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Improved Detection of Kidney Stones Using an Optimized Doppler Imaging Sequence

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Abstract—**Kidney stones have been shown to exhibit a "twinkling artifact" (TA) under Color Doppler ultrasound. Although this technique has better specificity than conventional B-mode imaging, it has lower sensitivity. To improve the overall performance of TA as a diagnostic tool, Doppler output parameters were optimized *in vitro*. The collected data supports a previous hypothesis that TA is caused by random oscillations of multiple micron-sized bubbles trapped in the cracks and crevices of kidney stones. A set of optimized parameters were implemented such that the acoustic output remained within the FDA approved limits. Several clinical kidney scans were performed with the optimized settings showing improved SNR relative to the default settings.**

Keywords— Ultrasound, Doppler, kidney stone, detection, cavitation, optimization

I. Introduction

Kidney stone disease affects 11% of the population in the US [1] with a recurrence rate of 35-50% within 5 years [2]. Typical diagnostics for kidney stone disease include computed tomography (CT) and KUB x-ray, and can lead to a considerable radiation exposure, particularly in vulnerable populations (e.g. children) and recurrent stone formers. Though ultrasound is also used, it suffers from a broad range of sensitivity (78%-96%) and specificity (31-100%) in the detection of stones [3,4].

A method for improving the sensitivity and specificity of ultrasound is to leverage an imaging artifact that kidney stones viewed under Color Doppler appear to "twinkle". With this phenomenon, the color-coded velocity estimation fluctuates randomly throughout the entire Doppler color map. Studies have shown that although the sensitivity of the twinkling artifact (TA) is lower than B-mode (56% vs 71%), the specificity is much greater (74% vs 48%) [5]. However, one could use B-mode to find a suspected region of a possible kidney stone and then test the region with Doppler to see that it twinkles to improve the overall accuracy of detection.

In addition to testing the efficacy of TA as a diagnostic tool, there has been research to determine the mechanism of the TA with the intention of improving the sensitivity. Theories have ranged from phase jitter or saturation of the hardware to motion or reverberation of the stone. Our group has hypothesized the existence of micron-sized bubbles trapped in the cracks and crevices on the stone [6]. To understand why this would cause TA, one needs to understand that Color Doppler measurements are designed to be sensitive to weakly scattering blood cells and filter out the strongly scattering vessel wall. This is typically done by first filtering the Doppler pulse ensemble with a high-pass wall filter and then the velocity is calculated from the phase difference between the pulses. If the target has some randomness in the scattering, then the phase and amplitude will have randomness as well. In the case of a kidney stone, multiple bubbles trapped in cracks or crevices can oscillate from a strong incident wave, such as a Doppler pulse. Since a Doppler pulse is multiple cycles in each pulse, the initial part of the pulse excites the bubbles and then the latter part of the wave scatters back randomly with the collective random growth and collapse of the bubbles. The wall filter removes the bright scattering signal from the stone itself, leaving only the random backscatter signal from the bubbles. This leads to random phase delay between pulses, a random velocity estimate, and thus a random color representation or "twinkling" when displayed. This hypothesis was tested by the disappearance of TA as ambient pressure was increased to suppress bubble activity during stone imaging [6].

Under this hypothesis, we aimed to improve the sensitivity and specificity of TA as a diagnostic tool for kidney stone detection. The approach has two-parts:

1. Enhance the random bubble activity without exceeding FDA acoustic output limits. This will improve the sensitivity of TA.
2. Filter out blood flow and motion artifacts that typically appear with probe motion during a Color Doppler imaging. This will improve the specificity.

II. Materials and Methods

To allow for full control over the Doppler imaging hardware and software, we used a V-1 Verasonics Data Acquisition System (VDAS, Verasonics Inc., Redmond, WA, USA). The device is programmed and controlled through a host computer (HP Z820, Hewlett Packard, Palo Alto, CA,

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USA) using MATLAB R2011b (Mathworks, Waltham, MA, USA). The system is programmed to work with the ATL HDI C5-2 ultrasound imaging probe (Philips Ultrasound, Andover, MA, USA).

An agar-glycerol based soft-tissue mimicking phantom (fig. 1) was made per IEC guidelines [7]. The phantom had 5 cm of material between the probe and the targets, and a 1 cm fluid filled void around the targets. This was sandwiched with 4 cm of material and an acoustic absorber on the bottom to prevent reflections. A 4 mm calcium oxalate monohydrate stone extracted from a kidney stone patient and a 4 mm glass sphere were used as targets. The probe was aligned with the targets such that the brightest hyperecho from both targets was achieved in a B-mode scan. The glass sphere was used as a reference Doppler power value, as its smooth surface does not have any bubbles trapped on its surface, and therefore, is a stable backscattering target. Therefore, as parameters were changed (Table 1), any change in the Doppler power of the glass sphere would be due to some other effect.

