the effect of motion artifacts
  - Relative strain information of the tissue being imaged
- Design phantom simulation environment to test methodology
- Perform phantom simulation and in vivo testing of techniques

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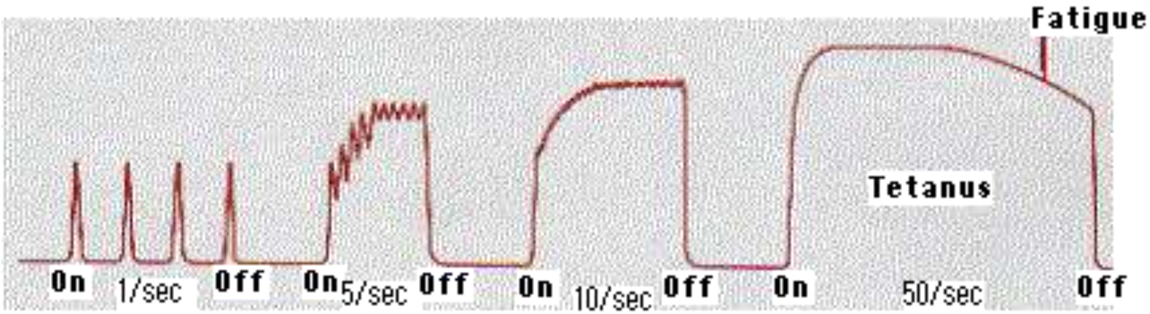
Skeletal Muscle

- Attached to the bones of the skeleton and control body movement
- Contraction of muscle fibers occur in response to an electrical stimulus from either a motor neuron or external source
  - At a rate of 1 Hz the muscle responds with a single twitch
  - At a rate between 5 Hz and 10 Hz the twitches begin to fuse together in a phenomenon called clonus
  - At a rate greater than 50 Hz the muscle goes into smooth, sustained contraction called tetanus



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Current Methods to Measure Muscle Motion

- Electromyography (EMG)
  - Measures electric current resulting from muscle contraction with skin surface or needle electrode
  - EMG data is a summation of the electrical signals from under the electrode
- Mechanomyography (MMG)
  - Measures the mechanical motion resulting from a muscle contraction
  - Most commonly done with skin surface mounted accelerometers
- Magnetic Resonance Imaging (MRI)
  - Most commonly used to measure tissue strain
- Ultrasound (US)
  - Most commonly used to track tissue motion and measure strain
  - Theoretically capable of both internal and external measurements
    - » Not limited to surface motion or summations

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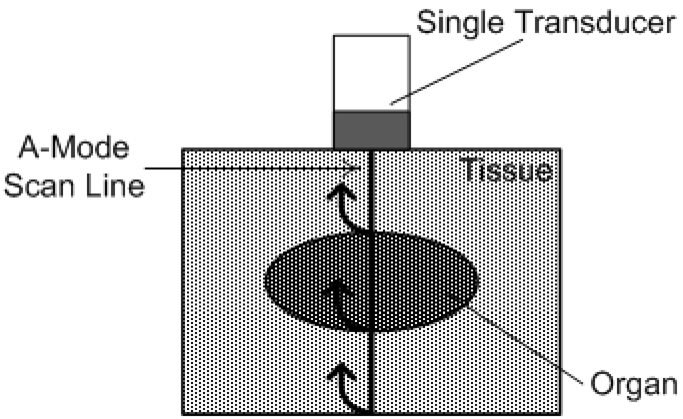
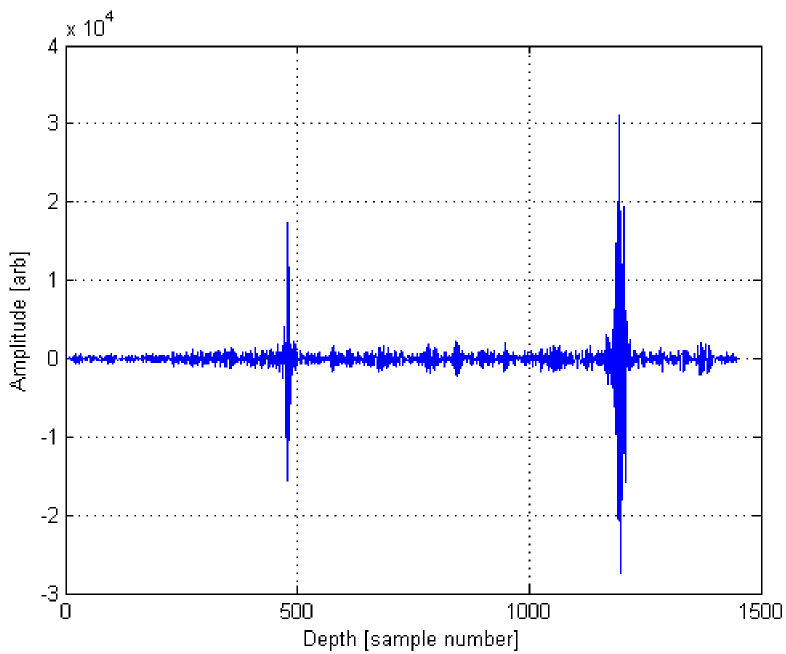
Ultrasound Principles

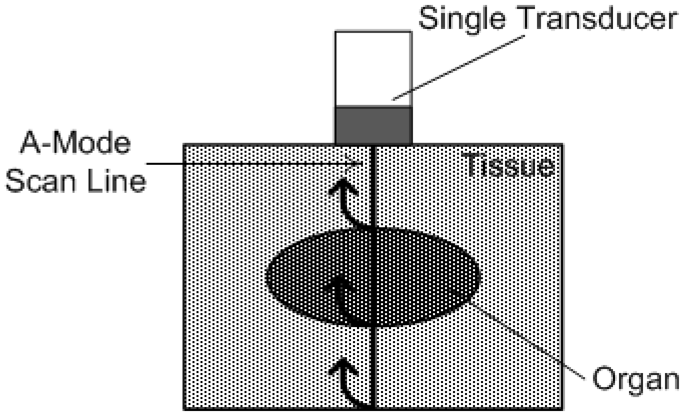


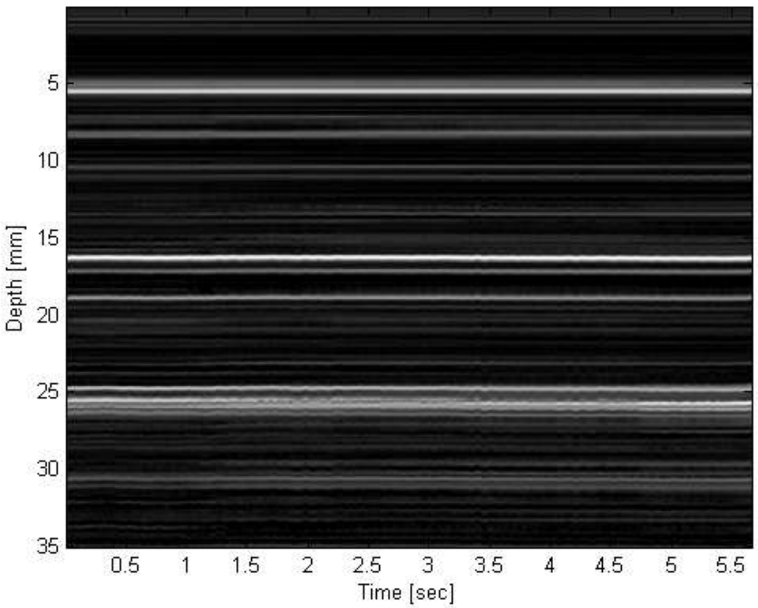


- Transducer is excited on a repetitive basis (PRF)
  - Image created based on received echo
- Three major imaging modalities:
  - A-mode: amplitude or envelope signal from a single transducer usually displayed on an oscilloscope
  - M-mode: A-mode scan lines from a single transducer are converted into grey scale and used as columns of an image
  - B-Mode: A-mode scan lines from multiple adjacent transducers are converted into grey scale and used as columns of an image

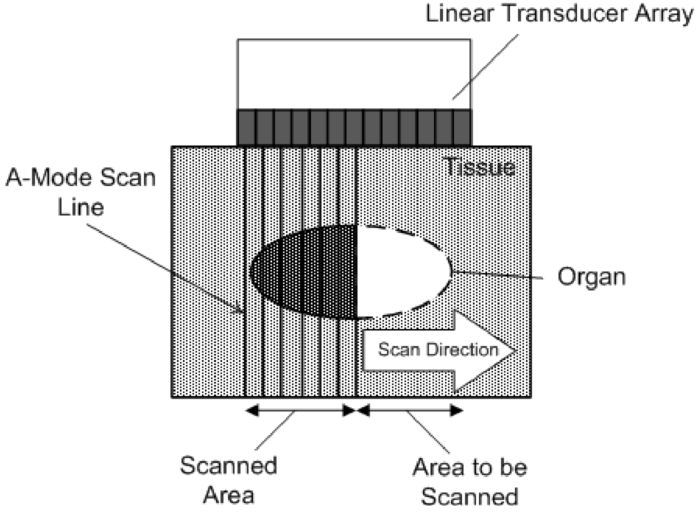
Ultrasound Principles

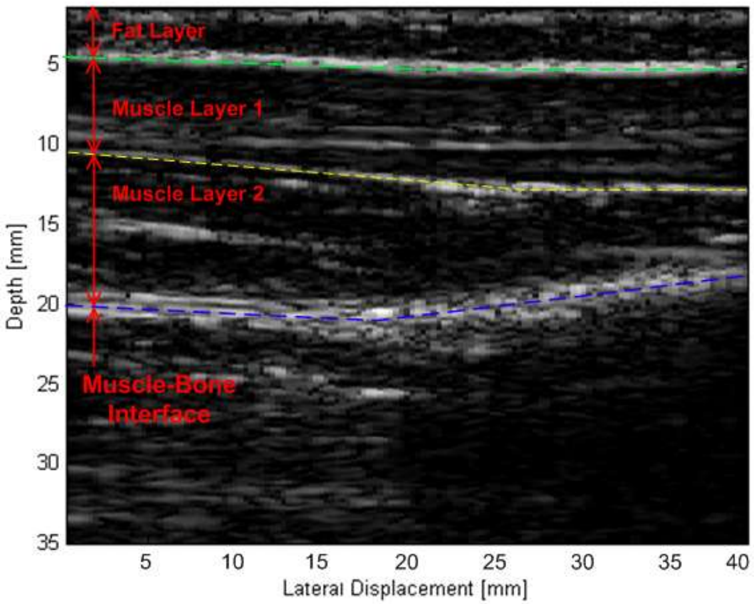
M-Mode





B-Mode





Ultrasound Hardware

- Ultrasound machine:
  - Picus system
  - Clinically FDA approved
- Ultrasound probe:
  - L10-5 40 mm linear array probe
  - 127 transducers separated by 315 µm

