April 2019

Ultrasound for Knee Osteoarthritis Screening: A Panoramic Reconstruction of the Knee Joint

Olivia M. Leavitt  
Worcester Polytechnic Institute

Rosanna S. Heidt  
Worcester Polytechnic Institute

Shion Matsumoto  
Worcester Polytechnic Institute

Follow this and additional works at: https://digitalcommons.wpi.edu/mqp-all

Repository Citation
Ultrasound for Knee Osteoarthritis Screening: 
A 3D Reconstruction of the Knee Joint

A Major Qualifying Project

Authors: Rosanna Heidt, Olivia Leavitt, Shion Matsumoto
Advisor: Dr. Karen Troy
Co-Advisor: Dr. Haichong Zhang

Department of Biomedical Engineering & Mechanical Engineering
Faculty of Biomedical Engineering
Worcester Polytechnic Institute

April 24, 2019
Abstract

Osteoarthritis (OA) is a significant and growing disease. Ultrasound (US) imaging provides an accessible method of imaging soft and hard tissue in the assessment of musculoskeletal morphology, particularly in screening for OA. The team created a device, protocol, and reconstruction software to acquire images of and measure the knee articular cartilage thickness, a proxy for joint space width. The resulting device can be used to detect and monitor progress of joint space narrowing. Using the device, the femoral articular cartilage thickness was measured with up to \( \pm 5 \) mm of resolution as compared to that of the gold standard, MRI.
Acknowledgements

We would like to thank our advisor Professor Karen Troy for her continued expertise in biomechanics, and co-advisor Professor Haichong Zhang for his expertise in ultrasound and image processing.

Thank you to Bryan Choate, who graciously provided use of his knee, as well as his MRI scan.

We would also like to acknowledge Professor Thomas Clancy and Professor Moinuddin Bhuiyan for lending us the HP ImagePoint ultrasound machine for the duration of our project, and for providing a work space for us in Atwater Kent to conduct testing.

We would also like to thank Lisa Wall for providing materials, a work space in Goddard Labs, and answering our many questions.
Authorship

Shion was primarily responsible for the development and testing of the reconstruction algorithm. He also led the preliminary design of the curvilinear rail sub-assembly. Shion sourced the initial inertial measurement unit and integrated it with MATLAB. He aided in the development of the 3D-printed components and provided input on the mechanical fixture design. Shion was the primary author for the reconstruction portion of the report. He also led the conversion of the report to Overleaf.

Rosie was the primary contributor to the design of the final mechanical system. Given the curvilinear rail design concept, she iterated through multiple designs to interface the rail with the fixture. Rosie also led the testing and analysis of the inertial measurement unit - both the accelerometer and the gyroscope. She was the point person for acquiring scans from the machine. Rosie was the primary author for the mechanical design sections of the report and supported the effort to convert the report over to Overleaf. Rosie took initiative for a majority of the communication between the team and with the advisors.

Olivia was in charge of developing various coupling media designs. She contributed her anatomical knowledge to the generated scans to help contextualize the scanning window. She supported testing procedures and design ideation.
## Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>ASP</td>
<td>Acoustic Standoff Pad</td>
</tr>
<tr>
<td>CAD</td>
<td>Computer-Aided Design</td>
</tr>
<tr>
<td>CT</td>
<td>Computed Tomography</td>
</tr>
<tr>
<td>DOF</td>
<td>Degree of Freedom</td>
</tr>
<tr>
<td>EPS</td>
<td>Electromagnetic Tracking System</td>
</tr>
<tr>
<td>IMU</td>
<td>Inertial Measurement Unit</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
</tr>
<tr>
<td>OA</td>
<td>Osteoarthritis</td>
</tr>
<tr>
<td>OPS</td>
<td>Optical Positioning System</td>
</tr>
<tr>
<td>PNN</td>
<td>Pixel Nearest Neighbor</td>
</tr>
<tr>
<td>SSIM</td>
<td>Structural Similarity Index</td>
</tr>
<tr>
<td>US</td>
<td>Ultrasound</td>
</tr>
</tbody>
</table>
# Contents

## 1 Introduction 1

## 2 Background 3

### 2.1 Osteoarthritis 3

#### 2.1.1 Definition and Epidemiology of Osteoarthritis 3

#### 2.1.2 Clinical Features of Osteoarthritis 5

#### 2.1.3 Functional Anatomy of the Knee 5

#### 2.1.4 Knee Osteoarthritis 7

### 2.2 Imaging Techniques for Diagnosing Osteoarthritis 10

#### 2.2.1 Radiography 11

#### 2.2.2 Magnetic Resonance Imaging 12

#### 2.2.3 Computed Tomography 13

#### 2.2.4 Nuclear Imaging 13

#### 2.2.5 Ultrasound 14

#### 2.2.6 Knee Osteoarthritis Imaging Modality Summary 14

#### 2.2.7 Musculoskeletal Ultrasound 15

#### 2.2.8 Joint Space Width Measurement 16

#### 2.2.9 Physics of Ultrasound 20

#### 2.2.10 US System Settings, Probe Maneuvers, and Artifacts 21

### 2.3 3D Ultrasound and Reconstruction 24

### 2.4 Musculoskeletal Ultrasound Accessories 27

#### 2.4.1 Acoustic Standoff Pad 28

#### 2.4.2 Water as Coupling Medium 29

#### 2.4.3 External Fixtures 30

### 2.5 Ultrasound Regulations 31

#### 2.5.1 Bioeffects of Ultrasound 31

#### 2.5.2 Regulation and Safety Guidelines 32

#### 2.5.3 Scope of Practice & Clinical Standards for Diagnostic Sonographers 33

#### 2.5.4 Ultrasound Imaging Protocol 34

## 3 Project Approach 36

### 3.1 Initial Client Statement 36

### 3.2 Objectives 37

### 3.3 Constraints 39
3.4 Revised Client Statement ........................................... 40
3.5 Problem Statement ..................................................... 41
  3.5.1 Problem Statement Breakdown .................................. 41
3.6 Project Approach ....................................................... 43
3.7 Project Timeline ......................................................... 45

4 Component Selection & Design Alternatives 46
  4.1 Needs Analysis .......................................................... 46
  4.2 Functional Requirements ............................................ 46
  4.3 Component Selection & Experimentation ........................... 51
    4.3.1 3D Ultrasound Method ........................................ 51
  4.4 Probe Localization Method ......................................... 53
  4.5 Coupling Medium ...................................................... 56
  4.6 Conceptual Designs .................................................. 63
    4.6.1 Sliding Water Cup Design ..................................... 63
    4.6.2 Cast Protector Design ................................--------- 64
    4.6.3 Curvilinear Rail Design ...................................... 65
    4.6.4 Conceptual Design Decision Matrix .......................... 66
  4.7 Curvilinear Rail Design ............................................. 67
    4.7.1 Plastic Bag Design ............................................ 83
    4.7.2 Saline Bag Design ............................................ 86
  4.8 Image Reconstruction ................................................. 88
    4.8.1 B-Scan Cropping Window ..................................... 89
    4.8.2 Reconstruction ................................................. 90

5 Results ................................................................. 94
  5.1 Ultrasound Imaging Setup .......................................... 94
    5.1.1 Image Acquisition ............................................ 97
    5.1.2 Imaging Conventions ........................................ 99
  5.2 Imaging Experimentation ............................................ 101
    5.2.1 Phase 0: Experimentation with Probe Angle, Knee Angle, and
          System Settings .............................................. 102
    5.2.2 Phase 1: Repeatability Without ASP .......................... 109
    5.2.3 Phase 2: Repeatability with ASP ................................ 112
    5.2.4 Reliability Comparison ...................................... 115
  5.3 Pose Sensing and Displacement Algorithm ........................ 118
    5.3.1 Numerical Double Integration Method Testing .............. 118
List of Figures

1. Incidence of OA over various age groups by gender and joint location. Arthritis is most commonly found in the knees, and females are at an increased risk for all joints [Neogi & Zhang, 2013] .................................. 4
2. Bones, cartilage, and ligaments of the knee. ................................................. 6
3. Lateral cross section of the knee joint showing synovial joint structures. ........ 7
4. Radiographs of joint space in normal knee (A), and mild (B), moderate (C), and severe (D) OA [Altman & Gold, 2007] ........................................ 9
5. Healthy knee (left) vs. knee showing severe osteophytes on the medial and lateral femur and medial tibia (right) [Altman & Gold, 2007] ........... 10
6. 2D US image of femoral articular cartilage depicted as the monotonous hypoechoic band between the soft tissues and femur [Faisal, Ng, Goh, & Lai, 2018] ................................................................. 17
7. US transverse images of femoral articular cartilage with (left to right) lateral condyle, intercondylar notch, and medial condyle cartilage thickness measurements. The left image shows articular cartilage with hyperechoic, sharply-defined interfaces. This knee was later classified as normal during dissection. The right image shows articular cartilage that appears thinner and has less defined interfaces. This knee was later classified as severely damaged during dissection [Naredo et al., 2009] .................................................. 17
8. Transverse A) and longitudinal B) scans of knee joint to measure femoral articular cartilage thickness [Yoon et al., 2008] .................................. 19
11. Aquaflex ultrasound gel pad (ASP) ............................................................. 28
13. Volumetric reconstruction of 2D scan using probe orientation sensor [Dahl, 2018] .......................................................... 30
15. National and international US regulations [Ng, 2002] .................................. 33
16 Aquaflex Ultrasound Gel Pad used as a baseline of commercial quality for a coupling medium. This pad was 9 cm in diameter with a thickness of 2 cm. .................................................. 58

17 Comparison of B-scans without ASP and with ASP. These images were of the lateral side of the knee and were used to obtain data during later testing of sample repeatability. A) Scan with no ASP. Surface of the knee is not identified, and a much shallower depth was used to acquire the image. B) Scan with ASP increased ease of image acquisition. Surface of the skin is shown, and the added 2 cm ASP was account for in increased depth set for acquisition. .................. 59

18 A) 16 g/L Agar sample. Best mechanical properties of all the pure agar mixes. B) 40 g/L gelatin sample. Stiffest gelatin sample, yet still too soft and resulted in tearing easily. .................. 60

19 Imaging a wooden dowel submerged in a Pyrex container filled with water. A) A wooden dowel was submerged underwater and a transducer was similarly submerged roughly an inch above the dowel. B) B-scan of submerged wooden dowel. The outline of the wooden dowel can be easily identified. .................. 61

20 Methods tested to degas water. A) Bagged tap and distilled water and exposed still water. B) Vacuum chamber used to degas tap and distilled water in Pyrex glass. .................. 62

21 Simple gimbal design to hold water and rotation around two axes of interest. A) Cross-sectional view of the gimbal that allowed for rotation around two axes of interest. B) Image of how gimbal would fit on knee. Rubber coated around edges would form a watertight seal and would secure around a patient’s leg. .................. 64

22 Cast protector that could be adapted to submerge the knee in water and provide a window through which the probe could image the knee (VBESTLIFE, 2019). .................. 64

23 Initial concept schematic of curvilinear rail design. The hinge (orange) would allow the rotation and locking of the rail at discrete (or continuous) positions. The rail (black) traces out the path that the probe would follow (Busti, 2015). .................. 65

24 Screw clamp curvilinear rail design (Wilkerson, 2016). .................. 68
<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
</tr>
</thead>
</table>
| 25 | A) Isometric view of initial CAD model of curvilinear rail design.  
B) Front view of design with rail diameter dimension shown as 170 millimeters. |
| 26 | A) Isometric view of car.  
B) Isometric view of car assembled with heat-set brass inserts and V-rollers. |
| 27 | Probe rendering based on caliper measurements and visual inspection.  
A) Isometric view.  
B) Front view with dimensions shown in inches. |
| 28 | Probe sleeve CAD model.  
A) The window at the bottom allows the imaging surface of the probe to pass through and the side walls secure the probe in place. The back wall of the sleeve and the two arms on the side provide additional support to prevent unwanted probe motion.  
The clearance between the two side arms is slightly undersized for the probe such that they compress the sides of the probe.  
B) The prismatic joint, commonly referred to as a dovetail, was designed to secure the sleeve to the car design. |
| 29 | A) Soldering iron used to insert heat-set insert into 3D-printed car.  
B) Car design with three adjustable shoulder V-rollers. The rollers can be tightened or loosened to adjust the distance between it and the other two rollers depending on the desired fit onto the curvilinear rail. |
| 30 | A) Fully assembled curvilinear rail design with car and probe sleeve.  
B) By adjusting the shoulder of the rollers, they could be further tightened onto the rail and prevent the probe from sliding. |
| 31 | Final probe sleeve-car CAD model.  
A) Isometric view shows that holes were added to the sleeve for the heat-set inserts, reducing the overall profile of the sleeve part.  
B) Side view to show that the final design incorporates the previously designed car and sleeve.  
Vertical alignment of the car along the sleeve was adjusted with a variety of sizes from 0cm to 15 cm offset.  
C) CAD of the final probe sleeve with heat-set inserts and wheels attached to the rail. |
| 32 | The final probe sleeve with heat-set inserts fit around the linear probe.  
Note that this is the 0cm vertical offset fit. This design reduced instability and tolerance issue as experienced in the two-part car-sleeve system shown earlier. |
| 33 | Here, the final probe sleeve with the IMU, mounted with simple command strips.  
A) Side view.  
B) Isometric front.  
C) Isometric back. |
34 Initial rail cap design to be attached to the ends of the rails in A, B, C, and D. Rail cap was left transparent for viewing of the perpendicular hole to be drilled into the cap to allow for hinge about a perpendicular axis. A) Isometric view. B) Top view. C) Side view. D) Cap magnified view. 74

35 Initial rail cap prototypes in A and B. A) Initial rail caps relying on push-fit from rail profile. Printed shape was found to be too complex to print multiple caps, and was thus not used. B) A push-fit slot rail cap idea that relied on a single slot to be hammered into the rail for final fit. This prototype was not secure enough, and was discarded. This idea was modified for use in the final design. 75

36 Final rail cap concept and prototypes. A) CAD model of the t-slotted rail cap. B) CAD model of the t-slotted rail cap, but transparent to emphasize the negative space at the end of the rail. Note that both ends have a slot each. C) Printed rail with t-slot hole. D) Printed rail fitted with the t-slot rail cap. E) Full view of the rail caps in a printed rail. 75

37 Final rail design and various prototypes. A) CAD of the final rail to be used. This rail profile cupped the sides of the v-roller wheels for a more secure fit. B) Final rail design on the final assembly to indicate proper fit and actual use. C) From top to bottom: 150cm, 160cm, 170cm, 170cm, 180cm, and 190cm sized rails. 76

38 Comparing the rail-wheel interaction between the old (blue) and new (black) rails. A) The wheel glides along pretty easily with the original design, though the lack of material to support the sides of the wheel result in a loose fit. B) The new rail supports the wheel along its entire profile, not just the negative ‘v’. This results in a more secure fit of the wheel along the path. C) Comparison between the two profiles with old shown in blue and new shown in black. 76

39 Rail mount frame design. A) Isometric view of the frame. B) Base dimensions of the design, which is made from wood and has a dimension of 500 mm x 408 mm. C) Height of the usable slider portion of the design, approximately 60 cm. 78
<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>40</td>
<td>Full assembly concept of rail design and frame. Knee model inserted to provide a reference for the dimensions of the fixture. Note that this knee model is not bent at the correct angle at which it would be imaged, and is placed to show how the fixture would interact with a real knee.</td>
</tr>
<tr>
<td>41</td>
<td>Constructing the wooden base and drilling holes for the 80/20 frame. A) Making sure the holes are level and of the proper depth. B) Drilled holes before hammering in the t-nuts. C) Transparent CAD view of how the t-nuts interact with the flange-head bolt and the wooden base.</td>
</tr>
<tr>
<td>42</td>
<td>Circuit holder design. A) The initial concept in which an IMU would be placed to stabilize during calibration. B) The holder attached to the rest of the 80/20 with two wing-nuts, as shown. C) The printed circuit holder on the frame, with the IMU in the pocket to maintain a vertical position throughout calibration.</td>
</tr>
<tr>
<td>43</td>
<td>CAD models of the final rail assembly. A) Isometric view, front, to show how wheels interface with the rail. B) Isometric view, back, to show the probe fit within the sleeve. C) Side view of the assembly. D) Magnified view of how the rail mounts to the frame with the elevator bolt, washer, and hex bolt.</td>
</tr>
<tr>
<td>44</td>
<td>Final assembly design is as shown from various views. A) Isometric view. B) Side view. C) Front view.</td>
</tr>
<tr>
<td>45</td>
<td>Final frame assembly is shown with knee.</td>
</tr>
<tr>
<td>46</td>
<td>Heat-sealed plastic bags containing tap water. Bags of two different widths and a variety of lengths were made to experiment with the bags of various dimensions. A) Bags were first filled with tap water and left open to degas as much as possible. B) Bag was heat-sealed to desired length. C) Two of heat-sealed plastic bags made.</td>
</tr>
<tr>
<td>47</td>
<td>Images obtained using heat-sealed plastic bags filled with tap water. Image quality obtained using the plastic method was superior to that of agar.</td>
</tr>
<tr>
<td>48</td>
<td>Variety of saline bags bought for imaging. The volumes of the bags from left to right are as follows: 1000 mL, 500 mL, 250 mL, 100 mL, 50 mL. The height and width from left to right are as follows: (12” by 4.5”), (8” by 4.5”), (6.5” by 4.5”), (5” by 3”), (4” by 2.5”).</td>
</tr>
</tbody>
</table>
49  Suprapatellar femoral cartilage thickness imaged through 50 mL saline bag. A) B-scan resulting from imaging suprapatellar to the knee, with knee at 90° flexion, and probe approximately 60° with the 50 mL saline bag. B) Same imaging method as A but with anatomical regions labeled.

50  Lateral femoral articular cartilage imaged through 50 mL saline bag. A) B-scan resulting from imaging medial knee with knee at 90° flexion, and probe approximately longitudinally oriented to the medial with the 50 mL saline bag. B) Same imaging method but anatomical regions labeled.

51  Closed-loop saline bag design. 6.5” by 4.5” and 4” by 2.5” saline bags were connected with 4 feet of 3/16” ID tubing. A barbed Y-connector was used to join the two ends of the tube to provide a bleeding port, capped with the red stopper in the image. Clamps on either side of connector prevent flow of saline once bag filled with desired amount of saline.

52  Saline bag buckled to the knee. A) 6.5” by 4.5” saline bag for suprapatellar imaging and B) 4” by 2.5” bag meant for medial or lateral imaging.

53  (left) Individual B-scans are only able to provide limited views of the structure of interest. (right) By combining multiple B-scans into one reconstructed image, a large portion of the underlying structure can be visualized. For this, the orientation at which each B-scan was taken needs to be provided to the reconstruction algorithm.

54  Identification of cropping window for depth of 5 cm using MATLAB’s `bwboundaries` function. Each depth setting had a slightly different cropping window; therefore, the process was repeated for all depth settings.

55  (left) Bounding box for set of rectangles with bottom left corner at (0,0) rotated by six equispaced angles between -45° and 45° around a center point (10, 30). (right) Bounding box shifted such that bottom left corner is at (1,1).
56 Demonstration of reconstruction in yaw with ten 50-by-50 matrices of 100s spaced equally between -90 and 90 degrees with a rail radius of 50 pixels and nearest-neighbor interpolation method. Center of rotation is shown as the red crosshair. Axes are in pixels. (left) Grayscale. (right) Color scale set to parula. Regions separated into four regions: no frame (navy), one frame (blue), two intersecting frames (teal), three intersecting frames (gold), and four intersecting frames (yellow).

57 Demonstration of interpolation methods in reconstruction in yaw. Close-up of reconstructed image at the boundary of a region with two intersecting frames and only one frame. (left) Nearest-neighbor interpolation. (center) Bilinear interpolation. (right) Bicubic interpolation. Bilinear and bicubic interpolation result in a gradient at the boundary, whereas nearest-neighbor interpolation results in a clear boundary between two regions of differing pixel intensities.

58 (Left) SIEMENS Sonoline Adara. This machine had a working system but there were no compatible linear probes available within the budget to purchase. (Right, Top) Back of the SIEMENS machine. The yellow converter cable, BNC-S-Video was removed and taken to use for HP Image Point for data acquisition. (Right, Bottom) SIEMENS endocavity probe.

59 Later model of SIEMENS linear probe. It was not compatible with the SIEMENS Sonoline Adara model and was used instead as a dummy probe for test purposes.

60 GE RT 3200 Advantage III. This machine did not have a working monitor, and its working condition is unknown. The two probes available are an endocavity and a curvilinear probe, neither of interest for this project.

61 HP Image Point Machine. This machine was the model used in this project, as it had both a working system and the appropriate probe type for imaging superficial structures – a linear transducer probe.

62 Schematic of components needed to export image from ultrasound system to MATLAB on personal laptops.

63 BNC-to-BNC, S-Video, Hauppauge Analog Video Digitizer to USB connector setup. Within MATLAB, the “MATLAB Support Package for USB Webcams” hardware support package in conjunction with the Image Acquisition Toolbox was used to acquire video in real-time.
<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>64</td>
<td>Displaying image acquired from US system in MATLAB using Image Acquisition Toolbox function, <code>imaqtool</code>.</td>
</tr>
<tr>
<td>65</td>
<td>A) Knee at full extension, 0° B) Knee at 45° flexion C) Knee at 90° flexion D) Knee at 135° flexion E) Knee at extreme flexion, 150°.</td>
</tr>
<tr>
<td>66</td>
<td>Demonstration of probe planar angle measurement. All three axes of interest: x, y, z, and the three associated rotations: yaw, pitch, roll, are shown. Note that this origin is not local to the probe itself, and this diagram is to illustrate general motion of the probe with respect to the anatomy.</td>
</tr>
<tr>
<td>67</td>
<td>Graphics depicting the anatomical locations imaged with the suprapatellar position of the probe on the knee, with the knee at 90° flexion. This was setup was used as the initial baseline protocol for determining femoral articular cartilage thickness, as it resulted in the clearest image on the US machine. A) Probe at 45° pitch. B) Probe at 30° pitch. C) Close up of probe on femoral head on frontal plane. D) Probe at 0° pitch.</td>
</tr>
<tr>
<td>68</td>
<td>Preliminary testing of the ultrasound machine with different knee angles and probe angles. Here, suprapatellar to the knee, specifically at a knee angle of 90° was found to be the most consistently good orientation. A) Knee at 90° flexed, probe suprapatellar at 0° yaw, 0° pitch, 0° roll B) Corresponding B-scan.</td>
</tr>
<tr>
<td>69</td>
<td>Stitched medial, center, and lateral scans completed by hand to illustrate the anatomical regions of interest. This image was stitched together by hand to illustrate different anatomical regions of interest observed with the suprapatellar view.</td>
</tr>
<tr>
<td>70</td>
<td>Preliminary testing of the US machine with ASP and extreme flexion of the knee, with probe lateral to patella A) Knee at extreme flexion, roughly 150° with the probe lateral to the patella, at 45° yaw, 45° pitch, 0° roll. B) corresponding B-scan.</td>
</tr>
<tr>
<td>Figure</td>
<td>Description</td>
</tr>
<tr>
<td>--------</td>
<td>-------------</td>
</tr>
<tr>
<td>71</td>
<td>Image of the HP Image Point settings. Relevant settings and buttons for imaging the knee are boxed in different colors. Red: Gain dial and Time Gain Compensation sliders. These settings were subject to high variation throughout testing. These settings were adjusted until appropriate contrast was achieved. Blue: (left to right) Map / Smoothing / Persist dials. Map controls RGB output of the monitor, while smooth and persist control averaging and pixel lingering. Green: (top to bottom) Dynamic Range, Depth, and Focus of the image. Controls depth, range, and focus of the sound waves. Orange: (Top to bottom, left to right) Caliper, Trace, Erase, Enter, and Freeze buttons. Used to obtain measurements post-image acquisition for distance and area calculations on the system.</td>
</tr>
<tr>
<td>72</td>
<td>Three measurements of the trochlear femoral articular cartilage thickness using the system’s built-in “Freeze” and “Caliper” function. The distance between any two points imaged were provided with an accuracy of three decimal places in centimeters.</td>
</tr>
<tr>
<td>73</td>
<td>Graphic of suprapatellar probe placement on knee at 90° flexion and each paired ultrasound image. A) Knee at 90° and probe at -45° yaw, 0° pitch, 0° roll B) Corresponding scan of the medial condyle C) Knee at 90° and probe at 0° yaw, 0° pitch, 0° roll of the trochlear surface, or centerline D) Corresponding scan of the trochlear surface E) Knee at 90° and probe at 45° yaw, 0° pitch, 0° roll F) Corresponding scan of the lateral condyle.</td>
</tr>
<tr>
<td>74</td>
<td>Graphic of suprapatellar probe placement on knee at 90° flexion and each paired ultrasound image with Aquaflex Acoustic Standoff Pad (ASP). A) Knee at 90° and probe at -45° yaw, 0° pitch, 0° roll with ASP B) Corresponding scan of the medial condyle C) Knee at 90° and probe at 0° yaw, 0° pitch, 0° roll of the trochlear surface, or centerline with ASP D) Corresponding scan of the trochlear surface E) Knee at 90° and probe at -45° yaw, 0° pitch, 0° roll with ASP F) Corresponding scan of the lateral condyle.</td>
</tr>
<tr>
<td>75</td>
<td>For Patient A, the normalized standard deviations were lower for the measurements using ASP than for those without, except in the trochlear region, indicating that the ASP increased the reliability of condylar cartilage measurements.</td>
</tr>
</tbody>
</table>
76 For Patient B, there was no clear trend when comparing the ASP and non-ASP measurements. However, there was greater variability at the edges than in the center, most likely due to varying measurement locations in regions where cartilage was thicker.

77 For Patient C, there was no clear trend when comparing ASP to non-ASP measurements, nor was there a clear trend across the cartilage surface.

78 Overall, the coefficient of variation was lower with the ASP than without it.

79 Double trapezoidal numerical integration scheme using MATLAB function, `cumtrapz`, tested on function whose analytic solutions are known. Values of $\sin(t)$, $\cos(t)$, and $\sin(t/2)$ were assigned to $a_x$, $a_y$, and $a_z$, respectively, with nonuniform spacing for $t$.

80 Comparison of displacements results from numerical double integration scheme to analytic solution for acceleration. Though the numerical solutions for displacement in the y- and z-directions are accurate, the displacement in the x-direction drifts from about $t=4s$.

81 Relative error of numerical solution for each direction of displacement for every time step for the x, y, and z axis. Though the relative error is high at the beginning, this can be disregarded as a small error produces large errors given the small values being used to calculate relative error.

82 (left) Acceleration in the x-, y-, and z-directions while inertial measurement unit at rest. Slight offset can be seen for all directions. Gradual stabilization, particularly in the y-direction, can be seen as well. (right) The effect of the offset and gradual stabilization on the resulting displacement are made evident by the fact that the displacement is continually increasing or decreasing despite the inertial measurement unit being at rest.

83 (left) Acceleration in the x-, y-, and z-directions while inertial measurement unit at rest. The signal is much more stable except for the large negative acceleration around 25 seconds. (right) Displacement calculated from the stabilized acceleration signal.
Results of leaving the inertial measurement unit at rest on the table. This figure includes the component breakdown of the acceleration in the X, Y, Z (top to bottom). Each of these subplots shows the original data (cyan), band-pass filtered data (dark blue), median-filtered data (red), and the expected data (black). Here, as the IMU was simply at rest, the acceleration should reflect 0 on each of the subplots. The median filter simply removed the high frequency noise, and was thus eliminated as a potential filter. The bandpass filter was able to remove the high and low frequency noise, and was used. Here, it is shown that the bandpass filtered data most closely bounces around the true expected data of 0.

Filtered results of the no movement test using a band-pass. Though capable of decreasing the magnitude of the acceleration, the filtered data still resulted in a calculated displacement of about 6.96mm when it should have reflected 0 (or a result closer to 0).

a) Drawn square path of known dimensions (5” by 5”) on a piece of clean computer paper taped to a flat table top. b) Starting and ending placement of the IMU. One hand was used to move the IMU along the set path while the other hand held the cord.

Results of the linear displacement test. Original data is plotted, which includes acceleration, gyration, and calculated linear displacement. The raw data is very noisy and incorrectly results in three different displacements, where it should result in a net 0 mm.

Testing the IMU’s reliability.

The motion of the IMU for rotation. A) 0 degrees. This is the position in which the IMU was calibrated. B) 45°. C) 135°. D) 180°. Here, the motion was paused for roughly 1 second before returning back to the 0° position.

MATLAB plots used to obtain the resulting accuracy and precision of testing the IMU’s gyroscopic reliability. (left) Raw rotational data from one trial, or 3 cycles of the two known positions. Because the IMU was rotated about 1 axis, data is only shown along one of the three components. This is as expected. The IMU experienced three cycles per trial. (right) The relevant positions, the 0 and 180 degree positions were then extracted. From this, accuracy and precision was calculated, as the expected positions were known.
91 (left) Example of a square with a side length of 20 pixels and (right) circle with a radius of 20 pixels used for simulated B-scan reconstruction in yaw. Blue and yellow pixels have a value of 0 and 1, respectively.

92 (left) B-scan to be extracted in rotated position. (right) B-scan rotated to vertical position to prepare for extraction.

93 Simulated B-scan planes of circle at (left to right) -90°, -45°, 0°, 45°, and 90°. Blue pixels correspond to the background, light blue to the region of the circle not within the window of the B-scan, yellow to the region of the circle within the window of the B-scan, and green to the background region within the window of the B-scan. The red cross is the center of rotation.

94 Simulated B-scan planes of square at (left to right) -90°, -45°, 0°, 45°, and 90°. Blue pixels correspond to the background, light blue to the region of the circle not within the window of the B-scan, yellow to the region of the circle within the window of the B-scan, and green to the background region within the window of the B-scan. The red cross is the center of rotation.

95 Comparison of (left) simulated circle and (right) partially reconstructed circle. The pixel intensity in order of least to greatest is navy blue, light blue, teal, green, orange, light orange, and yellow. A higher intensity represents a larger number of intersections of distinct B-scans. In theory, regions with higher intensity values in this test will result in higher resolution images as the region is imaged from multiple viewpoints.

96 Comparison of (left) simulated square and (right) partially reconstructed square. The pixel intensity in order of least to greatest is navy blue, light blue, teal, green, orange, light orange, and yellow. A higher intensity represents a larger number of intersections of distinct B-scans. In theory, regions with higher intensity values in this test will result in higher resolution images as the region is imaged from multiple viewpoints.
97 Comparison of partial reconstructions of a circle with imaging radii of (left to right) 30, 40, and 50 pixels. At r=30, the center of the reconstructed geometry has the most intersecting B-scans and the number of intersections decrease radially. A similar gradient is seen in r=40, though the region with the most intersections is slightly above the center. Reconstruction with r=50 results in a horseshoe-like geometry. The region with the most intersections in this case lies along top half of the curve. (Note: Shapes may appear to have different dimensions, but this is the result of differing axes) ........................................ 136

98 Comparison of partial reconstructions of a square with imaging radii of (left to right) 30, 35, and 40 pixels. At r=30, the square is reconstructed in its entirety with a majority of it imaged by multiple B-scans. r=35 presents similar results, though the bottom corners of the reconstruction appear rounded and there appear to be fewer intersecting B-scans overall. With r=40, the limited field of view becomes apparent as the bottom half of the reconstructed square is either imaged once or twice or not at all. (Note: Shapes may appear to have different dimensions, but this is the result of differing axes) ..................... 136

99 (top left) Original square for which B-scans were simulated and reconstruction was performed. Reconstruction performed using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values. ........................................ 138

100 (top left) Original circle for which B-scans were simulated and reconstruction was performed. Reconstruction performed using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values. ........................................ 139

101 (top left) Original composite shape for which B-scans were simulated and reconstruction was performed. Reconstruction performed using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values. ........................................ 140

XVI
102 (top left) Original square for which B-scans were simulated and reconstruction was performed. Reconstruction performed with noise introduced into the angular displacement data using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values.