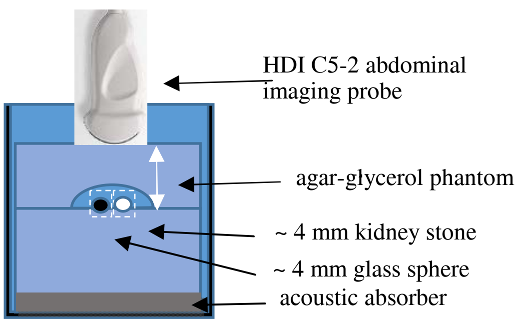

Fig. 1. Experimental Setup

A plane-wave Doppler imaging sequence was used with each parameter tested individually. The digitized signal was monitored to make sure that the A/D acquisition was not saturated, since this can also cause a twinkling artifact.

TABLE I. OPTIMIZATION PARAMETERS

Parameter:	Default:	Range:
# Cycles/Pulse	3	0.5 – 7.5 cycles
# Pulses/Ensemble	14	4 – 25 pulses
Transmit Angle	0 deg	-45deg – +45deg
Pulse Repition Frequency (PRF)	4000Hz	500Hz – 4000Hz
Doppler TX Voltage	20Vp	5Vp – 35Vp

IQ data was collected after the Verasonics software beamforming process. The first two pulses in the ensemble were dropped and the remaining pulses were high-pass wall-filtered by a quadratic regression curve fit method. Since the magnitude of the TA is required for optimization, Doppler power was calculated for each pixel over the entire imaging plane. The stone and glass sphere positions were then manually selected and the average Doppler power/pixel was calculated for a 5 mm x 5 mm square region centered on the selected target. A 10 mm x 10 mm square region, also centered on the target but excluded pixels from the target's ROI, was used as the "noise" value for calculating the effective SNR of the TA. Three acquisitions were collected for each set of parameters and the SNR of the stone was plotted along with that of the glass sphere as reference.

III. RESULTS

A. # Cycles/Pulse:

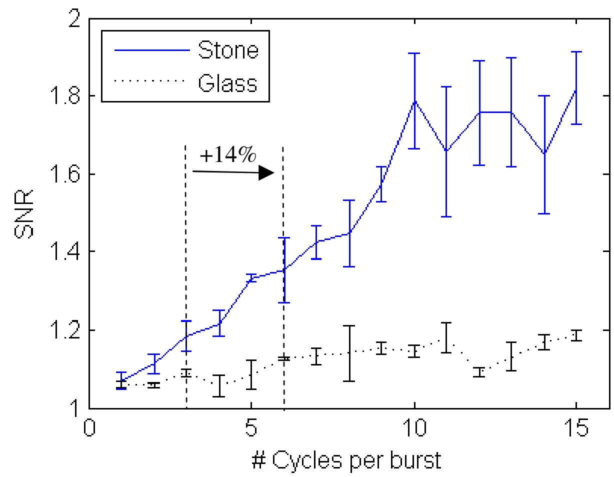

Fig. 2. # Cycles/Pulse. An increase in the number of cycles from 3 to 6 results in a 14% improvement in SNR. The effect seems to saturate when the pulse is 10 cycles or longer.

Increasing the number of cycles for each pulse improved the SNR linearly. This effect supports the theory of micron-sized bubbles since increasing the number of cycles of ultrasound would generate an increase in random bubble activity. The downside to longer pulses is a decrease in axial resolution, but this is an acceptable sacrifice since we are using twinkling to detect the stone and B-mode for the actual imaging.

B. Doppler Transmit Angle:

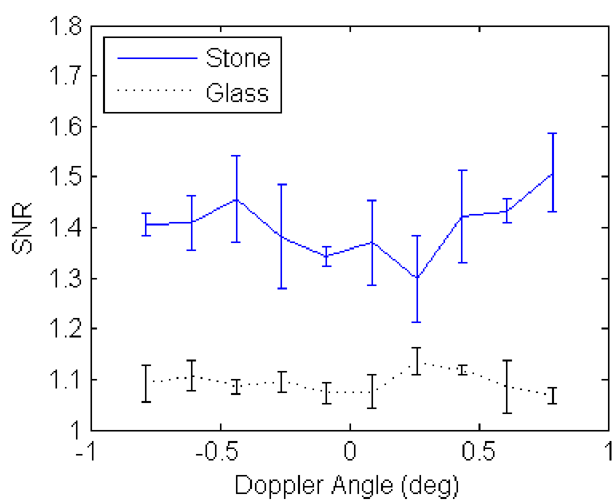

Fig. 3. Doppler Transmit Angle. Angle does not affect SNR.

