Published in final edited form as:

J Ultrasound Med. 2019 May ; 38(5): 1361–1365. doi:10.1002/jum.14802.

The effect of a thin fluid layer on surface wave speed measurements: a lung phantom study

Jinling Zhou, PhD and Xiaoming Zhang, PhD

Department of Radiology, Mayo Clinic, 200 First St. SW, Rochester, MN 55905, USA

Summary

Lung ultrasound surface wave elastography (LUSWE) is a novel technique that measures superficial lung tissue elastic properties. A thin pleural fluid layer covers a lung but its effect on lung measurements in LUSWE is unknown. We modeled a lung and pleural fluid with sponges and a thin layer of ultrasound transmission gel. Sponge surface wave speeds measured from LUSWE were compared for sponges without and with the thin ultrasound gel layer, at three wave excitation frequencies. The comparison showed that the sponge surface wave speed measurements were not affected by the thin layer gel.

Keywords

LUSWE; surface wave speed; pleural fluid; lung; lung phantom

Pulmonary fibrosis is a chronic progressive lung disease where interstitial lung tissues between air sacs (alveoli) are damaged, scarred and hardened.^{1,2} A hardened lung causes difficulties in breathing and few patients survive the disease after 3 to 5 years of diagnosis.³ The disease currently is affecting millions of people worldwide.⁴ High-resolution computed tomography (HRCT) is the clinical standard for the diagnosis⁵ but the ionizing radiation from HRCT⁶ may increase the risk of cancer.⁷ In search of a non-invasive alternative, we have invented lung ultrasound surface wave elastography (LUSWE),^{8–11} which assesses pulmonary fibrosis by measuring the surface wave speed of superficial lung tissue.

In LUSWE, a local, short (e.g. 0.1 second) harmonic vibration is generated on the skin of chest wall in an intercostal lung space with a handheld shaker. An ultrasound probe is aligned about 5 mm away from the local vibration excitation in the same intercostal lung space to measure the generated surface wave propagation along the lung. The radio-frequency (RF) data of ultrasound echo from the tissues are obtained first. By demodulation of the RF data using quadrature detection, the IQ data of ultrasound signals are processed. The IQ data consist of 2D intensity information for the duration of the vibration excitation. Using cross-correlation analysis of the ultrasound IQ data, the motion at a location in the conventional ultrasound B-mode imaging can be calculated and the wave propagation in the ultrasound imaging area can be analyzed.^{12,9} The surface wave propagation is used to study

the surface tissues; and the shear wave propagation is used to study the subcutaneous tissues. We have demonstrated that the shear wave propagation in the subcutaneous tissues was directly related to the surface wave propagation but 5% faster.¹³ Lung is an organ filled with air and covered by a thin layer of fluid (i.e. pleural cavity or fluid).¹⁴ However, the effect of the thin pleural fluid layer on lung measurements is unknown. Our objective in the current research is to model the thin pleural fluid layer effects on lung for the LUSWE measurement.

A lung phantom model was constructed, for which the lung tissues and the pleural fluid layer were modeled by sponges and a thin layer of ultrasound transmission gel. Low cost sponges have been used in multiple researches to model a lung.^{15–18} Lung has a spongy porous structure; sponges, as demonstrated by Christian Blüthgen *et al.*, can reproduce A- and B-line artifacts observed in lung ultrasound images.¹⁷ A polyurethane sponge was reported to be similar to a lung in acoustic impedance, porosity, sound velocity, etc. by Soldati *et al.*¹⁸ Cellulose sponge pieces likewise have been used by Molinari *et al.* for pulmonary edema modeling.¹⁵ Capable of reproducing similar surface wave speeds in patient study, two cellulose sponges were used as our lung phantoms. Similarly as in LUSWE, a shaker was used to generate the 0.1 second local harmonic vibration on the sponge surfaces and to initiate the wave propagation in the sponges. The wave was captured by an ultrasound probe and recorded as IQ data. The movement of the locations in the 0.1 second was Fourier transformed on to the inversed space, where the phase shift was calculated from cross-spectra and used to determine the sponge surface wave speeds.^{8–11} The surface wave speeds for sponges with and without a thin layer of gel were compared to model the effect of plural fluid layer on lung surface wave measurements in LUSWE.

