Automatically-optimized local phase features of ultrasound images: first clinical study

HACIHALILOGLU IH[1], ABUGARBIEH RA[1], HODGSON AH[2], ROHLING RR[1,2], GUY PG[3]

[1]*Department of Electrical and Computer Engineering, University of British Columbia, Vancouver, BC, Canada*
[2]*Department of Mechanical Engineering, University of British Columbia, Vancouver, BC, Canada*
[3]*Department of Orthopaedics, University of British Columbia, Vancouver, BC, Canada*

ilkerh@ece.ubc.ca

Introduction: The most common imaging techniques used to guide orthopaedic interventions are X-ray-based [1], but ultrasound (US) imaging offers several advantages for intraoperative imaging as it is portable and produces realtime 2D or 3D images without exposing either the patient or surgical team to ionizing radiation. However US imaging has several limitations: user-dependent image quality, limited field of view, orientation dependence, and high levels of speckle. Bone boundaries typically appear as blurred bands with a thickness of 2-4mm which makes it difficult to accurately and automatically detect the bone surface [2]. Previously, we have shown it is possible to extract clear bone surfaces from 3D B-mode ultrasound images based on local phase features [3-5] with an accuracy of less than 0.4mm [5]. The local phase images are extracted by filtering the B-mode US image in the frequency domain with a Log-Gabor filter. Although successful results were achieved, the choice of filter parameters do affect the local phase method's sensitivity to typical US artifacts. We have therefore recently proposed a framework to automatically select these parameters using the bone surface information obtained from the B-mode US images [6].

In this study we show our first clinical results using local phase information to identify fractures from B-mode images using automatically-selected filter parameters. A standard pre-operative CT image was also available during the study which provided the gold standard surface match.

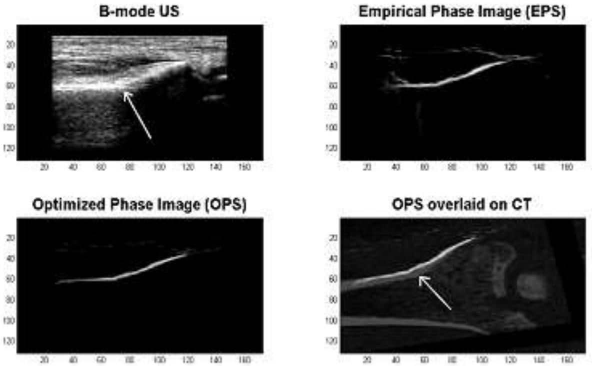
Methods: We obtained both CT scans and US images from a patient admitted to Vancouver hospital for a suspected radial fracture (this is the first case in a more extended study now underway). The CT scan was obtained during the standard clinical assessment. Once the presence of a fracture was confirmed, the patient was informed about the study and invited to participate. Informed consent for the use of 3D US and access to the previously-acquired CT scan was obtained. All US examinations in this clinical study were performed with a commercially-available real-time scanner (Voluson 730, GE Healthcare, Waukesha, WI) with a 3D RSP5-12 transducer. This is a mechanized probe in which a linear array transducer is swept through an arc range of 20°. During the scan, standard US coupling gel was applied to the skin over the scan sites for dorsal, volar, and radial views. The US image was processed using the algorithm described in [6].

The reference bone surface was extracted from the CT image by a relatively simple thresholding method as described in [7] and the US image was matched to the CT surface by matching selected anatomical landmarks (note: in our previous ex vivo bovine study, we used implanted fiducials to perform the registration, but fiducials could not be used in this clinical study) and computing the rigid body transformation using the AMIRA (TGS, San Diego, USA) landmark-based rigid registration algorithm. The algorithm transforms the input image (CT dataset) by applying a global translation and rotation by minimizing the sum of the squared distances between the corresponding fiducial points from the US dataset. The resolution of the CT volume was 0.35mm´0.35mm´0.9mm and the one of the US was 0.19mm in all directions.