103 (top left) Original circle for which B-scans were simulated and reconstruction was performed. Reconstruction performed with noise introduced into the angular displacement data using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values.

104 (top left) Original composite for which B-scans were simulated and reconstruction was performed. Reconstruction performed with noise introduced into the angular displacement data using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values.

105 Example of reconstruction algorithm but frames acquired in pitch as opposed to yaw motion of probe. The region towards the center of rotation is densely imaged; however, regions farther radially are very sparsely imaged.

106 Final curvilinear rail sub-assembly design. A) Front view with the probe, IMU, and saline-bag system mounted onto a patient’s knee. B) Top view of the probe mounted onto the frame. The patient’s knee would enter the frame such that the top of the image represents the anterior and the bottom the posterior.

107 Final design with a subject in place is shown. Components include: the mechanical frame & wooden base, the curvilinear rail sub-assembly, and the saline-bag system.

108 A wooden dowel of known diameter was suspended in bag of US gel for the phantom reconstruction testing. The dowel was imaged roughly 20 to 25 times with an angular span of approximately 120°. Note that the scan window obtains a cross-sectional view of the wooden dowel. Total system verification was performed by acquiring the dowel’s radius from the resulting reconstruction and comparing it with the known diameter.
<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>109</td>
<td>CAD model of the final curvilinear rail sub-assembly used to determine the imaging radius. This was taken to be the radius formed between the center of rotation of the probe and the probe surface. The imaging radius is shown in blue, about 39.9 mm. The center of rotation is shown as a red dot. (left) Front view. (right) Side view.</td>
</tr>
<tr>
<td>110</td>
<td>Diameter of the reconstructed measured using MATLAB’s <code>imdistline</code> tool. Pixel distance was found to be 107.35. Based on the mm per pixel ratio of 0.0838 for a depth setting of 3 cm, this pixel distance corresponds to 9.7 mm.</td>
</tr>
<tr>
<td>111</td>
<td>Four trials of human subject testing were performed. (A) Two trials were performed with the saline bag system and (B) two trials were performed without the saline bag system. For all trials, a total of 20 to 25 B-scans were obtained with an angular span of approximately 120° beginning at the suprapatellar and ending at the medial region.</td>
</tr>
<tr>
<td>112</td>
<td>The femoral articular cartilage thickness was measured at the locations labeled A through E for every reconstructed image. Measurements at their equivalent locations were taken from the MRI and compared.</td>
</tr>
<tr>
<td>113</td>
<td>MRI scan of region of interest taken using Mimics. Femoral articular cartilage thickness is labeled along the entire knee. The measurements corresponding to A through E are the five measurements from the right. The scan is oriented with the four letters on the border A, L, P, R corresponding to anterior, left, posterior, and right, respectively. Since the scan is of the right knee, left is medial and right is lateral.</td>
</tr>
<tr>
<td>114</td>
<td>Plane along which MRI image was taken. The red line corresponds to the plane along which the US scans were taken.</td>
</tr>
<tr>
<td>115</td>
<td>Saline bag construction summary.</td>
</tr>
<tr>
<td>116</td>
<td>Identification of cropping window for depth of 5 cm using MATLAB’s <code>bwboundaries</code> function. Each depth setting had a slightly different cropping window; therefore, the process was repeated for all relevant depth settings.</td>
</tr>
<tr>
<td>117</td>
<td>Identification of cropping window for depth of 5 cm using MATLAB’s <code>bwboundaries</code> function. Each depth setting had a slightly different cropping window; therefore, the process was repeated for all relevant depth settings.</td>
</tr>
<tr>
<td>118</td>
<td>Test setup for determining reliability of the IMU’s gyroscope.</td>
</tr>
</tbody>
</table>
119  Test setup for determining reliability of the IMU’s gyroscope . . . . 186
## List of Tables

<table>
<thead>
<tr>
<th>Table</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Literature sources for articular cartilage measurements</td>
<td>8</td>
</tr>
<tr>
<td>2</td>
<td>Literature values for joint space width, femoral articular cartilage, and tibial articular cartilage</td>
<td>9</td>
</tr>
<tr>
<td>3</td>
<td>Comparing the Advantages and Disadvantages of Knee OA Imaging Types</td>
<td>15</td>
</tr>
<tr>
<td>4</td>
<td>Image acquisition methods for measuring femoral articular cartilage thickness using US</td>
<td>18</td>
</tr>
<tr>
<td>5</td>
<td>Probe manipulation maneuver effects</td>
<td>22</td>
</tr>
<tr>
<td>6</td>
<td>Typical system settings on US systems</td>
<td>23</td>
</tr>
<tr>
<td>7</td>
<td>Common Sonographic Artifacts in MSK US</td>
<td>24</td>
</tr>
<tr>
<td>8</td>
<td>Description, advantages, and disadvantages of common 3D US methods</td>
<td>25</td>
</tr>
<tr>
<td>9</td>
<td>Descriptions, advantages, and disadvantages of common probe localization methods</td>
<td>26</td>
</tr>
<tr>
<td>10</td>
<td>Attenuation indices of various ultrasound coupling media (Shigemura et al., 2017)</td>
<td>29</td>
</tr>
<tr>
<td>11</td>
<td>Literature values for joint space width, femoral articular cartilage, and tibial articular cartilage</td>
<td>48</td>
</tr>
<tr>
<td>12</td>
<td>3D US method decision matrix weighting scheme</td>
<td>51</td>
</tr>
<tr>
<td>13</td>
<td>3D ultrasound decision matrix</td>
<td>53</td>
</tr>
<tr>
<td>14</td>
<td>Probe localization method decision matrix weighting scheme</td>
<td>54</td>
</tr>
<tr>
<td>15</td>
<td>Probe localization decision matrix</td>
<td>55</td>
</tr>
<tr>
<td>16</td>
<td>MPU-6050 sampling rates for gyroscope and accelerometer</td>
<td>56</td>
</tr>
<tr>
<td>17</td>
<td>Coupling medium decision matrix weighting scheme</td>
<td>56</td>
</tr>
<tr>
<td>18</td>
<td>Coupling medium decision matrix</td>
<td>63</td>
</tr>
<tr>
<td>19</td>
<td>Conceptual design decision matrix</td>
<td>67</td>
</tr>
<tr>
<td>20</td>
<td>Cropping window for depths of 3 to 7 cm</td>
<td>90</td>
</tr>
<tr>
<td>21</td>
<td>Pixel to millimeter conversion for depth settings of 3, 4, 5, 6, and 7 cm</td>
<td>90</td>
</tr>
<tr>
<td>22</td>
<td>Optimal setting determined experimentally for imaging femoral articular cartilage with probe suprapatellar</td>
<td>108</td>
</tr>
<tr>
<td>23</td>
<td>Optimal range for each setting determined experimentally for imaging femoral articular cartilage with probe suprapatellar, 0° yaw, 0° pitch, 0° roll.</td>
<td>109</td>
</tr>
<tr>
<td>24</td>
<td>Person A results from repeatability phase 1</td>
<td>111</td>
</tr>
<tr>
<td>25</td>
<td>Person B results from repeatability phase 1</td>
<td>111</td>
</tr>
<tr>
<td>Table</td>
<td>Description</td>
<td>Page</td>
</tr>
<tr>
<td>-------</td>
<td>------------------------------------------------------------------------------</td>
<td>------</td>
</tr>
<tr>
<td>26</td>
<td>Person C results from repeatability phase 1</td>
<td>112</td>
</tr>
<tr>
<td>27</td>
<td>Optimal range for each setting determined experimentally for imaging</td>
<td></td>
</tr>
<tr>
<td></td>
<td>femoral articular cartilage with probe suprapatellar, 0° yaw, 0° pitch,</td>
<td></td>
</tr>
<tr>
<td></td>
<td>0° roll with ASP to improve image quality. While settings were ideally</td>
<td></td>
</tr>
<tr>
<td></td>
<td>left unaltered, they had to be altered for subjects B and C to obtain</td>
<td></td>
</tr>
<tr>
<td></td>
<td>readable femoral articular cartilage thickness measurements.</td>
<td>114</td>
</tr>
<tr>
<td>28</td>
<td>Person A results from repeatability phase 2</td>
<td>114</td>
</tr>
<tr>
<td>29</td>
<td>Person B results from repeatability phase 2</td>
<td>114</td>
</tr>
<tr>
<td>30</td>
<td>Person C results from repeatability phase 2</td>
<td>115</td>
</tr>
<tr>
<td>31</td>
<td>Summary of phase 1 &amp; 2 results averaged over three trials</td>
<td>115</td>
</tr>
<tr>
<td>32</td>
<td>IMU Global Positioning Results</td>
<td>128</td>
</tr>
<tr>
<td>33</td>
<td>Results from testing the IMU's gyroscope accuracy and precision</td>
<td>131</td>
</tr>
<tr>
<td>34</td>
<td>SSIM and 2D cross-correlation values for reconstructed square, circle, and</td>
<td></td>
</tr>
<tr>
<td></td>
<td>composite using nearest neighbor, bilinear, and bicubic pixel interpolation</td>
<td>145</td>
</tr>
<tr>
<td></td>
<td>methods</td>
<td></td>
</tr>
<tr>
<td>35</td>
<td>SSIM and 2D cross-correlation values for reconstructed square, circle, and</td>
<td></td>
</tr>
<tr>
<td></td>
<td>composite using nearest neighbor, bilinear, and bicubic pixel interpolation</td>
<td>145</td>
</tr>
<tr>
<td></td>
<td>methods with artificial noise introduced into the angular displacement data</td>
<td></td>
</tr>
<tr>
<td>36</td>
<td>Cost breakdown of components of final fixture design</td>
<td>149</td>
</tr>
<tr>
<td>37</td>
<td>Measurements of right knee femoral articular cartilage thickness measure-</td>
<td></td>
</tr>
<tr>
<td></td>
<td>ments taken from four trials of US imaging and reconstruction</td>
<td>161</td>
</tr>
<tr>
<td>38</td>
<td>Frequency at given depth</td>
<td>177</td>
</tr>
<tr>
<td>39</td>
<td>Cropping window for corresponding depth setting</td>
<td>178</td>
</tr>
</tbody>
</table>
1 Introduction

Osteoarthritis (OA) is the most common joint disease in the United States, and is a known precursor to physical disability in the elderly (Y. Zhang & Jordan, 2010). Roughly 10% of men and 13% of women over the age of 60 nationally and about 10% of people over 60 years of age internationally are afflicted, making it a major public health issue (World Health Organization, 2003; Y. Zhang & Jordan, 2010). OA diagnosis, treatment, and management costs the United States about $28.1 billion annually, which will likely increase as a result of the obesity epidemic and shifting age demographics in the coming years (Bitton, 2009).

OA is characterized by the degeneration of the bones, cartilage, menisci, ligaments, and synovial tissue in and around joints (Braun & Gold, 2012). Traditionally, radiography has been used to image OA-related joint morphology such as joint space width (JSW) narrowing and osteophytes, though other imaging modalities such as magnetic resonance imaging (MRI), computed tomography (CT), nuclear imaging, and ultrasound (US) have been used as well (Braun & Gold, 2012).

In clinical settings, OA is diagnosed using the Kellgren-Lawrence (KL) scale, a grading system based on subjective assessments of radiographs by trained individuals, which scores and establishes OA severity on a five-point scale from zero to four (Braun & Gold, 2012; Kellgren & Lawrence, 1957). Advantages of radiographs include ease of use, low cost, and high accessibility; however, radiographs are unable to directly image soft tissues, an important factor to consider in knee OA diagnosis (Hayashi, Roemer, & Guermazi, 2016; Wenham, Grainger, & Conaghan, 2014). MRI, on the other hand, can observe direct changes in joint structure and is instrumental to OA research (Hayashi et al., 2016). Soft tissues can be visualized with high resolution, allowing for improved diagnosis and treatment (Tanamas, Wluka, Jones, & Cicuttini, 2010). However, MRI can be impractical in smaller clinical settings as machines are expensive to install and operate (Naredo et al., 2009). CT, which is capable of obtaining many cross-sectional digital images like MRI, is also limited in its accessibility due to relatively high costs (Wenham et al., 2014). Nuclear medicine imaging and CT both rely on radiotracers to image active metabolism and bone structural changes (Hayashi et al., 2016). Limitations of nuclear imaging include poor anatomical resolution and the use of relatively toxic ionizing radiation (Tanamas et al., 2010).

Compared to the previously listed imaging techniques, US is user-centric and avoids the use of ionizing radiation while benefiting from the comfort of portabil-
ity and the ability to provide dynamic imaging capabilities (Braun & Gold, 2012). Considerable effort has already been placed into imaging musculoskeletal structures using US with a large portion dedicated to knee joint morphology imaging and OA diagnosis (Bevers, Bijlsma, Vriezekolk, van den Ende, & den Broeder, 2014; Faisal et al., 2018; Naredo et al., 2009; Slane, Slane, & Scheys, 2017). US holds promise due to its quick image acquisition time, convenience, low-cost accessibility, and sensitivity towards synovitis and joint effusion (Braun & Gold, 2012; Tanamas et al., 2010). One major limitation of US is its partial tissue and lack of bone penetration, which may pose a challenge for imaging deeper structures in a bony region such as the knee (Tanamas et al., 2010). However, with the growing prevalence of knee OA on both a national and international scale, there exists an increasing need for a portable, low-cost method of diagnosis.

A variety of fixtures have been designed to adapt US for imaging musculoskeletal structures, however, none have been specifically designed for the measurement of JSW and articular cartilage thickness in the knee (Herickhoff, Morgan, Broder, & Dahl, 2018; H. K. Zhang, Finocchi, Apkarian, & Boctor, 2016). The goal of this project was to create a prototype of a fixture and develop a method to use such fixture to image knee joint morphology to aid in the diagnosis of OA using US. These two clinical features have been identified as indicators of OA progression and severity. The objectives of this project were to:

- Determine the optimal orientation, and location of the probe on the knee to measure joint space width and femoral articular cartilage thickness.
- Design a fixture to maintain optimal orientation and location of an US probe on the knee.
- Stabilize the joint in at least one standard pose in which the joint space width reliably predicts a healthy or diseased state.
- Provide at least one image of the joint from which the joint space can be accurately, precisely, reliably, and repeatedly measured.
- Track the location and orientation of the probe during imaging and, through post-processing, create a reconstruction of the imaged region.
2 Background

2.1 Osteoarthritis

OA is a significant health concern in the United States with 27 million patients treated for OA annually (Bitton 2009). Approximately 10% of men and 13% of women over 60 years of age experienced knee OA in 2011 (Y. Zhang & Jordan, 2010). Around 10% of people over 60 years of age worldwide experience some form of OA. This number has been increasing in recent years due to the obesity epidemic and changing age demographics in the country (Bitton, 2009). It is the most common joint disease in the United States and is a known antecedent to physical disability in the elderly (Y. Zhang & Jordan, 2010). OA diagnosis, treatment, and management costs Americans $28.1 billion per year and accounts for 1 to 2.5% of the GDP of developed nations (Bitton, 2009). At a cost of $3.2 to 13.4 billion per year, OA is the costliest occupation-related disability (Bitton, 2009).

2.1.1 Definition and Epidemiology of Osteoarthritis

OA is a disease state characterized by degeneration of the bone, cartilage, menisci, ligaments, and synovial tissue covering and lubricating the ends of bones in joints, thinning the joint space and eventually allowing the bones to rub together (Braun & Gold, 2012). Symptoms of the disease include pain, swelling, and limited range of motion in the affected joint. OA may occur in any synovial joint in the body but most commonly affects the knees, hands, and hips (Y. Zhang & Jordan, 2010) (Figure 1).
BACKGROUND

Figure 1: Incidence of OA over various age groups by gender and joint location. Arthritis is most commonly found in the knees, and females are at an increased risk for all joints (Neogi & Zhang, 2013).

There are several important risk factors for OA, which can be broadly categorized as person-level and joint-level (Allen & Golightly, 2015). One major person-level risk factor for OA is sex. Females are significantly more likely to be treated for OA and are more likely to experience severe OA in their lifetimes (Allen & Golightly, 2015; Y. Zhang & Jordan, 2010). It is thought that hormonal changes lead to an increased risk of OA, as women treated with estrogen replacement after menopause are 15% less likely to require a total joint replacement (Y. Zhang & Jordan, 2010). Genetic factors are another major person-level factor (Y. Zhang & Jordan, 2010). It is estimated that genes influence 50-65% of the symptoms of OA, though this is more common for hand and hip OA than in the knee (Y. Zhang & Jordan, 2010). Obesity is a major and growing risk factor for OA (Allen & Golightly, 2015; Y. Zhang & Jordan, 2010). Individuals who are obese are 2.96 times as likely to develop knee OA in their lifetime (Neogi & Zhang, 2013). Low bone mineral density is often comorbid with OA, but a definitive cause and effect relationship between the two has not been established (Neogi & Zhang, 2013).

In addition to person-level risk factors, several joint-level risk factors are also important in the development of OA. Type of occupation is a major risk factor for OA that is generally joint-specific. Occupations that involve repetitive joint use, particularly repetitive bending and kneeling, account for a 1.6-fold increase in knee OA. However, there is some evidence that regular exercise and subsequent muscle
and joint strengthening can protect against OA. Normal person-to-person anatomical differences also account for some OA risk \cite{Neogi & Zhang 2013}.

2.1.2 Clinical Features of Osteoarthritis

OA is generally diagnosed with a combination of physical examination and image-based tests. In some cases, diagnosis occurs based on symptoms alone, in which it is called symptomatic OA; it can also be diagnosed solely based on radiographic findings and is called radiographic OA. The most recognizable symptom of OA is joint pain and stiffness \cite{Walker 2009}; however, there are several key physical and radiographic features that can be used to characterize the disease.

2.1.3 Functional Anatomy of the Knee

The knee is a complex joint with a wide range of motion and is typically classified as a modified hinge joint. The bones that meet at the knee joint are the femur, the tibia, and the patella, which are held in place by a collection of ligaments (Figure 2). The articular surfaces of the bones are covered with a layer of hyaline cartilage, which lubricates the joint and protects the sensitive bone ends from coming into contact. The medial and lateral condyles of the femur are covered with this cartilage. The medial and lateral menisci are found on the tibial plateau and form a cushioned platform for the femoral condyles. The femur-tibia connection is stabilized by the anterior cruciate ligament, posterior cruciate ligament, lateral collateral ligament, and medial collateral ligament. The patella is held in place by the patellar ligament \cite{Harris, Ranson, & Robertson 2014}.  


As with all movable joints of the body, the knee joint is surrounded by a cavity filled with fluid that further lubricates the cartilage interface, known as the synovial cavity (Figure 3).
Figure 3: Lateral cross section of the knee joint showing synovial joint structures.

Major muscles that attach at or around the knee include the quadriceps femoris, the main knee extensor, which inserts through the quadriceps tendon into the patella and then to the tibial tuberosity via the patellar ligament; the hamstrings muscles, which insert into the lateral tibia; and the vastus lateralis and medialis (Harris et al., 2014).

2.1.4 Knee Osteoarthritis

OA is defined as the degeneration of soft, lubricating tissues in the joint space. In particular, the articular cartilage becomes thinner due to wear and age (Kwok et al., 2011). The joint menisci also decrease in thickness and tend to lose their structural integrity over time, eventually becoming so thin that they no longer fulfill their function of protecting the joint (Verdonk et al., 2016). Through this process, the joint space becomes smaller (Kwok et al., 2011) (Figure 4). Thus, one of the key radiographic findings in OA is known as joint space narrowing, or a reduction in the joint space width (Kwok et al., 2011). Reliably measuring joint space width is a key first step in radiographic diagnosis of OA because accurate measurements can be used to diagnose OA before the onset of symptoms (Kwok et al., 2011). Indeed, the joint space width is remarkably invariable patient-to-patient in healthy individuals and only shows minimal reduction due to age in the absence of OA (Beattie et al., 2008).
The joint space is composed of the femoral articular cartilage, tibial articular cartilage, and the lubricating synovial fluid layer. The femoral and tibial articular cartilage can be measured independently. The medial and lateral condyles of the femur and the medial and lateral aspects of the tibial articular cartilage should be considered somewhat separately, as sex, joint stability, and anatomical differences may cause thinning of the femoral articular cartilage but not the tibial articular cartilage, or of the medial aspects without damage to the lateral aspects. Advanced knee OA is typically treated with total joint replacement surgery, but partial joint replacements, which reconstruct individual pieces of cartilage, are a recent addition to the market. Since partial knee replacements are preferable to total, it is important to know whether there is a difference in cartilage thickness and wear between the two condyles of a knee so that clinicians and patients can make more informed choices about which option is likely to be most effective. Human cadavers with and without OA have been assessed for femoral and tibial articular cartilage thickness as well as composition and stiffness (Adam et al., 1998; Shepherd & Seedhom, 1999). Tables 1 and 2 summarize femoral and tibial articular cartilage thickness and joint space width measurements reported by various sources.

Table 1: Literature sources for articular cartilage measurements

<table>
<thead>
<tr>
<th>Reference</th>
<th>Measurements</th>
<th>Purpose</th>
<th>Subject</th>
<th>Method</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adam et al. 1998</td>
<td>Articular cartilage &amp; joint space width</td>
<td>OA</td>
<td>Cadaver</td>
<td>A-mode US to directly measure cartilage perpendicular to articular surface</td>
</tr>
<tr>
<td>Shepherd &amp; Seedhom 1999</td>
<td>Articular cartilage thickness</td>
<td>Old age individuals</td>
<td>Cadaver</td>
<td>Direct measurement of dissected specimens</td>
</tr>
<tr>
<td>Beattie et al. 2008</td>
<td>Joint space width</td>
<td>Healthy individuals</td>
<td>Living subjects</td>
<td>X-ray</td>
</tr>
<tr>
<td>Tuna et al. 2016</td>
<td>Femoral articular cartilage thickness</td>
<td>OA</td>
<td>Living subjects</td>
<td>B-mode US, suprapatellar</td>
</tr>
</tbody>
</table>
Table 2: Literature values for joint space width, femoral articular cartilage, and tibial articular cartilage

<table>
<thead>
<tr>
<th>Physiology of Interest</th>
<th>Beattie et al.</th>
<th>Adam et al.</th>
<th>Tuna, Balci, &amp; Ozacazar</th>
<th>Shepherd &amp; Seedorphom</th>
</tr>
</thead>
<tbody>
<tr>
<td>Joint space width (mm)</td>
<td>5.25</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Medial femoral articular cartilage thickness (mm)</td>
<td>N/A</td>
<td>1.86 ± 0.36</td>
<td>1.925 ± 0.306</td>
<td>1.65 - 2.65</td>
</tr>
<tr>
<td>Lateral femoral articular cartilage thickness (mm)</td>
<td>N/A</td>
<td>1.94 ± 0.46</td>
<td>1.924 ± 0.398</td>
<td></td>
</tr>
<tr>
<td>Medial tibial articular cartilage thickness (mm)</td>
<td>N/A</td>
<td>1.75 ± 0.19</td>
<td>N/A</td>
<td>2.07 - 2.98</td>
</tr>
<tr>
<td>Lateral tibial articular cartilage thickness (mm)</td>
<td>N/A</td>
<td>2.08 ± 0.43</td>
<td>N/A</td>
<td></td>
</tr>
</tbody>
</table>

Figure 4: Radiographs of joint space in normal knee (A), and mild (B), moderate (C), and severe (D) OA (Altman & Gold, 2007)

Over time, increased friction between bones in synovial joints causes inflammation, which in turn causes the formation of osteophytes (Figure 5), sometimes called “bone spurs” (Altman & Gold, 2007). Osteophytes increase in size and number with the progression of OA and are an important measure of the severity of OA (Kellgren & Lawrence, 1957).
There are several other important clinical features of OA that are not as widely used for diagnosis. An increase in the density of the bone directly underneath the cartilage on the epiphysis of the bone, or subchondral sclerosis, may contribute to the progression of OA. Subchondral cysts – excess bone growth under cartilage, is another concern (Cox, van Donkelaar, van Rietbergen, Emans, & Ito, 2012).

2.2 Imaging Techniques for Diagnosing Osteoarthritis

Radiography (X-ray) serves as the ‘gold standard’ for diagnosing OA, mainly by measuring knee joint space width and visualizing osteophytes (Naredo et al., 2009). Newer methods of image diagnosis, such as MRI, CT, positron emission tomography, and US have all advanced OA treatment and management (Braun & Gold, 2012).
2.2.1 Radiography

Radiography is still one of the most accessible methods of evaluating the knee joint to diagnose OA. In this method, the patient’s knee is extended to obtain a bilateral anteroposterior image while the patient is weight-bearing. X-rays can detect OA-associated bony features like osteophytes, subchondral sclerosis, and cysts, and provide a qualitative picture of overall joint alignment (Hayashi et al., 2016). Joint space width measurement is used as a proxy for cartilage thickness and meniscal integrity, as soft-tissues cannot be imaged through radiography.

The KL classification is a five-point grading scale typically used to diagnose knee OA and has standardized radiograph interpretation since 1957 (Kellgren & Lawrence, 1957). The KL provides healthcare providers guidance on treatment, specifying which patients may best suited for surgical intervention (Kohn, Sassoon, & Fernando, 2016). It rates radiographs on a scale from 0 to 4, with Grade 0 indicating no presence of OA and Grade 4 indicating severe OA. Each grade is paired with detailed radiographic descriptions, though it is important to note that the time progression between stages defined by the scale is unclear (Kohn et al., 2016). The KL scale has been criticized due to its emphasis on osteophyte formation with respect to worsening OA, and has not deviated much since its creation (Altman & Gold, 2007). Despite its clear limitations, it remains one of the most widely used radiographic OA diagnosis schemes.

In 1996, another atlas developed by the Osteoarthritis Research Society International (OARSI) was introduced to aid in diagnosing OA progression (Braun & Gold, 2012). Instead of establishing a semi-quantitative grading scheme, OARSI provides image examples for specific features of OA for each grade. Additional efforts have been made to increase availability of some atlases for OA diagnosis by converting some radiographs to electronic format. While these scales provide useful baselines, they are but a part of a greater clinical assessment of medical history, physical exams, laboratory testing, and imaging testing.

Radiography, although an older method, is still relied upon due to its high accessibility, low cost, and simplicity. Its main limitation is that it does not allow for direct visualization of soft tissues, which can make early diagnosis of OA quite difficult. It also presents images of the knee in only one position, while joint space width may change with joint angle. Radiography has its workarounds, as OA symptoms like joint space narrowing and osteophytes still appear on X-ray images. Still, anteroposterior radiographic images detect only about 56% of OA cases (Wenham et al., 2014). Radiography is insensitive to structural changes, and once it is detected,
OA is already well-established. For example, knees with Grade 1 joint space narrowing have already lost 11-13% of cartilage, as well as additional bone marrow lesions, meniscal extrusion, and tibiofemoral cartilage defects (Tanamas et al. 2010).

### 2.2.2 Magnetic Resonance Imaging

Though the gold standard of imaging OA typically involves examining the joint space and osteophyte formation, recent evidence has encouraged clinicians to perceive the entire joint and multiple tissues. MRI specializes in soft tissue contrast and provides excellent visualizations of joint morphology and biochemistry. Despite its expense, the main advantage of MRI is its ability to visualize soft tissues directly, revealing some of the drawbacks of the gold standard radiography.

MRI allows for direct observation of changes in the joint structure from early/pre-OA through the established disease, both cross-sectionally and three-dimensionally. Additionally, MRI has also been able to link OA’s characteristic pain with changes in joint structure, as well as cartilage morphology and composition (Hayashi et al. 2016; Tanamas et al. 2010). MRI can assess cartilage damage through arthrography, as its high resolution provides delineation of subchondral bone marrow lesions, as well as central osteophytes, which are more pivotal to OA progression (Hayashi et al. 2016). When compared with CT and radiography, MRI was consistently more accurate and sensitive to detecting tricompartmental disease, cartilage loss, meniscal and ligamentous abnormalities characteristic to knee OA, and even capable of detecting early OA signs (Chan et al. 1991). These findings only support MRI’s stance as the gold standard in knee OA imaging, due to its use in noninvasive longitudinal monitoring of OA in both research and diagnosis.

Cartilage defects, bone marrow lesions, subchondral bone cysts, trabecular bone changes, synovitis and joint effusion, meniscal pathology can all be observed, quantified, and classified using MRI. Soft tissue visualization allows clinicians to more accurately diagnose patients in need of knee joint replacement, as well as identify and reduce risk factors in OA. While MRI has contributed much to the research in understanding the natural history of OA and may be useful in early detection, practical use in smaller clinical settings is limited due to its high cost and time-consuming nature (Naredo et al. 2009; Tanamas et al. 2010).
2.2.3 Computed Tomography

CT images physiological structures by obtaining multiple cross-sectional images and is on par with images obtained via MRI. CT is similar to radiography in that it is excellent at depicting cortical bone and soft tissue calcification (Hayashi et al., 2016). It can offer 3D imaging of a joint, and is typically cheaper, more accessible and has faster scan acquisition times compared to MRI (Wenham et al., 2014). CT can also be used to assess cartilage damage through arthrography. It is currently used for evaluating superficial and focal cartilage damage, which is particularly useful in differentiating between tissue boundaries (Hayashi et al., 2016). CT has also provided improved visualization of the knee joint particularly with pre-surgical planning and risk assessment.

The major limitation of CT is its use of ionizing radiation, though with peripheral joints like the knee, these doses are typically lower. In addition, CT relies on contrast agents to delineate cartilage, though it is able to visualize surface lesions better than MRI (Tanamas et al., 2010). In terms of diagnosing OA, CT is used as an alternative to MRI. A study found that CT reported similar findings to that of radiography in that it only displayed bicompartamental cartilage loss, as opposed to MRI’s tricompartmental loss (Chan et al., 1991). In fact, CT displayed less cartilage loss, about 25%, as compared to MRI’s 60% and radiography’s 35%. However, CT was able to display more osteophytes in the medial compartment than radiography. Overall findings establish that MRI is more sensitive than both radiography and CT in assessing the extent and severity of osteoarthritic changes.

2.2.4 Nuclear Imaging

Nuclear medicine imaging techniques such as PET, rely on radiotracers to image active metabolism and bone turnover changes exemplified in osteophyte formation, subchondral sclerosis, subchondral cyst formation, and bone marrow lesions – all of which are characteristic of OA (Hayashi et al., 2016). Bone scintigraphy, which involves capturing changes in bone metabolic activity, provides full-body images which can assist with locating soft tissues and bone origin of pain. While PET offers insight towards bone turnover and metabolic changes, it does not provide the necessary resolution for diagnosis (Tanamas et al., 2010).

Limitations of this method include poor anatomical resolution, but this can be overcome with hybrid technologies. This hybrid imaging (PET-CR and PET-MR) combines functional imaging, provided by PET, with high resolution anatomical
imaging, provided by either CT or MRI. It is additionally not a practical imaging method, as it can be expensive and is not portable.

### 2.2.5 Ultrasound

US relies on the emission, reflection, and detection of high frequency mechanical waves to identify underlying anatomical structures. US probes, also known as transducers, enable this by emitting short bursts of US into tissue and detecting the reflections, or echoes, to determine the structure of the underlying anatomy (Mikla I & Mikla V 2014). Current US technology provides multiplanar image acquisition and dynamic structures in real-time while maintaining low-cost accessibility and minimally invasive techniques (Braun & Gold 2012). In addition to its accessibility, US enables dynamic studies and weight-bearing examinations of patients with OA.

US has demonstrated sensitivity towards the presence of synovitis and joint effusion, but is limited in soft tissue assessment of tibial cartilage and subchondral joint structures (Tanamas et al. 2010). In addition, US is helpful for diagnosing painful joints (Tanamas et al. 2010). One European study of 600 patients found a positive and linear relationship between US-detected inflammatory features and knee pain in motion, which suggests a relationship between synovitis and knee pain (Hayashi et al. 2016). One review of US in managing knee OA found that US may have the potential to provide significant benefits in treating and managing knee OA, but literature on US lacks consistencies and its errors have been well documented (Chang, Fenster, Kathuria, Molinari, & Suri 2008). If anything, the field of US on managing knee OA requires stricter methodologies and more thorough reporting of parameters and application techniques. Overall, US is a generally safe form of imaging ideal for daily use in clinical settings due to its cost-effectiveness, portability and ease of use (Chang et al. 2008).