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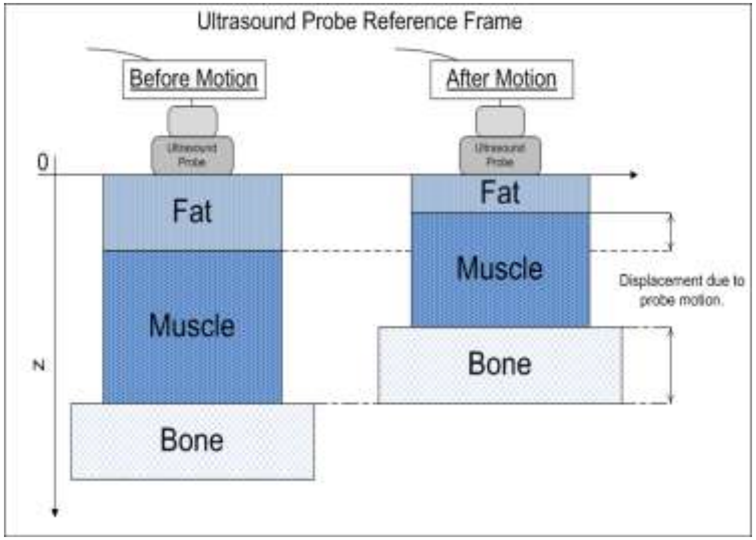
Reference Frame and Coordinate System

- Ultrasound probe reference frame places coordinate system origin at the surface of probe
  - All motion is relative to the surface of the probe
    - Displacement in a direction towards the probe is negative and away from the probe is positive
    - Displacement magnitude increases with respect to depth for a uniform object under a constant, external force applied at the surface of the probe



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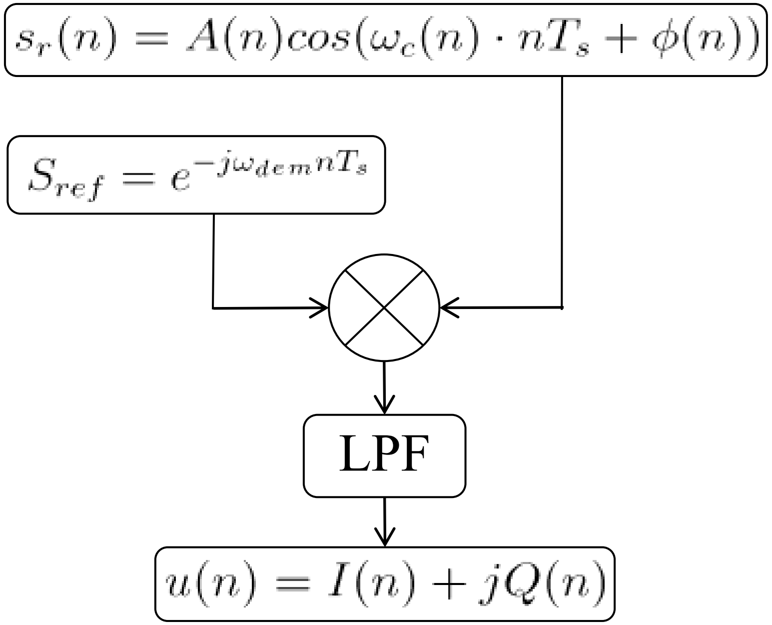
Displacement Estimation

- Target motion toward or away from the probe will cause consecutive received pulses to experience a shift in phase
  - This phase shift is recovered by quadrature detection and converted to an estimation of displacement using autocorrelation

Quadrature Detection



$$s_r(n) = A(n)cos(\omega_c(n) \cdot nT_s + \phi(n))$$

$$S_{ref} = e^{-j\omega_{dem} nT_s}$$

$$u(n) = I(n) + jQ(n)$$

Autocorrelation

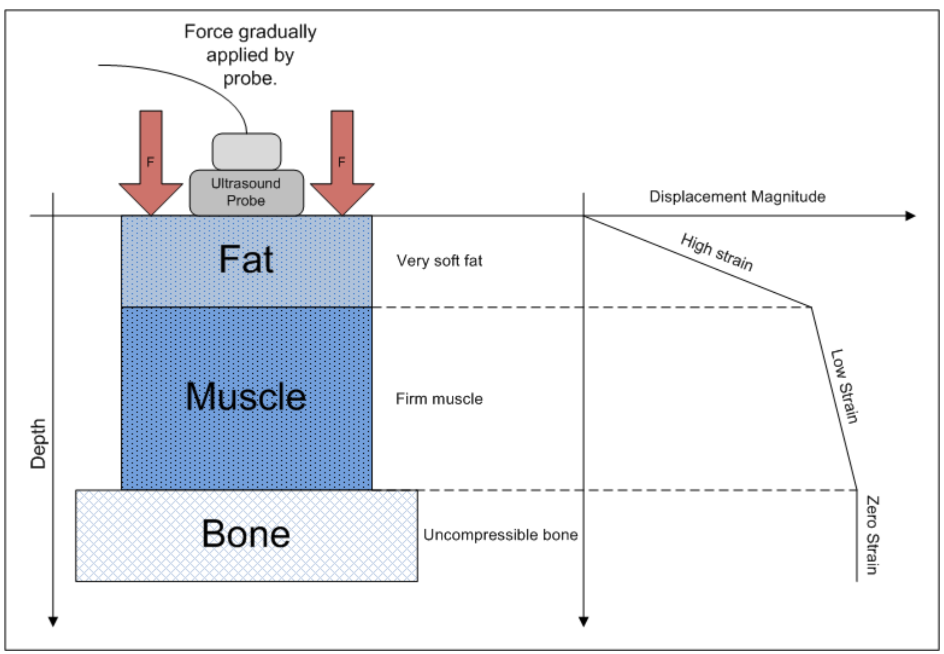
$$r_{n,m}(N,M) \equiv u_z(n,m) \cdot u_z^*(n+N, m+M)$$

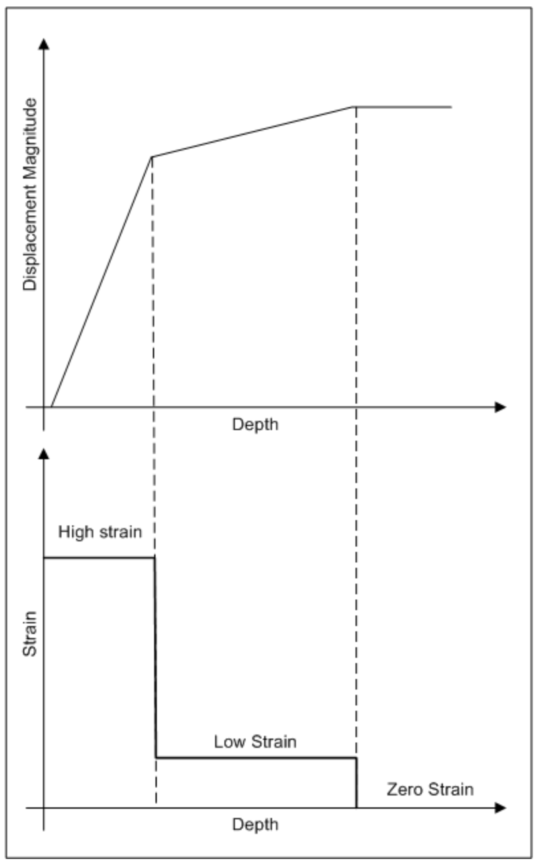
$$\Delta d_z(n,m) = \frac{c\Delta\phi_z(n,m)}{4\pi \bar{f}_c(n,m)} = \frac{c}{4\pi \bar{f}_c(n,m)} \angle r_{n,m}(0,1)$$

$$d_z(n,m+1) = d_z(n,m) + \Delta d_z(n,m)$$

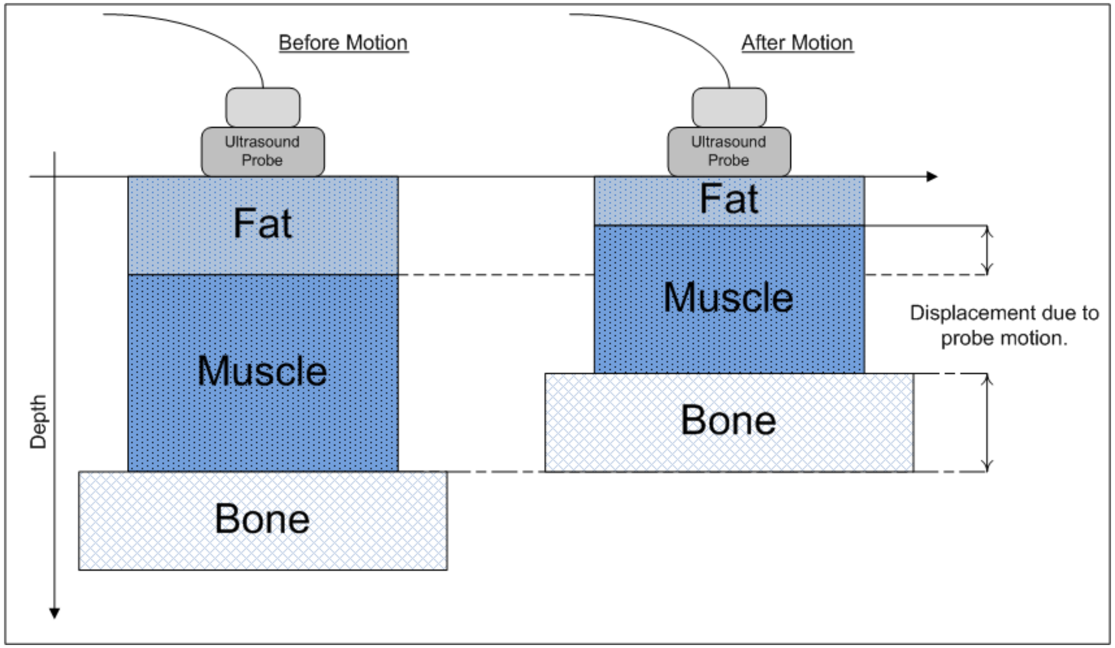
Strain Estimation

- Axial strain is a measure of the relative deformation of an object and can be estimated by the spatial gradient of displacement in the axial direction