Varying the transmit angle of the Doppler ensemble did not have a significant effect on increasing the SNR. This also supports the bubble theory because micron sized bubbles should have no angle dependence on their backscatter.

C. Pulse Repitition Frequency (PRF)

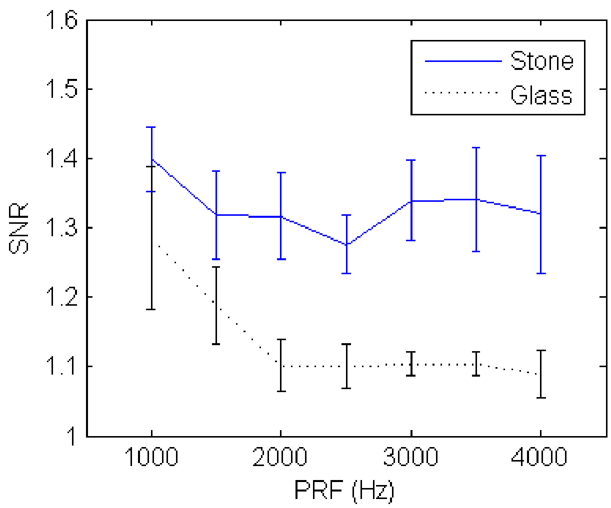

Fig. 4. Pulse Repetition Frequency (PRF). PRF does not affect SNR.

SNR remained constant over the tested PRF range. This is explained by the decay time for a micron-sized bubble being much shorter than the period between pulses. Therefore, no pulse should interfere with a prior or subsequent pulse. This independence of the PRF on SNR allows for a maximum PRF setting dependent on imaging depth. This would increase the range of the velocity measurement, which will improve the efficacy of the wall filter for removing motion artifact and low velocity blood flow.

D. Amplitude:

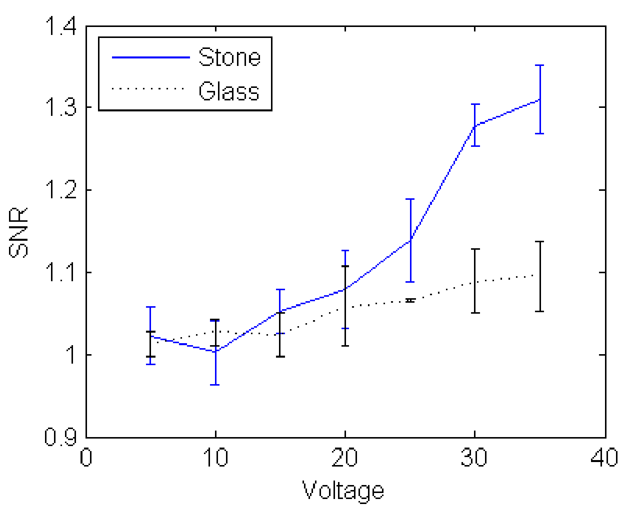

Fig. 5. Doppler transmit voltage.

Increasing the transmit amplitude of the Doppler signal increases the SNR for both the stone and the glass sphere, though the stone has more significant improvement. The overall increase in SNR is due to the increase of the backscatter signal over thermal noise. The greater increase from the stone compared to the glass sphere is due to the bubbles having a greater response to the change in the acoustic output.

E. # Pulses/Ensemble

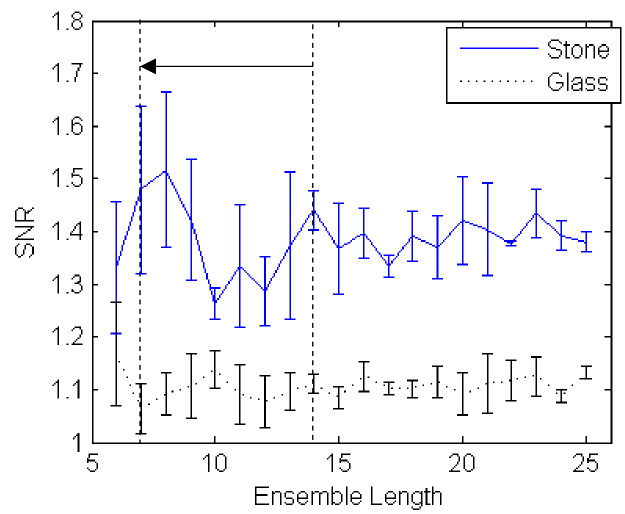

Fig. 6. Ensemble Length. Decreasing from number of pulses from 14 to 7 per ensemble does not change the SNR.