### 2.2.6 Knee Osteoarthritis Imaging Modality Summary

The summary of all imaging modalities can be seen in Table 3. Each of the imaging modalities discussed previously is briefly outlined with the given indicators: cost per scan, portability, radiation, ionizing agent, and overview of imaging capabilities.
Table 3: Comparing the Advantages and Disadvantages of Knee OA Imaging Types

<table>
<thead>
<tr>
<th>Image Modality</th>
<th>Cost per Scan</th>
<th>Portable</th>
<th>Radiation</th>
<th>Ionising Agent</th>
<th>Imaging Capabilities</th>
</tr>
</thead>
<tbody>
<tr>
<td>X-Ray</td>
<td>$50 *</td>
<td>Yes</td>
<td>Yes</td>
<td>No</td>
<td>Intended for imaging cortical bone and soft tissue calcifications</td>
</tr>
<tr>
<td>MRI</td>
<td>$400 *</td>
<td>No</td>
<td>No</td>
<td>No</td>
<td>Intended for imaging visualizing soft tissue, capable of 3D morphology</td>
</tr>
<tr>
<td>CT</td>
<td>$200 *</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
<td>Intended for imaging cortical bone and soft tissue calcification, capable of 3D morphology</td>
</tr>
<tr>
<td>Nuclear, PET</td>
<td>Not available</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
<td>Intended for imaging bone turnover and metabolic changes, generally not good enough for diagnosis, more capable when supplemented with other types of imaging like MRI/CT</td>
</tr>
<tr>
<td>Ultrasound</td>
<td>Not available</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
<td>Included for imaging soft tissues, poor bone penetration, capable of 3D morphology</td>
</tr>
</tbody>
</table>

(Koplas et al., 2008)
(The Center for Medical Imaging, 2018)

2.2.7 Musculoskeletal Ultrasound

The ability of US to create static and dynamic images non-invasively at relatively inexpensive costs has shown promise for applications such as joint morphology. In addition, “it has been shown to be more sensitive than clinical examination in picking up peri- and intra-articular soft tissue lesions” which allows the visualization of structural, mechanical, and inflammatory properties of a region of interest all with the same instrument (Bevers et al., 2014). Although a validated method for imaging musculoskeletal structures has yet to be developed, various groups have explored US as a potential alternative. US was used as an alternate method to quantify medial knee instability by measuring medial knee gap width (Slane et al., 2017). US provided measurements statistically similar to those from CT scans with high inter-rater reliability (Slane et al., 2017). Furthermore, the study explored the ability to measure knee gap width with an applied load, which, if validated, would provide valuable clinical insight into joint morphology with a noninvasive form of measurement (Slane et al., 2017). In another study, US was used to visualize and correlate cartilage and soft tissue structures with pain resulting from knee OA (Bevers et al., 2014). US
was used to identify several characteristics that were hypothesized to be associated with knee OA pain: effusion, synovial hypertrophy, meniscal protrusion, infrapatellar bursitis, Baker’s cyst, and cartilage thickness (Bevers et al., 2014). Although the study did not find correlations between ultrasonographic findings and knee OA pain, it suggests that US may be a viable alternative to more involved forms of medical imaging such as MRI.

In low-to-middle-income countries, diagnostic imaging is infrequent to almost lacking. As such, the demand for a more affordable, accessible image option has greatly increased. US has gained recent popularity because of its affordability, durability, and portability. Machine design has also become increasingly user-oriented, targeting more novice users. Not only are they becoming easier to use for untrained users, they may also be easier to use for non-native speakers. Fewer knobs and emphasis towards designing key features have improved overall usability of newer machines (Sippel, Muruganandan, Levine, & Shah, 2011).

US has a wide variety of applications, and is especially useful for both smaller clinic diagnoses, as well as pre-, post-, and during operation. This is because of US’s advantageous real-time image acquisition. Several smaller studies have found that US has successfully influenced the treatment and expectations of patient diagnoses in low-to-middle-income countries. In Ghana, 40% of all clinical diagnoses were improved, along with the approximate 28% of patients in the Amazon. Larger studies have also benefited from the introduction of US in clinics. Approximately 67.8% of US systems implemented in western Cameroon were found to be useful in diagnosis, and all the images, about 31.6% provided new diagnoses while 36.2% confirmed previous diagnoses. US was found to significantly impact diagnosis and management of patients in low-to-middle-income countries, adding to overall benefit and proof of general advantage of use (Sippel et al., 2011).

2.2.8 Joint Space Width Measurement

Though US may be able to identify features associated with OA pain, such information may only reflect on the analgesic properties of a given intervention and provide insufficient details on the anatomic features driving OA. One clinically relevant feature of knee OA is femoral articular cartilage thickness, which can be measured at the lateral condyle, intercondylar notch, and medial condyle. The thickness of the cartilage is measured as the distance from the soft tissue-cartilage interface and cartilage-bone interface (Naredo et al., 2009) (Figure 6).
US images of the femoral articular cartilage obtained from the knee in the transverse plane displayed the stark contrast between a healthy knee and a severely damaged knee (Figure 7). A significant correlation between US and direct, anatomic measurements of femoral articular cartilage thickness was found further supporting the use of US in OA diagnosis (Naredo et al., 2009).

When measuring femoral articular cartilage thickness in the knee, there has yet to be a consensus regarding the positioning of knee joint and the transducer...
neuvers employed. While all the studies assessed have used linear transducers for their aptness in visualizing superficial structures, they used various degrees of knee joint flexion and varied between transversely and longitudinally scanning the femoral articular cartilage (Table 4).

Table 4: Image acquisition methods for measuring femoral articular cartilage thickness using US

<table>
<thead>
<tr>
<th>Authors</th>
<th>Transducer</th>
<th>Transducer Pose</th>
<th>Knee Joint Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>Faisal et al., 2018</td>
<td>Linear transducer (8-12 MHz)</td>
<td>Placed transversely to leg and perpendicular to bone surface</td>
<td>Supine position with knee fully flexed (120°)</td>
</tr>
<tr>
<td>Bevers et al., 2014</td>
<td>35 mm linear transducer (8-15 MHz)</td>
<td>Immediately above patella in transverse plane</td>
<td>Knee in maximum flexion</td>
</tr>
<tr>
<td>Naredo et al., 2016</td>
<td>Linear transducer, 14 MHz, 72 dB dynamic range, 37 dB gain, 28 mm depth</td>
<td>Transducer placed transversely to leg just above upper pole of patella</td>
<td>Knee in maximum flexion (124° to 141°)</td>
</tr>
<tr>
<td>Schmitz et al., 2017</td>
<td>Linear transducer (10 MHz)</td>
<td>1) Transversely over medial femoral condyle 2) Longitudinally in sagittal plane midway between medial borders of patella and medial femoral condyle</td>
<td>1) 90° of flexion 2) Maximal flexion (133° to 151°)</td>
</tr>
<tr>
<td>Yoon et al., 2008</td>
<td>Linear transducer (12.5 MHz)</td>
<td>1) Transversely immediately above patella 2) Longitudinally along midline of medial or lateral condyles</td>
<td>Supine position under maximum flexion of knee joint</td>
</tr>
</tbody>
</table>

Though inconclusive, one study suggests that longitudinal scans are superior to suprapatellar transverse scans for measuring femoral articular cartilage thickness. The synovial space-cartilage interface was less defined in images obtained using the suprapatellar transverse scan and did not allow for the differentiation between maximum and minimum femoral articular cartilage thickness measurements which were clearly discernible using a longitudinal scan (Yoon et al., 2008). Figure 8 shows the suprapatellar transverse scan and longitudinal scan and images obtained from each probe maneuver.
US has been used to measure the tibial articular cartilage in cadavers but not in live humans. The presence and orientation of the patella and the cupped shape of the tibiofemoral joint make it difficult to measure tibial articular cartilage thickness without breaking apart the knee joint \cite{Kuroki2008}. However, both the thickness and sound conduction of the tibial articular cartilage have been used to assess osteoarthritic damage postmortem \cite{Adam1998, Kuroki2008}. Sound conduction allows the measurement of the stiffness and integrity of the tibial articular cartilage, which is often indicative of the strength of the joint and therefore predisposition to OA \cite{Kuroki2008}.

The promise of US, however, is equally matched by the challenges of adapting the technology for imaging joint morphology. For example, US is limited in tissue imaging and incapable of bone penetration, which presents major difficulties in imaging bony regions like the knee, which is frequently obstructed by the patella \cite{Bevers2014, Naredo2009}. For the knee, this issue can be mostly addressed by inducing hyperflexion; however, individuals with severe degenerative changes due to OA and other joint ailments may not be able to achieve extreme degrees of knee flexion \cite{Naredo2009}. In addition, the lack of a consensus regarding the anatomical significance of ultrasonographic features presents another obstacle in the acceptance of US as a viable imaging method for joint morphology. Though the interpretation of images obtained by various medical imaging systems
provides a challenge, this is further compounded by the comparatively high dependence on the sonographer for image acquisition in US compared to other imaging techniques such as MRI and CT. Despite these current limitations, US is still being explored as a potential imaging modality as it can be used to examine a variety of clinical indications and has no absolute contraindications [American Institute of Ultrasound in Medicine, 2012].

2.2.9 Physics of Ultrasound

Musculoskeletal (MSK) US is based on the same physical phenomena as other applications of diagnostic US; however, the structures of interest in joint morphology dictate additional considerations that may not otherwise be present in obstetrics or echocardiography. Diagnostic US is based on non-invasive techniques and is predicated upon the principle of piezoelectricity. This concept describes a property present in only a subset of materials that produces mechanical pressure when stimulated by an electric field and an electric field when stimulated by mechanical pressure. US probes harness this property by emitting short bursts of US into tissue and detecting the reflections, or echoes, to determine the structure of the underlying anatomy – this mechanism is referred to as the “pulse-echo principle” [Mikla I & Mikla V, 2014].

Clinical US is categorized into brightness-mode (B-mode) and Doppler mode. These modalities are based on differing types of US waves emitted by the transducer and are used for different applications. B-mode is typically used to display morphological features of vessels using a linear array of transducers to image a plane through the body. Doppler mode, on the other hand, can be split into several different categories of imaging applications, though its two main types include Pulsed-Wave Doppler and Continuous-Wave Doppler. The former involves sampling from only a small sample volume, while the latter involves sampling along a line through the body. Continuous-Wave Doppler does not provide the depth of the wave, as different crystals are used in sending and receiving the signals. Pulsed-Wave Doppler is particularly useful in determining depth estimates of the target site, as it uses only one piezoelectric crystal [Meairs & Hennerici, 2011].

In B-mode, US pulses are emitted at a fixed rate with each pulse emitted only after the probe has detected the reflection of the previous pulse [Lemoigne, Caner, & Rahal, 2007]. The reflected waves produce a cross-sectional view of the plane along which the pulse is emitted [Lemoigne et al., 2007]. The depth of penetration, thus the depth of structures that can be imaged, is determined by the frequency of the waves emitted [Lemoigne et al., 2007]. Higher frequencies (7-12 MHz) have
shorter wavelengths, which provide a higher imaging resolution. However, soft tissue attenuation of waves increases proportionally with frequency, thus decreasing imaging depth (Abu-Zidan, Hefny, & Corr, 2011). Because of this, diagnostic US for superficial structures is performed with linear array transducers, which typically emit higher frequencies (Abu-Zidan et al., 2011).

2.2.10 US System Settings, Probe Maneuvers, and Artifacts

Like all medical imaging modalities, images obtained from US must be interpreted with preexisting knowledge of the physiological structures of interest; however, with US, the interpretation of images in 2D is particularly challenging for untrained users. The differentiation of tissue types in US is based on the ability of the structure to reflect or transmit US waves in relation to the surrounding tissue – a property known as echogenicity. Tissues can be characterized as hyperechoic (white), hypoechoic (gray), or anechoic (black) depending on its color in the image. For example, bone appears anechoic with a hyperechoic rim accompanied with an acoustic shadow, a type of imaging artifact, behind it as it is impenetrable to US waves. Tissues such as cartilage, muscles, ligaments, and tendons appear hypoechoic and are distinguished based on unique characteristic features (Ihnatsenka & Boezaart, 2010).

Though the composition of the tissue dictates how it is perceived by US, a multitude of external factors can also heavily influence US images. These external factors can be broadly classified into US system settings and probe manipulation and orientation (Ihnatsenka & Boezaart, 2010) (Table 5 and Figure 9). US system settings refers to the manipulation of controls on the instrument to alter imaging to the desired quality: in a common US system, this can be achieved by modifying gain, time gain compensation, focus, depth, and scanning mode settings (Table 5).
Table 5: Probe manipulation maneuver effects

<table>
<thead>
<tr>
<th>Maneuver</th>
<th>Effect/Use</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pressure (P)</td>
<td>• Decrease distance to structure of interest</td>
</tr>
<tr>
<td></td>
<td>• Affects echogenicity of tissue</td>
</tr>
<tr>
<td>Alignment / Sliding (A)</td>
<td>• Scan broad region to find structure of interest</td>
</tr>
<tr>
<td>Rotation (R)</td>
<td>• Attain true axial view of target with long axis parallel to surface but not perpendicular to current US plane</td>
</tr>
<tr>
<td>Tilt (T)</td>
<td>• Change angle of incidence to make probe perpendicular to structure of interest</td>
</tr>
</tbody>
</table>

*Ihnatsenka & Boezaart (2010)*

Figure 9: US probe manipulation maneuvers *Ihnatsenka & Boezaart (2010)*
Table 6: Typical system settings on US systems

<table>
<thead>
<tr>
<th>Settings</th>
<th>Effect</th>
</tr>
</thead>
</table>
| Gain                         | • Increasing gain allows processing of more incoming echoes, creating a brighter image  
                               • Can be adjusted in nearfield, farfield, or overall field of screen display  
                               • Increasing gain also increases image noise and artifacts with loss of contrast and finer details                                       |
| Time Gain Compensation       | • Adjust image brightness at specific depths to address the fact that deeper structures experience more attenuation resulting in lower resolution                                                                 |
| Focus                        | • Region between nearfield and far field, known as “focal zone”, has highest lateral resolution  
                               • Manually adjust depth of scan to visualize structure of interest in focal zone                                                                 |
| Depth                        | • Adjusted by manipulating probe frequency and depth/penetration setting on US machine  
                               • Scanning at increased depth reduces frame rate and lowers image quality                                                                 |
| Scanning mode setting        | • Can change standard presets on most modern-day systems for common applications (abdomen, obstetrics, vascular, musculoskeletal)  
                               • B-mode, M-mode, Doppler                                                                                                                                               |

*(Enriquez & Wu, 2014)*

Even with the ideal parameters, however, sonographic artifacts can arise and display false structures or fail to display existing structures *(Abu-Zidan et al., 2011)*. Some common artifacts in MSK US are acoustic shadows, lateral shadows, anisotropic effect, refraction, and reverberations. It is important to note that these artifacts, though failures to accurately depict the underlying structures, are commonly used to identify tissue types *(Table 7)*.

---

23
Table 7: Common Sonographic Artifacts in MSK US

<table>
<thead>
<tr>
<th>Artifact</th>
<th>Effect/Use</th>
<th>Description</th>
<th>Likely Structures</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acoustic shadow</td>
<td>- US cannot transmit through solid structures that are strong reflectors</td>
<td>- Produces “shadow” behind structure</td>
<td>- Bone</td>
</tr>
<tr>
<td></td>
<td>- Can also occur as a result of refraction</td>
<td></td>
<td>- Calcified structure</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>- Injured and curled fibrous tissue</td>
</tr>
<tr>
<td>Lateral shadow</td>
<td>- Forms on flanks of curved structures</td>
<td>- Similar to an acoustic shadow but the shadows appear lateral to the</td>
<td>- Tendons</td>
</tr>
<tr>
<td></td>
<td>- No large difference in acoustic impedance at tissue interface, yet</td>
<td>structure of interest</td>
<td>- Cysts</td>
</tr>
<tr>
<td></td>
<td>insonating angle is adherence to tissue’s curvature</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Reverberation</td>
<td>- US beam reflected between two strong reflectors</td>
<td>- Mirror image</td>
<td>- Curved cortical bone tissue</td>
</tr>
<tr>
<td>Comet tail</td>
<td>- Presence of a metallic object embedded in tissue</td>
<td>- Strong, dense linear reflections deep to reflecting surface</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>- Intensity tapers like a comet’s tail</td>
<td></td>
</tr>
<tr>
<td>Refraction</td>
<td>- Difference in US propagation speed between two tissue types</td>
<td>- Lesions deep to interface appear displaced</td>
<td>- Interface between fat tissue and muscle tissue</td>
</tr>
<tr>
<td></td>
<td>- Wave direction changes deep to interface</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Anisotropic effect</td>
<td>- Tissues show abnormal echogenicity, typically in an oblique insonating angle</td>
<td>- Tissue echogenicity changes with slight rotation of probe</td>
<td>- Curved tendon and ligament insertions</td>
</tr>
</tbody>
</table>

*Ihnatsenka & Boezaart (2010)*

### 2.3 3D Ultrasound and Reconstruction

Although 2D US is well-established within the medical community, it has multiple limitations: 1) effective use and interpretation of 2D US requires extensive knowledge of the underlying anatomy and 2) how it is represented in US, familiarity with the effects of various probe maneuvers, and an understanding of the multitude of artifacts present. The appeal of 3D US is indisputable as it is much easier to interpret compared to 2D images and less dependent on the skill of the sonographer.
In addition, it allows the visualization of planes otherwise impossible in 2D US and improves the field of view, thus providing clinicians a more complete view of underlying anatomic structures (Huang & Zou, 2015; Mozaffari & Lee, 2017). In addition, it is particularly useful for applications such as joint morphology as it enables accurate distance measurements in any orientation (Fenster, Downey, & Cardinal, 2001).

Within 3D US, there are three approaches to constructing 3D images of the anatomy: 2D array scanning, mechanical scanning, and freehand scanning (with and without position sensing) (Huang & Zou, 2015; Chang et al., 2008) (Table 8).

Table 8: Description, advantages, and disadvantages of common 3D US methods

<table>
<thead>
<tr>
<th>Method</th>
<th>Description</th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
</thead>
<tbody>
<tr>
<td>2D Array Scanning</td>
<td>2D-array transducer allows imaging of volume as opposed to 1D-array which only images a plane</td>
<td>• Real-time 3D imaging (4D US)</td>
<td>• Expensive</td>
</tr>
<tr>
<td></td>
<td></td>
<td>• Does not require 3D reconstruction software</td>
<td>• Requires specialized US system</td>
</tr>
<tr>
<td>Mechanical Scanning</td>
<td>Motorized mechanism used to translate, rotate, and/or tilt probe while acquiring 2D images</td>
<td>• Consistent, repeatable motions</td>
<td>• Added size and weight</td>
</tr>
<tr>
<td></td>
<td></td>
<td>• Position can be tracked easily</td>
<td>• Decreased flexibility in imaging regions of interest</td>
</tr>
<tr>
<td>Freehand Scanning (with position sensing)</td>
<td>Operator manipulates position and orientation of probe over anatomy and sensor on probe used to orient 2D images in 3D reconstruction</td>
<td>• Increased flexibility in imaging regions of interest</td>
<td>• Must track motion in 3D</td>
</tr>
<tr>
<td></td>
<td></td>
<td>• Only requires addition of small sensor (i.e. IMU)</td>
<td>• Increased computation required to orient 2D images in 3D</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• Frequent calibration required</td>
</tr>
<tr>
<td>Freehand Scanning (without position sensing)</td>
<td>Assume predefined scanning geometry and gather set of 2D images for 2D reconstruction</td>
<td>• Does not require addition of any physical components</td>
<td>• Accuracy not guaranteed because position and orientation not tracked</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>• Cannot be used for measurements</td>
</tr>
</tbody>
</table>

When using a 2D US probe to image in 3D, multiple 2D scans need to be processed to create a 3D composite in a method known as volume reconstruction. Volume reconstruction presents another challenge in the adaptation of US for imaging joint morphology as it requires accurate input on the location and orientation (pose)
of the probe to correctly orient the 2D images in 3D space. This substantially increases the computational load placed on the system (Huang & Zou, 2015). Various methods, or localization devices, can be used to determine the pose of the probe: electromagnetic positioning system (EPS), optical positioning system (OPS), and mechanical positioning system (MPS) (Table 9).

Table 9: Descriptions, advantages, and disadvantages of common probe localization methods

<table>
<thead>
<tr>
<th>Localization Method</th>
<th>Description</th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
</thead>
<tbody>
<tr>
<td>EPS</td>
<td>Use magnetic fields (AC or DC) to determine location and orientation (Mozaffari &amp; Lee, 2017)</td>
<td>• Most cost effective and convenient (Mozaffari &amp; Lee, 2017) &lt;br&gt; • Not limited by line of sight</td>
<td>• Susceptible to disruption by metallic materials (Mozaffari &amp; Lee, 2017)</td>
</tr>
<tr>
<td>OPS</td>
<td>Use multiple cameras to track markers on structure (Cenni et al., 2016)</td>
<td>• Systems with large number of cameras (&gt;10) have high precision (Cenni et al., 2016) &lt;br&gt; • Inexpensive, portable versions exist (Cenni et al., 2016)</td>
<td>• Limited by line of sight (Mozaffari &amp; Lee, 2017) &lt;br&gt; • Requires large set of additional equipment</td>
</tr>
<tr>
<td>MPS</td>
<td>Motorized mechanism used to translate/rotate/tilt probe while acquiring frames (Chang et al., 2008)</td>
<td>• Consistent, repeatable motions (Chang et al., 2008) &lt;br&gt; • Position track easily (Chang et al., 2008)</td>
<td>• Added size and weight &lt;br&gt; • Decreased flexibility in imaging regions of interest</td>
</tr>
</tbody>
</table>

Once the desired scans have been delivered to the processing system with their respective pose information, one of several reconstruction methods is utilized to create a 3D reconstruction of the 2D scans: voxel-based method (VBM), function-based method (FBM), pixel-based method (PBM), and hybrid method. Though these methods employ different algorithms to perform reconstruction, their goal is to translate pixels into voxels. This process involves selecting the most appropriate, representative 3D grid to fit the imaging data into, then implementing an interpolation technique to determine voxel values for coordinates that were imaged once, imaged multiple times, and not imaged (Mozaffari & Lee, 2017). The use of 3D freehand US requires the integration of multiple systems that can be categorized chronologically into localization, image acquisition, pre-processing, reconstruction, post-processing, and visualization (Figure 10).
2.4 Musculoskeletal Ultrasound Accessories

Although US technology has experienced breakthroughs, other methods to improve US imaging quality have been developed. Many accessories that have been developed have allowed the adaptation of US for a broadened field of use. Furthermore, these accessories are often much more cost-effective compared to purchasing an US imaging system and can be used to augment imaging systems that may otherwise be considered obsolete.
2.4.1 Acoustic Standoff Pad

The acoustic standoff pad (ASP) is a packaged sonolucent gel that provides a flexible, minimally attenuative platform to separate the probe from body surfaces intended to improve visualization of superficial structures (Civco, 2006) (Figure 11). In addition, the flexibility of the ASP allows it to conform to irregularly shaped body surfaces – this is particularly useful when scanning bony regions such as the knee (Biller & Myer, 1988). ASPs are non-sterile and reusable, making them an economic option for adapting US for imaging superficial structures. It only requires acoustic gel to be applied to both the surface interfacing with the anatomy and the probe (Civco, 2006).

![Aquaflex ultrasound gel pad (ASP).](image)

Figure 11: Aquaflex ultrasound gel pad (ASP).

One immediately noticeable limitation of the ASP is the limited surface area that can be imaged while maintaining the position of the ASP as they are typically 10 cm by 15 cm with varying thicknesses. Additionally, the user is required to fix the ASP in place with one hand while scanning the region with the other. Though it greatly improves the imaging capability of US, it compromises in ease of use. A patented design to address this issue has been issued and is comprised of a holder that houses a cutout of the gel (Wendelken & Pope, 2005) (Figure 12).
2.4.2 Water as Coupling Medium

Similar to the ASP, water has also been used as a coupling medium for US (Huang & Zou, 2015; Shigemura et al., 2017; Michalek, Donaldson, McAleavey, Johnston, & Kiska, 2013; H. K. Zhang et al., 2016). In addition to the low attenuation index of water compared to various media that US is propagated through, water conforms to the topology of the underlying anatomy, thus mitigating any difficulties related to interfacing the probe with the anatomy (Table 10).

Table 10: Attenuation indices of various ultrasound coupling media (Shigemura et al., 2017)

<table>
<thead>
<tr>
<th>Medium</th>
<th>Attenuation Index (dB/cm/Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water (20°C)</td>
<td>0.002</td>
</tr>
<tr>
<td>US Gel</td>
<td>0.05</td>
</tr>
<tr>
<td>Blood</td>
<td>0.2</td>
</tr>
<tr>
<td>Fatty Tissue</td>
<td>0.6</td>
</tr>
<tr>
<td>Muscle</td>
<td>2.3</td>
</tr>
<tr>
<td>Air</td>
<td>12.3</td>
</tr>
<tr>
<td>Bone</td>
<td>13.0</td>
</tr>
</tbody>
</table>

The speed of US in water (1480 m/s), however, is slightly lower than the speed of US in tissue, which ranges from roughly 1540 to 1590 m/s (Azhari, 2010). This difference can be diminished by adding salt to deionized water. Ultrasonic speed, $v$,
(m/s) as a function of the saline concentration, \( C \), (g/L) at a temperature of 22°C can be expressed as the following (Al-Nassar, Al-Halal, Khan, & Al-Kaabi, 2006):

\[
v(C) = 0.94C + 1480.5
\]

2.4.3 External Fixtures

Like many fields in medicine, the augmentation of US imaging systems with robotic technology is an active field of research. One particularly promising project is the use of a six degree-of-freedom robotic arm combined with a concept known as synthetic tracked aperture US (STRATUS) to detect the movement and orientation of the probe so that low-resolution images can be superimposed to create higher resolution composites (H. K. Zhang et al., 2016). This system uses low-frequency US to image deep structures and was shown to successfully improve image quality in comparison to images obtained using the conventional B-mode. Although this technology is highly promising, the use of a robotic arm greatly decreases its accessibility and increases the complexity of the overall system.

In contrast to the previously mentioned complex co-robotic US system, projects have also been dedicated to developing low-cost, highly-accessible US systems. The main difficulty with US is its dependence on the positioning and orientation of the probe and the lack of cardinal orientation, as used in more complex imaging systems like CT or MRI. By designing a fixture mounted with an orientation sensor, individual 2D scans can be loaded into voxels at the appropriate orientation, allowing even unskilled users to intuitively visualize anatomic features (Morgan et al., 2018) (Figure 13).

Figure 13: Volumetric reconstruction of 2D scan using probe orientation sensor (Dahl, 2018)

Another low-cost design for the adaptation of 2D US systems for 3D US features a
fixture that allows the user to choose from one of three rotational degrees of freedom (Figure 14). The fixture secures the probe in a cradle and tracks its orientation using an inertial measurement unit (IMU) so that it can tag 2D scans with information on the orientation of the image when performing 3D reconstruction. The system was able to rapidly acquire and produce reconstructions of phantoms and in vivo structures accurately (Morgan et al., 2018).

Figure 14: Low-cost volumetric US fixture (Morgan et al., 2018)

2.5 Ultrasound Regulations

As with other medical devices, US imaging must also undergo similar testing and regulations. One major concept all imaging must bear in mind is ALARA, which is exposure As Low As Reasonably Achievable (Ng, 2002). Common guidelines provided by major US organizations similarly assist in reducing exposure by teaching users to diagnose as safely and as effectively as possible.

2.5.1 Bioeffects of Ultrasound

The effects of US through human tissue has been extensively researched, namely through dose-effect studies. Those that found US-induced negative effects have occurred at higher intensities than diagnostic US (Ng, 2002). While US results in fewer adverse effects than other ionizing radiation types, it still can affect the body thermally and mechanically by producing pockets of gas in body fluids or tissue, called cavitation (Turnbull, 2014). Mechanical interactions of US on tissues can affect overall development and vibration of microbubbles in the tissue. US on biological tissues
can result in stable cavitation, which is the creation of bubbles, and transient cavitation, which refers to the collapse of microbubbles. These bioeffects of US can be measured with a thermal index and a non-thermal index, known as the mechanical index. Applying a sound beam to a biological tissue results in attenuation due to absorption, but for low power US, heat is rapidly dissipated (Ng, 2002).

Imaging modes, particularly B-mode, pulsed and color flow Doppler, can be characterized by different lengths of the pulses used, repetition frequency, and pressure in the pulses (Hamper, DeJong, Caskey, & Sheth, 1997). Both pulsed Doppler and color flow imaging have narrow, focused beams, which may result in larger temperature gradients experienced across the skin as a result of absorption of sound waves by the tissue scanned. However, major takeaways, as with all imaging modalities, emphasize that scanning should only be carried out when there is a clinical need particularly in applications of higher risk, like obstetrics or ophthalmometry.

2.5.2 Regulation and Safety Guidelines

US, unlike ionizing radiation, has fewer risks and, subsequently, fewer regulations on diagnostic equipment. The United States Food and Drug Administration (FDA) introduced acoustic output limits, based on the application. The two most conservative applications for US include ophthalmic and fetal exposure (Ng, 2002). Additional developments for safety include displaying the real-time acoustic output for both thermal index and mechanical index. Thermal index is specific to soft tissue and bones, while mechanical index estimates the risk of producing mechanical effects in the overall tissue (Turnbull, 2014). It should be noted that US probes are rated as a Class 2 medical device (Turnbull, 2014).

Overall benefits and risks of US can be optimized per location using operator controls that toggle output intensity and transmitted US field. Some examples include system mode, pulse repetition frequency, focusing depth, pulse length, and transducer choice. Thresholds of bioeffect intensity is not yet known, so operators are highly encouraged to administer personal judgement to determine optimal intensity output of the equipment while using the lowest output power. Both national (FDA) and international standard are outlined in Figure 15 for each of the US modes and exposure levels (Ng, 2002).
2.5.3 Scope of Practice & Clinical Standards for Diagnostic Sonographers

The Scope of Practice and Clinical Standards for the Diagnostic Medical Sonographer document, created in 2015, is a collaborative effort of 16 different organizations to describe and outline the current community standard of care [Ng, 2002]. This document sets the scope of practice for diagnostic medical sonographers, as
well as their role in the clinic. It emphasizes the patient as the central point of interest, to maximize patient safety above all else. Facilities that offer US imaging are highly encouraged to participate in accredited and certification programs that teach ALARA by the American Institute of Ultrasound in Medicine (AIUM) and the American Registry of Diagnostic Medical Sonographers (ARDMS).

Devices should additionally use proper patient positioning, tools, devices, and ergonomic adjustment in addition to appropriate scanning techniques. Sonographers must also follow defined standards, which involve patient assessment techniques, patient education and communication, diagnostic protocol and implementation, diagnostic exam results, and documentation. Sonographers are expected to provide utmost quality of care, with minimal dosage, and effective communication and assessment of the results.

2.5.4 Ultrasound Imaging Protocol

The ARDMS has developed a guideline for the examination of various joints and anatomic regions with the examination of the knee divided into four quadrants (American Institute of Ultrasound in Medicine, 2012). The guideline was developed to examine the knee for a variety of clinical indications and comprehensively covers the knee. It contains specific instructions on the placement of the probe and the position of the patient. It states that MSK US should be performed with a high-resolution linear array transducer with a large bandwidth at frequencies between 7.5 and 12 MHz (American Institute of Ultrasound in Medicine, 2012). Though a protocol for knee examination specifically for OA has not yet been agreed upon, this guideline provides a thorough method for examining the knee.