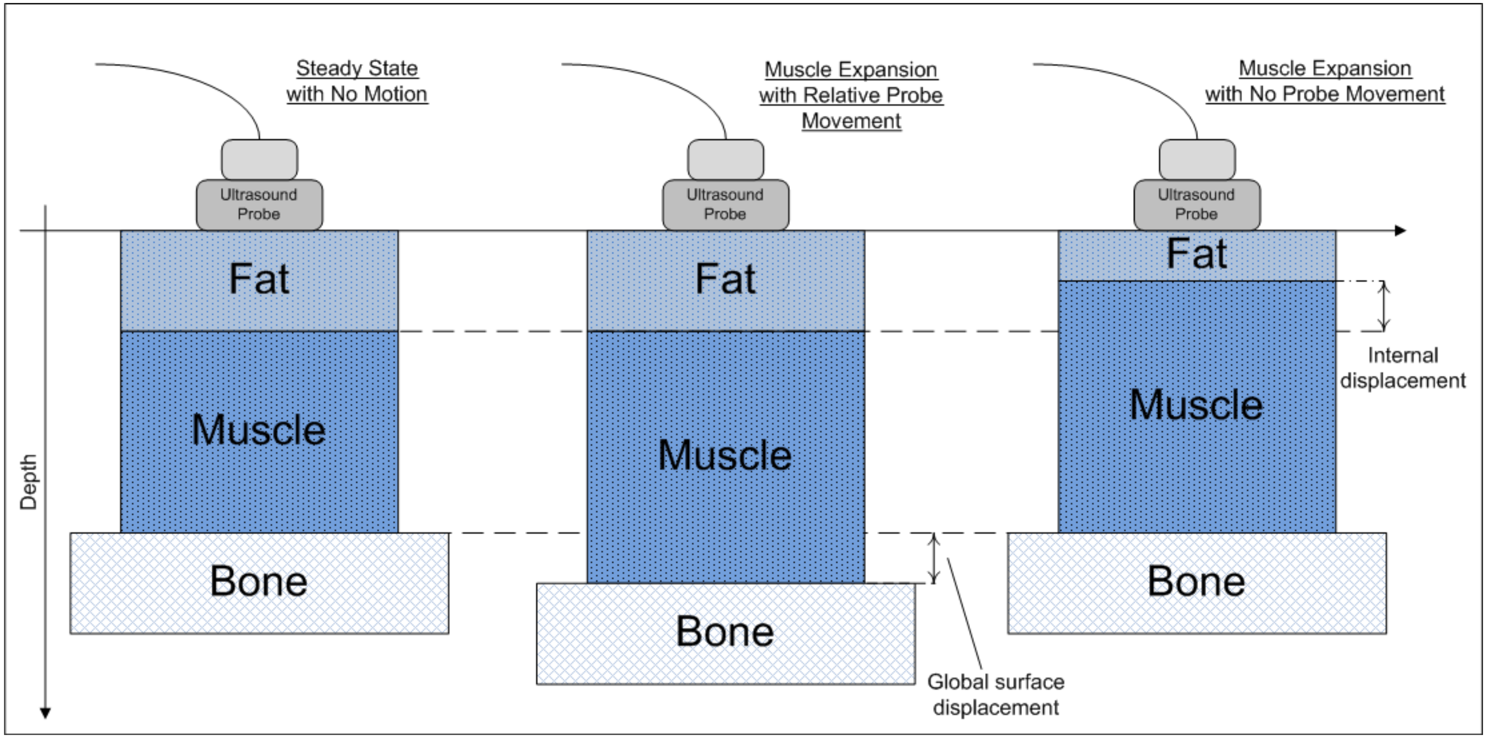


Global Surface Motion



- External motion occurs due to motion of the probe or object being imaged
  - Classified as artifact motion
- Estimated by ideally fixing the probe to the object surface and measuring the change in distance between the probe and bone surface

Internal Tissue Motion



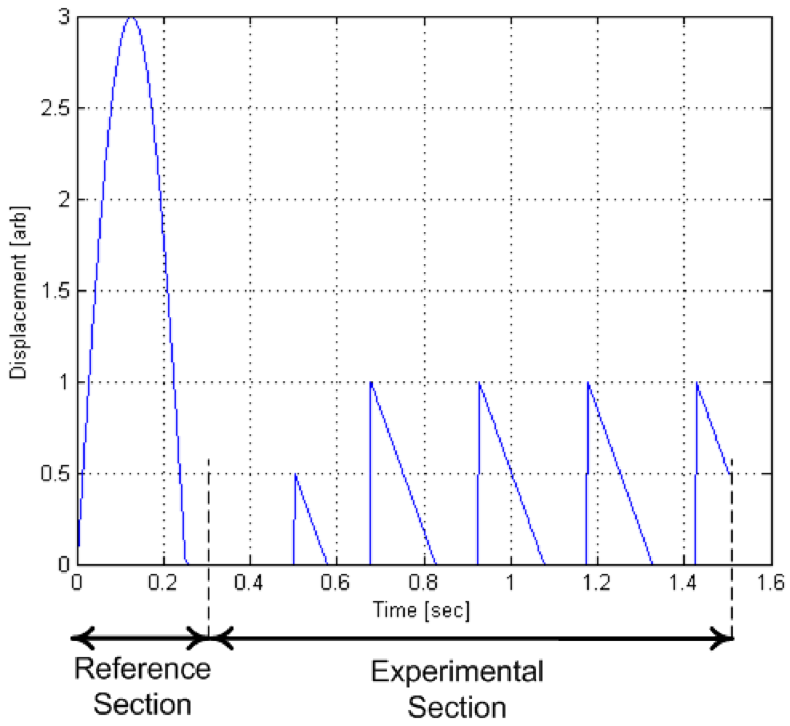
- Internal motion occurs and originates within the object being imaged
  - Ideally no probe or external object motion should occur
- The distance between the probe and bone surface must remain constant in order to accurately estimate internal motion

Removing Motion Artifacts

- Practically very difficult to keep a fixed distance between probe and bone surface
  - Bone boundary algorithm for motion artifact removal fixes the distance between probe and bone surface regardless of the motion occurring during data acquisition
  - Bone is uncompressible and should not experience motion due to muscle contraction. Any motion measured at the surface of the bone can be assumed to be a result of a motion artifact
- The three main components of the algorithm are:
  - Measurement procedure
  - Bone boundary tracking
  - Depth scaling and motion artifact subtraction

Measurement Procedure

- Divided the experimental procedure into two different sections of collected data:
  - Reference Section
    - Contains only external motion caused by pushing probe down into the object being imaged
    - Used for depth scaling
  - Experimental Section
    - Contains desired experimental signals



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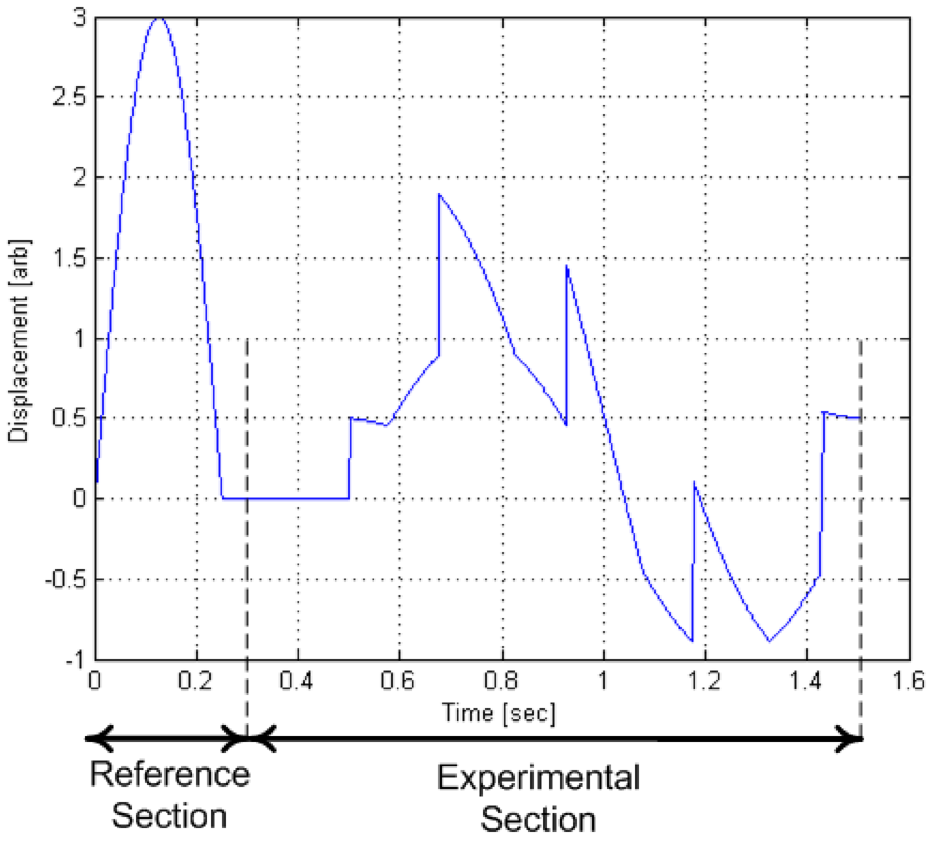
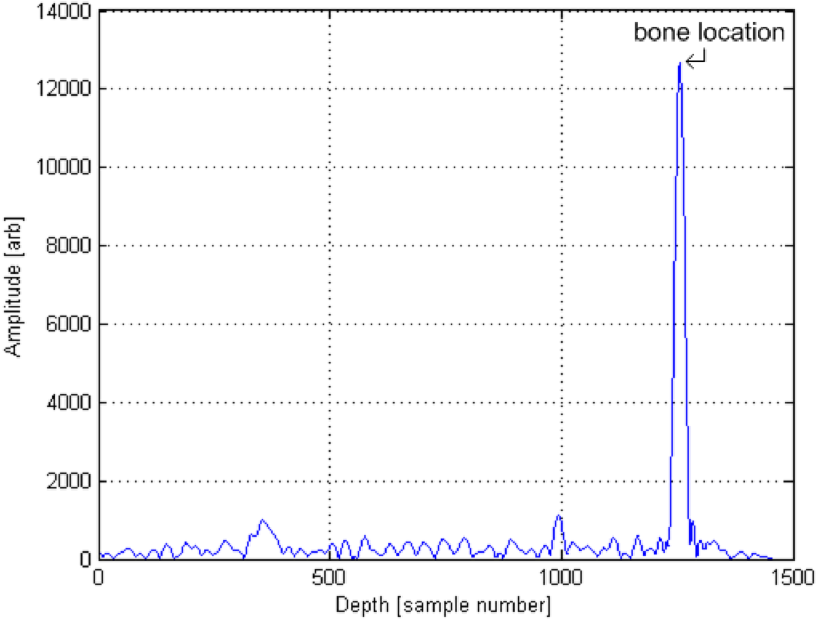
Measurement Procedure with Motion Artifact