Among the sponges that were tested, cellulose sponges resembled the human lung LUSWE results the most. Two sizes of cellulose sponges were used to confirm the repeatability of our results. The sizes of the two sponges are 6"x3.6"x0.9" and 7.7"x4.2"x1.5" (Ocelo utility sponge, 3M, St. Paul, MN, USA). Aquasonic 100 ultrasound transmission gel (Parker laboratories, Inc., Fairfield, New Jersey, USA) was applied to the sponge surfaces to model the pleural fluid. Similar amount of gels were smoothed out on a similar sized area of the two sponges to reduce the gel thickness difference on the two sponges. These lung phantoms are put into a similar setup as used in LUSWE.⁸ Figure 1 illustrates the experiment setup for measuring the phantoms instead of a patient. A large rubber was put under the sponges to reduce the reflection of sound waves from the underneath hard surface. Aquaflex ultrasound gel pad of 15 mm thick (Parker laboratories, Inc., Fairfield, New Jersey, USA) was placed atop the sponges to simulate the intercostal muscle. The moisturized gel pad was soft which matched the sponge well. A shaker (Model: FG-142, Labworks Inc., Costa Mesa, CA, USA) was positioned on the sponge surface to generate harmonic vibrations on the sponges. Three excitation frequencies 100, 150 and 200 Hz were used and each vibration lasted 0.1 s. A linear array ultrasound transducer L11–4V with a central frequency of 6.428 MHz was put above the gel pad to measure the surface displacement of sponge in response of the mechanical excitation. The sponge displacement data were collected for the whole duration of the shaker vibration through a vantage ultrasound system by Verasonics (Verasonics, Kirkland, WA, USA) at a frame rate of 2 thousand per second and stored as the IQ data. The information recorded for each frame can be plotted as a B-mode (B stands for brightness)

image. The motion at each pixel of the B-mode image can be measured using the ultrasound tracking technique. Through analyzing the correlations of the motions at several locations, the surface wave speed of the phantom is obtained.

Two typical ultrasound B-mode images for a sponge without and with the thin layer gel were shown in Figure 2. The image for sponge without the thin layer gel showed multiple A-lines, which are caused by the reverberation of sound waves in the gel pad between the ultrasound probe and the sponge surface. Adding a thin layer of gel erased the A-line artifacts and introduced B-lines as demonstrated in Figure 2 (B). These B-lines were possibly created by the ultrasound trapped in between the air sacs and the thin layer gel. The presence and behaviors of A- and B-line artifacts are common features observed in lung ultrasound images, which suggests that these sponges can be used as lung phantoms.

The harmonic vibrations from the shaker initiated the surface wave propagation on the sponge. B-mode images were taken consecutively by the ultrasound probe to capture the motion of the waves. The behaviors of the waves in the real space were Fourier transformed onto the inversed space for information such as frequencies, phase shift etc. Because the surface wave speed is determined not only by the wave frequency but also by the stiffness and density of the traveling media,⁸ we can study the ultrasound gel's effect on the sponge stiffness through its effects on the surface wave speeds. The surface wave speeds were estimated by analyzing the phase change of the waves $c_s = \omega|\Delta r/\Delta\phi|$ across a distance. The angular frequency ω is derived from the shaker frequency. ϕ is the phase change over the distance r of two measured locations, which is calculated through a cross-spectrum method⁹. The accuracy of the surface wave speed estimation can be improved by involving more locations. Eight locations, marked by blue dots in the ultrasound images in Figure 2, were used for each speed analysis. The r/ϕ ratio among these locations were calculated using a linear regression model $\Delta\phi = -\alpha\Delta r + \beta$, where $\Delta\phi$ was the linear regression value of multiple ϕ , α and β were the regression coefficients. The surface wave speed therefore is $C_s = \omega/\alpha$. Figure 3 is a typical graph we used for speed analysis, which also well illustrates the regression process. The phase delay and distance of each selected location with regards to the first location was plotted. The phase delay and the distance were then fitted to a linear relationship. The slope coefficient for the linear fit is α .