US can be used to measure several different types of knee pathologies with the use of multiple probes; these include muscle injuries, ligament and tendon injuries, and bone injuries and pathologies. The most common muscle injuries detectable by US are muscle tears, particularly of the quadriceps and gastrocnemius muscles, though damage to other muscles can also be detected. Tendon and ligament injuries detectable by US include ACL and PCL tears, as well as damage to the patellar tendon. While bone fractures and calcification can be detected by US, there is no established method to image these structures for the diagnosis of OA via US (Connell & Comin, 2014).

For imaging of the knee joint via US, several probes are typically needed. A high frequency linear probe is generally used to image superficial structures, while lower frequency probes are used to assess deeper structures, such as the ACL. Images
are first taken from an anterior view with the knee at 30 degrees of flexion, which provides the best view of the joint capsule without interference from the patella. Then, the sonographer may move to lateral views, particularly with the knee fully extended and the probe angled up toward the joint capsule past the lateral femoral condyle. It is recommended to use color Doppler US imaging for tendon, ligament, and cartilage pathologies (Connell & Comin, 2014).
3 Project Approach

3.1 Initial Client Statement

The following initial client statement was provided by Professor Karen Troy, from Worcester Polytechnic Institute:

*Ultrasound measurement of joint morphology to create 3D models: development and validation of tools to improve validity and reproducibility.*

Ultrasound is a relatively low-cost and portable imaging modality that has the potential for broad applications in musculoskeletal health. A key advantage of ultrasound is that it can be used during functional and weight-bearing movements, which are often difficult to capture with other types of imaging, such as MRI. Ultrasound has been broadly used for surgical guidance but has the potential for generating 3-dimensional images and measurements, and image-based models for simulation of mechanical behavior.

A key challenge for ultrasound imaging is that many intrinsic and extrinsic factors affect measurements of musculoskeletal tissues. For example, extrinsic factors such as probe angle, transducer/skin pressure, and other operator-dependent factors can affect image quality and measurements. Intrinsic factors such as the amount of synovial fluid present in a joint space, tissue-specific deterioration, and morphology, can all affect measurements of joints and tissues. These factors collectively hamper the validity and reproducibility of measurements. To make ultrasound measurements more broadly useful, improved tools combined with standardization and validation of data collection and measurement should be developed.

**Project Goals:**

**Project 1: 3D mapping of the knee joint – methods and sensitivity analysis**

Goal 1: Determine the sensitivity of ultrasound measurements of joint space width to extrinsic factors such as probe angle, skin pressure, and location.
Goal 2: Develop methods and tools to minimize variability due to extrinsic factors. These may include robotic-assist systems, feedback tools, or other methods to maximize repeatability of measurements.

Goal 3: Determine the sensitivity of ultrasound measurements of joint space width to intrinsic factors such as skin and subcutaneous fat thickness, amount of synovial fluid, and cartilage condition.

Goal 4: Validate ultrasound measurements of joint space width against other 3D imaging modalities (MRI or CT).

3.2 Objectives

Objectives for the project and their rationales were developed based on the initial client statement, literature review, and discussions with the client. The goal of this project was to create a prototype of a fixture and construct a method to use the fixture to image knee joint morphology to aid in the diagnosis of OA using US. A variety of fixtures have been designed to adapt US for imaging musculoskeletal structures, however, none have been specifically designed for the measurement of joint space width and cartilage thickness in the knee [Morgan et al., 2018; H. K. Zhang et al., 2016]. These two clinical features have been identified as indicator of OA progression and severity. The objectives of this project are the following:

Determine optimal orientation and location of the probe on the knee to measure joint space width and femoral and tibial articular cartilage thickness.

The orientation and location of the US probe can have significant effects on the measurements obtained from US. To address this issue, a specific imaging protocol is needed to control these variables. A protocol that can be generalized to image patients’ knees regardless of their state must be developed to standardize acquisition of data relevant to OA diagnosis. This protocol will dictate the system settings of the US machine and the probe maneuvers required to image physiologic structures of interest with repeatability, reliability, accuracy, and precision. These factors will be additionally supplemented by values obtained in literature research.

Design a fixture to maintain optimal orientation, and location of an US
probe on the knee to image regions and features of interest.
Joint space width measurements should be repeatable not only across different patients but also across sonographers with varying levels of experience. Therefore, it is insufficient to simply provide a protocol with a chosen orientation and location – a mechanical fixture or feedback mechanism is necessary to incorporate into the design. Once the imaging protocol has been developed, a fixture can be designed to aid in the standardization of the probe maneuvers to be employed. This will broaden the target user demographic by allowing both experienced and non-experienced users to utilize the system to image knee joint morphology and obtain similar joint morphological data.

Stabilize the joint in at least one standard pose in which measurements obtained from imaging reliably predicts a healthy or diseased state of the knee joint.
The degree of flexion in the knee determines the visibility of underlying structures in the knee to US as well as possibly affecting the measured joint space width, because the femoral articular cartilage, tibial articular cartilage, and synovial fluid thickness are not constant for all regions of the condyles, nor for all knee angles or poses. The posterior aspect of the femoral condyle, for instance, may not be exposed to as much shear stress, and so may not display the same degree of degradation as the inferior portion. Stabilizing the knee for specific imaging procedures ensures that the structures of interest are unobstructed across a variety of sonographers and patients.

Provide at least one image of the joint from which the joint space width, femoral articular cartilage thickness, and tibial articular thickness can be accurately, precisely, reliably, and repeatedly measured
While controlling all the variables determined by the previous objectives, it remains necessary to image the knee using US in order to measure the joint space width in at least one location. With the determined US protocol obtained in Objective 1, mechanical fixture from Objective 2, and pose from Objective 3, an US image can then be generated. This will be accomplished to ensure repeatability, reliability, accuracy, and precision, which will be verified on both a non-human and human test subject. These indicators will be additionally supplemented by values obtained in literature research.

Track the location and orientation of the probe during imaging and,
through post-processing, create a stitched visualization or reconstruction of the imaged region.

Although B-scans can be used to approximate joint space width or femoral articular cartilage and tibial articular cartilage thickness, the limited perspective inhibits the ability to visualize complex topology. The location and orientation of the probe can be used to orient B-scans in space. Multiple acquired B-scans can be combined to form a single reconstructed image, which is much more visually intuitive and allows for measurements of joint space width and femoral articular cartilage and tibial articular cartilage thickness along a wider region, as well as allowing for the identification of the thinnest joint space width on a given knee.

3.3 Constraints

Similarly, a list of constraints was developed for the project based on the limitations regarding time, resources, function, regulations, and other external factors:

Compatible with a variety of US models and with all linear transducers
The imaging protocol, fixture, and software must be compatible for any US system and the fixture must be compatible, with minimal extra parts, with all linear transducers. The purpose of the project is to provide users with an effective, relatively inexpensive alternative to purchasing a separate US system for imaging joint morphology by adapting traditional, 2D US technology; therefore, the design must accommodate a variety of systems that may seek to use this product as an accessory.

Cost less than $750 to prototype
The MQP budget of $250 per person limits the project budget to $750. The combined cost of research, development, prototyping, and testing should remain below this value.

Comply with FDA and ISO regulations regarding ultrasound
The FDA classifies diagnostic US transducers as a class 2 device (Health, n.d.). To function within the class, any methods or fixtures developed and designed must comply with regulations outlined by this class.

Fixture weight under 2 pounds and maximum dimension of one foot in each direction
The goal to provide an accessible method of imaging joint morphology requires that the fixture be easy to handle for the clinician and be comfortable for the patient. Although the values for what can be considered “usable” are subjective, the design must be as light and as compact as reasonably possible to promote portability and ease of use.

**IEC compliance**

Should the fixture draw electrical current, it must comply with IEC 60601 standards, which establish the current safety and essential performance of medical electrical equipment. The device will be both safe to operate with minimal risk to the clinician and patient.

### 3.4 Revised Client Statement

Based on the objectives and constraints identified and discussions with the client, the initial client statement was revised as follows, with notable portions underlined:

**Ultrasound measurement of the knee joint to identify and measure clinical features of osteoarthritis: Development and verification of an imaging protocol and/or mechanical fixture to standardize diagnosis.**

Ultrasound is a relatively low-cost, portable, and safe imaging modality that has the potential for applications in musculoskeletal health. It has been explored as an alternative to traditional medical imaging technologies in the diagnosis of knee osteoarthritis such as radiography, magnetic resonance imaging, and computed tomography. The ability to image functional and weight-bearing movements also makes it a promising technique for those in practice and in academia.

A key challenge in musculoskeletal ultrasound imaging is that a variety of intrinsic and extrinsic factors affect the images produced. Extrinsic factors such as probe location, probe angle, probe pressure and other operator-dependent factors can affect imaging quality and measurements. Intrinsic factors such as the amount of synovial fluid present in joint space, tissue-specific deterioration, and joint morphology can affect measurements of joints and tissues.

Though musculoskeletal ultrasound has shown success in academia, it has
yet to be widely adopted in medicine due to the lack of verification on the accuracy, precision, repeatability, and reliability of measurements. This can be attributed to the lack of a consensus on the imaging procedure and interpretation for the diagnosis of a musculoskeletal disease such as osteoarthritis. To encourage the adoption of ultrasound for musculoskeletal applications, improved tools combined with standardization and verification of data collection and measurement should be developed.

**Project Goal:** Develop and design an imaging protocol and/or mechanical fixture to reliably, accurately, precisely, and repeatedly measure the knee joint width space to diagnose knee osteoarthritis with ultrasound.

### 3.5 Problem Statement

A variety of fixtures have been designed to adapt US for imaging musculoskeletal structures; however, none have been specifically designed for the measurement of joint space width and articular cartilage thickness in the knee (Morgan et al., 2018; H. K. Zhang et al., 2016). The goal of this project is to create a prototype of a fixture and develop a method to use such fixture to image knee joint morphology to aid in the diagnosis of OA using US. These two clinical features have been identified as indicators of OA progression and severity. The objectives of this project are to:

- Determine the optimal orientation of the probe on the knee to measure joint space width by determining femoral articular cartilage thickness and tibial articular cartilage thickness.
- Stabilize the joint in at least one standard pose in which the joint space width reliably predicts a healthy or diseased state.
- Provide at least one image of the joint from which the joint space can be accurately, precisely, reliably, and repeatedly measured.
- Track the location and orientation of the probe during imaging and, through post-processing, create a stitched visualization or 3D reconstruction of the imaged region.

#### 3.5.1 Problem Statement Breakdown

The following section describes the breakdown of the problem statement as described above, which is important in developing a project approach for the design.
Each of these topics is essential when formulating an initial design, as each provides context and presents a need for specific features.

**Determine the optimal orientation of the probe on the knee to measure joint space width by determining femoral articular cartilage thickness and tibial articular cartilage thickness.**

To obtain the measurements of the joint space width, by gathering femoral and tibial articular cartilage thickness, it is first necessary to understand how the anatomy shifts at different views, locations, and angles of the knee. The movement of the patella with respect to changing knee angles must be understood in determining where best to locate and orient the probe. Knowing how best to angle the probe such that the anatomical landmarks of interest are exposed is essential not only for optimal ultrasound scans, but also for later image reconstruction.

**Stabilize the joint in at least one standard pose in which the joint space width reliably predicts a healthy or diseased state.**

As the knee joint may be considered a complex joint, it may be necessary to limit the degrees of freedom in the model to reduce overall variation. This method of reconstruction comes at a cost, however, as it may sacrifice overall portability, usability, and manufacturing price. Multiple poses may be considered for a more complete 3D image reconstruction, or at the very least, a higher degree of accuracy when approximating the joint space width. When evaluating poses, is may also be helpful to consider the method from the patient’s perspective in that a realistic pose should be agreed upon for a typical knee OA patient; in other words, no outlandish, unrealistic poses that may put the patient at risk.

**Provide at least one image of the joint from which the joint space can be accurately, precisely, reliably, and repeatedly measured.**

This methodology should at the very least obtain one specific view of the knee that can be reliably obtained in the most accurate and precise way. as possible. A single view may not even constitute both the femoral and tibial articular cartilage, in fact, as the anatomical feature of interest changes with changing orientation of the probe on the knee. The degree to which the joint space may be deemed accurate, or to the extent at which progress can be detected, is ultimately dependent on either a) obtained literature values or b) self comparison from the beginning, or no method whatsoever.
Track the location and orientation of the probe during imaging and, through post-processing, create a stitched visualization or reconstruction of the imaged region.

To properly reconstruct an image of the knee joint, it will be necessary to determine where exactly the probe lies in space, particularly with respect to anatomical features of the knee. One method that may be considered is to track the orientation of the probe in real time so that B-scans and their corresponding angles can be passed into a reconstruction algorithm.

3.6 Project Approach

To achieve the project goal and determined objectives, a timeline was developed. These steps can be broadly categorized as follows: determine morphological structures of interest, determine anatomical pose, determine transducer orientation, design a mechanical fixture to stabilize and guide the transducer probe, verify design with a non-human subject, and verify design with a human subject.

Determine morphological structures of interest (Chapter 5.2.1)
The first step in our project was to determine the morphological structures of interest that could be imaged using US. This was achieved by experimenting with various system settings, probe positions, and knee positions. We identified several regions that have been shown to be clinically significant in the diagnosis of OA and were easily located using three major anatomical landmarks: intercondylar notch, medial condyle, and lateral condyle.

Determine ideal anatomical and probe poses (Chapter 5.2.1)
Once we determined the morphological structures of interest, we sought out the ideal anatomical and probe poses to image these structures. For this, we experimented with various knee angles ranging from 0° full extension to 150° flexion. We also varied probe placements to image each of the three regions, as described in Project Approach Part 1. Through this, we determined that a knee flexion of 60° or greater provided the clearest view of the underlying anatomy. As for the orientation and placement of the probe, we found that orienting the flat side of the probe to be perpendicular to the surface of the skin to be the most reliable method to image clearly.
Evaluate reliability of ultrasound in imaging morphological structures of interest (Chapter 5.2.2)
Using the ideal anatomical and probe poses determined, we evaluated the reliability of US to provide measurements of femoral articular cartilage thickness. We imaged and measured femoral articular cartilage thickness at the intercondylar notch, and lateral and medial condyles using the ultrasound system on three human subjects with the same knee. We used this study as an investigation into the reliability of the current method used to image knee joint morphology and as an indicator for the need for an improved method.

Experiment with acoustic standoff pad to increase image quality and ease of use of system (Chapter 5.2.3)
After the reliability study, we tested the reliability of a commercial ASP as it was found to increase the ease with which irregular surfaces (i.e. bony regions around the patella) could be maneuvered. We simultaneously pursued and will continue pursuing the creation of our own ASP using materials such as gelatin and agar. If successful, this would allow us to create ASPs of similar if not improved characteristics specific to our application at a fraction of the cost of its commercial counterparts.

Determine ideal method to perform reconstruction (Chapter 4.6)
Given our goal of reliably imaging features of OA, we arrived at the conclusion that 3D reconstruction was a requirement to gain a complete, intuitive visualization of the morphological structures being imaged. Of the methods currently employed to reconstruct 2D B-scans into a 3D volume or surface, we determined that the freehand US with probe localization method to be the most ideal option.

Implement probe localization and reconstruction (Chapters 4.4 and 4.8)
In order to complete reconstruction with the freehand probe localization method, we will need to synchronize readings from an IMU with the images acquired through the US system. This will allow us to orient each frame in a representative volume and reconstruct the imaged structure. To allow the end user to obtain measurements from the reconstructed volume, the software implemented must provide the ability to measure the distance between any two points given the spatial resolution of the original B-scans.

Component and Design Testing (Chapter 5.3, 5.4, and 7.2)
During the design process, individual components will be tested to verify they are functioning as expected within the design. The two key components of the design include the IMU and reconstruction algorithm. The IMU will be tested by evaluating its ability to track predetermined paths of known angular displacements. The reconstruction algorithm will first be tested using artificially created data to simulate B-scans. Once completed, the system will be coupled with the IMU by imaging geometrically simple objects of known dimension, referred to as phantoms. These tests will be used to advise any necessary changes to the existing design.

Verify design with human test subject (Chapter 7.3)

Once the design has been tested, it will be used to image a human subject’s knee. The reconstructed volume will be compared to an existing MRI scan of the subject segmented in Mimics. This will allow us to evaluate the ability of the design to image and subsequently provide quantitative measurements of the structures of interest.

3.7 Project Timeline

A-term consisted of a thorough literature review into the current clinical practices of OA diagnosis and existing designs to improve them. A-term accomplishments correspond to Chapters 1, 2, and 3 of the report. B-term was mainly focused on imaging studies, where we began to integrate the IMU with the final design and the development of a reconstruction algorithm. Chapter 1, 2, and 3 were revised, Chapter 4 and Chapter 5 were partially completed. C-term was focused on the development of an imaging protocol, creation of a fixture, and some initial testing of individual components and software. Chapters 5 and 6 were partially completed and 7 and 8 were outlined. D-term consisted of more formal testing with non-human test subjects and a human test subject. D-term was mostly dedicated to finalizing the report and preparing a presentation on the project. Chapter 8 was completed and the report was reviewed and finalized for submission.
4 Component Selection & Design Alternatives

4.1 Needs Analysis

The needs of the final design can be separated into the following categories in order of highest to lower priority: imaging (primary), usability (secondary), accessibility (secondary), compliance (secondary), manufacturability (tertiary), serviceability (tertiary), and environment (tertiary). These categories were ordered based on the perspective of the user - those thought to be more important were given higher precedence than the others. The intended user for the device will likely be experienced sonographers or academic researchers who are familiar with US. With respect to the patient, the device must be most importantly safe to use, provide accurate readings for the clinician to interpret, and accommodate a range of patients. For the clinician, the device must be easy to handle, simple to interpret, and provide accurate readings.

The system must first obtain images of sufficient quality such that the desired measurements and features can be extracted with accuracy, precision, and repeatability. These features must be prioritized without significantly compromising usability, accessibility, and compliance. Factors such as ease of use; timeliness; portability; and spatial, electronic, and user-related needs will serve as indicators for success. Ideally, these will be met while satisfying financial, manufacturing, and environmental constraints. Of the listed categories, only the regulatory requirements will be dictated by external bodies. Finance, manufacturing, and environment-related needs of the final design will be determined and justified by those used for similar designs.

4.2 Functional Requirements

A set of overarching functions of the product were determined to guide the definition of specific, functional requirements. Based on the functions, a set of functional requirements was defined. These requirements outlined criteria that individual components, sub-assemblies, and the final product should satisfy. Each specification is accompanied by its justification:

Joint gap width measurements of knee at any knee flexion angle between fully flexed and fully extended.

The joint space width may be different for a given individual at different flexion angles, as there may be uneven degradation of the femoral articular cartilage. There
may be a particular angle, such as 90° flexion, that provides the best image or the joint width that is most predictive of a diseased state. However, for any given individual there should be the opportunity to image the anatomy at several knee angles. For this, the team agreed that a device capable of adjusting between positions would be best, as it could be adjusted across patients and for each patient. Preliminary designs for this approach may not be able to achieve a continuous data acquisition, and as such, should be left for later iterations. Discrete, or ‘mode-type’ solutions will likely be a good starting point to determine baseline reliability.

Joint gap width measurements of knee in both standing and seated positions.

An important component of OA is limited joint motion as well as bone-bone contact when there is a load on the joint. Current imaging methodologies, including X-ray and MRI, can only provide images of an unloaded joint. One area where US can provide more versatility is imaging of knees under loaded conditions. Thus, the device should be able to provide images of the knee such that the areas of interest are exposed in a position osteoarthritic adults would be capable of assuming. The design will incorporate changes in physiology with changing knee position.

Accommodate knees with circumference between 31.0 cm and 41.2 cm.

In order for the device to work on most individuals, it must be able to perform all of its functions on knees between the 5th percentile of circumference for women and the 95th percentile for men [National Aeronautics and Space Administration, 2000]. As many individuals with OA experience inflammation, it may be advantageous to extend the maximum circumference up to 50 cm. Adjustability may be attained by incorporating a movable part within a design, if necessary, to allow for a range of knee sizes.

Repeatable joint space width measurements.

While there is natural variability of knee joint gap width for healthy individuals of the same size, joint gap width rarely varies more than 0.35 mm between healthy individuals [Adam et al., 1998]. The device must be able to detect differences in joint space gap width greater than 0.35 mm in order to detect the disease state, so the variation between measurements of the same healthy individual must not exceed this value. Known parameters which may affect consistency include applied pressure, orientation and movement of the probe. With this in mind, a design capable
of detecting pressure involved at the skin-probe interface may be useful, as well as a design able to monitor real-time location and orientation.

Accurate joint space width measurements
The device must provide measurements of the femoral articular cartilage thickness that reflect values found in literature. The ranges for medial and lateral femoral articular cartilage thickness found in literature are 1.87 ± 0.37 mm and 1.94 ± 0.46 mm, respectively. Table 11 is also found in Chapter 2 but is included below for convenience.

Table 11: Literature values for joint space width, femoral articular cartilage, and tibial articular cartilage

<table>
<thead>
<tr>
<th>Physiology of Interest</th>
<th>Beattie et al.</th>
<th>Adam et al.</th>
<th>Tuna, Balci, &amp; Ozacazar</th>
<th>Sheperd &amp; Seed-hom</th>
</tr>
</thead>
<tbody>
<tr>
<td>Joint space width (mm)</td>
<td>5.25</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Medial femoral articular cartilage thickness (mm)</td>
<td>N/A</td>
<td>1.86 ± 0.36</td>
<td>1.925 ± 0.306</td>
<td>1.65 - 2.65</td>
</tr>
<tr>
<td>Lateral femoral articular cartilage thickness (mm)</td>
<td>N/A</td>
<td>1.94 ± 0.46</td>
<td>1.924 ± 0.398</td>
<td></td>
</tr>
<tr>
<td>Medial TAC thickness (mm)</td>
<td>N/A</td>
<td>1.75 ± 0.19</td>
<td>N/A</td>
<td>2.07 - 2.98</td>
</tr>
<tr>
<td>Lateral TAC thickness (mm)</td>
<td>N/A</td>
<td>2.08 ± 0.43</td>
<td>N/A</td>
<td></td>
</tr>
</tbody>
</table>

Fix knee at an angle between 30° and 180°
To obtain both repeatable and accurate joint space width measurements, the knee joint must be kept stable throughout the procedure. Due to individual anatomical differences and an interest in varying degrees of knee flexion and extension, the design will allow the knee to be fixed at any angle between 30° and 180°. As the lower leg and the knee can be quite irregular, a ± 10° tolerance will be given. For static poses, especially for preliminary testing, a goniometer may be used.

Image acquisition rate of at least 14 Hz
A sufficient image acquisition rate is necessary to accurately reconstruct the region of interest given the speed with which the probe will be maneuvered across the body surface. A similar system had a minimum acquisition rate of 14 Hz imaging phantoms and in vivo structures and was able to obtain visualizations of sufficient resolution (Herickhoff et al., 2018). Though higher image acquisition rates may be used, their effects on the computational requirements of the system will have to be evaluated to obtain a balance between resolution and efficiency.
**Total acquisition time below 60 seconds**

One of the primary advantages of US is its rapid image acquisition ability. A similar adaptation of a 2D US system for 3D US has been able to acquire images of phantoms and *in vivo* structures in roughly 30 seconds or less (Herickhoff et al., 2018). Therefore, the design, so as not to compromise the time efficiency of US, must have a maximum acquisition time below 60 seconds per medial or lateral side of the knee.

**Complete offline image reconstruction**

Though individual B-scans may provide information on joint space width, they are limited in the field-of-view in which measurements can be taken. Since the location of features such as minimum joint space width differ on an individual basis, reconstruction based on individual B-scans and pose information is necessary to reliably obtain such critical measurements. Real-time reconstruction is the ultimate goal; however, this is outside of the scope of this project. Offline reconstruction is sufficient as the goal of this project is to develop and validate a method that measures joint space width.

**Allow measurement of distance between any two points in domain**

Given a reconstructed image, the software must be able to calculate the distance between any two points chosen in the image with an accuracy of ±0.1 mm. Typical values of femoral articular cartilage thickness range from 0 to 6 mm; therefore, a 0.1 mm tolerance would provide cartilage thickness measurements with sufficient accuracy (Favre, Scanlan, Erhart-Hledik, Blazek, & Andriacchi, 2013).

**Used by unskilled sonographers**

The device is intended to be used by both skilled and unskilled sonographers given that US is one of the more accessible medical imaging modalities (Sippel et al., 2011). Given instructions on the use of the system, any user should be able to operate the system.

**Cost to produce less than $750**

Given that the user has additive manufacturing capabilities, the cost to produce and purchase all the required components should be less than $750. This does not include the cost of 3D printers or any part of an US machine.

**Complete spatial and temporal calibration in less than 30 seconds**
IMU’s are known to require frequent calibrations as a slight offset can lead to the gradual accumulation of error in position and orientation. To minimize this effect, the IMU must be easily and rapidly calibrated prior to each use.

**Time to set up the fixture in less than 3 minutes**
The time required to set up the device shall remain with a time constraint of 3 minutes. This is to increase both convenience and overall usability of the device.

**Complies to relevant regulations**
The device must comply with all relevant FDA regulations and *IEC 60601*, if reliant on an input power source, to maintain the safety and well being of all who may interact with the device. This aligns with all other devices intended for medical use.

**Materials, procedure, and structure do not cause patient discomfort**
The device should not only be usable on most patients but should provide the maximum comfort to the patient. It should not be too tight or loose, may need to include padding, and should not use materials that may trigger an allergic reaction. The device and procedure should also not force the patient into an uncomfortable or unnatural position.

**Image acquisition and positioning system can be powered with laptop/desktop (excluding US)**
The system’s image acquisition and positioning system should only require a laptop or desktop for power. This is to increase portability of design, as well as maximize accessibility for all users.

**Built with off-the-shelf and 3D-printable components**
A large contributor to the manufacturability of the final design will be the use of only off-the-shelf and 3D-printed components. This greatly reduces the cost of and variability between individual assemblies of the final design. Furthermore, it allows the end user to easily replace broken components or accommodate a modification. The following design will be generated around the parts currently under possession, but the base design idea may be translated across machinery of different dimensions.

**Accommodate both knees**
The system will accommodate both knees in order to maximize usability and ac-
cessibility. By extending functionality to both knees, portability is also increased, while minimizing overall cost and time. It benefits both the user and clinician, who are first-line direct users, as well as more indirect players, like manufacturing and marketing.

**Withstand specified environmental conditions such as temperature and humidity**
The system will accommodate a range of temperatures and humidity, for both shipping and operation. For a portable, accessible system intended for the international market, it should withstand a larger range of working and shipping conditions than most.

### 4.3 Component Selection & Experimentation

Several decision matrices were developed to quantitatively evaluate various components in the design. Each matrix had a basic set of criteria upon which potential candidates were evaluated: price and ease of use. In addition, component-specific parameters were added per the discretion of the team to further inform the decision-making process. Each of these parameters were given a specific weight, which was then multiplied by the assigned score in the decision matrix. The total for each candidate was compared and was used to advise component selection.

#### 4.3.1 3D Ultrasound Method

The team considered various methods of 3D US as explained in Chapter 2. 2D array scanning was not considered as the US system in our possession was not capable of it. A mechanical scanning system, freehand with positioning system, and freehand without positioning system was evaluated based on five parameters: ease of use, ease of implementation, image quality and utility, portability, and cost, shown in Table 12.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Weight</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ease of Use</td>
<td>1</td>
</tr>
<tr>
<td>Image Quality and Utility</td>
<td>3</td>
</tr>
<tr>
<td>Portability</td>
<td>2</td>
</tr>
<tr>
<td>Cost</td>
<td>2</td>
</tr>
</tbody>
</table>

Table 12: 3D US method decision matrix weighting scheme

The parameters were scored on a scale from least (1) to most (3) important.
Image quality and utility was given the highest weight, as the team believed it to be most important to be able to acquire readable and usable images. Next, portability and cost were assigned equal weights of 2, as the former affects overall convenience in use, and the latter drives the overall function of the design. Ease of use was assigned the lightest weight of 1, as it was not as essential for data processing or creation. The description and the scoring for each parameter are as follows:

Ease of Use: Level of knowledge regarding medical ultrasound and human anatomy required by the user to successfully prepare, configure, calibrate, and operate the system.

1. Requires extensive knowledge of medical ultrasound and human anatomy to use system. User has at least 2 years of working experience using an ultrasound with live patients and comfortable knowledge of anatomy

2. Requires moderate knowledge of medical ultrasound and human anatomy to use system User has at least 1 year of working experience using an ultrasound with live patients and familiarity with anatomy.

3. Requires minimal to no knowledge of medical ultrasound or human anatomy to use system. User has less than 1 year of working experience using an ultrasound with live patients and little knowledge of anatomy.

Image Quality and Utility: Quality and utility of B-scans and 3D reconstruction produced by the system given that it is used correctly. Image quality refers to the resolution of B-scans and corresponding reconstruction. Image utility refers to the usefulness of the B-scans and corresponding reconstruction provided and serves as a reflection of the ease with which the system can be used to image structures and regions of interest.

1. System unable to provide neither high image quality nor utility

2. System able to provide high image quality but not image utility or vice versa

3. System able to provide both high image quality and image utility

Portability: Ease with which the system can be disassembled and moved to another location, excluding US system.

1. Requires multiple people to transport the system
2. Can be transported by one able-bodied person but may require multiple trips

3. Easily transported by one able-bodied person and can be carried by hand in few trips

Cost: Projected total cost of any hardware or software that needs to be purchased
to assemble the system (excluding the cost of an US system and any necessary
accessories).

1. > $750

2. ≥ $250 and ≤ $750

3. < $250

Based on the criteria specified above, the decision matrix was used to calculate
a score for each of the 3D US methods (Table 13).

Table 13: 3D ultrasound decision matrix

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Weight</th>
<th>Mechanical</th>
<th>Freehand with Positioning</th>
<th>Freehand without Positioning</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ease of Use</td>
<td>1</td>
<td>3</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Image Quality and Utility</td>
<td>3</td>
<td>3</td>
<td>3</td>
<td>2</td>
</tr>
<tr>
<td>Portability</td>
<td>2</td>
<td>2</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Cost</td>
<td>2</td>
<td>2</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Total</td>
<td>18</td>
<td>23</td>
<td>20</td>
<td></td>
</tr>
</tbody>
</table>

Based on the evaluation of the four parameters, the freehand method with position-
ing was determined to be the most appropriate 3D US method for the project.
Upon discussion, the team agreed that this was a reasonable choice given that a
mechanical system would limit the freedom of the sonographer to image regions of
interest compared to a freehand system. Additionally, a freehand without position-
ing system was deemed not as reliable as a freehand system with positioning as the
former depended on the correlation of successive B-scans and the latter on probe
localization measurements to identify the orientation of the B-scan during recon-
struction.

4.4 Probe Localization Method

After deciding to design a freehand US system with position sensing, the team
considered various probe localization devices that could identify the position and
orientation of the probe during imaging. They were evaluated according to four parameters: accuracy, ease of use and setup, portability, and price (Table 14).

Table 14: Probe localization method decision matrix weighting scheme

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Weight</th>
</tr>
</thead>
<tbody>
<tr>
<td>Accuracy</td>
<td>2</td>
</tr>
<tr>
<td>Ease of Use &amp; Setup</td>
<td>2</td>
</tr>
<tr>
<td>Portability</td>
<td>1</td>
</tr>
<tr>
<td>Price</td>
<td>1</td>
</tr>
</tbody>
</table>

The parameters were scored on a scale from 1 (worst) to 3 (best). The parameters chosen for probe localization were all assigned a weight of 1, as the team believed each of the parameters to share almost equal importance. The description and the scoring for each parameter is as follows:

**Accuracy:** Spatiotemporal resolution of the localization device. Also, the device must be able to measure the angular displacement in addition to translational displacement of the probe.

1. Spatial resolution of cm scale, angular resolution of single-degree scale, and temporal resolution of second scale
2. Spatial resolution of mm scale, angular resolution of single-degree scale, and temporal resolution of millisecond scale
3. Spatial resolution of sub-mm scale, angular resolution of sub-degree scale, and temporal resolution of millisecond scale

**Ease of Use and Setup:** Ease with which user can prepare and use the device. This includes calibration and data acquisition.