- Simulated collected data observed at an arbitrary depth
  - Reference section contains only probe motion
  - Experimental section contains simulated muscle contraction and motion artifact



Bone Boundary Echo Tracking and Depth Scaling

- Bone boundary tracked to determine motion artifact
  - Initial location found with demodulated baseband envelope and B-mode image
  - Windowed peak tracking algorithm tracks bone boundary over all time
- Magnitude of displacement at bone boundary scaled with respect to depth
  - Scaled by peak-to-peak displacement magnitude comparison in reference section
  - Scaled displacement subtracted to obtain internal displacement estimation



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Simulation System: Phantom Development

- Compressible internal tissues simulated with agar
  - Agar powder mixed with water
    - The higher the agar concentration, the stiffer the solid
      - » 1 w% agar resembles fat
      - » 2 w% agar resembles resting muscle
      - » 3 w% agar resembles contracted muscle
  - Carbon particles added to act as ultrasound scatterers
- Uncompressible bone simulated with plexiglas

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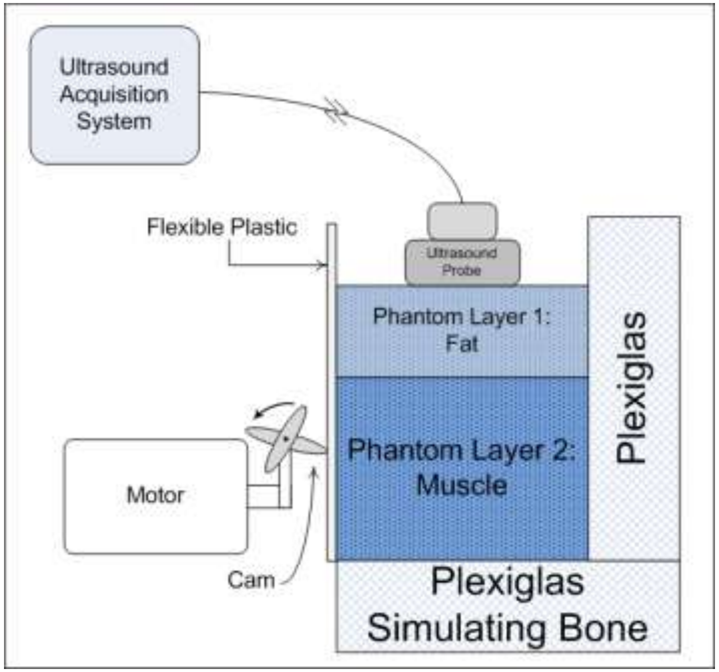
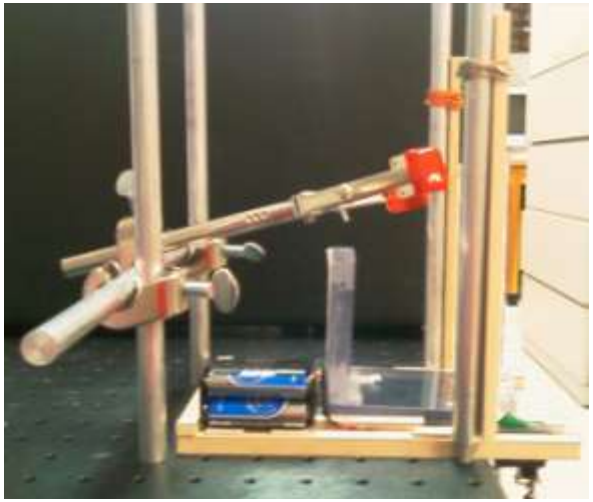
Simulation System: Hardware

- Phantoms placed in hardware used to simulate internal muscle stimulation





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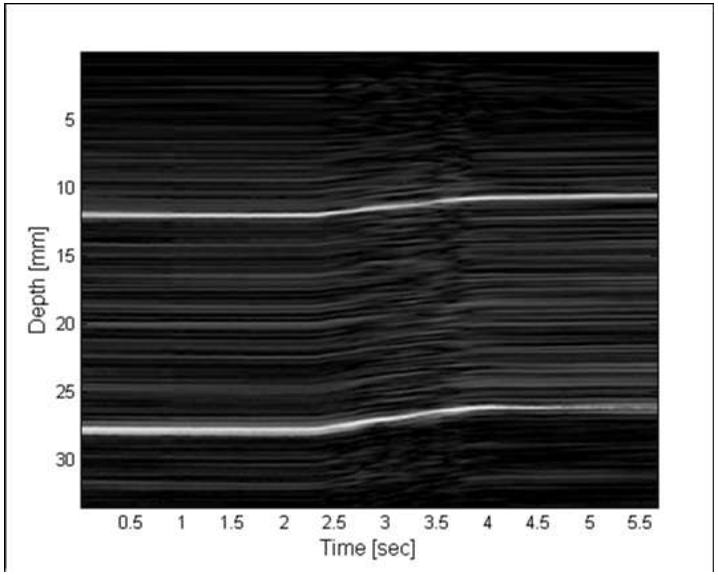
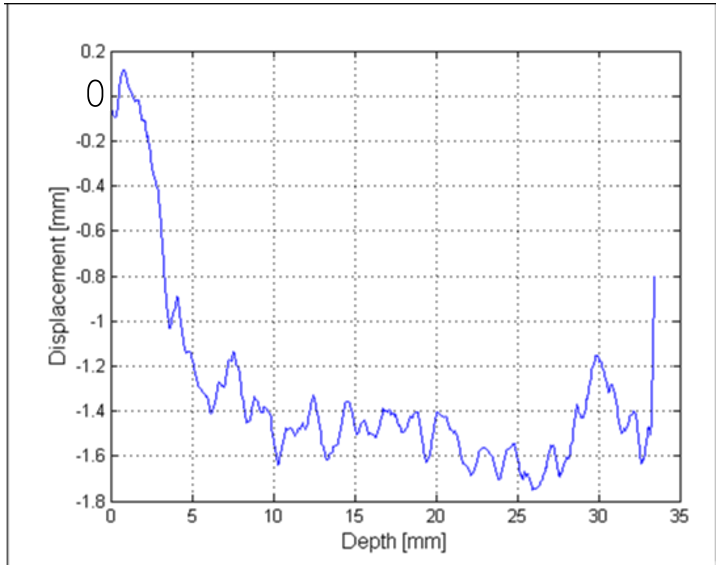
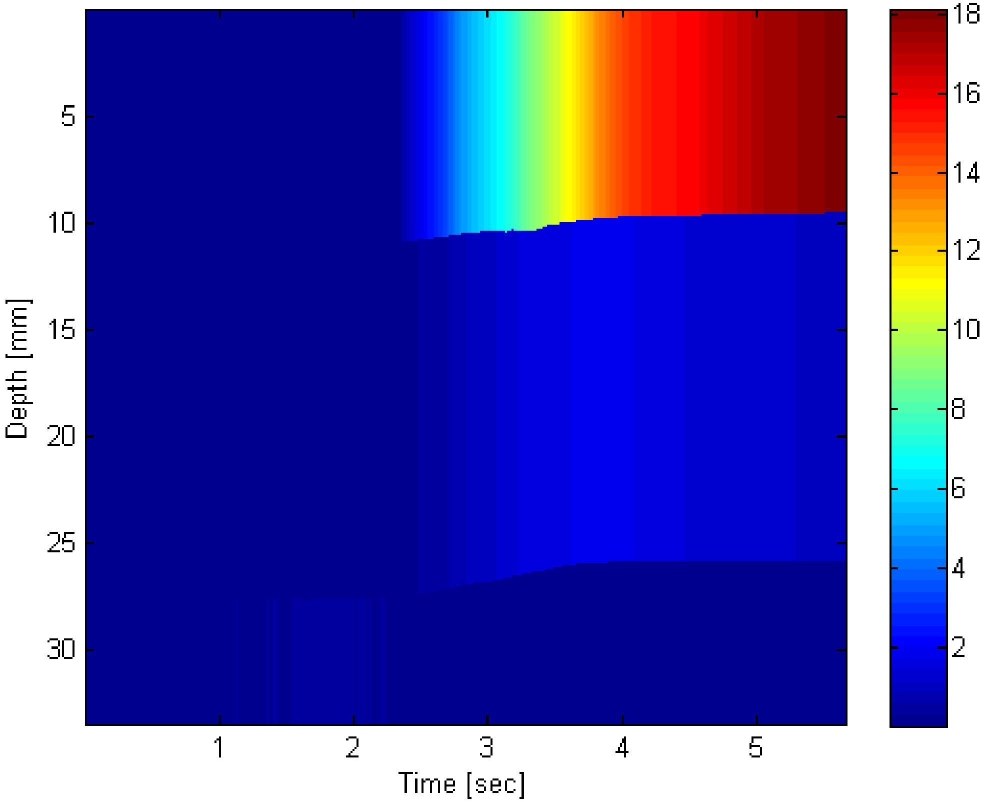
Simulation Experimental Strain Results