Ensemble length also did not have an effect on the SNR since the period between pulses is longer than the bubble decay time. However, the wall filter operates differently depending on the number of samples, and too short of an ensemble will begin to filter out the random bubble activity signal as well. Further research could be directed in selecting a wall filter to work better with a minimal ensemble length.

F. Clinical data with preliminary optimized settings

Based upon the *in-vitro* work, new output parameters were programmed in to the Doppler imaging sequence for human trials. Doubling the number of transmit cycles from the default of 3 to 6 increased the SNR 14%. To maintain the same acoustic output intensity, the number of pulses in the ensemble was reduced from 14 to 7, which had no effect on the SNR. A total of five frames of data were collected from two different human patients; two from one patient and three form another (Table II). An example frame for both B-mode and TA are shown in Fig. 7.

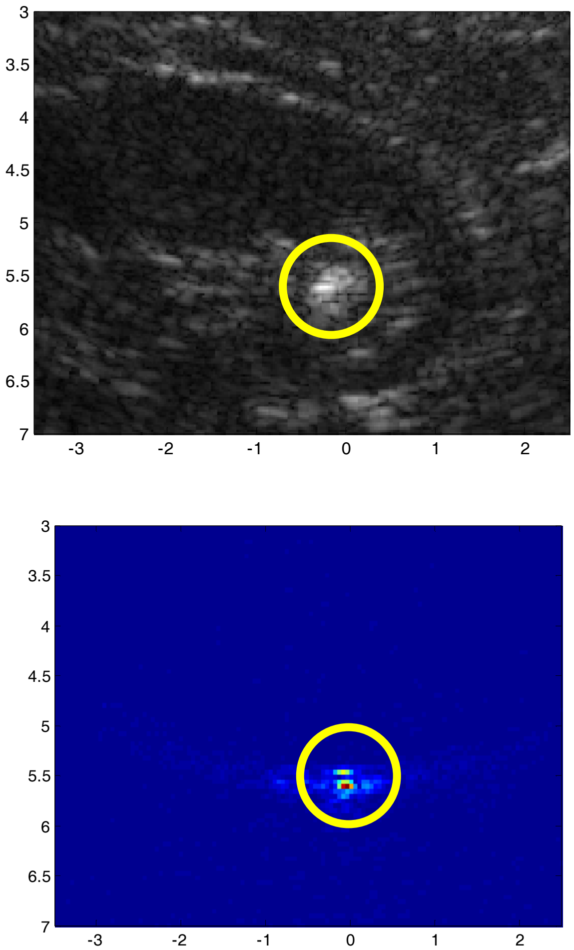

Fig. 7. (Top): B-mode image showing a kidney outline and a 3 mm stone appears as a bright hyperecho. (Bottom): the Doppler power map with optimized parameters. Only the kidney stone is visible in this image with some slight banding artifact.

TABLE II. PRELIMINARY CLINICAL RESULTS

SNR	TA	B-mode
Mean	11.1	2.6
Std. dev.	11.4	0.3
Min.	2.11	2.3
Max.	29.4	3.1

The large standard deviation of the TA SNR needs to be investigated further; it is much higher than the measurements *in vitro*. Possible factors include motion from patient breathing or variability in how well the stone is positioned in the imaging plane. However, it should be noted that the frame with the lowest SNR was on the same order as the B-mode detection and the maximum SNR was an order of magnitude higher. Therefore, it can be suggested that depending on the acquisition frame, TA has similar if not much better sensitivity compared to B-mode.

IV. Conclusion

This work was developed on the underlying theory that TA is due to micron-sized bubbles trapped in the cracks of kidney stones. By systematically varying all of the Doppler output parameters, we have seen a parameter sensitivity which supports the trapped bubble theory. Additionally, by using an optimized set of parameters we are able to collect data in human scans that suggest an increased sensitivity of the TA for kidney stone detection. Future work involves continuing to collect data samples from kidney stone patients and comparing the sensitivity of the new parameters to the default parameter set, as well as further refinement of the parameters. Additionally, changing the wall filter was not investigated in this study. Further research would involve filtering methods potentially decrease or eliminate motion artifact and blood flow from the estimation. Another improvement might include be using a broadly focused Doppler beamforming method. This would increase the energy directed at the stone and enhance bubble activity. This would need to be remain within the FDA acoustic output limits as before.

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