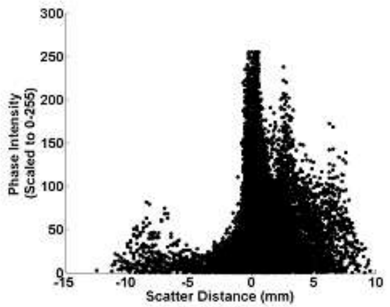
Following registration, we computed a signed distance map around the bone surface contour extracted from the CT image (positive indicates higher in the image than the bone surface). We then transformed each non-zero value in the phase-processed US image to its corresponding location in the CT image and identified the distance value associated with this location. This produced a set of intensity/distance pairs. High intensity values confined to a zone near zero distance would indicate an accurately located surface.

Results: Figure 1a shows a 2D slice of the pre-operative CT image and the corresponding B-mode US image on the top row. The bottom row in Fig1.a shows the extracted local phase images obtained using the Log-Gabor filter with empirical filter parameters (bottom left image) and the optimized filter parameters (bottom right). From the images we can clearly see the importance of filter parameter selection. The local phase symmetry method with the optimized filter parameters is less sensitive to typical US artifacts and extracts sharper bone boundaries.

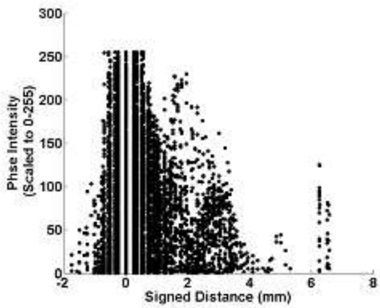
This observation is also confirmed by comparing the scatter plots presented in Figure 1b and c. It is visually apparent that the PS image obtained using the optimized parameters is essentially free of typical US artifacts or soft tissue interfaces compared to the PS obtained using the empirically-set parameters.



(a)



(b)



(c)

Fig. 1. (a) B-mode US image(top left), local phase image obtained using empirical Log-Gabor filter parameters (top right), local phase image obtained using automatic Log-Gabor filter parameters (bottom left), and overlay of optimized local phase image with gold standard CT surface. White arrows show the location of the fracture. (b) Scatter plot of local phase intensities obtained using empirical filter parameters and corresponding distance measure on registered CT image. (c) Scatter plot of local phase intensities obtained using optimized filter parameters and corresponding distance measure on registered CT image.

We also evaluated the correspondence between the two surfaces by identifying the location of the peak intensity pixel in each vertical column of the 3D US data set. This value was 0.94mm (std 1.54) for PS surfaces extracted using empirical filter parameters and 0.37mm (std 0.74) using the optimized filter parameters.

Discussion: We have demonstrated that a 3D US image processed using automatically-selected filter parameters can produce bone surfaces which lie well within 1 mm, on average, of the bone surface estimated from CT images. Our current study is investigating the hypothesis that US images processed in this way can be used to detect fractures reliably in the emergency room.

References

1. Sugano N. Computer assisted orthopaedic surgery, Journal of Orthopaedic Science, 2003: 442-448,
2. Jain A.K., Taylor R.H. Understanding bone responses in B-mode ultrasound images and automatic bone surface extraction using a bayesian probabilistic framework. Proc. of SPIE Medical Imaging, 2004: 131-142.
3. Hacihaliloglu I et al. Bone segmentation and fracture detection in ultrasound using 3D local phase features. MICCAI 2008: 287-295
4. Hacihaliloglu I et al. Enhancement of bone surface visualization from 3D ultrasound based on local phase information, IEEE International Ultrasonics Symposium 2006: 21-24
5. Hacihaliloglu I et al. Bone Surface Localization in Ultrasound Using Image Phase Based Features, Ultrasound in Medicine and Biology, 2009, volume 35, number 9: 1475-1487.
6. Hacihaliloglu I. et al. Automatic Data-Driven Parameterization for Phase-Based Bone Localization in US Using Log-Gabor Filters, International Symposium on Visual Computing, 2009: 944-954.
7. Hacihaliloglu I. et al. Assessing Surface Localization Accuracy in 3D Local Phase Ultrasound Images Using CT Reference Images, Computer Assisted Orthopaedic Surgery (CAOS), 2009: 137-140.