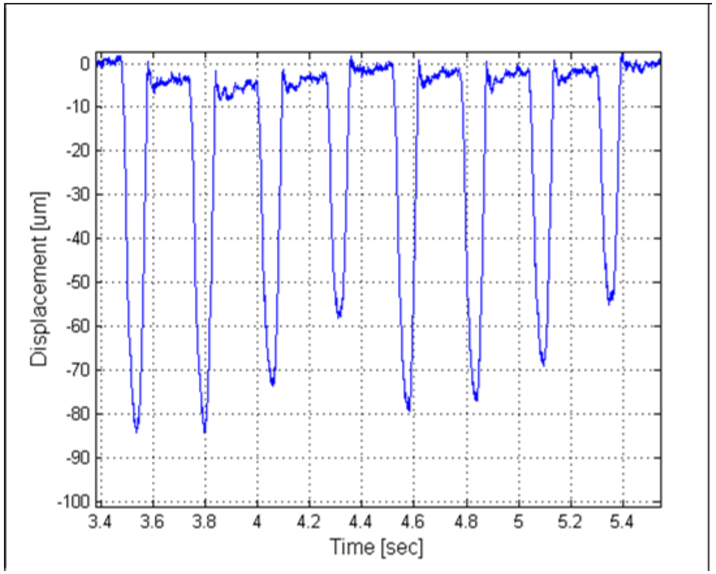


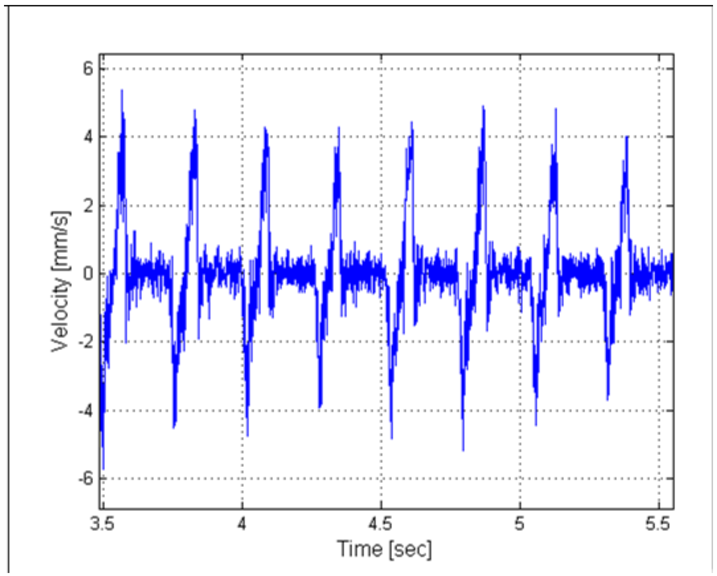


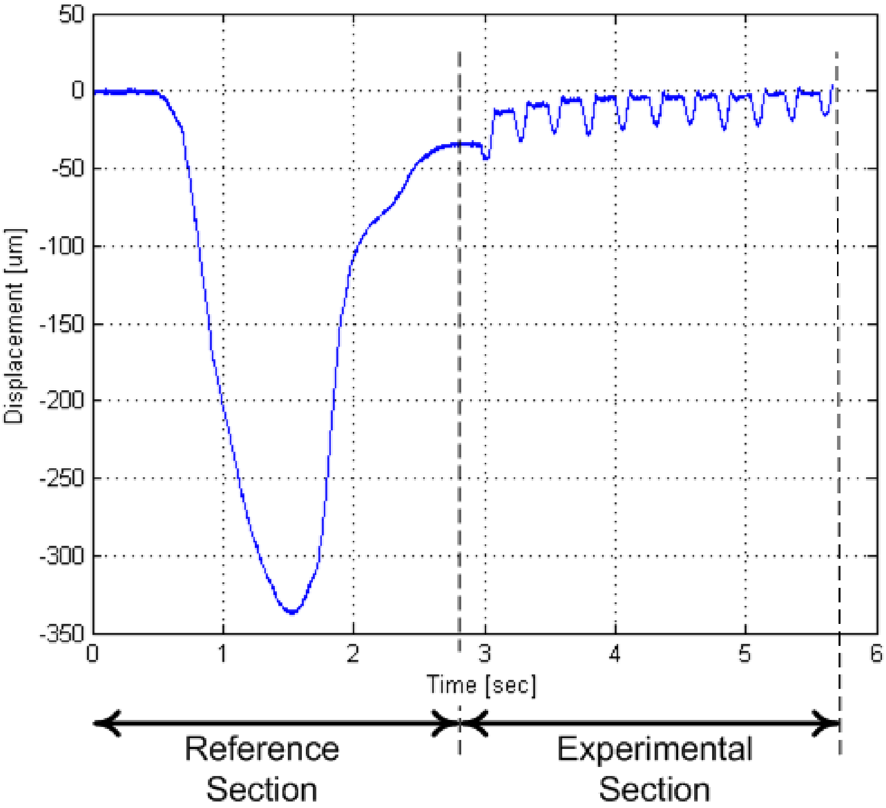


- M-mode data of 1 w% and 3 w% two layer agar phantom

Simulated Muscle Contraction Experimental Results



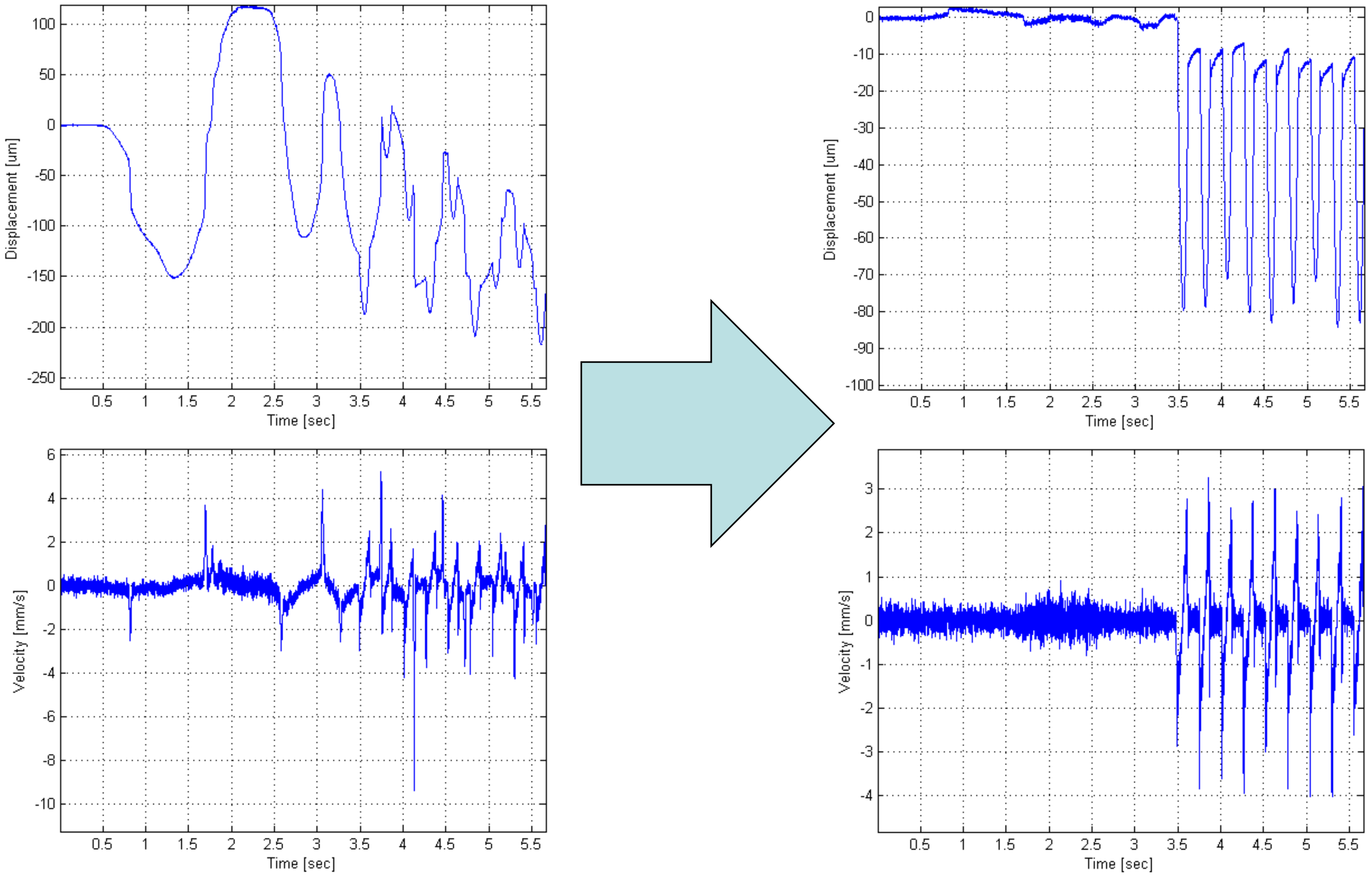




- M-mode data of 19 mm thick 3 w% single layer agar phantom

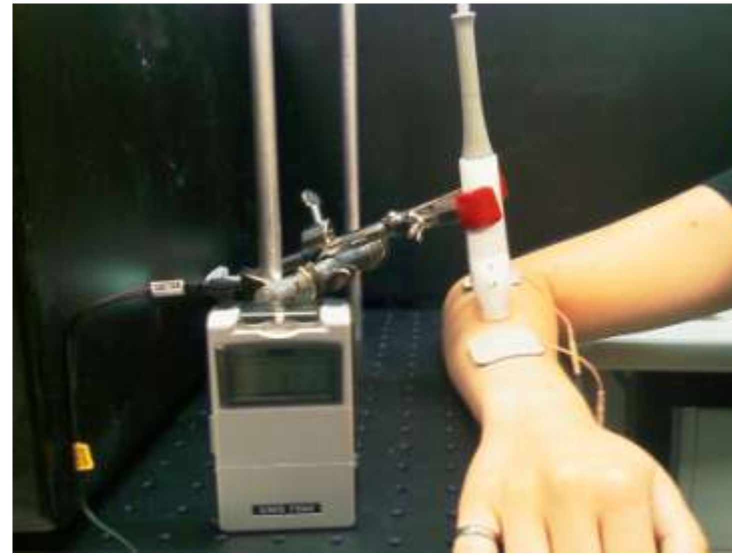
Effectiveness of Motion Artifact Removal

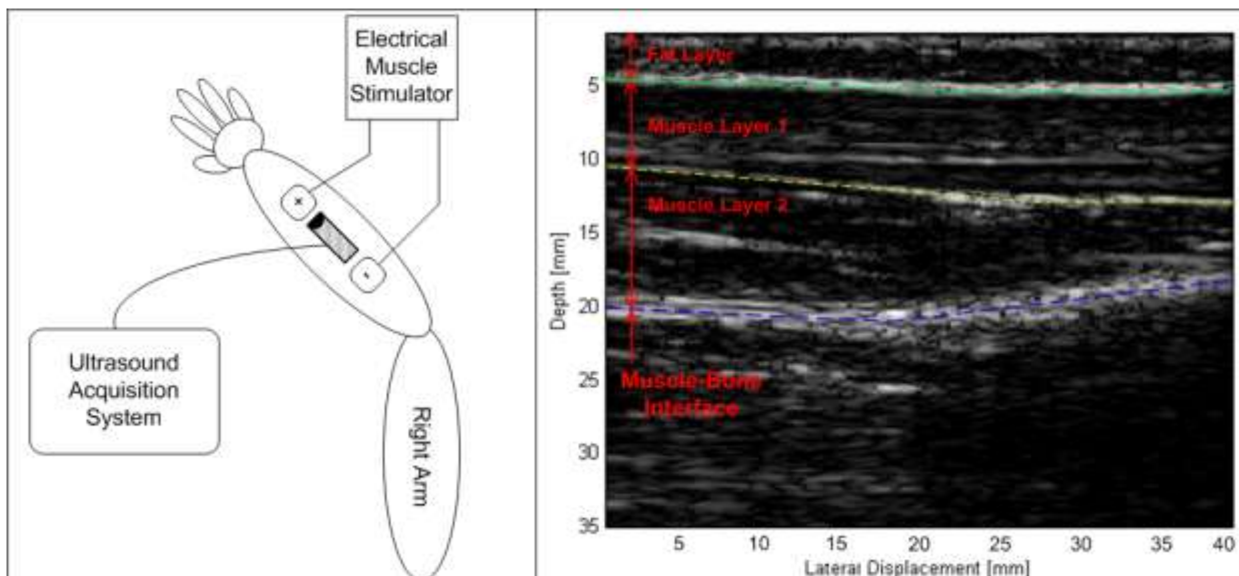
- Probe motion intentionally induced



In Vivo Experimental Design

- Forearm muscle stimulated with EMS at a variety of repetition rates ranging from 2 Hz to 12 Hz