1. Substantial preparation and calibration are necessary to configure system. Device-specific software is needed to acquire data.
2. Requires moderate preparation and calibration process to set up system. Multiple components may need to be configured, synchronized and/or calibrated. Device-specific software may be required to acquire data.
3. System can be set up easily and requires only a one-step calibration procedure. Data can be acquired using a personal device.
Portability: Ease with which the device can be disassembled (if necessary) and moved to another location

1. Assembling and disassembling is involved and requires care. Transportation of device in one trip by an individual is challenging.

2. Assembling and disassembling is simple but may take some time. Though possible to transport device by hand, the size and/or weight of the device may provide some challenges.

3. Can be easily assembled/disassembled by one person and transported by hand with moderate ease.

Price: Price per unit (or set) of component and any necessary accessories

1. > $100

2. ≥ $25 and ≤ $100

3. < $25

The components compared were inertial measurement units, optical positioning systems, and electromagnetic positioning systems. The components evaluated for each type of localization device were as follows:

IMU: GY-521 MPU-6050

OPS: Three-integrated cameras

EPS: Polhemus electromagnetic tracking system

Based on the decision matrix, the IMU was determined to be the ideal localization device for the design the (Table 15). The team agreed with this evaluation as the IMU provided the necessary spatiotemporal resolution and was far superior compared to an optical or electromagnetic positioning system in ease of use, portability, and cost. Given the focus on accessibility of the final design, the spatial and financial advantages associated with the IMU justified this decision.

Table 15: Probe localization decision matrix

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Weight</th>
<th>IMU</th>
<th>OPS</th>
<th>EPS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Accuracy</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Ease of Use</td>
<td>1</td>
<td>3</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Portability</td>
<td>1</td>
<td>3</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Price</td>
<td>1</td>
<td>3</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>11</strong></td>
<td><strong>8</strong></td>
<td><strong>8</strong></td>
<td><strong>55</strong></td>
</tr>
</tbody>
</table>
The MPU-6050 obtains data on six DOF’s (three-axis gyroscope and three-axis accelerometer) and temperature. The unit includes a digital motion processor (DMP) that is capable of calculating its position and orientation in real-time, thereby offloading a large computational requirement from the system processor (InvenSense, 2013). The sampling rate of the MPU-6050 is shown in Table 16.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Sample Rate (kHz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gyroscope (Slow)</td>
<td>1</td>
</tr>
<tr>
<td>Gyroscope (Fast)</td>
<td>8</td>
</tr>
<tr>
<td>Accelerometer</td>
<td>1</td>
</tr>
</tbody>
</table>

### 4.5 Coupling Medium

One accessory currently in use by sonographers to improve image quality is the acoustic standoff pad. The team decided to evaluate the effectiveness of the commercial ASP (Aquaflex), Agar 16 g/L, Gelatin 40 g/L, as compared to none. These modes were compared in terms of 4 parameters: ease of use, lifetime, flexibility, and image quality, shown in Table 17. All materials chosen for testing were biologically inert and were found to be biocompatible.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Weight</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ease of Use</td>
<td>2</td>
</tr>
<tr>
<td>Lifetime</td>
<td>2</td>
</tr>
<tr>
<td>Flexibility</td>
<td>2</td>
</tr>
<tr>
<td>Durability</td>
<td>3</td>
</tr>
<tr>
<td>Image Quality</td>
<td>3</td>
</tr>
</tbody>
</table>

The parameters were scored on a scale from 1 (worst) to 3 (best). The parameters chosen for were all assigned a weight of 1. The description and the scoring for each parameter is as follows:

**Ease of Use**: simplicity of use while acquiring US images. Usage includes the application of US gel and moving the probe over the surface of the material.

1. Difficult to use while using the probe to image the knee. Slides out of place if not held constantly with one hand.
2. Somewhat difficult to use. Requires smooth movement of the probe to prevent slippage.

3. Very easy to use. Sticks to skin well enough that no hand is needed to secure the material in place while scanning.

**Durability:** Number of sessions the ASP can be used to image the knee without a noticeable decrease in image quality, flexibility, or breakage

1. < 5 separate scan sessions
2. ≥ 5 and ≤ 15 scan sessions
3. > 15 scan sessions

**Flexibility:** Angle to which the material is capable of bending without failure

1. Not at all flexible, < 60 degrees of bending
2. Relatively flexible, ≥ 60 and ≤ 135 degrees of bending
3. Very flexible, > 135 degrees of bending

**Image Quality:** Degree to which the material is capable of improving the image quality of the US scan in that more of the knee may be scanned. The bony structure of the knee can be difficult to image with a stiff, flat linear probe.

1. Detracts from quality of the image
2. Does not improve quality of the image much at all
3. Improves the quality of the image

**Acoustic Standoff Pad Experimentation** The ASP purchased was an antibacterial sonolucent bacteriostatic disposable US pad. These pads were easily found and purchased through Amazon in a box of 7 (Figure 16).
Overall this pad was durable, easy to use, and lasted much longer than that of the homemade agar-gelatin pad. The pad itself was encased in a plastic holder, and was immersed in a thin oily fluid, likely to prevent the pad from drying out. This pad proved to provide decent images for visualizing the knee space and was much better than no coupling medium at all. Overall, the ASP helped identify the surface of the anatomical region of interest, and improved ease of image acquisition. While it did not improve the image quality significantly, it did make imaging objects of smaller radii easier. The probe itself when imaging over curved surfaces could push into the pad and acquire images with faster speed. A comparison of the quality of images obtained with and without an ASP are shown in Figure 17.
Figure 17: Comparison of B-scans without ASP and with ASP. These images were of the lateral side of the knee and were used to obtain data during later testing of sample repeatability. A) Scan with no ASP. Surface of the knee is not identified, and a much shallower depth was used to acquire the image. B) Scan with ASP increased ease of image acquisition. Surface of the skin is shown, and the added 2 cm ASP was account for in increased depth set for acquisition.

**Agar and Gelatin Experimentation** Common materials used in image modalities must consider the mechanics of US, which in brief, relies on sound wave propagation through biological tissue to create an image. One material typically used to create phantoms for US purposes is agar – it is particularly useful as its acoustic properties are very similar to that of biological tissue ([Khera & Keshava, 2014](#)). Another, perhaps more commercially accessible material also commonly used, is gelatin. Homemade phantoms can be made quite easily, with just mixing the powder, boiling in water, and setting in the mold of interest. Agar and gelatin were explored as homemade alternatives to the commercially available acoustic standoff pad, Aquaflex, to test feasibility of use in later designs (Figure 18).
After initial testing with the commercial standoff pad, pure agar, and pure gelatin, the commercial pad and the agar were the most sonolucent of the three. Mechanical differences changed the quality and usefulness of the pad; for example, agar was found to be stiffer than the Aquaflex, while gelatin was noticeably softer. This difference in flexibility and strength affected the overall usability of the acoustic pad. Gelatin, even at a high concentration of 40 g/L, was too soft and could not be used to image the knee. It also had an unpleasant smell, particularly during fabrication. Agar, on the other hand, was nearly identical to the Aquaflex, though was still somewhat stiff. Higher concentrations of agar reported similar breakage at high enough probe pressures. Even though their mechanical properties contrasted, with one being stiff while the other was too soft, both agar and gelatin were quite promising materials. They were both easy to work with and set relatively quickly – no more than 6 hours per pad. While the agar exhibited good sonolucent properties, its mechanical properties needed to be altered. For this, mixtures of different concentrations of these two materials were explored: one with the ideal concentrations of both, and one with a total solute concentration of 20 g/L.

While the exploration with different sonolucent materials was fruitful, it was difficult to find an ideal concentration of either agar or gelatin without resulting in breakage. Furthermore, even the best concentrations of agar-gelatin from an imaging and mechanical properties standpoint quickly led to bacterial growth. This not only introduced a hazard, it also compromised the mechanical integrity of the material itself. From general use, the agar-gelatin samples lasted approximately two weeks. Incorporating a material that lasts only a couple weeks at best would become
costly over time and may be unfeasible in some parts that cannot readily access the necessary materials.

**Water Experimentation** Water was presented as a potential coupling medium given its favorable acoustic properties for US imaging. Biological tissue is primarily comprised of water, so it followed that water may provide the appropriate sonolucent properties as an acoustic standoff pad without the biological hazard the agar-gelatin mixture carried. Water, in particular, saline, has demonstrated to be an effective coupling medium for US (Bottenus et al., 2016; Shigemura et al., 2017). Initial testing of water as a coupling medium was performed with wooden dowels in a Pyrex glass (Figure 19).

![Figure 19: Imaging a wooden dowel submerged in a Pyrex container filled with water. A) A wooden dowel was submerged underwater and a transducer was similarly submerged roughly an inch above the dowel. B) B-scan of submerged wooden dowel. The outline of the wooden dowel can be easily identified.](image)

Water, as opposed to agar or gelatin, did not suffer from bacterial growth, was easily accessible, and inexpensive. Initial challenges of working with the water were the incredible noise and overall amplification of the speed of sound through water as compared to what had been observed without the ASP, agar, or gelatin. Water was not as relied upon as a coupling material for US, and few products seemed to be on the market that relied upon water. To amend this, the limitations and benefits of water needed to be explored further in an effort to improve US image quality.

Once water was confirmed as a viable coupling medium, a variety of experiments were performed to observe the water’s effect on the US scans. Initial testing was
performed with water in a Pyrex container, and, later, Ziploc bags filled with water. After plastic bags were used to encase the water, the rippling observed in earlier tests no longer became a problem. This is thought to be a result of visible water currents that resulted from moving the probe too quickly in the glass Pyrex with water obtained by a nearby, and likely aerated, water fountain.

In the first few imaging scans, there were many bubbles and ripples throughout the water during scanning. To determine the causes for this, the team prepared a variety of water samples to image. Namely, tap water and distilled water were compared in two tests: newly obtained water and water left to sit out on the counter. There was less of a difference between tap water and distilled water and found that water with bubbles proved to be noisiest.

To address the noisy images, a variety of methods were tested to minimize bubbles. The first involved physically tapping the container of water to remove bubbles, while the second used water that had been sitting out for a while. “Older” water proved to have the best image quality, as there were fewer bubbles and noise. A third, more vigorous method used a degassing chamber to remove as much air as possible. This water was found to have a similar quality to that of water left on the counter to degas itself (Figure 20).

![Figure 20: Methods tested to degas water. A) Bagged tap and distilled water and exposed still water. B) Vacuum chamber used to degas tap and distilled water in Pyrex glass.](image)

When testing with Ziploc bags, the wrinkling of the plastic bag, particularly around regions with small radii of curvature, posed a potentially significant design challenge. In addition, Ziploc bags were both prone to breakage at the edges, closures, and corners. Table 18 shows the result of the decision matrix for the various coupling media.
Table 18: Coupling medium decision matrix

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Weight</th>
<th>ASP*</th>
<th>Agar (16g/L)</th>
<th>Gelatin (40g/L)</th>
<th>Ziploc</th>
<th>Custom Bag</th>
<th>Saline Bag</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ease of Use</td>
<td>2</td>
<td>2</td>
<td>2</td>
<td>1</td>
<td>2</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Durability</td>
<td>2</td>
<td>2</td>
<td>2</td>
<td>1</td>
<td>2</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td>Flexibility</td>
<td>2</td>
<td>2</td>
<td>1</td>
<td>3</td>
<td>2</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td>Image Quality</td>
<td>3</td>
<td>3</td>
<td>3</td>
<td>1</td>
<td>1</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>21</strong></td>
<td><strong>19</strong></td>
<td><strong>13</strong></td>
<td><strong>13</strong></td>
<td><strong>18</strong></td>
<td><strong>25</strong></td>
<td></td>
</tr>
</tbody>
</table>

*Commercially purchased ASP: Aquaflex

4.6 Conceptual Designs

Based on the previously selected 3D US method, probe localization method, and coupling medium, several conceptual designs to assemble them together were considered. These designs were broadly categorized as the following: sliding water cup design, cast protector design, and curvilinear rail design. Each design was evaluated subjectively for its advantages and disadvantages, and a decision matrix was used to choose the design moving forward.

4.6.1 Sliding Water Cup Design

An initial idea to contain the water was the use of a barrel surrounding the leg. Initial design ideas considered placing a patient in a larger container full of water to allow for imaging the knee. These ideas included placing the patient face-down on a massage chair that submerged only the knee, placing the patient’s leg into a bucket of water, to even placing the patient into a bath of water. These ideas, however, were disregarded as they required too much water for such a small region of interest.

The team then went on to consider a design that contained water in a closed region. The probe was based on a prototype created by another group would lock into a gimbal that allowed for rotation in potentially three axes of interest (Morgan et al., 2018). A CAD model and simple mock-up for this ‘encapsulated water’ idea is shown below in Figure 21. First, the device would Velcro to the patient’s knee, the gimbal barrel would be filled with the coupling medium of choice, and the probe would lock into the gimbal. For this particular design, the probe would rotate around two axes of interest, and would be ‘lockable’ at the position of interest. Disadvantages of this design include the risk of spilling water on and around the patient, not being able to control the height of the probe within the gimbal, and only being able to measure the knee at discrete locations.
Figure 21: Simple gimbal design to hold water and rotation around two axes of interest. A) Cross-sectional view of the gimbal that allowed for rotation around two axes of interest. B) Image of how gimbal would fit on knee. Rubber coated around edges would form a watertight seal and would secure around a patient’s leg.

4.6.2 Cast Protector Design

Another design that was considered was the adaptation of watertight cast protectors to contain water (Figure 22). Originally designed to keep water outside of a region, the intention was to submerge the entire leg or just the knee in water or US gel. A watertight window could be made through which the probe could image the region of interest.

Figure 22: Cast protector that could be adapted to submerge the knee in water and provide a window through which the probe could image the knee (VBESTLIFE, 2019).

Obvious disadvantages of this design include the need for a large volume of coupling media, risk of spillage, and lack of freedom and comfort for the patient. Though
submerging the knee in water allows the imaging of the knee at any angle, the potential hazards and inefficiencies of the design were thought to outweigh its benefits.

4.6.3 Curvilinear Rail Design

The curvilinear rail design proposed the mounting of an US probe on a fixed track or rail fixed along the anterior side of the knee (Figure 23). Unlike the previous designs, the fixture itself would not contain the coupling medium, therefore this design would require either the direct application of a coupling medium, such as US gel, or another design capable of retaining a coupling medium.

Figure 23: Initial concept schematic of curvilinear rail design. The hinge (orange) would allow the rotation and locking of the rail at discrete (or continuous) positions. The rail (black) traces out the path that the probe would follow (Busti, 2015).

The hinge, shown in orange, would allow the rotation of the rail to discrete angles and allow the probe to trace a path along the anterior side of the knee. Depending on the position of the rail, such a design would allow the probe to traverse along various regions of the FAC. Because the hinge (when locked) and the rail constrain five degrees of freedom, the probe would only be able to travel along the path of the rail. Though this limits the imaging window, it allows the calculation of the orientation as well as the position of the probe relative to the knee with a single angular measurement – this is because a given tilt in the probe can only be achieved
at a single location along the rail.

### 4.6.4 Conceptual Design Decision Matrix

A decision matrix was composed for the three conceptual designs proposed above. The matrix evaluated the ease of use, safety, image quality. The parameters were scored on a scale from 1 (worst) to 3 (best). The criteria for each parameter is defined as follows:

#### Ease of Use

- **1. Requires engagement from the patient or requires at least one sonographer and another person to aid in one of the steps.**
- **2. May require some participation on the part of the patient. Can be operated by one sonographer, though some steps may pose slight difficulty.**
- **3. Requires little to no engagement from the patient. Requires one sonographer to easily operate the system without difficulty.**

#### Safety

- **1. The device poses some risk to the patient and sonographer. Sharp edges in design and moving parts may result on minor scratches and little skin breakage.**
- **2. The device poses little risk to the patient and sonographer. Design includes pinch points but is unlikely to result in skin breakage.**
- **3. The device poses almost no risk to the patient and sonographer. There is very little chance of scraping, and any harm to the patient would be slight discomfort.**
Image Quality: Degree to which the fixture is capable of providing B-scans at the desired locations with the necessary feedback to reconstruct the underlying anatomy accurately. In addition, the field of view capable with the design must be considered.

1. It is difficult to view the region of interest in the reconstructed image, and the two lines indicating the femur and tibial plateau are barely shown. The measured joint space exceeds the literature range of femoral cartilage thickness by 2 mm.

2. There is some noise in the reconstructed image and some of the regions of interest are cut out of frame. Some of the measurements fall within the defined literature range for femoral cartilage thickness.

3. There is very little noise in reconstructed image acquired and both the lateral, medial, and center points of the articular cartilage can be captured. The image accuracy falls within the values as described in the literature for femoral cartilage thickness.

Cost: Total cost of the materials chosen to construct the design.

1. > $750
2. $100 and ≤ $750
3. < $100

Based on the decision matrix (Table 19), the curvilinear rail was chosen to be the base design moving forward.

Table 19: Conceptual design decision matrix

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Weight</th>
<th>Sliding Water Cup</th>
<th>Cast Protector</th>
<th>Curvilinear Rail</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ease of Use</td>
<td>3</td>
<td>2</td>
<td>1</td>
<td>3</td>
</tr>
<tr>
<td>Safety</td>
<td>2</td>
<td>2</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td>Image Quality</td>
<td>3</td>
<td>2</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td>Cost</td>
<td>1</td>
<td>3</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>19</strong></td>
<td><strong>14</strong></td>
<td><strong>26</strong></td>
<td></td>
</tr>
</tbody>
</table>

4.7 Curvilinear Rail Design

The first concept for the design combined the rail with clamps at the ends of the rail that would compress both sides of the knee (Figure 24). This design was aimed
at being configurable to a variety of knee widths; however, its obvious limitations involved the instability of the rail. If the probe were to be mounted and the rail were rotated such that it was parallel with the ground, the resulting moment was thought to destabilize the rail. Furthermore, the compressive forces needed to supply the static friction to maintain the position of the rail was predicted to be a source of discomfort for patients, particularly with patients suffering from knee OA onset inflammation.

![Screw clamp curvilinear rail design](Wilkerson, 2016).

A preliminary CAD model of the rail and car assembly was created to better visualize the dimensions of the design and how best to interface it with the knee (Figure 25). The diameter of the rail was chosen based on the typical sizing of knee braces for a given knee width and anthropomorphic data on knee circumference. Current knee braces and knee sleeves intended for orthopedic applications were referenced for sizing. Leg circumferences here for XL sleeves were found to be approximately 457 and 402 mm, with a knee width of about 127 mm.
The design consisted of a car with three adjustable shoulder V-rollers that mounted onto an angle rail (Figure 26). The probe would be secured between two mirrored components that could be tightened by Velcro and mounted onto the car using a prismatic joint. The main concern with the initial design was the distance that the mounted part extended past the center of the rail: this would result in a large torque prior to any application of external forces and was cause for concern of derailment or fracture of the rail.

Further iterations of the design focused on reducing the profile of the design in order to reduce the torque applied to the rail from the weight of the probe. To aid in this process, a CAD rendering of the probe was created using caliper measurements and visual inspection (Figure 27).
To reduce the profile of the car assembly, a custom probe sleeve was designed to secure the probe without the need for two separate components (Figure 28). The component was designed such that the probe would snap in place and the slightly undersized dimensions would secure the probe in place.

The curvilinear rail, car, and probe sleeve were 3D-printed at the Prototyping Lab in the Foisie Innovation Studio and assembled with the off-the-shelf components such as the heat-set brass inserts and V-rollers (Figure 29 & 30).
4 COMPONENT SELECTION & DESIGN ALTERNATIVES

Figure 29: A) Soldering iron used to insert heat-set insert into 3D-printed car. B) Car design with three adjustable shoulder V-rollers. The rollers can be tightened or loosened to adjust the distance between it and the other two rollers depending on the desired fit onto the curvilinear rail.

Figure 30: A) Fully assembled curvilinear rail design with car and probe sleeve. B) By adjusting the shoulder of the rollers, they could be further tightened onto the rail and prevent the probe from sliding.

A two-part probe sleeve would have allowed for multiple positions of the probe, and thus different types of reconstruction. Later prototyping and experimental fitting found that a single sleeve-car would reduce wobble and the tolerance issues associated with 3D printing. The final probe-sleeve-car concept is shown in Figure 31. The alignment of the car on the probe was varied with different vertical offsets, with four
variations: 0cm, 5cm, 10cm, and 15 cm. The printed rail-car with inserted heat-sets is shown in Figure 32.

Figure 31: Final probe sleeve-car CAD model. A) Isometric view shows that holes were added to the sleeve for the heat-set inserts, reducing the overall profile of the sleeve part. B) Side view to show that the final design incorporates the previously designed car and sleeve. Vertical alignment of the car along the sleeve was adjusted with a variety of sizes from 0cm to 15 cm offset. C) CAD of the final probe sleeve with heat-set inserts and wheels attached to the rail.

Figure 32: The final probe sleeve with heat-set inserts fit around the linear probe. Note that this is the 0cm vertical offset fit. This design reduced instability and tolerance issue as experienced in the two-part car-sleeve system shown earlier.
Simple rail caps were designed to fit on the ends of the rails for easy mounting on a separate fixture or frame. Initially, the team considered simple thumb screws that would screw in place at the femoral condyles at the knee. Perpendicular holes either drilled or printed into the device were made to allow the rail to rotate about a hinge. These caps would allow for exchangeable rail sizes and overall design flexibility. Figure 34 displays several views of the caps, shown in green, attached to the rail, cart, and probe.
Figure 34: Initial rail cap design to be attached to the ends of the rails in A, B, C, and D. Rail cap was left transparent for viewing of the perpendicular hole to be drilled into the cap to allow for hinge about a perpendicular axis. A) Isometric view. B) Top view. C) Side view. D) Cap magnified view.

After 3D printing rail caps of varying tolerances, the push-fit idea was found to be a poor fit. To address this, a variety of other designs were created. Two main ideas included a two part pin idea where an octagonal pin would lock the rail and cap in place (not shown) and a push-fit slot from the side that would be hammered in for a permanent fit in Figure 35. Initial concepts for the rail design involved a permanent push fit on part of the cap; these tolerances proved to be too difficult for the 3D printers available. It is important to note that rail design occurred simultaneously, as it was necessary to modify the rail’s ends to match the design of the cap. To account for the slot-based ideas, negative slot-holes were added for a smooth fit.
The final rail-cap design built upon the previously mentioned push-fit slot idea. Instead of a permanent fit, however, this cap allowed for a looser interface and relied on the compression of the sourced elevator bolt at the rail-mount connection. The CAD model and the printed prototype are shown in Figure 36.

Rails of varying sizes were also explored, in an effort to find an ideal width size. These sizes ranged from 150 cm to 190 cm. After printing these rails, it was found that the 150cm was far too small, with reasonable widths varying between 160 cm and 180 cm. Full assembly of the probe sleeve on the rail was found to have a loose
fit. To address this, a rail of higher profile for the wheels was designed. This is shown in Figure 37 and a comparison between the old and new profiles is shown in Figure 38.

Figure 37: Final rail design and various prototypes. A) CAD of the final rail to be used. This rail profile cupped the sides of the v-roller wheels for a more secure fit. B) Final rail design on the final assembly to indicate proper fit and actual use. C) From top to bottom: 150cm, 160cm, 170cm, 170cm, 180cm, and 190cm sized rails.

Figure 38: Comparing the rail-wheel interaction between the old (blue) and new (black) rails. A) The wheel glides along pretty easily with the original design, though the lack of material to support the sides of the wheel result in a loose fit. B) The new rail supports the wheel along its entire profile, not just the negative ’v’. This results in a more secure fit of the wheel along the path. C) Comparison between the two profiles with old shown in blue and new shown in black.
A separate fixture to mount the rail was designed using standard 80/20 components (Figure 39). The design consisted of two vertical pillars that allowed the vertical translation of the rail sub-assembly. Each end of the rail was mounted onto sliders on the pillars using a pivot joint that allowed locking at discrete positions. By creating an external fixture, the stability of the rail was independent of the patient’s ability to maintain their leg in a set position and also allowed the imaging of the knee in a variety of anatomical positions: both sitting and standing as well as the posterior of the knee.

Dimensions for the overall footprint of the rail frame were based around the dimensions of a typical patient of knee OA – specifically, a patient that fell between the 20\textsuperscript{th} and 80\textsuperscript{th} percentile of the target population. Finding exact dimensions of the knee were somewhat difficult, so assumptions were made based on current devices on the market intended for osteoarthritic populations. Here, it can be assumed that typical knee osteoarthritic patients are above the age of 60. Knee dimensions were anticipated to be larger than average, with increasing BMI and increasing age, as well as a potentially swollen knee from inflammation. Average knee height was found to be close to 53 ± 2 cm. To account for this, two 40-series T-Slot supports were purchased that were 60 cm in height.

This design was particularly advantageous as it would allow for vertical adjustment. A taller slider would account for patients exceeding the anticipated knee height. Short patients would, of course, be accounted for, as this design is capable of vertical adjustment. The wooden base was made to fit around both the rail that was previously fitted to work with the cart, as well as a typical patient’s knee and leg width. While width is harder to adjust for in this design, most knee widths do not vary much between patients. A heel cup, not shown in the CAD model, would additionally allow for better leg placement and help the user maintain steady posture during scanning. Fewer pieces in this design allow for easier assembly, disassembly, and replacement of parts, should they fail with overuse or age.
4 COMPONENT SELECTION & DESIGN ALTERNATIVES

Figure 39: Rail mount frame design. A) Isometric view of the frame. B) Base dimensions of the design, which is made from wood and has a dimension of 500 mm x 408 mm. C) Height of the usable slider portion of the design, approximately 60 cm.

Assembled, the fixture is shown in Figure 40 where the rail mount can be adjusted vertically, can hinge about the two points of contact with and tightened into place. A simple knee model was imported into SolidWorks and placed into the assembly to display the scale of the fixture.
Figure 40: Full assembly concept of rail design and frame. Knee model inserted to provide a reference for the dimensions of the fixture. Note that this knee model is not bent at the correct angle at which it would be imaged, and is placed to show how the fixture would interact with a real knee.

Actual construction of the parts, along with sourcing the materials for the wooden base resulted in modifying the interaction between the 80/20 parts and the base. A sourced base of a shipping pallet was modified for use in this design. To add thickness to the central part of the wooden base, another flat wooden baseboard was glued onto the pallet; this resulted in a thickness of about 1 inch. Holes of varying widths to account for the different sized rails were drilled into the base (Figure 41). T-nuts were hammered in the bottom to provide threading for simple flange-head bolts.
Another concern with the frame was how the rail was to interface with the sliders. Initial concepts involved drilling bolts directly into the plastic, though this was discarded. Following the design of the t-slotted rail caps, they were paired with elevator bolts, flat washers, and hex nuts. The hex nuts are tightened to lock the rail in place during scanning. CAD models for this are shown in Figure 43.

To hold the IMU still during calibration, a simple holder was designed and 3D printed. Holes on the top and bottom allowed it to be fixed to the frame with wing-nuts. The initial holder was found to be too short, so it was elongated. The final design and printed piece is shown in Figure 42.
Figure 42: Circuit holder design. A) The initial concept in which an IMU would be placed to stabilize during calibration. B) The holder attached to the rest of the 80/20 with two wing-nuts, as shown. C) The printed circuit holder on the frame, with the IMU in the pocket to maintain a vertical position throughout calibration.

Figure 43: CAD models of the final rail assembly. A) Isometric view, front, to show how wheels interface with the rail. B) Isometric view, back, to show the probe fit within the sleeve. C) Side view of the assembly. D) Magnified view of how the rail mounts to the frame with the elevator bolt, washer, and hex bolt.

The final CAD of the assembly, updated with used parts along with their corresponding appearance, is shown in Figure 44. The final assembled frame is shown in
Figure 44: Final assembly design is as shown from various views. A) Isometric view. B) Side view. C) Front view.

Figure 45: Final frame assembly is shown with knee.
4.7.1 Plastic Bag Design

To further explore the use of water as a coupling medium, a variety of water-filled bags were created using the bag maker at Troy lab (Figure 46). Bags were first filled with tap water, then sealed with a heat sealer. Here, plastic bag making techniques were explored to minimize overall air bubbles and optimize bag dimensions for the knee. Longer bags were thought to image better due to draping and more material to pull taught against the skin surface, whereas square bags would not provide enough contact area and may be more difficult to control.

Figure 46: Heat-sealed plastic bags containing tap water. Bags of two different widths and a variety of lengths were made to experiment with the bags of various dimensions. A) Bags were first filled with tap water and left open to degas as much as possible. B) Bag was heat-sealed to desired length. C) Two of heat-sealed plastic bags made.

During imaging, both sides of the bag were covered in US gel. The main issues with the plastic bags were determined to be wrinkling, leakage, and bubbles. The lack of elasticity in the material disallowed the bags from conforming to the irregular surface of the knee and created gaps in the images where the bag was not in contact with the skin. In addition, the integrity of the bags was inconsistent, and leakage was commonly observed with the bags at the corners of the heat-sealed end. The bubbles were seen as speckles superficial to the surface of the knee in the image, however, the time gain compensation could be adjusted to eliminate them in the image (Figure 47).
Having confirmed the viability of a plastic bag of water as a coupling medium, salt was added to the water to match the speed of sound in the medium and tissue. Commercial saline bags were purchased to test their viability as a coupling medium from DiaMedical USA (Figure 48). All five sizes were purchased to determine the best size for image quality. The best shape would provide minimal wrinkling and a proper depth.
Imaging with the saline bags was successful as the image quality was comparable to that of water (Figures 49 & 50). Here, different bag sizing was again explored for imaging both above and medial to the knee. Valves at the bottom of each bag, as well as the precut holes in the plastic material itself would allow for simple adjustment of bag thickness and integration into the mechanical frame already in place.

Figure 49: Suprapatellar femoral cartilage thickness imaged through 50 mL saline bag. A) B-scan resulting from imaging suprapatellar to the knee, with knee at 90° flexion, and probe approximately 60° with the 50 mL saline bag. B) Same imaging method as A but with anatomical regions labeled.

Figure 50: Lateral femoral articular cartilage imaged through 50 mL saline bag. A) B-scan resulting from imaging medial knee with knee at 90° flexion, and probe approximately longitudinally oriented to the medial with the 50 mL saline bag. B) Same imaging method but anatomical regions labeled.

Apart from the image quality, the saline bag was found to be superior at conforming to the irregular surface of the knee to the heat-sealed plastic bags. The addition of plasticizers in the saline bags provided them with elasticity; also, the
material used was thicker and better sealed thereby reducing the risk of leakage. For water as a coupling medium and through the designs that were explored, leakage, convenience of use, and simplicity of design were the main factors. Other ideas that involved the patient completely exposed to a body of open water could adversely affect mechanical parts and overall usability of the design. A saline bag encased in already softened plastic, with saline already prepared, as well as being a common object present in most hospitals made this idea the most attractive option.

4.7.2 Saline Bag Design

Once the saline bags were found to be a viable coupling medium, a system was designed to allow the adjustment the thickness of the bags. The design was inspired by the inflation of a cuff used in a sphygmomanometer. The main findings when experimenting with the Ziploc, custom, and saline bags were that beyond having malleable plastic able to bend smoothly over an object with a small radius, it is also necessary to pull the material taught to prevent excessive wrinkling. While bubbles and water impurities were originally thought to cause unwanted noise in the image, it was later found to be somewhat helpful in locating the surface of the skin. Bubbles appeared in images as white speckles that moved when applying pressure to the bag of water.