The above measurement and speed analysis were repeated three times for each shaker frequency. The averaged results are summarized and shown in Figure 4 along with our previously reported patient study data (The human study here was approved by the Mayo Clinic Institutional review board).¹⁰ The surface wave speeds for sponge without a thin layer gel are in blue, the gel-covered ones are in yellow, and the statistical average for ten patients are in green. The uncertainties of the speeds were marked by black vertical bars. As it is shown in the figure, the surface wave speed of both sponges increased with the shaker frequency. This trend has been observed in our patient study,¹⁰ which further validates the use of these sponges as lung phantoms. T tests have been carried out at $p < 0.05$ between measurements done at different frequencies to test the significance of the increase. The increase with frequency was found significant for all measurement pairs except for the 150 and 200 Hz pair for the sponge 1 without gel. Whereas the increase of the surface wave

speeds with shaker frequency was significant, t test between the without and with gel cases revealed that the surface wave speeds for sponges without and with the thin layer gel were not significantly different ($p > 0.05$). The thin layer ultrasound gel applied on the sponge surface therefore does not noticeably affect the sponge surface wave speed measurements. This phantom modeling supports the indication that the pleural fluids might not alter significantly the lung surface wave speeds. However, the structure of a lung is more complex than the sponge. The material makeup for the lung and pleural fluids are different from our sponges and the ultrasound gel. Further investigations are needed to confirm this result for human lung.

To summarize, we have modeled the pleural effects on the lung through a phantom study. Two sponges of different sizes were used as the lung phantom and a thin layer of ultrasound transmission gel was applied on top of these phantoms to model the pleural fluid. We have observed ultrasound A- and B- line artifacts in these phantoms, as well as the increase of the surface wave speed with the shaker frequency. These observations and results are consistent with clinic ultrasound images of lung and our previous patient study, which validated our use of these sponges as lung phantoms. There were no significant differences between the surface wave speeds for a sponge without and with the thin layer gel, from which we conclude that the thin layer gel does not affect the surface wave speed measurements of the sponge. The surface wave speed of the sponge is mainly determined by its own elastic properties. Since the phantom model is a simplified version of a lung and a thin pleural fluid layer, the broad indications about the thin pleural fluid layer's effect on the lung needs further investigation and validation.

Acknowledgement

This work is funded by NIH R01HL 125234. The authors would like to thank Boran Zhou for assisting the experiments.

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Figure 1:

Experiment setup. An ultrasound probe (transducer) was used to measure the wave generated on the sponge surface by a shaker.

Figure 2:

Two typical ultrasound B-mode images for a sponge without (A) and with (B) the thin layer gel. The uppermost horizontal line in image (A) is the sponge surface. The parallel white lines below are the ultrasound artificial A-lines caused by the reverberation of ultrasound in between the sponge surface and the ultrasound probe. The reverberations of ultrasound between the thin layer gel and air sacs, however, eliminated A-lines and produced B-lines instead, one of which is circled in image (B).

Figure 3:

Phase delay—distance relationship for the selected locations. The surface wave speed was calculated based on the regression results of the slope. Note that the first location is set at the origin. The phase delay and distance of the other locations are relative to the first location.

Figure 4:

Sponge surface wave speeds measured for (A) sponge 1 and (B) sponge 2, at three shaker frequencies 100, 150 and 200 Hz. The patient data are the statically average for ten patients with interstitial lung disease, a broad category of lung diseases which includes pulmonary fibrosis.