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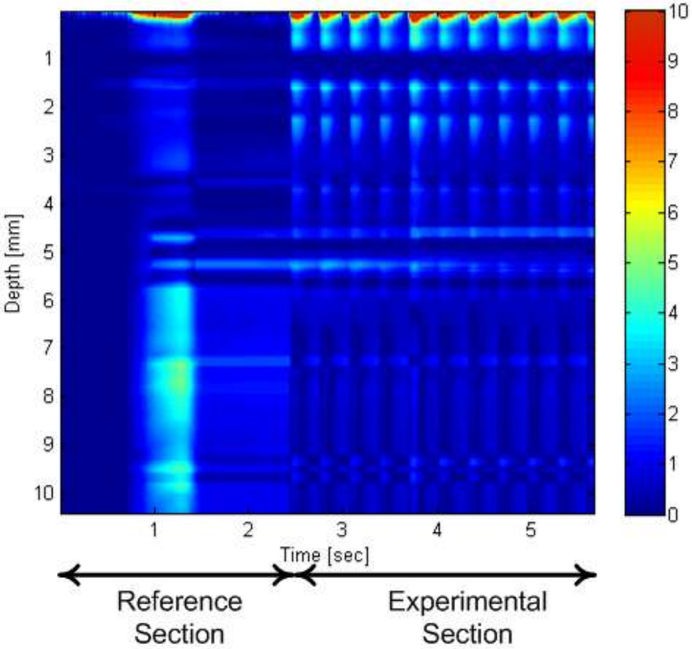
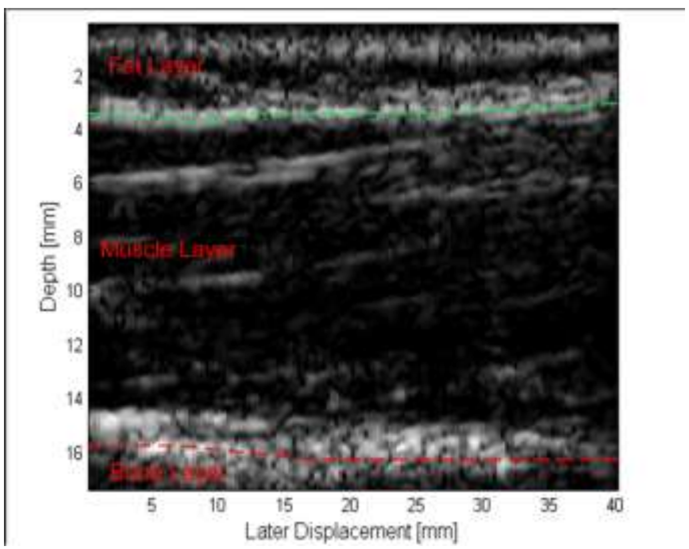
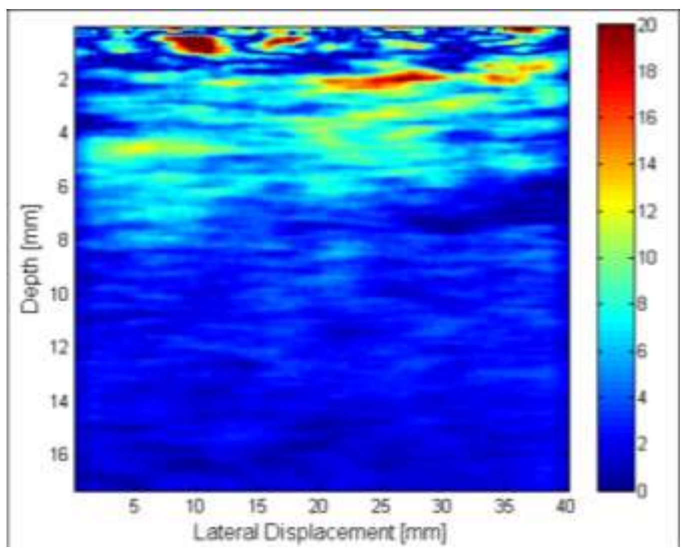
In Vivo Strain Estimation



- M-mode strain profile during 3 Hz EMS
  - Contracted muscle layer stiffer than fat layer