At the time of ideation and experimentation, a variety of sizes were used to determine the ideal bag thickness and dimensions for both above and to the sides of the knee. To address the differing locations and bag thicknesses required, a closed loop of saline within two different saline bag sizes was created (Figure 51). Clampable tubes allow for control of flow between bags, which can be controlled simply through applying pressure on one end to the other, or changing the height of the bags. This system worked well as it allowed for customizability within a closed system. Appendix A contains instructions on creating the saline bag system.
Figure 51: Closed-loop saline bag design. 6.5” by 4.5” and 4” by 2.5” saline bags were connected with 4 feet of 3/16” ID tubing. A barbed Y-connector was used to join the two ends of the tube to provide a bleeding port, capped with the red stopper in the image. Clamps on either side of connector prevent flow of saline once bag filled with desired amount of saline.

Regardless of the movement required of the probe, constrained or freehand, high image quality was necessary to visualize deeper structures. For this, stabilizing the saline bag with respect to the knee was necessary. As typically only one sonographer is present during imaging, having a component that could fix the bag to the patient was important, as it could be slippery after applying US gel. To do so, simple Velcro loops were created for both bags. Two slits were cut into the top and bottom tabs of the saline bags so that a strap could be looped through. With the addition of the buckle, the saline bag could be strapped onto the knee (Figure 52). Additionally, the pre-cut hole in the larger bag will allow for easy mounting on a simple hook made available on the rail frame.
4 COMPONENT SELECTION & DESIGN ALTERNATIVES

Figure 52: Saline bag buckled to the knee. A) 6.5” by 4.5” saline bag for suprapatellar imaging and B) 4” by 2.5” bag meant for medial or lateral imaging.

4.8 Image Reconstruction

With only one B-scan at a time, US is only able to image a limited portion of the structure of interest. However, by combining multiple B-scans into one reconstructed image, a larger portion of the structure of interest can be visualized (Figure 53).

Figure 53: (left) Individual B-scans are only able to provide limited views of the structure of interest. (right) By combining multiple B-scans into one reconstructed image, a large portion of the underlying structure can be visualized. For this, the orientation at which each B-scan was taken needs to be provided to the reconstruction algorithm.
By combining frames obtained during imaging with pose data obtained through the IMU, the structure imaged was reconstructed. The reconstruction algorithm can be broken down into two steps: bin dimension calculation and distribution. The entire algorithm was written in MATLAB, and the functions and scripts written to obtain US images, IMU data, and complete reconstruction are available on the Ultrasound-Joint-Morphology-MQP GitHub repository.

4.8.1 B-Scan Cropping Window

The image acquisition script used to capture frames was not able to discern between the region containing the B-scan and the surrounding region containing items for the menu. Because only the pixels concerning the B-scan were desired, a cropping window for each depth setting was determined as each depth setting resulted in a different imaging window. MATLAB’s `bwboundaries` function was used to identify the cropping window for various depth settings on the US (Figure 117 & Appendix B).

Figure 54: Identification of cropping window for depth of 5 cm using MATLAB’s `bwboundaries` function. Each depth setting had a slightly different cropping window; therefore, the process was repeated for all depth settings.
The cropping window for most commonly used depth settings of 3, 4, 5, 6, and 7 cm are summarized in Table 20.

Table 20: Cropping window for depths of 3 to 7 cm

<table>
<thead>
<tr>
<th>Depth (cm)</th>
<th>Cropping Window [xmin xmax ymin ymax]</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>[81 532 60 418]</td>
</tr>
<tr>
<td>4</td>
<td>[138 477 60 418]</td>
</tr>
<tr>
<td>5</td>
<td>[173 443 60 418]</td>
</tr>
<tr>
<td>6</td>
<td>[195 420 60 418]</td>
</tr>
<tr>
<td>7</td>
<td>[210 403 60 418]</td>
</tr>
</tbody>
</table>

From the selected depth setting and the dimensions of the cropping window, the pixel to millimeter resolution was calculated. The pixel to millimeter resolution for depths of 3 to 7 cm are shown in Table 21.

Table 21: Pixel to millimeter conversion for depth settings of 3, 4, 5, 6, and 7 cm

<table>
<thead>
<tr>
<th>Depth (mm)</th>
<th>mm/pixel</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>0.0838</td>
</tr>
<tr>
<td>40</td>
<td>0.1117</td>
</tr>
<tr>
<td>50</td>
<td>0.1397</td>
</tr>
<tr>
<td>60</td>
<td>0.1676</td>
</tr>
<tr>
<td>70</td>
<td>0.1955</td>
</tr>
</tbody>
</table>

4.8.2 Reconstruction

Given the orientation of the probe with respect to its motion across the curvilinear rail, images were not reconstructed into a panorama of the underlying anatomy.

Bin Dimension Calculation The bin dimensions were calculated using the size of the frames and a vector of angular displacement data, \( \theta \), using the following steps:

1. Find x- and y-coordinates of four corners of frame with bottom left corner at (0,0).
2. Calculate x- and y- coordinates of center of rotation given the radius of the path traced by the imaging surface of the US probe
3. Calculate the x- and y-coordinates of the frame after rotation by the first angle in \( \theta \) around center of rotation.
4. Repeat step 3 for every angle in $\theta$

5. Find the minimum and maximum values for the rotated x- and y-coordinates

6. Find the convex hull (bounding rectangle) of the points found in step 5

7. Shift the coordinates of the bounding rectangle such that the bottom left corner is at (1,1)

8. Return coordinates of frame at $0^\circ$ rotation (default location) and bin dimensions

The culmination of steps 1 to 7 and the final bounding box are shown in Figure 55.

![Figure 55](image)

Figure 55: (left) Bounding box for set of rectangles with bottom left corner at (0,0) rotated by six equispaced angles between $-45^\circ$ and $45^\circ$ around a center point (10, 30). (right) Bounding box shifted such that bottom left corner is at (1,1).

**Distribution Step** Reconstruction resulted in a 2D image as all the frames were taken in the same plane. This, however, resulted in substantial overlap between each frame. Once the frame was placed into its default position in a placeholder bin, it was rotated around the previously defined center of rotation. This bin was then added to a tracking bin. Once every frame was rotated and added to the tracking bin, the average of each pixel was calculated according to the number of times each pixel was imaged. Counterclockwise rotation was defined as positive rotation. The steps are summarized as follows:
1. Initialize tracking, placeholder, mask, and mask tracking bins of zeros with bin dimensions determined from bin dimension calculation

2. Place frame into placeholder bin at default location

3. Place mask of frame (all ones) into mask bin at default location

4. Rotate frame in placeholder and mask bin by corresponding angle in $\theta$

5. Add placeholder bin to tracking bin

6. Add mask bin to mask tracking bin

7. Repeat steps 2 to 5 for every frame

8. Calculate average pixel value by performing element-wise division of tracking bin by mask tracking bin

The reconstruction was tested with ten 50-by-50 frames of 100s to visualize the effect of intersecting frames. Though the input images were originally grayscale, the reconstructed image was shown in both grayscale and with a built-in MATLAB colormap (Figure 56).

![Grayscale and Color Reconstruction](image)

Figure 56: Demonstration of reconstruction in yaw with ten 50-by-50 matrices of 100s spaced equally between -90 and 90 degrees with a rail radius of 50 pixels and nearest-neighbor interpolation method. Center of rotation is shown as the red crosshair. Axes are in pixels. (left) Grayscale. (right) Color scale set to parula. Regions separated into four regions: no frame (navy), one frame (blue), two intersecting frames (teal), three intersecting frames (gold), and four intersecting frames (yellow).

Through the demonstration, it is obvious that regions closer to the center of rotation have more overlapping frames. For example, the regions in yellow are the
averages of four intersecting frames, thus, theoretically, they are more accurate representations of their respective region than the gold, teal, and blue regions, which have fewer intersecting frames. Ideally, the anatomic feature being imaged lies in this band of intersecting frames – this can be achieved by adjusting the US depth, the thickness of the saline bags, or the offset of the probe from the surface of the knee. The results of the previous reconstruction were also used to demonstrate the difference in interpolation methods (Figure 57). Though the results may not appear significant, it is important to note the difference in the gradient of pixel values at the boundaries for each method. Whereas the nearest-neighbor interpolation method results in a distinct boundary between the teal and blue region, the bilinear and bicubic interpolation methods produce a gradient between the two regions.

Figure 57: Demonstration of interpolation methods in reconstruction in yaw. Close-up of reconstructed image at the boundary of a region with two intersecting frames and only one frame. (left) Nearest-neighbor interpolation. (center) Bilinear interpolation. (right) Bicubic interpolation. Bilinear and bicubic interpolation result in a gradient at the boundary, whereas nearest-neighbor interpolation results in a clear boundary between two regions of differing pixel intensities.
5 Results

5.1 Ultrasound Imaging Setup

Prior to working with the knee and determining anatomical poses, it was first necessary to obtain and set up a working US machine. The first two models under consideration for this project, SIEMENS Sonoline Adara GM6705A2A00 (Figure X) and GE RT 3200 Advantage III (Figure 58), were passed over due to hardware issues. The two problems encountered were finding a working machine and a working compatible linear probe, which was essential for scanning superficial structures in the knee.

Figure 58: (Left) SIEMENS Sonoline Adara. This machine had a working system but there were no compatible linear probes available within the budget to purchase. (Right, Top) Back of the SIEMENS machine. The yellow converter cable, BNC-S-Video was removed and taken to use for HP Image Point for data acquisition. (Right, Bottom) SIEMENS endocavity probe.

The SIEMENS Sonoline Adara did not have a compatible linear transducer probe, and the GE RT 3200 Advantage II lacked both a working monitor and a linear probe. The former was obtained from Professor Gregory Fischer of the WPI Mechanical Engineering and Robotics Engineering Department. While the machine itself worked, it lacked a linear probe. No compatible probe was found, and this particular model was found to be an end-of-life product, which meant that there was neither technical support from the company nor compatible parts being sold. Additionally, no service
or user manual was found. The one seller who did respond said that they had neither a user manual, nor a linear probe available for purchase. A later model of the linear transducer probe was found on eBay and purchased by the BME Department; however, this model was also not compatible with the SIEMENS machine. This probe is shown in Figure 59.

Figure 59: Later model of SIEMENS linear probe. It was not compatible with the SIEMENS Sonoline Adara model and was used instead as a dummy probe for test purposes.

The second machine, the GE RT 3200 Advantage III, was older than the SIEMENS Sonoline Adara. This machine was salvaged from another project, where it had been let go previously. This machine did not have a working monitor, as the CRT monitor was broken. The two probes available for use, shown in Figure 60, were curvilinear and endocavity. The machine’s working condition was not confirmed, as a compatible monitor was not found. This model had neither a complete working system nor a linear probe.
Figure 60: GE RT 3200 Advantage III. This machine did not have a working monitor, and its working condition is unknown. The two probes available are an endocavity and a curvilinear probe, neither of interest for this project.

The third system, a modified HP Image Point (Figure 61), was found in working condition, courtesy of the WPI ECE Department. This machine, manufactured in 1989, had both a working system and a linear probe, and was the model used for the testing in this project.
Figure 61: HP Image Point Machine. This machine was the model used in this project, as it had both a working system and the appropriate probe type for imaging superficial structures – a linear transducer probe.

5.1.1 Image Acquisition

The US system was initially owned by Professor Peder Pedersen of the WPI ECE Department and was received by Professors Edward Clancy and Moinuddin Bhuiyan. With the machine came safety and service user manuals and a quick-access guide. The 15-pin VGA output from the US system was connected to a Hauppauge USB-Live 2 analog video digitizer via a female-to-female BNC connector, S-Video to card-capture, card-capture to USB, which was then connected to a laptop. This exported the images seen on the monitor of the US system onto our personal laptops through MATLAB (Figures 62 & 63).
Figure 62: Schematic of components needed to export image from ultrasound system to MATLAB on personal laptops.

Figure 63: BNC-to-BNC, S-Video, Hauppauge Analog Video Digitizer to USB connector setup Within MATLAB, the “MATLAB Support Package for USB Webcams” hardware support package in conjunction with the Image Acquisition Toolbox was used to acquire video in real-time.

The built-in `imaqtool` function launched an interactive GUI that allowed us to preview and acquire a specified burst of frames and export them into desired file formats (compressed and uncompressed). Within the GUI, the UYVY_720x480 option under the Hauppauge Cx23100 Video Capture folder was chosen in the Hardware Browser (Figure 64). Videos were saved as MPEG-4 (.mp4) file formats with a frame rate of 30 and quality of 75 as specified in the “Logging” tab as the other options (Archival, Motion JPEG 2000, Motion JPEG AVI, and Uncompressed AVI) resulted in difficulties with compatibility or extremely large file sizes.
5.1.2 Imaging Conventions

Prior to imaging with the US system, the team established conventions to identify knee angle and probe angle to avoid confusion and maintain consistency throughout all imaging trials. First, knee angle was defined using the femur as the axis. The full extension of the lower leg was defined as $0^\circ$ and any flexion from that point was defined as a positive angular displacement, with measurements obtained via a goniometer. Figure 65 depicts the five knee angles explored, namely full extension, $45^\circ$ flexion, $90^\circ$ flexion, $135^\circ$ flexion, $150^\circ$ flexion.
Figure 65: A) Knee at full extension, 0° B) Knee at 45° flexion C) Knee at 90° flexion D) Knee at 135° flexion E) Knee at extreme flexion, 150°.

Probe angle was defined in in three parts: yaw, pitch, and roll (Figure 66). These directions were decided while accounting for ease of image reconstruction during post-processing, by aligning basic probe maneuvers with the inertial measurement unit. The right-hand rule was applied to verify correct angular direction with positive axes. The origin for the probe’s movement was assumed to be along the same side as the locating notch. This origin, however, is not local to the probe and movement in later designs was assumed to be local to the stationary piece.
FIGURE 66: Demonstration of probe planar angle measurement. All three axes of interest: x, y, z, and the three associated rotations: yaw, pitch, roll, are shown. Note that this origin is not local to the probe itself, and this diagram is to illustrate general motion of the probe with respect to the anatomy.

It should be noted that the locating notch on the side of the probe should always be oriented towards the patient’s right side for more lateral measurements and upwards for longitudinal measurements. This is to help the user orient the US image, especially when locating the structures of interest. It is not an uncommon technique used among US technicians and was recommended in the HP Image Point Ultrasound user guide.

5.2 Imaging Experimentation

Three phases of imaging trials were conducted to determine parameters such as knee angle, probe angle, and system settings. Meanwhile, the system was also evaluated to determine overall system capability and effectiveness in obtaining desired anatomical measurements. The purpose and summary of each phase are outlined below:

Phase 0 Determine optimal range of knee angle, probe angle, and system settings.
Unstructured experimentation with various knee angles, probe angles, and system settings.
Phase 1 Determine reliability of US in measuring femoral articular cartilage thickness. Measurement of medial, centerline, and lateral femoral articular cartilage thickness using system.

Phase 2 Determine effect of ASP on reliability of US in measuring femoral articular cartilage thickness. Measurement of medial, centerline, and lateral femoral articular cartilage thickness using system with ASP.

5.2.1 Phase 0: Experimentation with Probe Angle, Knee Angle, and System Settings

To find the optimal system settings for superficial imaging of the knee, the team experimented with the US system to find optimal ranges of knee angle, probe angle, and system settings as well as imaging regions. We found that during full extension, $0^\circ$, to approximately $60^\circ$ flexion of the knee, the patella obscured the physiological structures of interest. The joint space width was best imaged with angles upwards of $60^\circ$ flexion. The team thought this was due to both the movement of the patella and natural widening of the joint space during flexion. This movement of the patella was confirmed with a foam injection anatomical model of the knee (Figure 67).
Through imaging, the team identified that a knee angle in the range of 90° to 150° (maximum flexion) was most suitable for imaging the femoral articular cartilage and joint space width with the probe positioned transversely and superior to the patella. Knee angles smaller than 60° were found to be too difficult to image, however 90° and above was found to work the most consistently. While a knee angle closer to 135° revealed more of the femoral articular cartilage, we believed that a patient experiencing limited range of motion due to OA would find this position uncomfortable and difficult to maintain for a long period of time.

While overall knee angles were tested, so were the probe locations on the knee. The team identified five main locations to place the probe: suprapatellar, lateral to the patella, medial to the patella, superior to the patella, and posterior to the patella. Overall the team found the suprapatellar, lateral to patella, and medial to patella to work the best. Superior to patella and posterior to patella, shown in Figure 60 were too difficult to locate the anatomical regions of interest, even when modifying the US settings.

Due to the consistency and ease of locating anatomical regions of interest with the probe suprapatellar, the team chose this location on the knee for later testing.
Not only was it easy to find the joint, it was also the most forgiving in terms of what settings to adjust to see the gap (Figure 68). Lateral and medial to the patella were trickier to image as the joint space showed up quite faded in comparison to the suprapatellar image.

Figure 68: Preliminary testing of the ultrasound machine with different knee angles and probe angles. Here, suprapatellar to the knee, specifically at a knee angle of 90° was found to be the most consistently good orientation. A) Knee at 90° flexed, probe suprapatellar at 0° yaw, 0° pitch, 0° roll B) Corresponding B-scan.

To help illustrate the anatomical regions represented via the US scans, see Figure 69. This image, however, was constructed by manually stitching together adjacent images for a more complete view of the region, as is evident by the sharp lines where the images meet. When scanning the knee with the protocol as outlined in Figure 69 it is possible to quite easily locate the joint space. It was not until later imaging until the team realized that the space being measured was not the entire space width. When scanning the knee with the probe located suprapatellar, the gap measured was actually the femoral articular cartilage thickness.
Figure 69: Stitched medial, center, and lateral scans completed by hand to illustrate the anatomical regions of interest. This image was stitched together by hand to illustrate different anatomical regions of interest observed with the suprapatellar view.

In clinical settings, acoustic standoff pads are sometimes used to help image more difficult structures within the anatomy. In an attempt to obtain the cleanest side view of the patella, the team tried an extreme scenario where the knee was the most flexed, approximately 150°, with the probe twisted at approximately 45° yaw, 45° pitch, 0° roll, as shown in Figure 70. The team recognized that this was an unrealistic scenario for osteoarthritic patients in particular, but it was interesting to find that the joint space width was, in fact, locatable.
Figure 70: Preliminary testing of the US machine with ASP and extreme flexion of the knee, with probe lateral to patella A) Knee at extreme flexion, roughly 150° with the probe lateral to the patella, at 45° yaw, 45° pitch, 0° roll. B) corresponding B-scan.

Smaller knee flexion angles yielded poor image quality as the position of the patella obstructed the path of the waves into the joint space width whereas obtuse knee angles increased ease of locating the joint space width, as the movement widened the gap. 90° was found to be a good baseline for the majority of initial testing and was easiest to image while sitting. We also experimented with various probe angles and found that a probe roll of ± 20° provided a clear view of the FAC.

Throughout imaging sessions, the team also experimented with various system settings such as gain, time gain compensation, depth, focus, dynamic range, persist, and smooth. The controls for each of these parameters are shown in Figure 71.
Figure 71: Image of the HP Image Point settings. Relevant settings and buttons for imaging the knee are boxed in different colors. Red: Gain dial and Time Gain Compensation sliders. These settings were subject to high variation throughout testing. These settings were adjusted until appropriate contrast was achieved. Blue: (left to right) Map / Smoothing / Persist dials. Map controls RGB output of the monitor, while smooth and persist control averaging and pixel lingering. Green: (top to bottom) Dynamic Range, Depth, and Focus of the image. Controls depth, range, and focus of the sound waves. Orange: (Top to bottom, left to right) Caliper, Trace, Erase, Enter, and Freeze buttons. Used to obtain measurements post-image acquisition for distance and area calculations on the system.

A particularly useful function of the system was the built-in caliper tool. During imaging, the frame could be “frozen” by pressing the “Freeze” button and the caliper tool could be used to measure the distance between any two points on the monitor to an accuracy of three decimal points (Figure 72). Multiple measurements in the same image could be obtained by pressing the “Caliper” button again.
Figure 72: Three measurements of the trochlear femoral articular cartilage thickness using the system’s built-in “Freeze” and “Caliper” function. The distance between any two points imaged were provided with an accuracy of three decimal places in centimeters.

Though the settings varied on an individual basis, a range of settings that provided the clearest images was determined (Table 22). For future imaging sessions, these parameters did not serve as hard constraints but rather as initial settings that were then modified according to the discretion of the sonographer to provide the clearest image.

Table 22: Optimal setting determined experimentally for imaging femoral articular cartilage with probe suprapatellar

<table>
<thead>
<tr>
<th>System Setting</th>
<th>Optimal Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (Frequency)</td>
<td>3 cm (15 Hz)</td>
</tr>
<tr>
<td>Gain</td>
<td>20-80</td>
</tr>
<tr>
<td>TGC</td>
<td>Variable</td>
</tr>
<tr>
<td>Dynamic Range</td>
<td>35-65 dB</td>
</tr>
<tr>
<td>Map</td>
<td>A</td>
</tr>
<tr>
<td>Smooth</td>
<td>6-7</td>
</tr>
<tr>
<td>Persist</td>
<td>3-7</td>
</tr>
</tbody>
</table>
Overall, the team found that the knee angle, probe angle, and system settings were highly dependent on the preferences of the sonographer. We failed to identify a linear relationship between any of the parameters and image quality though a recommended range was found for all.

5.2.2 Phase 1: Repeatability Without ASP

The ability of the US system to measure femoral articular cartilage thickness was determined by performing a repeatability test per the Repeatability Without ASP Protocol (Appendix C). For each subject, three views of the femoral articular cartilage surface were taken with a 90° knee flexion and 0° roll and 0° pitch of the probe at a suprapatellar location on the knee: the medial condyle (“Medial”), the trochlear surface (“Centerline”), and the lateral condyle (“Lateral”) (Figure 66). The yaw was approximately -45°, 0°, and 45° when imaging the medial, centerline, and lateral FAC, respectively. US settings used in this testing are shown in Table 23.

Table 23: Optimal range for each setting determined experimentally for imaging femoral articular cartilage with probe suprapatellar, 0° yaw, 0° pitch, 0° roll.

<table>
<thead>
<tr>
<th>System Setting</th>
<th>Optimal Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (Frequency)</td>
<td>3 cm (15 Hz)</td>
</tr>
<tr>
<td>Gain</td>
<td>50</td>
</tr>
<tr>
<td>TGC</td>
<td>Variable</td>
</tr>
<tr>
<td>Dynamic Range</td>
<td>65 dB</td>
</tr>
<tr>
<td>Map</td>
<td>A</td>
</tr>
<tr>
<td>Smooth</td>
<td>7</td>
</tr>
<tr>
<td>Persist</td>
<td>3</td>
</tr>
</tbody>
</table>
Figure 73: Graphic of suprapatellar probe placement on knee at 90° flexion and each paired ultrasound image. A) Knee at 90° and probe at -45° yaw, 0° pitch, 0° roll B) Corresponding scan of the medial condyle C) Knee at 90° and probe at 0° yaw, 0° pitch, 0° roll of the trochlear surface, or centerline D) Corresponding scan of the trochlear surface E) Knee at 90° and probe at 45° yaw, 0° pitch, 0° roll F) Corresponding scan of the lateral condyle
For each location, three measurements of the negative space between the femoral surface and the soft tissue were taken using the caliper function on the US machine. The three locations, with three measurements each (9 measurements total), were taken three times per subject. The measurements from the table were entered into a comma-separated value file and read into MATLAB (MathWorks, 2018). The average measurements and standard deviations were found for each measurement. A “normalized” standard deviation was found by calculating the ratio of the standard deviation to the average for each measurement (Tables 24, 25, and 26). Anonymity was maintained between subjects, and were named A, B, C, correspondingly.

Table 24: Person A results from repeatability phase 1

<table>
<thead>
<tr>
<th>Person A</th>
<th>Trial #</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (Frequency)</td>
<td>1</td>
</tr>
<tr>
<td>Medial 1 (cm)</td>
<td>0.223</td>
</tr>
<tr>
<td>Medial 2 (cm)</td>
<td>0.175</td>
</tr>
<tr>
<td>Medial 3 (cm)</td>
<td>0.397</td>
</tr>
<tr>
<td>Central 1 (cm)</td>
<td>0.326</td>
</tr>
<tr>
<td>Central 2 (cm)</td>
<td>0.367</td>
</tr>
<tr>
<td>Central 3 (cm)</td>
<td>0.332</td>
</tr>
<tr>
<td>Lateral 1 (cm)</td>
<td>0.338</td>
</tr>
<tr>
<td>Lateral 2 (cm)</td>
<td>0.228</td>
</tr>
<tr>
<td>Lateral 3 (cm)</td>
<td>0.200</td>
</tr>
</tbody>
</table>

Table 25: Person B results from repeatability phase 1

<table>
<thead>
<tr>
<th>Person B</th>
<th>Trial #</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (Frequency)</td>
<td>1</td>
</tr>
<tr>
<td>Medial 1 (cm)</td>
<td>0.221</td>
</tr>
<tr>
<td>Medial 2 (cm)</td>
<td>0.219</td>
</tr>
<tr>
<td>Medial 3 (cm)</td>
<td>0.256</td>
</tr>
<tr>
<td>Central 1 (cm)</td>
<td>0.258</td>
</tr>
<tr>
<td>Central 2 (cm)</td>
<td>0.268</td>
</tr>
<tr>
<td>Central 3 (cm)</td>
<td>0.251</td>
</tr>
<tr>
<td>Lateral 1 (cm)</td>
<td>0.255</td>
</tr>
<tr>
<td>Lateral 2 (cm)</td>
<td>0.201</td>
</tr>
<tr>
<td>Lateral 3 (cm)</td>
<td>0.170</td>
</tr>
</tbody>
</table>
Table 26: Person C results from repeatability phase 1

<table>
<thead>
<tr>
<th>Person C</th>
<th>Trial #</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (Frequency)</td>
<td>1</td>
</tr>
<tr>
<td>Medial 1 (cm)</td>
<td>0.217</td>
</tr>
<tr>
<td>Medial 2 (cm)</td>
<td>0.308</td>
</tr>
<tr>
<td>Medial 3 (cm)</td>
<td>0.245</td>
</tr>
<tr>
<td>Central 1 (cm)</td>
<td>0.221</td>
</tr>
<tr>
<td>Central 2 (cm)</td>
<td>0.258</td>
</tr>
<tr>
<td>Central 3 (cm)</td>
<td>0.244</td>
</tr>
<tr>
<td>Lateral 1 (cm)</td>
<td>0.120</td>
</tr>
<tr>
<td>Lateral 2 (cm)</td>
<td>0.108</td>
</tr>
<tr>
<td>Lateral 3 (cm)</td>
<td>0.130</td>
</tr>
</tbody>
</table>

5.2.3 Phase 2: Repeatability with ASP

The acoustic standoff pad (ASP) used for reliability studies was an Aquaflex Ultrasound Gel Pad, with a 9 cm diameter and 2 cm thickness. The reliability tests performed without an ASP were repeated with the ASP to determine whether an ASP alone, without 3D reconstruction, would increase the reliability of the measurements. The test was performed per the Repeatability Without ASP Protocol (Appendix D).

While the same protocol was used from Repeatability Without ASP testing, graphics of the probe with ASP on the knee to obtain the measurements of interest are described in Figure 67. The settings used on the machine are shown in Table 27. Results for patients A, B, and C are shown in Tables 28, 29, and 30, respectively.
Figure 74: Graphic of suprapatellar probe placement on knee at 90° flexion and each paired ultrasound image with Aquaflex Acoustic Standoff Pad (ASP). A) Knee at 90° and probe at -45° yaw, 0° pitch, 0° roll with ASP B) Corresponding scan of the medial condyle C) Knee at 90° and probe at 0° yaw, 0° pitch, 0° roll of the trochlear surface, or centerline with ASP D) Corresponding scan of the trochlear surface E) Knee at 90° and probe at -45° yaw, 0° pitch, 0° roll with ASP F) Corresponding scan of the lateral condyle
Table 27: Optimal range for each setting determined experimentally for imaging femoral articular cartilage with probe suprapatellar, $0^\circ$ yaw, $0^\circ$ pitch, $0^\circ$ roll with ASP to improve image quality. While settings were ideally left unaltered, they had to be altered for subjects B and C to obtain readable femoral articular cartilage thickness measurements.

<table>
<thead>
<tr>
<th>System Setting</th>
<th>Setting</th>
<th>Subject A</th>
<th>Subject B</th>
<th>Subject C</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (Frequency)</td>
<td>4 cm (12 kHz)</td>
<td>4 cm (12 kHz)</td>
<td>5 cm (12 kHz)</td>
<td></td>
</tr>
<tr>
<td>Gain</td>
<td>32</td>
<td>21</td>
<td>61</td>
<td></td>
</tr>
<tr>
<td>TGC</td>
<td>Variable</td>
<td>0.265</td>
<td>0.167</td>
<td></td>
</tr>
<tr>
<td>Dynamic Range</td>
<td>60 dB</td>
<td>60 dB</td>
<td>35 dB</td>
<td></td>
</tr>
<tr>
<td>Medial 3 (cm)Map</td>
<td>A</td>
<td>A</td>
<td>A</td>
<td></td>
</tr>
<tr>
<td>Smooth</td>
<td>6</td>
<td>6</td>
<td>7</td>
<td></td>
</tr>
<tr>
<td>Persist</td>
<td>6</td>
<td>6</td>
<td>6</td>
<td></td>
</tr>
</tbody>
</table>

Table 28: Person A results from repeatability phase 2

<table>
<thead>
<tr>
<th>Person A</th>
<th>Trial #</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (Frequency)</td>
<td>1</td>
</tr>
<tr>
<td>Medial 1 (cm)</td>
<td>0.254</td>
</tr>
<tr>
<td>Medial 2 (cm)</td>
<td>0.246</td>
</tr>
<tr>
<td>Medial 3 (cm)</td>
<td>0.335</td>
</tr>
<tr>
<td>Central 1 (cm)</td>
<td>0.400</td>
</tr>
<tr>
<td>Central 2 (cm)</td>
<td>0.481</td>
</tr>
<tr>
<td>Central 3 (cm)</td>
<td>0.455</td>
</tr>
<tr>
<td>Lateral 1 (cm)</td>
<td>0.281</td>
</tr>
<tr>
<td>Lateral 2 (cm)</td>
<td>0.211</td>
</tr>
<tr>
<td>Lateral 3 (cm)</td>
<td>0.208</td>
</tr>
</tbody>
</table>

Table 29: Person B results from repeatability phase 2

<table>
<thead>
<tr>
<th>Person A</th>
<th>Trial #</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (Frequency)</td>
<td>1</td>
</tr>
<tr>
<td>Medial 1 (cm)</td>
<td>0.108</td>
</tr>
<tr>
<td>Medial 2 (cm)</td>
<td>0.133</td>
</tr>
<tr>
<td>Medial 3 (cm)</td>
<td>0.269</td>
</tr>
<tr>
<td>Central 1 (cm)</td>
<td>0.235</td>
</tr>
<tr>
<td>Central 2 (cm)</td>
<td>0.244</td>
</tr>
<tr>
<td>Central 3 (cm)</td>
<td>0.272</td>
</tr>
<tr>
<td>Lateral 1 (cm)</td>
<td>0.323</td>
</tr>
<tr>
<td>Lateral 2 (cm)</td>
<td>0.247</td>
</tr>
<tr>
<td>Lateral 3 (cm)</td>
<td>0.144</td>
</tr>
</tbody>
</table>
Table 30: Person C results from repeatability phase 2

<table>
<thead>
<tr>
<th>Person A</th>
<th>Trial #</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (Frequency)</td>
<td>1</td>
</tr>
<tr>
<td>Medial 1 (cm)</td>
<td>0.348</td>
</tr>
<tr>
<td>Medial 2 (cm)</td>
<td>0.252</td>
</tr>
<tr>
<td>Medial 3 (cm)</td>
<td>0.266</td>
</tr>
<tr>
<td>Central 1 (cm)</td>
<td>0.266</td>
</tr>
<tr>
<td>Central 2 (cm)</td>
<td>0.446</td>
</tr>
<tr>
<td>Central 3 (cm)</td>
<td>0.354</td>
</tr>
<tr>
<td>Lateral 1 (cm)</td>
<td>0.233</td>
</tr>
<tr>
<td>Lateral 2 (cm)</td>
<td>0.283</td>
</tr>
<tr>
<td>Lateral 3 (cm)</td>
<td>0.207</td>
</tr>
</tbody>
</table>

5.2.4 Reliability Comparison

Table 31 is a representative table that summarizes each of the findings per person, with averaged measurements per location, per person, per testing phase. From here, a standard deviation was also calculated of each measured location and divided by that region’s mean, normalizing the standard deviation against the value of the mean.