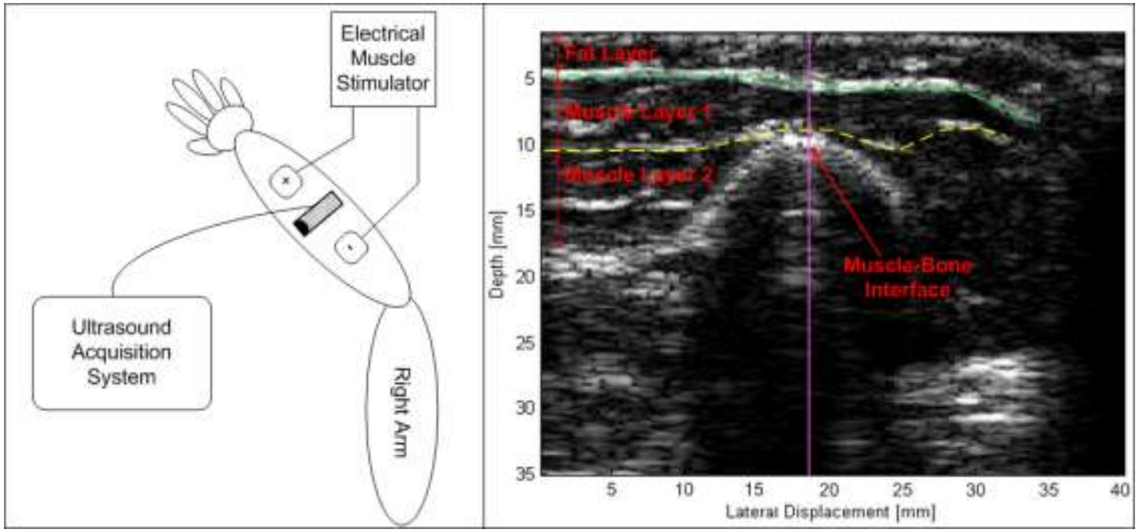


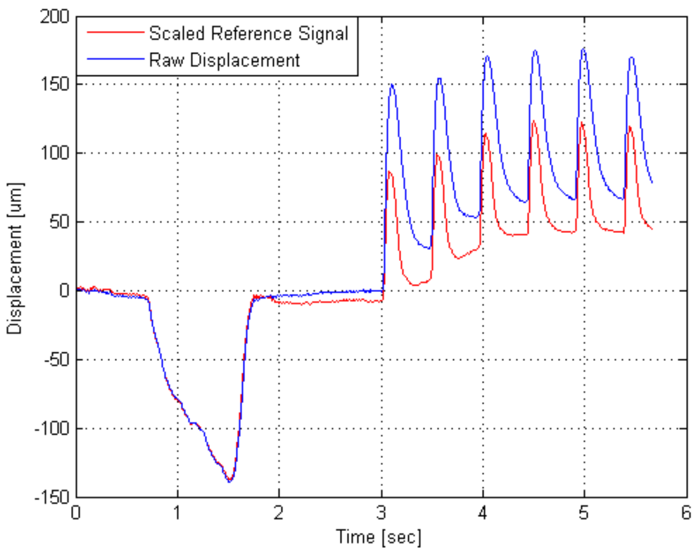


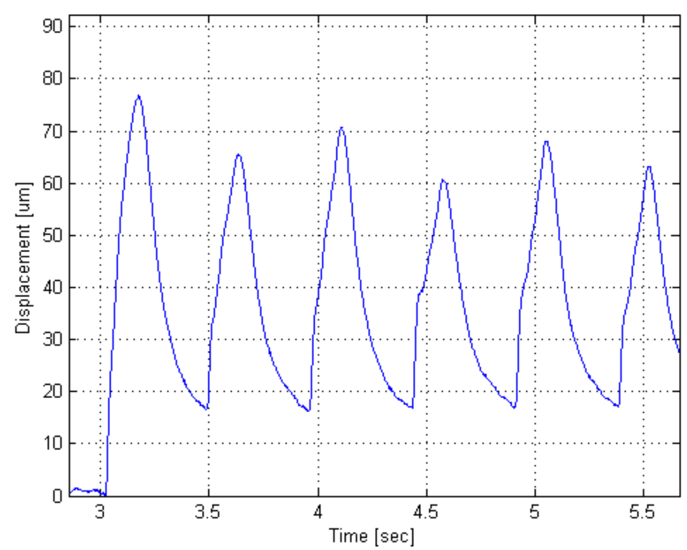
- B-mode strain image

In Vivo Displacement Estimation

- Displacement during 2 Hz EMS
  - Observed at a depth of about 7.16 mm

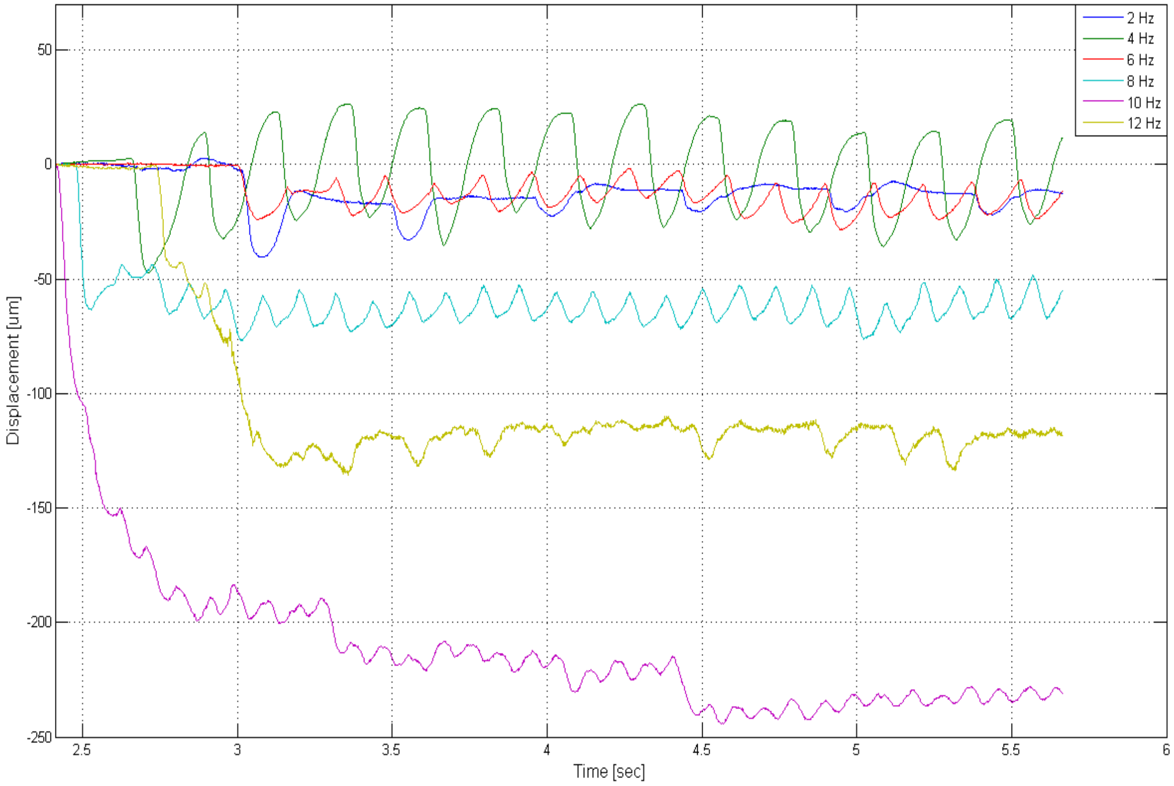






In Vivo Displacement Estimation

- Tissue displacement during 2Hz to 12 Hz EMS
  - Observed at a depth of about 6.47 mm
- Muscle contractions approach tetanus above 10 Hz



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Conclusion

- Designed and implemented a system to study human skeletal muscle capable of estimating:
  - Internal tissue displacement and velocity
  - Relative strain
- Designed and implemented an algorithm to remove the effects of motion artifacts during internal tissue measurements
- Designed and tested a phantom simulation system
  - Tissue mimicking phantoms
  - Hardware to simulate electrical muscle stimulation
- Performed phantom simulation and in vivo experiments using the developed system

Future Work

- Develop a more precise and controllable method to create tissue mimicking phantoms
- Perform additional in vivo experiments to better understand skeletal muscle
- Improve measurement accuracy

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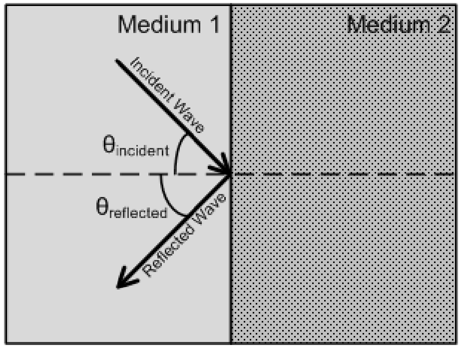
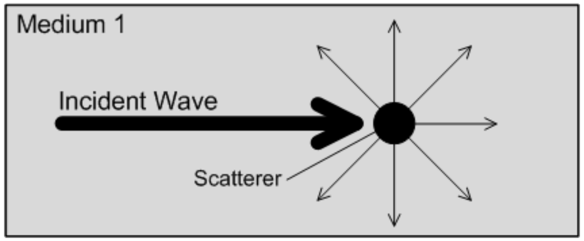
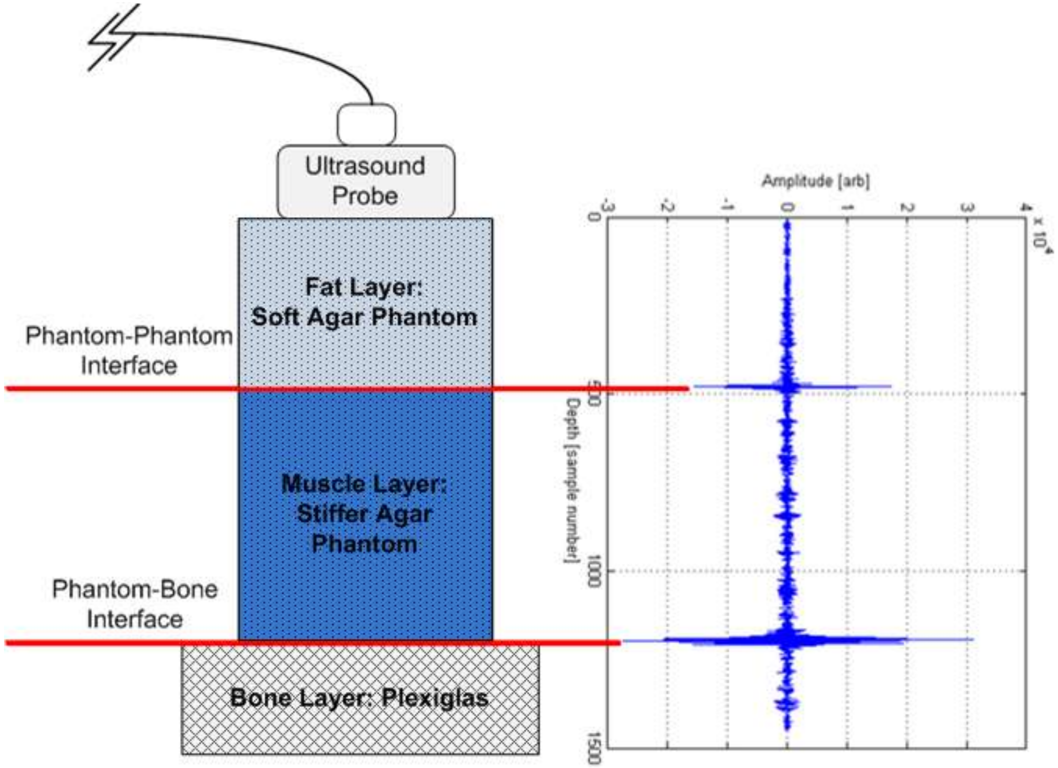
Questions?

Ultrasound Physics







- Two types of reflection:
  - Specular: occur at large interfaces and results in equal incident and reflected angles
  - Nonspecular: occur at interfaces of size comparable to or smaller than the ultrasound wavelength. Each small interface, referred to as a scatterer, acts as a new sound source and reflects sound in all directions
- Reflection and transmission from one medium to another is based on the acoustic impedance differences between the two media

Displacement Estimation

- A received ultrasonic signal can be thought of as a pure-tone frequency modulated (FM) signal
  - Ultrasound center frequency is analogous to FM carrier frequency
- Target motion toward or away from the probe will cause consecutive received pulses to experience a shift in phase
  - This phase shift is recovered by quadrature detection and converted to an estimation of displacement using autocorrelation

A received ultrasonic signal can be represented by:

$$s_r(t) = A(t)cos\left[\omega_c t + \phi(t)\right]$$

amplitude

US center frequency

phase shift

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Displacement Estimation: Quadrature Demodulation

A single received ultrasonic signal represented by

$$s_r(n) = A(n)cos(\omega_c(n) \cdot nT_s + \phi(n))$$

is multiplied by a reference sinusoid described by

$$S_{ref} = e^{-j\omega_{dem} nT_s}.$$

The quadrature demodulation process can be described as

$$I(n) = LPF\{s_r(n) \cdot \cos(\omega_{dem} nT_s)\} = \frac{1}{2}A(n)\cos[\Delta\omega(n) \cdot nT_s + \phi_n(n)]$$

and

$$Q(n) = LPF\{s_r(n) \cdot [-\sin(\omega_{dem} nT_s)]\} = \frac{1}{2}A(n)\sin[\Delta\omega(n) \cdot nT_s + \phi_n(n)]$$

where the complex baseband signal is given by

$$u(n) = I(n) + jQ(n)$$

and the phase shift can be found by

$$\angle u(n) = \Delta\omega(n) \cdot nT_s + \phi(n)$$

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Displacement Estimation: Two Dimensional Autocorrelation

The two dimensional complex autocorrelation function used to estimate phase shift is given by

$$r_{n,m}(N,M) \equiv u_z(n,m) \cdot u_z^{*}(n+N, m+M) .$$

The instantaneous displacement between two consecutive temporal samples can be calculated as

$$\Delta d_z(n,m) = \frac{c\Delta\phi_z(n,m)}{4\pi \bar{f}_c(n,m)} = \frac{c}{4\pi \bar{f}_c(n,m)} \angle r_{n,m}(0,1)$$