Table 31: Summary of phase 1 & 2 results averaged over three trials

<table>
<thead>
<tr>
<th>Phase</th>
<th>Person A</th>
<th>Person B</th>
<th>Person C</th>
</tr>
</thead>
<tbody>
<tr>
<td>No ASP</td>
<td>ASP</td>
<td>No ASP</td>
<td>ASP</td>
</tr>
<tr>
<td>Medial 1 (cm)</td>
<td>0.244</td>
<td>0.234</td>
<td>0.173</td>
</tr>
<tr>
<td>Medial 2 (cm)</td>
<td>0.228</td>
<td>0.238</td>
<td>0.174</td>
</tr>
<tr>
<td>Medial 3 (cm)</td>
<td>0.315</td>
<td>0.326</td>
<td>0.233</td>
</tr>
<tr>
<td>Central 1 (cm)</td>
<td>0.354</td>
<td>0.381</td>
<td>0.261</td>
</tr>
<tr>
<td>Central 2 (cm)</td>
<td>0.372</td>
<td>0.430</td>
<td>0.278</td>
</tr>
<tr>
<td>Central 3 (cm)</td>
<td>0.354</td>
<td>0.418</td>
<td>0.273</td>
</tr>
<tr>
<td>Lateral 1 (cm)</td>
<td>0.280</td>
<td>0.291</td>
<td>0.238</td>
</tr>
<tr>
<td>Lateral 2 (cm)</td>
<td>0.173</td>
<td>0.230</td>
<td>0.193</td>
</tr>
<tr>
<td>Lateral 3 (cm)</td>
<td>0.150</td>
<td>0.234</td>
<td>0.165</td>
</tr>
</tbody>
</table>

The standard deviations for each subject ranged between a value recorded as 0 by MATLAB (which we take as a value too small to calculate with MATLAB) and 0.08, which was between 5% and 35% of the mean (Figures 75 to 77). A lower standard deviation proportion was considered a more reliable result. While the ASP appeared to increase the reliability for one subject, the ASP alone only slightly increased repeatability over all subjects (Figure 78). However, since the ASPs consistently provided better ease of use (imaging bony regions of knee were easier with ASP) and
clearer images, the team decided to continue using them.

Figure 75: For Patient A, the normalized standard deviations were lower for the measurements using ASP than for those without, except in the trochlear region, indicating that the ASP increased the reliability of condylar cartilage measurements.

Figure 76: For Patient B, there was no clear trend when comparing the ASP and non-ASP measurements. However, there was greater variability at the edges than in the center, most likely due to varying measurement locations in regions where cartilage was thicker.
Figure 77: For Patient C, there was no clear trend when comparing ASP to non-ASP measurements, nor was there a clear trend across the cartilage surface.

Figure 78: Overall, the coefficient of variation was lower with the ASP than without it.
5.3 Pose Sensing and Displacement Algorithm

Earlier, the need for a reconstruction algorithm was established to visualize a larger region of the structure of interest. For this, and MPU 6050 IMU was used for probe pose sensing, as it was necessary to detect the orientation of each B-scan for image reconstruction. Though yaw, pitch, and roll were readily obtained using a digital motion processing library for the MPU6050_DMP6, displacement was not as easily obtained as only accelerations in the three orthogonal directions were provided. For this, a numerical double integration scheme was implemented in MATLAB to calculate the displacement of the probe from acceleration measurements during post-processing. The MATLAB \textit{cumtrapz} function was used to implement a double trapezoidal integration scheme and provided the displacement at all the provided time intervals in each direction. In addition, the function was able to integrate with nonuniform spacing, which was necessary as the measurements of acceleration were obtained with nonuniform time intervals.

5.3.1 Numerical Double Integration Method Testing

The integration scheme was first tested on functions of acceleration whose solutions for displacement could be calculated analytically. Acceleration was modeled as three different sinusoidal functions and the numerical double integration scheme was used to calculate the displacement for $t = 0$ to $t = 2\pi$ (Figure 79).
Figure 79: Double trapezoidal numerical integration scheme using MATLAB function, \texttt{cumtrapz}, tested on function whose analytic solutions are known. Values of $\sin(t)$, $\cos(t)$, and $\sin(t/2)$ were assigned to $a_x$, $a_y$, and $a_z$, respectively, with nonuniform spacing for $t$.

The numerical solutions for displacement were compared to the analytic solutions for displacement to evaluate the accuracy of the scheme (Figure 80). The scheme was able to calculate the displacement accurately with sufficient time resolution; however, the displacement drifted with a large time step. The results from this motivated the need for a sampling rate for acceleration to combat against drift in the calculated values for displacement.
Figure 80: Comparison of displacements results from numerical double integration scheme to analytic solution for acceleration. Though the numerical solutions for displacement in the y- and z-directions are accurate, the displacement in the x-direction drifts from about $t=4s$.

The relative error of the numerical solution shows that the majority of the results were accurate to within 10% (Figure 81). Because the scheme was able to provide sub-millimeter resolution even with such a low temporal resolution, we decided to use this scheme to calculate displacement from the linear acceleration recorded using the IMU.
RESULTS

Figure 81: Relative error of numerical solution for each direction of displacement for every time step for the x, y, and z axis. Though the relative error is high at the beginning, this can be disregarded as a small error produces large errors given the small values being used to calculate relative error.

The scheme was demonstrated to be accurate with sinusoidal accelerations at calculating displacement; however, it should be noted that the acceleration in this case was very smooth and for a short time span (6 seconds). Therefore, the effectiveness of the scheme at calculating displacement for noisier acceleration data and extended periods of time remained unclear at this point.

5.3.2 Experimental Testing of Numerical Method

Once the numerical scheme was tested on an artificial acceleration profile, it was tested on experimentally recorded values of acceleration. Though the scheme was shown to be accurate for a sufficient time resolution, initial offset was identified as an additional concern. In addition, the IMU required several seconds for the readings to stabilize once the MATLAB script was run. Both the offset in acceleration and gradual stabilization of the acceleration and theirs effects on the resulting displace-
ment are shown in Figure 82. Note that these data were obtained while the IMU was at rest. Instructions for calibrating and collecting data with the IMU are in Appendices E and F respectively.

Figure 82: (left) Acceleration in the x-, y-, and z-directions while inertial measurement unit at rest. Slight offset can be seen for all directions. Gradual stabilization, particularly in the y-direction, can be seen as well. (right) The effect of the offset and gradual stabilization on the resulting displacement are made evident by the fact that the displacement is continually increasing or decreasing despite the inertial measurement unit being at rest.

To combat the offset, calibration performed prior to each use of the IMU. The gradual stabilization was addressed by requiring the linear acceleration values to have an absolute value below 5 mm/s² while the IMU was at rest for the recording to begin. These additions somewhat reduced the error in displacement (Figure 83).
Figure 83: (left) Acceleration in the x-, y-, and z-directions while inertial measurement unit at rest. The signal is much more stable except for the large negative acceleration around 25 seconds. (right) Displacement calculated from the stabilized acceleration signal.

Though the acceleration appears to be consistent, a large amount of noise was detected. For this, a custom filter was applied to the data. Because filtering is a process typically completed after obtaining the data of interest, a few trials of known behavior and absolute positioning were obtained. The first was no movement, where known acceleration (x, y, z) and gyration (y, p, r) were completely still. For this, the IMU was calibrated and run on a table top and left undisturbed for about 10 seconds total. To streamline the filter design, a new script was written to plot the original signal with the calculated displacement, filter the data and calculate new displacement, and plot the final filtered data.

To begin filtering, it was first necessary to determine the characteristics of the raw signal. For this, a `pwelch` command was applied on the raw data. The peak natural frequency was found to be quite small, around a normalized frequency of 0.007813 \( \pi \) rad/sample. Understanding where the natural frequency is informs design process of the proper filter. The first filter designed was a low-pass filter, which was created using the `designfilt` function. Raw data was run through the filter using the built-in `filtfilt` function. The `designfilt` command allowed for simple customizability and the `filtfilt` command allowed for zero-phase distortion at the cost of squaring the magnitude of the transfer coefficients a and b and doubling
the overall filter order.

The original data and the resulting data from the first legitimate low-pass filter is shown in Figures 84 and 85. Note that this was the data acquired from just leaving the IMU flat on the table. The raw data should have been a steady 0 $mm/s^2$. As shown, both the raw and filtered data do not reflect the expected trend. The low-pass filter was capable of filtering out the high-frequency noise, though failed at removing baseline drift, which was to be expected as it results from low-frequency noise. As such, it is capable of finding the ‘average’ of the signal noise, though in Figures 84 and 85 any noise left unchecked in the acceleration was amplified during the double integration to acquire position.

In order to remove both the low-frequency and high-frequency noise, a band-pass filter was created. A Butterworth filter was chosen due to its maximally flat and smooth response. To remove the unwanted spikes, a median filter was applied. The median filter was used first to eliminate outliers before applying a bandpass Butterworth filter. With the median filter, overall signal smoothing can be achieved, and with the band-pass filter, unwanted high-frequency and low-frequency noise can be eliminated. It can be seen that the overall range of acceleration is lowered by at least half, and the resulting displacement decreases from over 200 mm to approximately 2 mm.
Figure 84: Results of leaving the inertial measurement unit at rest on the table. This figure includes the component breakdown of the acceleration in the X, Y, Z (top to bottom). Each of these subplots shows the original data (cyan), band-pass filtered data (dark blue), median-filtered data (red), and the expected data (black). Here, as the IMU was simply at rest, the acceleration should reflect 0 on each of the subplots. The median filter simply removed the high frequency noise, and was thus eliminated as a potential filter. The bandpass filter was able to remove the high and low frequency noise, and was used. Here, it is shown that the bandpass filtered data most closely bounces around the true expected data of 0.
5 RESULTS

Figure 85: Filtered results of the no movement test using a band-pass. Though capable of decreasing the magnitude of the acceleration, the filtered data still resulted in a calculated displacement of about 6.96mm when it should have reflected 0 (or a result closer to 0).

Another test was administered to see if the IMU could properly detect overall displacement. This would help determine the overall accuracy of the IMU. For this application, sub-millimeter resolution is necessary. If the IMU is capable of maintaining consistent global positioning, a simple test where the start and end positions of the IMU were the same would indicate its accuracy. For this, a square with a length of 100 mm was drawn on a piece of paper. The first known movement would be to move the IMU along the entire square, for one cycle. This setup is shown in Figure 86.
The results from this test indicated that the IMU was not capable of detecting a net displacement of 0 mm, with a distance of 400 mm via 2 degrees of freedom. Despite filtering, the overall results were indicative of a poor positioning system, as it showed a z-directional displacement of 100 mm. True displacement should have been along the x and y axis.

To verify the system inaccuracy, one more test was conducted: a linear displacement test. Here, only one degree of freedom was allowed. The IMU was moved along a known start, pause, and end point, with its end point being the same location as its start. For this, the IMU was moved along a single line segment from the previously drawn square, about 100 mm. The true displacement plot for this would be a positive trend to around 100 mm, flat line at 100 mm during the pause, then a downwards slope to return to its start position to about 0 mm. The results of the linear displacement test are shown in Figure 87.
Figure 87: Results of the linear displacement test. Original data is plotted, which includes acceleration, gyration, and calculated linear displacement. The raw data is very noisy and incorrectly results in three different displacements, where it should result in a net 0 mm.

Table 32: IMU Global Positioning Results

<table>
<thead>
<tr>
<th>Trial</th>
<th>No Movement</th>
<th>Square Test</th>
<th>Line Test</th>
</tr>
</thead>
<tbody>
<tr>
<td>Results (mm)</td>
<td>6.96</td>
<td>39.82</td>
<td>11.21</td>
</tr>
<tr>
<td>Expected (mm)</td>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>% Error</td>
<td>596%</td>
<td>3882%</td>
<td>1021%</td>
</tr>
</tbody>
</table>

Overall, the results from both the line-displacement test and the square test indicated that the IMU was not capable of measuring displacement with sub-millimeter accuracy (Table 32). All three trials should have resulted in close to 0 mm, though during error calculations there was a generous cushion of ±1 mm, though a sub-millimeter resolution would have been ideal. These results show that the IMU drifted even in the most basic case, where it simply rested on a table-top for 15 seconds. The square test and the line test additionally proved that even simple movement resulted in too much drift within the system even after 15-30 seconds of use. Aggressive filtering after these tests were administered found that no amount could reproduce...
the IMU’s movement without destroying the expected trends (most notably with the line displacement test).

There was too much drift in the system, and noise captured with the acceleration was amplified when integrating twice to obtain displacement. After researching additional models of inertial measurement units, the IMU’s that had the required precision, which were also of a suitable size for the application, were very expensive and out of the given budget for the project.

Given the inaccuracy of the IMU global positioning, a freehand US system using the IMU to provide both position and orientation was disregarded. While it was concluded that the IMU was incapable of providing accurate displacement data, it was capable of providing sufficiently accurate angular measurements.

Despite the inherent drift associated with the accelerometer, the IMU was still considered for its ability to detect changes in angle. Because this did not rely on any integration, it was tested for its gyroscopic reliability. For this, three trials of three cycles each were administered to test the gyroscope’s reliability. The test setup is shown in Figure 88. Two metal rulers were clamped to the edge of the table, to provide a straight-edge for which the IMU to rotate about. Because the anticipated curvature of the mechanical fixture was approximately 180° in total, the IMU was simply rotated about this. This motion is shown in Figure 89. Detailed protocol can be found in Appendix G. Raw data and the process of extracting the relevant data from these tests are shown in Figure 90.

![Figure 88: Testing the IMU’s reliability](image)
5 RESULTS

Figure 89: The motion of the IMU for rotation. A) 0 degrees. This is the position in which the IMU was calibrated. B) 45°. C) 135°. D) 180°. Here, the motion was paused for roughly 1 second before returning back to the 0° position.

Figure 90: MATLAB plots used to obtain the resulting accuracy and precision of testing the IMU’s gyroscopic reliability. (left) Raw rotational data from one trial, or 3 cycles of the two known positions. Because the IMU was rotated about 1 axis, data is only shown along one of the three components. This is as expected. The IMU experienced three cycles per trial. (right) The relevant positions, the 0 and 180 degree positions were then extracted. From this, accuracy and precision was calculated, as the expected positions were known.

Note that no filtering was applied to this data, as the raw data was found to be stable enough to read the current angle. To extract the relevant position data, a threshold boundary was applied. The positions of interest were the 0° and 180°. Accuracy and precision were calculated for these two positions of interest for each cycle for all three trials. The results for this reliability testing of the IMU’s gyroscope are as shown in Table 33. Overall accuracy was averaged to be approximately 0.96° with a precision of about 0.21°.
Table 33: Results from testing the IMU’s gyroscope accuracy and precision.

<table>
<thead>
<tr>
<th></th>
<th>Trial 1</th>
<th>Trial 2</th>
<th>Trial 3</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0°</td>
<td>180°</td>
<td>0°</td>
</tr>
<tr>
<td>Accuracy (°)</td>
<td>0.35</td>
<td>1.24</td>
<td>0.24</td>
</tr>
<tr>
<td>Precision (°)</td>
<td>0.149</td>
<td>0.245</td>
<td>0.171</td>
</tr>
</tbody>
</table>

For testing the gyroscope’s reliability, its precision was of higher concern. To produce an accurate reconstruction, it is more important to ensure that the angular displacement between scans is consistent; Reliability and consistency between subsequent scans is more important than being able to locate the true position of only a few scans of the bunch. Thus, it was a good indication that the overall precision was found to be of a higher degree than that of the accuracy. The resulting accuracy and precision of the IMU’s gyroscope was found to be good enough for reconstruction, especially when compared to that of the accelerometer.

5.4 Simulated B-scan Reconstruction Testing

Testing was performed with simulated data to verify the algorithm. B-scans for simple geometries were created and were used to reconstruct the original geometries of known dimensions (Figure 91).

Figure 91: (left) Example of a square with a side length of 20 pixels and (right) circle with a radius of 20 pixels used for simulated B-scan reconstruction in yaw. Blue and yellow pixels have a value of 0 and 1, respectively.

Once the simulated shapes were created, B-scans of the geometry at a range of
angles from a set radius were taken (Figure 93 & 94). The following steps (including
the initial creation of the geometry) were taken to accomplish this:

1. Initialize a bin of zeros (bin1)
2. Create a matrix of ones in the desired shape in bin1 (Figure 91)
3. Initialize a separate bin of zeros (bin2) with the same dimensions as bin1
4. Calculate the default location of the B-scan within bin2
5. Calculate the point of rotation of the B-scan within bin2
6. Create a matrix of twos in the default location of the B-scan in bin2
7. Rotate the B-scan in bin2 by the desired angle around the point of rotation
8. Add matrices bin1 and bin2 together (bin3)
9. Rotate bin3 by the equal and opposite amount as in step 7 (Figure 92)
10. Extract the B-scan from bin3

Figure 92: (left) B-scan to be extracted in rotated position. (right) B-scan rotated
to vertical position to prepare for extraction.
B-scans were simulated for a circle with a radius of 20 pixels and a square with a side length of 20 pixels. Each B-scan had dimensions of 40 pixels-by-20 pixels. The center of rotation was defined as the centers of the shapes and the imaging radius was set to 30 pixels. B-scans were simulated at five equip-distant angles from -90° to 90°. Figures 93 & 94 show the five B-scans simulated for the circle and square, respectively.

Figure 93: Simulated B-scan planes of circle at (left to right) -90°, -45°, 0°, 45°, and 90°. Blue pixels correspond to the background, light blue to the region of the circle not within the window of the B-scan, yellow to the region of the circle within the window of the B-scan, and green to the background region within the window of the B-scan. The red cross is the center of rotation.

Figure 94: Simulated B-scan planes of square at (left to right) -90°, -45°, 0°, 45°, and 90°. Blue pixels correspond to the background, light blue to the region of the circle not within the window of the B-scan, yellow to the region of the circle within the window of the B-scan, and green to the background region within the window of the B-scan. The red cross is the center of rotation.

5.4.1 Partial Reconstruction (Overlap Visualization)

Prior to full reconstruction, these simulated B-scans were partially reconstructed and used to demonstrate the regions and frequencies with which specific regions of the geometries were imaged (Figures 95 & 96).
Figure 95: Comparison of (left) simulated circle and (right) partially reconstructed circle. The pixel intensity in order of least to greatest is navy blue, light blue, teal, green, orange, light orange, and yellow. A higher intensity represents a larger number of intersections of distinct B-scans. In theory, regions with higher intensity values in this test will result in higher resolution images as the region is imaged from multiple viewpoints.

At first glance, the partial reconstruction in yaw of the circle appears incomplete as the lower region of the circle is missing. This is not the result of the reconstruction, rather it is due to the B-scans supplied for reconstruction. Looking at the B-scans supplied (Figure 93), it can be seen that none of the B-scans capture the lower third of the circle. Therefore, given the limited field of view of the B-scans, the circle cannot be reconstructed in its entirety. The pixel intensities from lowest to greatest are navy blue, light blue, teal, green, orange, light orange, and yellow. The center of the circle is mostly yellow indicating the highest pixel intensity - the region with the most overlapping B-scans. In contrast, regions in light blue were only captured by one B-scan.
Figure 96: Comparison of (left) simulated square and (right) partially reconstructed square. The pixel intensity in order of least to greatest is navy blue, light blue, teal, green, orange, light orange, and yellow. A higher intensity represents a larger number of intersections of distinct B-scans. In theory, regions with higher intensity values in this test will result in higher resolution images as the region is imaged from multiple viewpoints.

The partial reconstruction in yaw of the square resulted in a similar color gradient to that of the circle. Like the circle, regions toward the middle of the square have higher pixel intensities and vice-versa. Unlike the circle, however, every pixel of the square was scanned by at least one B-scan; therefore the square was reconstructed in its entirety. In both cases, the majority of the geometries intersected with the B-scans’ field of view. To further explore the effect of varying imaging setups, cases in which only small regions of the geometries were scanned were considered as well: the same shape geometry and size, and the angles of B-scans were kept, but the radius was varied (Figures 97 & 98).
5 RESULTS

Figure 97: Comparison of partial reconstructions of a circle with imaging radii of (left to right) 30, 40, and 50 pixels. At r=30, the center of the reconstructed geometry has the most intersecting B-scans and the number of intersections decrease radially. A similar gradient is seen in r=40, though the region with the most intersections is slightly above the center. Reconstruction with r=50 results in a horseshoe-like geometry. The region with the most intersections in this case lies along top half of the curve. (Note: Shapes may appear to have different dimensions, but this is the result of differing axes)

Figure 98: Comparison of partial reconstructions of a square with imaging radii of (left to right) 30, 35, and 40 pixels. At r=30, the square is reconstructed in its entirety with a majority of it imaged by multiple B-scans. r=35 presents similar results, though the bottom corners of the reconstruction appear rounded and there appear to be fewer intersecting B-scans overall. With r=40, the limited field of view becomes apparent as the bottom half of the reconstructed square is either imaged once or twice or not at all. (Note: Shapes may appear to have different dimensions, but this is the result of differing axes)

5.4.2 Full Reconstruction

Full reconstruction consisted of one additional step: averaging pixel values by the number of times each pixel was imaged. In theory, the pixel intensity of every pixel in the reconstructed image would match that of the original image; however, errors were accumulated during reconstruction due to the choice of pixel interpolation
methods. To evaluate the accuracy of the reconstructed image to the original image, the Structural Similarity Index (SSIM) and normalized 2D cross-correlation were used to evaluate the similarity of the reconstructed image to the original image. The SSIM calculates the similarity of an image to a reference image according to three metrics: luminance, contrast and structure. For reference, the SSIM of two of the exact same images is 1 (MathWorks, 2019).

Reconstruction was performed for three shapes: a square, circle, and a composite of the two shapes. The square had a side length of 20 pixels and the circle had a diameter of 20 pixels. The composite was created as a union of the two shapes, with the circle shifted 10 pixels to the left and up. Each simulated shape was scanned using a 60-by-20 pixel window at five equispaced angles from -90° to 90°. The center of rotation was set to the center of the shape (center of the square in the case of the composite) with an imaging radius of 30 pixels. The shapes were reconstructed with various pixel interpolation methods: nearest-neighbor, bilinear, and bicubic.

The square reconstructed using each of the three pixel interpolation methods had SSIM and normalized 2D cross-correlation values above 0.99 indicating that the reconstructed squares were nearly identical to the original square in both structure and pixel intensity (Figure 99).
Figure 99: (top left) Original square for which B-scans were simulated and reconstruction was performed. Reconstruction performed using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values.

The reconstructed circle showed similarly high values for both SSIM and normalized 2D cross-correlation (Figure 100).
Figure 100: (top left) Original circle for which B-scans were simulated and reconstruction was performed. Reconstruction performed using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values.

The reconstructed composites had a high degree of similarity as well, though the SSIM values for bilinear and bicubic interpolation dropped by roughly 0.01 (Figure 101).
5 RESULTS

Figure 101: (top left) Original composite shape for which B-scans were simulated and reconstruction was performed. Reconstruction performed using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values.

5.4.3 Noisy Reconstruction

Given that the IMU measurements had a certain degree of inaccuracy, the reconstruction algorithm was tested with the same geometries as in the previous section but with slightly offset angular displacement data. Such a test is important in the evaluation of the algorithm with random errors.
The B-scans were simulated using the same parameters; however, random noise was added to the angular displacement data input to the reconstruction algorithm. A random value within the range of ± 0.96° was added to the angular displacement data – this value was chosen based on the results of IMU testing. The reconstructed shapes, even with random noise retained a high degree of similarity (Figures 102 to 104).
Figure 102: (top left) Original square for which B-scans were simulated and reconstruction was performed. Reconstruction performed with noise introduced into the angular displacement data using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values.
Figure 103: (top left) Original circle for which B-scans were simulated and reconstruction was performed. Reconstruction performed with noise introduced into the angular displacement data using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values.
Figure 104: (top left) Original composite for which B-scans were simulated and reconstruction was performed. Reconstruction performed with noise introduced into the angular displacement data using the nearest neighbor, bilinear, and bicubic pixel interpolation methods are shown in the other three figures along with their SSIM and normalized 2D cross-correlation values.

The SSIM and normalized 2D cross-correlation values for each of the reconstructed shapes using each of the pixel interpolation methods with and without artificial noise introduced are summarized in Tables 34 and 35.
Table 34: SSIM and 2D cross-correlation values for reconstructed square, circle, and composite using nearest neighbor, bilinear, and bicubic pixel interpolation methods

<table>
<thead>
<tr>
<th>Shape</th>
<th>No Artificial Noise</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Nearest</td>
<td>Bilinear</td>
<td>Bicubic</td>
</tr>
<tr>
<td>Square</td>
<td>0.996/0.998</td>
<td>0.994/0.999</td>
<td>0.994/0.999</td>
</tr>
<tr>
<td>Circle</td>
<td>0.997/0.998</td>
<td>0.997/0.999</td>
<td>0.997/0.999</td>
</tr>
<tr>
<td>Composite</td>
<td>0.992/0.997</td>
<td>0.984/0.998</td>
<td>0.984/0.998</td>
</tr>
</tbody>
</table>

Table 35: SSIM and 2D cross-correlation values for reconstructed square, circle, and composite using nearest neighbor, bilinear, and bicubic pixel interpolation methods with artificial noise introduced into the angular displacement data

<table>
<thead>
<tr>
<th>Shape</th>
<th>With Artificial Noise</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Nearest</td>
<td>Bilinear</td>
<td>Bicubic</td>
</tr>
<tr>
<td>Square</td>
<td>0.955/0.988</td>
<td>0.963/0.991</td>
<td>0.966/0.993</td>
</tr>
<tr>
<td>Circle</td>
<td>0.971/0.992</td>
<td>0.972/0.991</td>
<td>0.975/0.993</td>
</tr>
<tr>
<td>Composite</td>
<td>0.919/0.985</td>
<td>0.931/0.991</td>
<td>0.934/0.992</td>
</tr>
</tbody>
</table>
6 Discussion

In this section, we will review and interpret findings presented in Chapter 5. Raw data is interpreted in a broader context, beyond the immediate design. The project’s impact on the economy, environment, society, politics, ethics, public health and safety, manufacturability and sustainability are discussed.

6.1 Interpretation of Results

The mechanical fixture, imaging method, and image reconstruction provide the user an easy way to measure a variety of patients and visualize the region of interest for knee OA screening. The initial objectives for this project are outlined below. For each objective, the accomplishments and limitations are discussed.

*Determine the optimal orientation and location of the probe on the knee to measure joint space width and femoral articular cartilage thickness.*

For this, a range of knee angles, probe angles, and probe positions on the knee were explored. A suprapatellar position of the probe on the knee was found to be most ideal as it allowed for the easiest identification and visualization of the femoral articular cartilage. For knee joints, the longitudinal axis that extends from the femur is taken to be a 0°, with clockwise rotation, or flexion of the knee interpreted as positive angular displacement. With respect to knee angle, 90° was found to be optimal, especially considering the limited range of motion in OA patients. Patients at risk for knee OA, or those who might already have the disease, are unlikely to have a wide range of motion of the knee. The best knee angle for imaging the joint space was approximately 100° to 135°, but due to the mobility required of the subject, it was not considered to be the ideal angle.

*Design a fixture to maintain optimal orientation and location of an US probe on the knee.*

A curvilinear rail and mounting fixture were designed to traverse the anterior region of the knee. Having identified the femoral articular cartilage as the primary region of interest in measuring articular cartilage thickness and knee joint space width, the rail allowed the user to consistently trace the necessary path. The rail only allows one degree of freedom – it fixes the offset of the probe from the knee surface as well as the tilt of the probe. Though the band being imaged can be changed by rotating the rail, the reconstruction is not able to account for such movement as it
reorients individual frames based only on one gyroscopic measurement. Therefore, in its current state, the curvilinear rail must be set to a angle angle prior to traversing the rail and acquiring images.

Stabilize the joint in at least one standard pose in which the joint space width reliably predicts a healthy or diseased state.

The design itself is capable of accommodating both a variety of patient knee sizes as well as sitting and standing positions. The heel cup at the base of the frame aligns the patient’s heel and helps keep the patient still. The design, does, however, rely upon the user’s ability to both operate an ultrasound machine and identify the correct anatomical region of interest on the scan.

Although the suprapatellar region was identified to be an optimal location to image the femoral articular cartilage, images of the region were not confirmed to be reliable predictors of health or diseased states. Similarly, 90° or greater of flexion was found to provide the clearest, most unobstructed view of the femoral articular cartilage, however, it, too, was not identified to be a pose that allowed the reliable prediction of a healthy or diseased state. Due the exploratory nature and findings of this design, it is better suited for pre-diagnostic applications, like screening for knee OA risk factors.

Additionally, the design and adjustability of the fixture allows for a variety of other knee poses. In particular, it is possible for a subject to stand up with their knee in the fixture (this would require a different configuration of the saline bag system), allowing it to be used for load-bearing measurements.

Provide at least one image of the joint from which the joint space can be accurately, precisely, reliably, and repeatedly measured.

Through the preliminary imaging studies (phases 1 and 2), the measurement of articular cartilage thickness was found to be highly inconsistent for both imaging alone and imaging with a commercially available acoustic standoff pad. This design sought to improve the accuracy and repeatability of imaging the knee joint space in a pre-diagnostic setting.

Track the location and orientation of the probe during imaging and, through post-processing, create a stitched visualization or reconstruction of the imaged region.

A curvilinear rail design was combined with an IMU to allow the tracking of the position and orientation of the probe during image acquisition. The orientation of
the probe was provided, from which the position of the probe along the rail could be calculated. Each frame was tagged with a corresponding pose, which allowed for the reconstruction of B-scans into a single panoramic image. The reconstruction algorithm was tested with simulated data in the form of basic geometric objects and showed that it could reconstruct simple shapes from B-scans with a high level of accuracy.

It should also be noted that, instead of developing a 3D reconstruction algorithm, a 2D panoramic reconstruction algorithm was developed instead. This decision was motivated by the fact that the curvilinear rail design would not be able to provide a reconstructed volume with sufficient resolution: inserting B-scans obtained using a probe motion in pitch results in a very sparsely filled volume (Figure 105). This means that the voxel values at empty voxels must be interpolated from neighboring scans.

![Figure 105: Example of reconstruction algorithm but frames acquired in pitch as opposed to yaw motion of probe. The region towards the center of rotation is densely imaged; however, regions farther radially are very sparsely imaged.](image)

6.2 Manufacturability

The manufacturability of the design is considerably high given the ease with which the necessary components can be made or ordered. The majority of the components are 3D-printed using standard printing filament in PLA with standard layer thickness and infill. In theory, if another user were to employ the design but had a different US probe, the only component that would have to be modified is the probe sleeve.
In addition to the ease with which the 3D-printed components can be obtained, all other purchased components are standard off-the-shelf products. Lastly, the cost of the printed and purchased components sum to $381.08 (Table 36). Given the durability of the components used, it is highly unlikely they will need to be replaced on a regular basis.