and accumulated displacement is given as

$$d_z(n,m+1) = d_z(n,m) + \Delta d_z(n,m)$$

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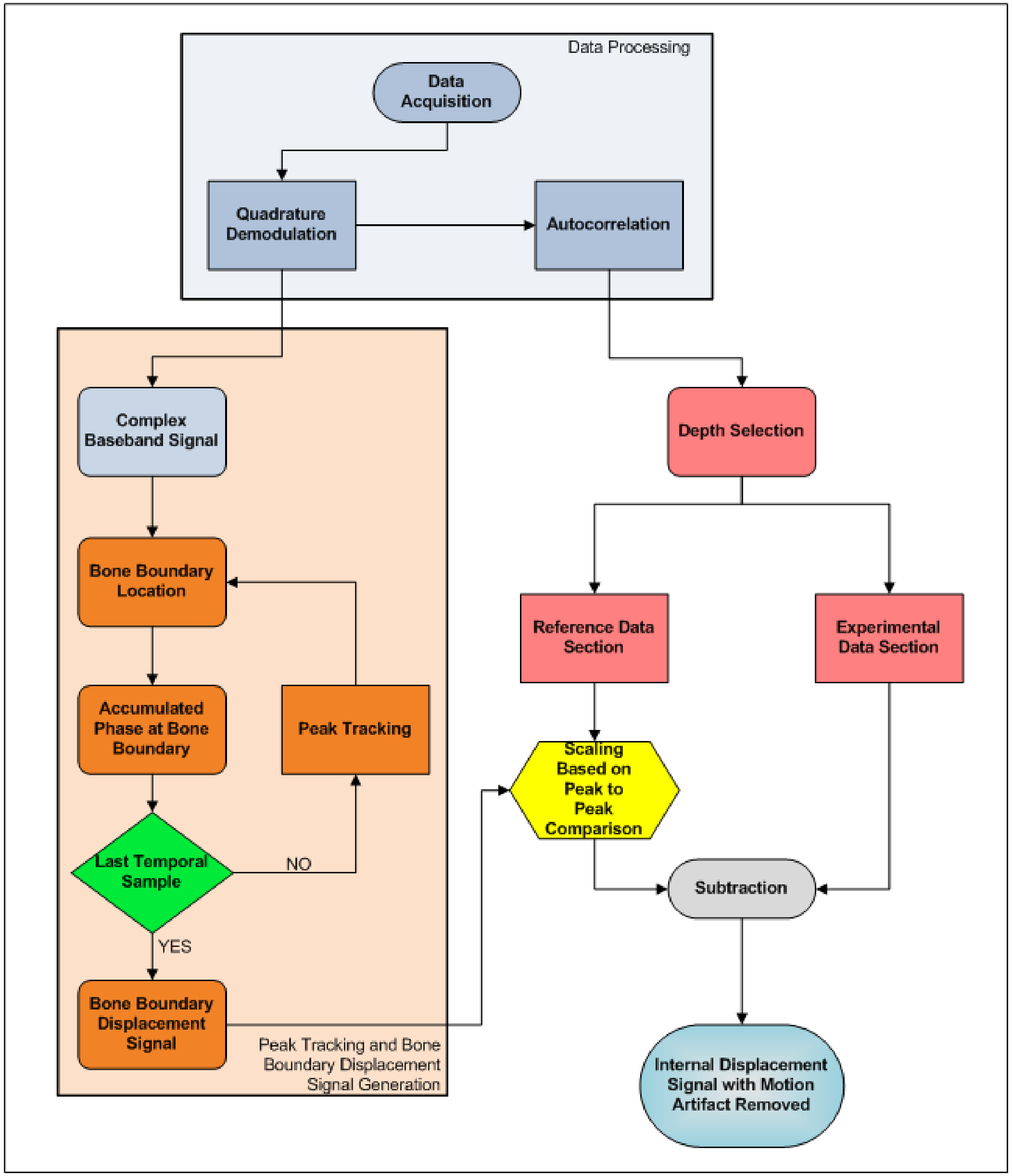
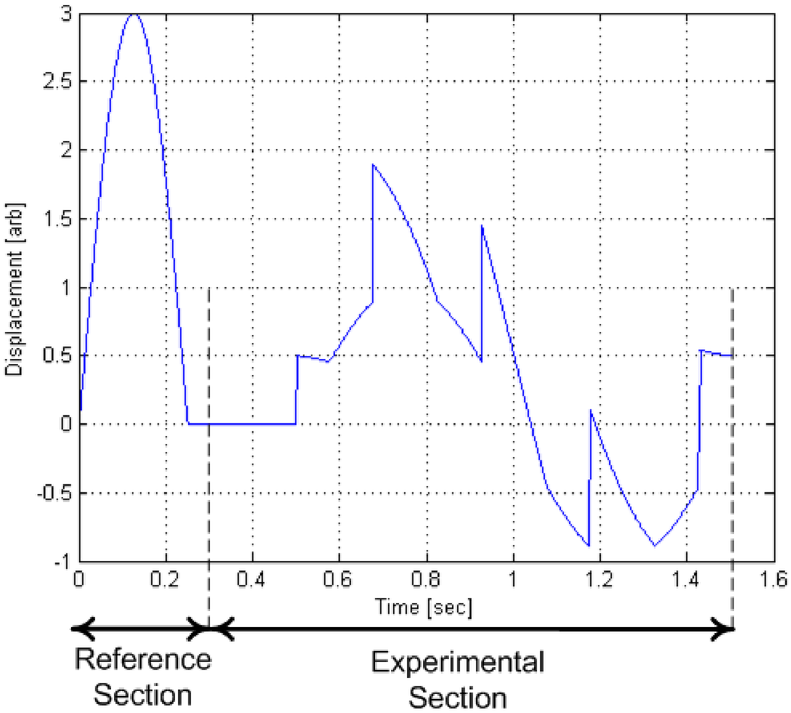
Bone Boundary Method of Motion Artifact Removal





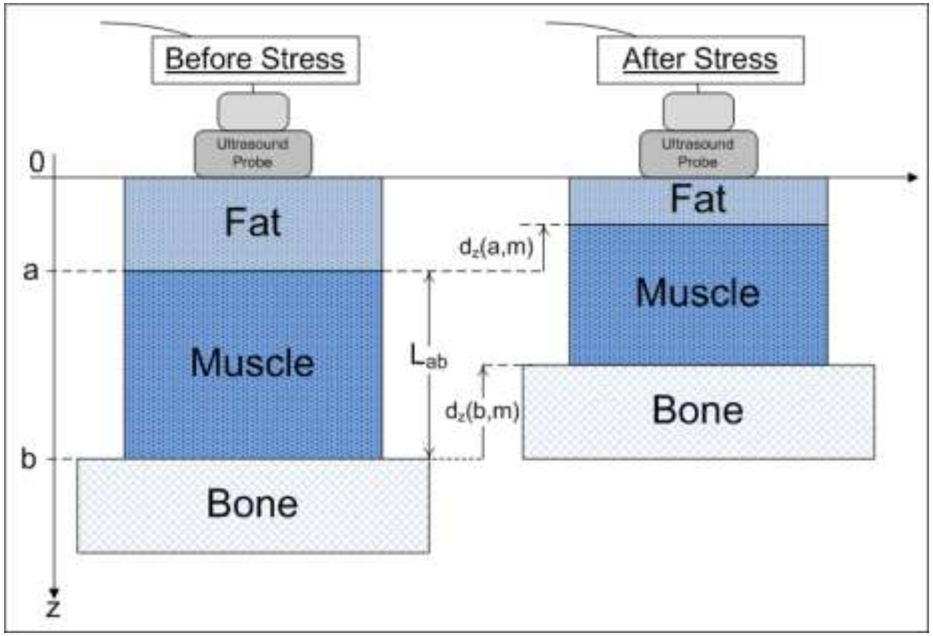
- Simulated collected data observed at an arbitrary depth
  - Reference section contains only probe motion
  - Experimental section contains simulated muscle contraction and motion artifact

Strain Estimation ®

- Axial strain is a measure of the relative deformation of an object and can be estimated by

$$\varepsilon_z(n_{ab}, m) = \frac{\Delta L_{ab}}{L_{ab}} = \left| \frac{d_z(a,m) - d_z(b,m)}{a - b} \right|$$

- Stiffer tissue shows less strain than softer tissue under equally applied forces



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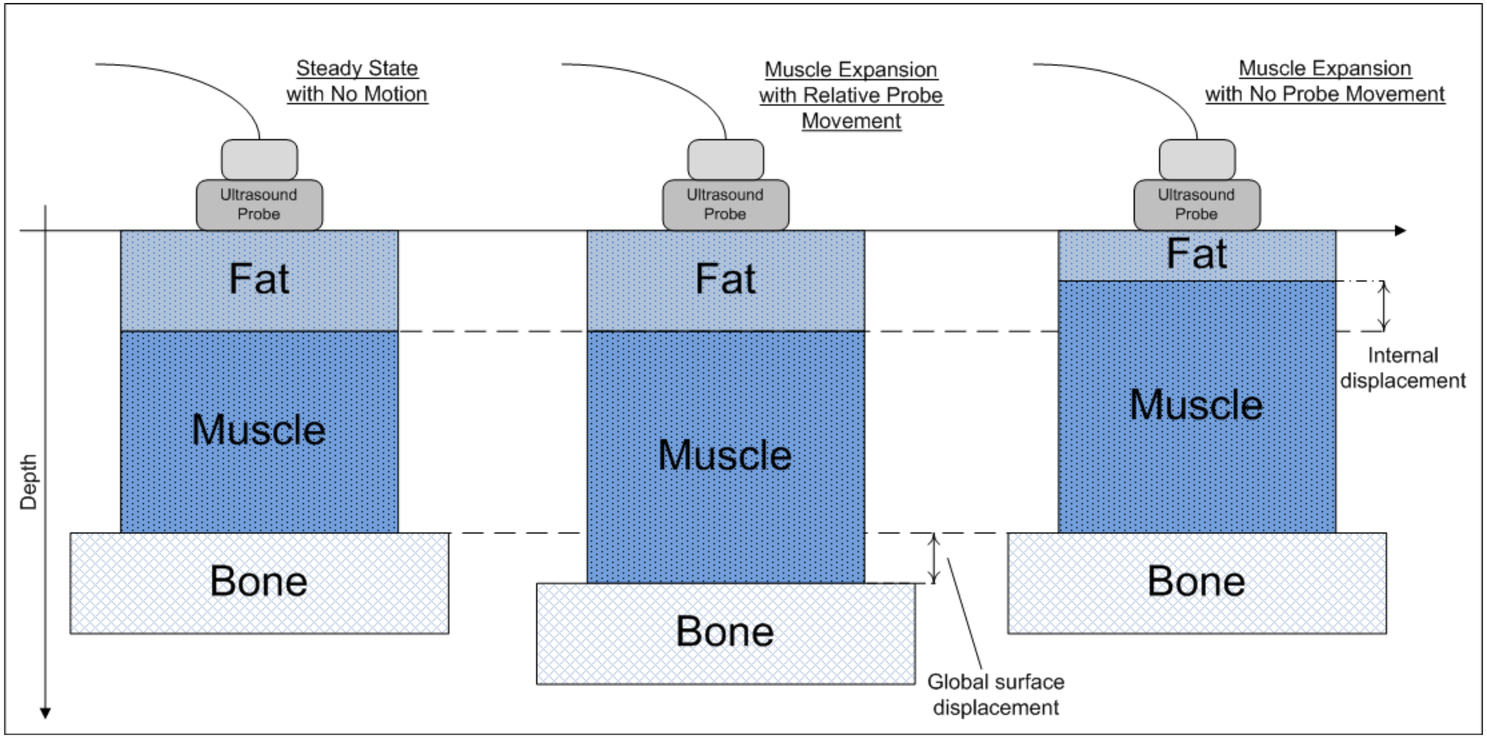
Motion Definitions ®



- External motion occurs due to motion of the probe or object being imaged
  - Classified as artifact motion
- Internal motion occurs and originates within the object being imaged
  - Ideally no probe or external object motion should occur

Extra Slide for explanation

Actual center frequency cannot be exactly known due to spatial and temporal fluctuation and is therefore estimated as

$$\bar{f}_c(n,m) = f_{dem} - \frac{\angle r_{n,m}(1,0)}{2\pi T_s}$$

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Extra slide for explanation

- Accuracy evaluation