Table 36: Cost breakdown of components of final fixture design

<table>
<thead>
<tr>
<th>Component</th>
<th>Cost (each)</th>
<th>Quantity</th>
</tr>
</thead>
<tbody>
<tr>
<td>Curvilinear Rail</td>
<td>$0.36</td>
<td>1</td>
</tr>
<tr>
<td>Rail Car</td>
<td>$0.63</td>
<td>1</td>
</tr>
<tr>
<td>Probe Sleeve</td>
<td>$0.85</td>
<td>1</td>
</tr>
<tr>
<td>V-Roller</td>
<td>$30.63</td>
<td>3</td>
</tr>
<tr>
<td>Rail Cap</td>
<td>$25</td>
<td>2</td>
</tr>
<tr>
<td>Heat-Set Inserts</td>
<td>$9.35</td>
<td>1 (pack)</td>
</tr>
<tr>
<td>8020 Aluminum Frame</td>
<td>$12</td>
<td>2</td>
</tr>
<tr>
<td>Linear Sliders</td>
<td>$67</td>
<td>2</td>
</tr>
<tr>
<td>Frame Accessories</td>
<td>$20</td>
<td>1</td>
</tr>
<tr>
<td>Plywood</td>
<td>$3</td>
<td>1</td>
</tr>
<tr>
<td>Saline Bag</td>
<td>$20</td>
<td>1 (assortment)</td>
</tr>
<tr>
<td>Velcro</td>
<td>$5</td>
<td>1</td>
</tr>
<tr>
<td>Tubing</td>
<td>$2</td>
<td>1</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>$381.08</strong></td>
<td></td>
</tr>
</tbody>
</table>

### 6.3 Health and Safety Issue

US has already been established as a safe form of medical imaging and our design does not alter the imaging system at all – it merely augments it with external fixtures. Given that the external fixture does not have any automated components such as a motor, there is no risk of machine-related injury as all movements are controlled by a human user. The design does, however, present pinch points at the point of connection between the curvilinear rail and linear slider, linear slider and the frame, and between the car sub-assembly and the edge of the linear slider when the car is moved laterally.

Additionally, as only trained medical professionals will be using this device, they will further prevent misuse or risks involved. Assuming this device will be used in its targeted location of only medical sites, patients who may suffer from pinched skin are also in the immediate area of first care. Besides these points, the design presents no major hazards to both the user and patient during use.
6.4 Economic Impact

The modality of B-mode US was chosen for its versatility in imaging soft and hard tissue with a portable, affordable system. While US offers neither the hard-tissue clarity of X-ray nor the soft-tissue precision of MRI, it is more portable than both systems and results from sonography with reconstruction are more quantitative than the Kellgren-Lawrence scale used for X-ray image processing. While the Kellgren-Lawrence scale provides only four degrees of severity on a largely subjective scale (Kellgren & Lawrence, 1957), our US system is capable of measuring the joint space width, and with higher resolution and better position reading would likely provide sub-millimeter accuracy. With even 0.5 mm accuracy across a joint space width of 5 mm, the sensitivity of the test is more than doubled, from 4 to 10 possible results.

The major economic impact of an US system for measuring knee joint space width is a reduction in the need for local MRI systems, particularly in developing countries. The US could be used to screen patients with early symptoms for joint space narrowing and refer those who need an MRI to a different facility. Typically, for those seeking care at low-cost facilities, as well as those with health insurance, screening tests lower risks of higher costs later, and are therefore often free. Considering that an x-ray of the knee for OA diagnosis costs approximately $200 and an MRI scan costs approximately $700, a preventive screening would significantly lower the costs to the patient, and is of minimal cost to health care facilities. After the disease has been diagnosed, US will also be useful in tracking progression of the disease on a regular basis. Most clinics and hospitals have access to donated obstetric ultrasound machines and low-cost linear probes are available. Our device could be used to retrofit these machines for musculoskeletal measurements of the knee.

6.5 Environmental Impact

Both the simplicity of design as well as machinery used in US acquisition contribute to the overall low environmental impact. This aim was to reduce its overall footprint and increase accessibility to regions that may not have the means to acquire expensive tools for first-pass diagnostic screening by augmenting technology that is likely to be readily available. For the duration of the project, a 30-year-old HP ImagePoint US machine was used, as it was representative of what many smaller clinics may rely upon for daily use. It would be simple for a clinic that uses US for obstetrics to obtain a linear probe and our system, adding to the functionality of the device in a likely under-served region. Total imaging time should not exceed 15-30
minutes, with a majority dedicated to post-acquisition image processing, which is completed on an external system.

Almost all of the pieces used in the design were second hand materials. The only newly purchased materials were the US gel and saline bags, which can also be acquired at a lower rate in bulk. The 3D-printed parts could be easily mass-produced through injection molding and added to the standard parts of the design. The fixture itself has a very low impact, as it is relatively compact and is anticipated to last years of use due to its durability and simple design. The mix of 3D-printing, recycled material, and easily sourced parts make this design environmentally friendly.

### 6.6 Societal Influence and Ethical Concern

The intention of this design is to promote the use of US in the pre-clinical setting for the identification of knee OA. Such a use would allow clinicians to more readily image patients suspected of knee OA and avoid the use of more involved medical imaging modalities such as CT, MRI, or X-ray. Screening is an essential process that most clinical practices offer for patients who may be fit the risk demographic for a particular disease [Koplas et al., 2008]. Early knee OA, for example, may not present easily identifiable symptoms, and screening offers an opportunity to identify the disease before further progression. In particular, opportunistic screening for obese and/or elderly patients may provide valuable insight into their knee joint health. It is important to note, however, that pre-screening is not used for final diagnosis, but rather to detect certain risk factors that contribute to later stages of the disease.

We anticipate that our system, with some modifications, could also be used to scan and reconstruct other regions prone to OA, particularly the hand and wrist, as well as commonly injured sites like the shoulder. While we did not explore the utility of US for non-OA injuries, the ACL and other ligaments are occasionally visible on US scans, so there may be some utility for rapid injury assessment as well.

The ethical concern, however, lies in the risk of producing incorrect results. A false positive may push clinicians toward recommending unnecessary actions moving forward, and a false negative may lead to a lack of treatment for a patient suffering from knee OA. Additional risks for imaging a patient may also include exposing the patient to unnecessary levels of US energy. The benefits to screening outweigh the negatives in that most believe that it is better determine one’s risk for an otherwise undetectable disease. In fact, many screening methods only produce approximately 50% true positives. This method would provide those in areas lacking available
screening a way to pre-screen for their risk of knee OA, and refer only at-risk patients for additional imaging studies. We do not anticipate any political ramifications of the design.
7 Final Design & Test Validation

7.1 Final Design

The final design consisted of two main sub-assemblies: the curvilinear rail and the fixture. The curvilinear rail sub-assembly included the following components (Figure 106).

1. Curvilinear rail: track to constrain motion of the probe
2. Probe sleeve: secure probe
3. V-rollers: secure probe sleeve onto the curvilinear rail and allow for motion along rail
4. Rail caps: joint the curvilinear rail to the fixture sub-assembly

![Figure 106: Final curvilinear rail sub-assembly design. A) Front view with the probe, IMU, and saline-bag system mounted onto a patient’s knee. B) Top view of the probe mounted onto the frame. The patient’s knee would enter the frame such that the top of the image represents the anterior and the bottom the posterior.](image)

The fixture consisted of two vertical 80/20 T-slotted extrusions bolted onto a wooden platform using right angle brackets. Sliders were mounted onto both rails to allow the vertical positioning of the curvilinear rail sub-assembly. The components for the fixture are as follows:
7.2 Wooden Dowel Phantom Reconstruction

The final design and reconstruction software were first tested on a wooden dowel phantom. A wooden dowel with a known diameter of 9.65 mm was submerged in a bag containing US gel and was imaged and reconstructed. The bag was placed in the center of the curvilinear rail and 20 to 25 B-scans were taken for each trial with an angular span of approximately 120° (Figure 108).
Figure 108: A wooden dowel of known diameter was suspended in a bag of US gel for the phantom reconstruction testing. The dowel was imaged roughly 20 to 25 times with an angular span of approximately 120°. Note that the scan window obtains a cross-sectional view of the wooden dowel. Total system verification was performed by acquiring the dowel’s radius from the resulting reconstruction and comparing it with the known diameter.

B-scans and angular displacement measurements were obtained through a custom MATLAB script and an Arduino sketch, respectively. The imaging radius was measured in the CAD model of the assembly to be approximately 39.9 mm, shown in blue in Figure 109.
Figure 109: CAD model of the final curvilinear rail sub-assembly used to determine the imaging radius. This was taken to be the radius formed between the center of rotation of the probe and the probe surface. The imaging radius is shown in blue, about 39.9 mm. The center of rotation is shown as a red dot. (left) Front view. (right) Side view.

Once reconstructed, the diameter of the dowel was measured using the \textit{imdistline} tool in MATLAB’s Image Processing Toolbox (Figure 110). The diameters for the three trials were 11.2, 9.4, and 9.7 mm yielding an error of 16.1, -2.6, and 6.6%, respectively.

Figure 110: Diameter of the reconstructed measured using MATLAB’s \textit{imdistline} tool. Pixel distance was found to be 107.35. Based on the mm per pixel ratio of 0.0838 for a depth setting of 3 cm, this pixel distance corresponds to 9.7 mm.
As shown in Figure 110, the reconstructed image only shows the outline of the top of the submerged dowel and is blurry. The partial outline is due to the lack of penetration of US waves through the dowel as well as the limited angular span at which the dowel was imaged. The blurriness is most likely due to the motion of the dowel within the phantom. Because the dowel was not set inside the gel-based phantom, the pressure from the probe during imaging may have displaced it. Such a displacement would have led to an inaccurate reconstruction of the dowel.

7.3 Human Knee Reconstruction

Following the dowel phantom, a human knee was imaged and reconstructed per the Human Subject Test protocol (Appendix H). The right knee of a subject, for whom we had an MRI of their right knee available was imaged and reconstructed. A total of four trials were conducted: two with US gel alone and two with the saline bag system in addition to gel (Figure 111).
Figure 111: Four trials of human subject testing were performed. (A) Two trials were performed with the saline bag system and (B) two trials were performed without the saline bag system. For all trials, a total of 20 to 25 B-scans were obtained with an angular span of approximately 120° beginning at the suprapatellar and ending at the medial region.

Once reconstructed, the femoral articular cartilage thickness was measured at five discrete locations (Figure 112).
Figure 112: The femoral articular cartilage thickness was measured at the locations
labeled A through E for every reconstructed image. Measurements at their equivalent
locations were taken from the MRI and compared.

Equivalent measurements of the femoral articular cartilage thickness were taken
from the MRI scan using Mimics (Figure 113).
The plane along which the image was taken is shown as the red horizontal line in Figure 114. It should be noted that the MRI was taken with the subject laying supine, thus his leg was fully extended, unlike the US scans where his knee was bent.
Measurements of the femoral articular cartilage thickness at all five locations for four trials of US imaging and reconstruction as well as the MRI are shown in Table 37. Because the articular cartilage thickness is not constant across the surface of the femur, each of the five distinct measurements was compared to the corresponding measurement from the reconstruction independently rather than acquiring a global average.

Table 37: Measurements of right knee femoral articular cartilage thickness measurements taken from four trials of US imaging and reconstruction and MRI.

<table>
<thead>
<tr>
<th>Location of Interest</th>
<th>Trial</th>
<th>A (mm)</th>
<th>B (mm)</th>
<th>C (mm)</th>
<th>D (mm)</th>
<th>E (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>MRI</td>
<td>5.41</td>
<td>5.07</td>
<td>3.57</td>
<td>2.21</td>
<td>1.89</td>
<td></td>
</tr>
<tr>
<td>Gel Only 1</td>
<td>*</td>
<td>*</td>
<td>*</td>
<td>*</td>
<td>*</td>
<td></td>
</tr>
<tr>
<td>Gel Only 2</td>
<td>4.0</td>
<td>2.7</td>
<td>*</td>
<td>*</td>
<td>*</td>
<td></td>
</tr>
<tr>
<td>Gel + Bag 1</td>
<td>3.1</td>
<td>2.8</td>
<td>2.8</td>
<td>1.8</td>
<td>1.5</td>
<td></td>
</tr>
<tr>
<td>Gel + Bag 2</td>
<td>3.4</td>
<td>2.6</td>
<td>2.4</td>
<td>1.8</td>
<td>1.6</td>
<td></td>
</tr>
</tbody>
</table>

[*] Reconstructed images were too blurry to take measurement

As indicated on the table, measurements were not obtainable for the majority of the reconstructed images from the gel only setup. This most likely resulted from unexpected motion of the probe during imaging. Given the irregular surface of the knee, the probe may have been displaced vertically as the probe was not constrained
in that direction. This would have resulted in a variable imaging radius, and, thus, an inaccurate reconstruction. In contrast, measurements were easily obtained for the gel and saline bag setup, albeit with large errors. This, again, was most likely due to the vertical movement of the probe, though not as severe as the setup with only the gel as the saline bag provided a deformable surface to somewhat standardize the imaging radius. In addition, because the subject’s knee was not constrained externally for both setups, the knee may have moved during imaging and introduced further errors into the reconstructed image.
8 Conclusion & Recommendations

In its current state, the final design is a proof of concept for the use of a curvilinear rail and panoramic reconstruction algorithm to visualize the FAC. In theory, the reconstruction algorithm is able to accurately reconstruct geometries given individual B-scans and their corresponding angles; however, because of inaccuracies arising from the fixture, the reconstructed images were not able to reconstruct the underlying geometry accurately. Given the current state of the design and accompanying software, there are several modifications that could be made to the current design:

Constrain probe vertically The first change to the design that should be made is to constrain the probe vertically in the probe sleeve. Though the sleeve prevents the probe from moving laterally, it does not vertically. This was most likely the cause of inaccurate reconstruction and needs to be addressed in future iterations.

Use of a modern US machine The US system used for this project was produced in the late 1980s and is currently a discontinued product. Because of this, the frames could not be exported efficiently, limiting the temporal resolution of B-scans. Furthermore, producers of modern research US systems, such as Verasonics, have built-in interfaces with MATLAB for ease of image acquisition and processing [“Verasonics” 2018]. Such a system would have alleviated many of the difficulties associated with exporting the output of the system into a readable format.

Replace IMU with a rotary encoder Through our component testing, the IMU was found to experience drift and did not have the sub-degree angular resolution desired. Compared to an IMU, a rotary encoder is able to provide more reliable feedback on angular displacement. However, a rotary encoder may present challenges given its increased bulkiness compared to an IMU.

Allow vertical adjustability of the probe sleeve In the current design, the probe sleeve holds the probe at a single offset from the rail. A design capable of vertically adjusting the probe would provide the user with the freedom to adjust the height of the probe to accommodate various knee sizes without having to adjust the sliders on the frame. For this, a rack and pinion system allowing discrete or vertical vertical positioning of the probe may be desirable.

Real-time reconstruction Because the current system performs post-acquisition
reconstruction, it is unable to provide the user with real-time feedback on regions that have already been imaged. Real-time reconstruction would allow the user to intuitively visualize regions that have not yet been imaged or need further imaging to increase the resolution. Given the large amounts of data that will need to be handled, such a software may need to be written in languages such as C or C++ instead of MATLAB.

**Additional degrees of freedom for rail** The current curvilinear rail design provides the probe with one degree of freedom along the path of the rail. In addition, the orientation of the probe only allows the reconstruction of a single plane of the knee. By tracking the rotation of the rail itself, the additional degree of freedom of the rail allows volumetric reconstruction as opposed to planar reconstruction.

**Motorized system for image acquisition** The current design requires the user to guide the probe along the rail by hand; such a system, however, is prone to introduce inaccuracies due to human errors and decrease the ease of use of the overall system. For this, the implementation of a servo or stepper motor into the design may alleviate any inaccuracies and difficulties associated with the use of the system.
References


REFERENCES


REFERENCES


doi: 10.1148/radiographics.17.2.9084086

doi: 10.1016/J.REHAB.2015.12.003


doi: [http://dx.doi.org/10.1016/j.sna.2015.09.025](http://dx.doi.org/10.1016/j.sna.2015.09.025)


doi: 10.1136/ard.16.4.494


Michalek, P., Donaldson, W., McAleavey, F., Johnston, P., & Kiska, R. (2013). Ultrasound imaging of the infraorbital foramen and simulation of the ultrasoun-
REFERENCES


National Aeronautics and Space Administration. (2000). *Anthropometry and Biomechanics.* Retrieved from [https://msis.jsc.nasa.gov/sections/section03.htm](https://msis.jsc.nasa.gov/sections/section03.htm)


170


REFERENCES


Appendix

The following appendices include all protocols used throughout the project.

A  Saline Standoff Bag System Construction Procedure

Materials:

1.5 m 4.7 mm inner diameter clear tubing

2 pcs 3.1 mm to 3.5 mm inner diameter straight tubing connectors

1 pc 3.5 mm inner diameter “Y” tubing connector

2 pcs Medium hose clamps

1 pc #10 septum cap

1 pc 1 L 0.9% saline solution bag

1 pc 250 mL 0.9% saline solution bag

Duct tape

Scissors

Tubing guillotine

Methods:

1. Cut tubing and assemble tube system.

   (a) Cut tubing about 30 cm from one end.
   (b) Attach the Y connector between the two pieces.
   (c) Feed each piece through its own hose clamp.
   (d) On the free end of each piece, connect the straight connectors.
   (e) Stopper the third port of the Y connector with the septum cap.
   (f) Close both hose clamps.

2. Connect the tubing to the saline bags.
(a) Over a sink, use a tubing guillotine to cut off the “snap-off” cap on the left hand port of the smaller saline bag.

(b) Quickly connect the straight connector on the longer segment of tubing to the port. Wrap with a couple of layers of duct tape.

(c) Repeat steps 2(a) - 2(b) with the larger saline bag.

3. Prime the tubing.

(a) Over a sink, open one hose clamp and remove the septum cap from the Y connector.

(b) Hold the bag upright with ports up on the side of the sink so most of the air is at the top of the bag.

(c) Gently squeeze the bag until fluid comes steadily out of the open port of the Y connector and no air is visible in the tubing.

(d) Quickly close the hose clamp and replace the septum cap.

(e) Repeat steps 3(a) - 3(e) with the other saline bag.

4. Bleed the larger bag.

(a) Open the hose clamp of the shorter segment (on the larger bag) and remove the septum cap.

(b) Hold the bag upright.

(c) Squeeze the bag until the desired thickness is reached, then close the clamp and replace the septum cap.

5. Test the system

(a) Place the bags on a flat surface.

(b) Open both hose clamps.

(c) Lift the larger bag above the smaller bag. The smaller bag should swell and there should be no leaks.

(d) Lower the larger bag below the smaller bag. The smaller bag should flatten with no leaking and minimal air entering the tubing.
Figure 115: Saline bag construction summary
B  Region of Interest Cropping Protocol

Protocol to determine cropping window for each depth on the HP ImagePoint. The protocol can be broadly categorized into image acquisition and cropping.

**Image Acquisition:**

1. Slide all Time Gain Compensation sliders to the highest setting.

2. Rotate the gain dial to the max setting.

3. Open and run the ‘imaqtool_vidaq.m’ file.

4. Wait until the small window appears to preview the ultrasound machine scan. It should mirror the ultrasound machine. If it does not appear, press the connectors together and try again.

5. Enter ‘1’ for frames to be acquired.

6. Check the quality of the frame by entering into the command window the following command:
   
   ```matlab
   imshow(rgb2gray(frames1));
   ```

7. If the entire frame is white, save as: ‘[DEPTH]_cm’. It will save as a ‘.mat’ file.

8. Repeat steps 2-7 for all depths.
Table 38: Frequency at given depth

<table>
<thead>
<tr>
<th>Depth (cm)</th>
<th>Frequency (Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>16</td>
<td>8</td>
</tr>
<tr>
<td>15</td>
<td>8</td>
</tr>
<tr>
<td>14</td>
<td>8</td>
</tr>
<tr>
<td>13</td>
<td>9</td>
</tr>
<tr>
<td>12</td>
<td>9</td>
</tr>
<tr>
<td>11</td>
<td>9, 10</td>
</tr>
<tr>
<td>10</td>
<td>10</td>
</tr>
<tr>
<td>9</td>
<td>10</td>
</tr>
<tr>
<td>8</td>
<td>11</td>
</tr>
<tr>
<td>7</td>
<td>12</td>
</tr>
<tr>
<td>6</td>
<td>12</td>
</tr>
<tr>
<td>5</td>
<td>13</td>
</tr>
<tr>
<td>4</td>
<td>14</td>
</tr>
<tr>
<td>3</td>
<td>15</td>
</tr>
<tr>
<td>2</td>
<td>16</td>
</tr>
<tr>
<td>1</td>
<td>17</td>
</tr>
</tbody>
</table>

Cropping

1. Binarize the image

![Original Image](image1.png) ![Binarized Image](image2.png)

Figure 116: Identification of cropping window for depth of 5 cm using MATLAB’s `bwboundaries` function. Each depth setting had a slightly different cropping window; therefore, the process was repeated for all relevant depth settings.

2. Use `bwboundaries` to identify boundaries of the cropping window
Figure 117: Identification of cropping window for depth of 5 cm using MATLAB’s `bwboundaries` function. Each depth setting had a slightly different cropping window; therefore, the process was repeated for all relevant depth settings.

3. Determine coordinates of four corners of cropping window

Table 39: Cropping window for corresponding depth setting

<table>
<thead>
<tr>
<th>Depth (cm)</th>
<th>Cropping Window [xmin xmax ymin ymax]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>[60 553 60 189]</td>
</tr>
<tr>
<td>2</td>
<td>[60 553 60 320]</td>
</tr>
<tr>
<td>3</td>
<td>[81 532 60 418]</td>
</tr>
<tr>
<td>4</td>
<td>[138 477 60 418]</td>
</tr>
<tr>
<td>5</td>
<td>[173 443 60 418]</td>
</tr>
<tr>
<td>6</td>
<td>[195 420 60 418]</td>
</tr>
<tr>
<td>7</td>
<td>[210 403 60 418]</td>
</tr>
<tr>
<td>8</td>
<td>[222 391 60 418]</td>
</tr>
<tr>
<td>9</td>
<td>[232 381 60 418]</td>
</tr>
<tr>
<td>10</td>
<td>[237 374 60 418]</td>
</tr>
<tr>
<td>11</td>
<td>[245 367 60 418]</td>
</tr>
<tr>
<td>12</td>
<td>[252 364 60 418]</td>
</tr>
<tr>
<td>13</td>
<td>[255 358 60 418]</td>
</tr>
<tr>
<td>14</td>
<td>[261 355 60 418]</td>
</tr>
<tr>
<td>15</td>
<td>[262 351 60 418]</td>
</tr>
<tr>
<td>16</td>
<td>[265 348 60 418]</td>
</tr>
</tbody>
</table>
C Phase 1 Imaging Study Protocol

1. Prepare US machine to the correct knob settings.
   (a) Toggle to ‘superficial mode’ on the US machine.
   (b) Press ‘B-Mode.’
   (c) Adjust the depth to between 2-5 cm depending on the probe location on the knee.
   (d) Adjust the gain dial such that the feature of interest can be seen. This, in combination with the time-compensation levers, can be adjusted to user preference.

2. Attach the ultrasound video USB prior to opening a MATLAB session. Type in `imaqtool` into the Command Window. Check that the session can recognize the video input and preview the Hauppauge window. A duplicate of the US machine screen should appear with real-time feedback. If not, unplug the USB, close MATLAB. Plug the USB back in, then open a new MATLAB session. Check that the session can recognize the video input.

3. Use goniometer to check angle of knee.
   (a) 90° from extension.
   (b) Have the subject sit on an adjustable height chair to ensure subject’s hips are level with knees, and feet are flat on ground.

4. Apply a generous layer of ultrasound gel to the probe and to the knee surface.

5. Use probe to identify the knee joint space while looking at monitor.

6. Locate the medial condyle and tap ‘Freeze.’

7. Using the caliper function, measure the distance for the medial AC.

8. Acquire a total of 3 measured distances.

9. Record these distances into the chart.

10. Acquire an image using MATLAB’s image acquisition toolbox.

11. Obtain a screenshot of the MPEG and insert into the chart.

12. Repeat step 5-7 for the center ridge and lateral AC.
D Phase 2 Imaging Study Protocol

Imaging study on same knee performed by same user with ASP:

1. Prepare ultrasound machine to correct knob settings.
   (a) Toggle to ‘superficial mode’ on the US machine.
   (b) Press ‘B-Mode.’
   (c) Adjust the depth to between 2-5 cm depending on the probe location on the knee.
   (d) Adjust the gain dial such that the feature of interest can be seen. This, in combination with the time-compensation levers, can be adjusted to user preference.

2. Attach the ultrasound video USB prior to opening a MATLAB session. Type in `imaqtool` into the Command Window. Check that the session can recognize the video input and preview the Hauppauge window. A duplicate of the US machine screen should appear with real-time feedback. If not, unplug the USB, close MATLAB. Plug the USB back in, then open a new MATLAB session. Check that the session can recognize the video input.

3. Use goniometer to check angle of knee
   (a) 90° from extension.
   (b) Have the subject sit on an adjustable height chair to ensure subject’s hips are level with knees, and feet are flat on ground.

4. Apply a generous layer of ultrasound gel to probe, knee surface, and both surfaces of ASP.

5. Place ASP on knee and press to ensure no air is trapped between knee-ASP and probe-ASP interface.

6. Use probe to identify the knee joint space while looking at monitor.

7. Locate the medial condyle and tap ‘Freeze.’

8. Using the caliper function, measure the distance for the medial AC.

9. Acquire a total of 3 measured distances.
10. Record these distances into the chart.

11. Acquire an image using MATLAB’s image acquisition toolbox.

12. Obtain a screenshot of the MPEG and insert into the chart.

13. Repeat step 5-7 for the center-ridge and lateral AC.
E  IMU Calibration Protocol

Proper calibration of the IMU is critical for obtaining accurate measurements of linear and angular displacement. Calibration of the IMU is performed as follows:

1. Connect IMU to laptop by USB.

2. Upload “MPU6050_calibration.ino” script to Arduino and wait for ‘done uploading.’

3. Open Serial Monitor (control+shift+m).

4. Set baudrate to 230400 (lower right-hand corner of the window).

5. Send any character to initialize calibration.

6. Copy and paste the gyroscope (3) and accelerometer (3) offsets displayed in the serial monitor into a temporary notepad / sticky note for safe keeping or take a snippet of the offsets.

7. Exit out of the Serial Monitor and close the calibration file

8. Open the “matlab_viz_6dof.ino” script to Arduino.

9. Enter the 6 offsets into the section starting at Line 11.

10. Upload the code onto the Arduino.

11. Open the Serial Monitor (control+shift+m).

12. Change the baudrate to 115200.

13. Press and send any random key to initialize the stabilization loop

14. Wait for the IMU to calibrate and verify it works.
F  IMU Data Collection Protocol

To acquire the measurements from the MPU-6050 in MATLAB, first follow the IMU calibration protocol. Once calibration is complete and the readings have been confirmed in the Serial Monitor, perform the following:

1. Close the Serial Monitor

2. Open MATLAB and run the “gyroviz.m” script

   (a) Check that the correct port is listed.

3. Note: If there is a port error, run the script again. If this does not work, unplug the USB, close and reopen MATLAB, and try steps 2-3 again.

4. To stop acquiring data, close the plot window or enter “Ctrl + C” in the command window.

5. Readings are stored in an n-by-4 matrix in the following format:

\[
\begin{bmatrix}
time & yaw & pitch & roll
\end{bmatrix}
\]
G  IMU Gyroscope Reliability Test Protocol

G.1  Angular Measurement Reliability Testing

The built-in gyroscope of the IMU MPU 6050 must be tested for accuracy and precision in calculating angular displacement. Prior to formal testing, it must first be tested for general repeatability. The axes of interest include the x-axis and the z-axis. Note that the IMU will be mounted onto the probe such that rotation about the x-axis of the IMU corresponds to when the probe aligns with the line made by the femur. Rotation about the z-axis of the IMU corresponds to when the probe is perpendicular to the line made by the femur. The follow protocol for initial reliability testing is as follows:

1. Choose one axis to rotate the IMU around: x or z.

2. Calibrate the IMU as per IMU Calibration Protocol.

3. Start acquiring IMU data as per IMU Data Acquisition in MATLAB protocol.

4. Once data acquisition is confirmed on the MATLAB command window, begin rotating the IMU about the axis of interest at a constant rate. Do not exceed an angular velocity 90 deg/s.
   (a) At 0, 90, 180, pause 1 second before continuing
   (b) Once the final angle of interest is acquired, reverse the movement and return to the start position.
   (c) Repeat Steps 4a and 4b for a total of 3 clockwise & counterclockwise cycles.

5. Exit out of the MATLAB program with a ctrl+c. Run the final section of the code with the appropriate labels.

6. Save the data of interest:

7. Save the pose vector as a .mat file.

8. Save the figure as a .fig and .png file for later access.

9. Repeat steps 1-6 for the other axis of interest

Once the figures have been acquired, analyze the results with the applied movement. Check that the angular displacement is about the correct direction, and that it matches somewhat consistently after visual inspection.
G.2 Formal Reliability Testing

Once the IMU gyroscope angle tracking was confirmed for a go-ahead with the initial reliability testing, formal testing could then be conducted. For this, the following setup was completed before beginning the tests. Materials include: a straightedge, two clamps, and a simple circuit holder.

![Test setup for determining reliability of the IMU's gyroscope.](image)

1. Align the IMU along the edge of the secure straight-edge. This will be the “starting” position.

2. Calibrate IMU per IMU Calibration Protocol. Align the IMU with the appropriate outline.

3. Start acquiring IMU data per IMU Data Acquisition in MATLAB protocol

4. Once data acquisition is confirmed on the MATLAB command window, begin rotating the IMU about the axis of interest at a constant rate. Do not exceed an angular velocity 90 deg/s.

   (a) Once the final angle of interest is acquired, reverse the movement and return to the start position.

   (b) Repeat Steps 4a and 4b for a total of 3 clockwise & counterclockwise cycles.

5. Exit out of the MATLAB program with "ctrl+c." Run the final section of the code with the appropriate labels.
6. Apply the custom MATLAB code to filter, extract the two angles of interest, and analyze for average error. Record the high-angle and low-angle average error.

7. Save the data of interest:

8. Save the pose vector as a .mat file.

9. Save the figure as a .fig and .png file for later access.

10. Repeat steps 3-7 for a total of 3 iterations.

Figure 119: Test setup for determining reliability of the IMU’s gyroscope.
H  Human Subject Test Protocol

The following protocol is intended for testing of the designed fixture with a human subject and comparison to MRI images of the same subject.

1. Place human subject in chair such that their knees rest comfortably at 90° flexion. If possible, use a chair with adjustable height and wheels with locks.

2. Without the fixture, use the probe to find the knee joint space and adjust the settings on the US machine. Though this varies per machine, make sure that the depth is around 3-5 cm. Select B-Mode, and change the gain dial such that the joint space is visible. If necessary, use the time-gain compensation sliders to adjust the scan as needed. Place the probe in the probe holder.

3. If using the saline standoff system, affix the system to the subject’s knee

   (a) Open the hose clamps to allow for flow between the two bags. Deflate the large saline bag as much as possible. Close the hose clamps.

   (b) Apply US gel to the subject’s knee and both sides of the saline bag.

   (c) Place the bag over the subject’s knee.

   (d) Wrap one Velcro strap up and around the thigh.

   (e) Wrap the second Velcro strap around the shank of the leg close to the knee.

   (f) Open the hose clamps to allow the larger bag to inflate to the desired thickness. Lock the hose clamps once this thickness is reached.

4. Adjust the height of the sliders on the fixture such that the hinges are aligned vertically with the subject’s femoral condyles. Lock the sliders in place by tightening the wing-nuts on the frame.

5. Align the subject’s heel to the edge of the wooden base, but check that the patient’s knee remains at 90°.

6. With an adjustable wrench, loosen the nuts on the mechanical frame responsible for holding the rail in position. Adjust the angle of the rail such that the probe is touching the surface of the bag. Tighten the nuts once the desired position is achieved.
7. Calibrate the IMU in the sleeve to the side of the fixture. Plug in the offsets acquired from the calibration script into the positioning script. Run the position script and do not continue until the yaw, pitch, and roll angles appear.

8. Affix the IMU to the probe.

9. Image the knee joint space width.
   
   (a) Start at the medial-most side of the joint.
   
   (b) Record the angle read by the IMU and acquire the image from the screen.
   
   (c) Move the probe 1-5° along the rail and repeat recording.
   
   (d) Record approximately 20-30 images, or until the structure of interest is fully imaged.

10. Remove the device from the patient’s knee and provide a towel to remove the US gel.