

Online First October 5, 2026

Research Paper

Image Quality Enhancement in Spectral-Domain Synthetic Transmit Aperture Ultrasound Imaging for 3D Volumetric Reconstruction Using Golay Coded Sequences

Jurij TASINKIEWICZ*, Ihor TROTS

*Institute of Fundamental Technological Research, Polish Academy of Sciences
Warsaw, Poland**Corresponding Author: yurijtas@ippt.pan.pl*Received September 17, 2025; revised March 26, 2026; accepted July 20, 2026;
version of record October 5, 2026; published issue XXXX.*

Spectral-domain (SD) ultrasound imaging techniques offer much faster reconstruction rates compared to conventional time-domain (TD) methods, such as synthetic transmit aperture (STA). Nevertheless, they often fail to deliver sufficient imaging depth, which remains a significant limitation. In this study, we propose employing coded transmission using Golay complementary coded sequences (GCCS) to enhance SD image reconstruction by improving imaging performance metrics such as signal-to-noise ratio (SNR) and visualization depth. Specifically, we introduce a modification to the 3D SD STA method (SD-STA) that integrates SD data processing with sub-aperture STA data acquisition and plane-wave transmission. This modification incorporates matched filtering and the compression of Golay-coded signals directly within the spectral domain, enhancing its efficiency in terms of transmission, detection, and processing of GCCS signals. The proposed imaging approach was validated using synthetic datasets generated with the Field II simulation program in MATLAB[®] and experimental data obtained from a perfect reflector and a commercial, general-purpose phantom using a Verasonics Vantage 256[™] research ultrasound system equipped with a dedicated 1024-element 2D matrix transducer. We generated 16-bit Golay-coded sequences and measured key imaging performance metrics – namely axial resolution, SNR, side-lobe level (SLL), and visualization depth – comparing them against parameters obtained using a conventional short-pulse transmit signal. Experimental results demonstrated a 14 dB increase in echo amplitude when using GCCS with matched filtering. Furthermore, phantom tests under low-voltage conditions showed a ~6 dB SNR gain and a twofold increase in imaging depth, both of which are consistent with simulation results.

Keywords: Golay complementary coded sequences (GCCS), synthetic transmit aperture (STA), spectral-domain ultrasound, Fourier transform, volumetric ultrasound imaging.



Copyright © 2026 The Author(s).
This work is licensed under the Creative Commons Attribution 4.0 International CC BY 4.0
(<https://creativecommons.org/licenses/by/4.0/>).

1. INTRODUCTION

Ultrasonography is one of the most commonly used medical imaging modalities due to its low cost, non-invasive nature, fast and simple data acquisition, and immediate availability of examination results, which makes it particularly advantageous for initial diagnostic assessments. In recent years, 3D ultrasound imaging has become increasingly widespread and clinically relevant (BERNAL *et al.*, 2020; HANSEN-SHEARER *et al.*, 2022; SALARI *et al.*, 2025). While traditional 2D ultrasound remains effective for many applications, it has limitations when visualizing volumetric anatomical structures due to its inherently planar nature. Three-dimensional techniques overcome these limitations by offering clinicians a more comprehensive and realistic representation of tissue morphology, facilitating improved diagnostic insight and more accurate assessment of complex geometries. This

is especially true in obstetric applications, where 3D ultrasound has recently been extensively used to support the diagnosis of fetal defects such as cleft lip and clubfoot (BÄUMLER *et al.*, 2011; CAMPBELL *et al.*, 2005; KIM *et al.*, 2024), and to improve evaluation and planning for potential interventions. It has also shown promise in significant reducing breast examinations times (PADILLA *et al.*, 2013; KOTSIANOS-HERMLE *et al.*, 2009) and improving the accuracy of attenuation and speed-of-sound measurements in biological tissues (WISKIN *et al.*, 2019). Moreover, recent advancements in ultrasound technology have enabled real-time 3D imaging, which has allowed dynamic visualisation, for example, in cardiology, including heart wall motion, valve function, and blood flow within vessels, to be achieved.

Three-dimensional ultrasound imaging usually requires using a 2D transducer array. Although a large number of different high-performance 1D transducer arrays are available, they are rarely used for 3D imaging because they require very complicated and time-consuming mechanical scanning of the measurement volume in order to acquire acoustic data, which is particularly important in the imaging of dynamic structures. Moreover, mechanical scanning tends to create motion-related artefacts and image distortions, and requires the use of advanced mechanical devices to control and precisely position the 1D probe, which significantly complicates the process of volumetric data acquisition. Therefore, volumetric imaging is now usually accomplished using 2D transducer arrays, and the most obvious example here is a fully sampled 2D array transducer, allowing the same underlying principles of ultrasound imaging in 3D to be applied. However, a significant limitation of this approach is the very complex electronics, high data bandwidth, and processing requirements resulting from a large independent signal channel count (e.g., 4096 for a 64×64 array transducer).

It is worth noting that several alternative strategies have been proposed to address the challenges of 3D data acquisition. One such approach is micro-beamforming (LOK, LI, 2018), in which transducer elements are grouped into smaller homogeneous sub-arrays that are managed independently, with only limited analog control within each group. Another strategy employs sparse arrays, where only a subset of elements from the full array is active during transmission and reception, thereby reducing system complexity while maintaining imaging performance (YEN *et al.*, 2000). While these approaches allow the number of active elements to be reduced, they typically demand more complex signal processing to compensate for drawbacks such as array undersampling, leading to increased side-lobes and the presence of grating lobes. Additionally, the smaller active transducer area leads to a lower SNR in the case of sparse arrays.

Another promising strategy for volumetric imaging involves row-column-addressed (RCA) array technology, which has recently garnered significant attention in the research community (MORTON, LOCKWOOD, 2003; HANSEN-SHEARER *et al.*, 2022; SALARI *et al.*, 2025). RCA arrays substantially reduce channel count – for an $N \times N$ 2D array, RCA requires only $2N$ connections instead of N^2 – by connecting all elements along rows for transmission and along columns for reception. However, this reduction in system complexity comes at a cost: RCA designs tend to suffer degraded beamforming capabilities, especially in the signal-channel direction, leading to diminished image quality (SALARI *et al.*, 2025).

Thus, when resolution and flexible, precise acoustic beam control are critical, a fully sampled 2D transducer array is often justified, despite its greater complexity, higher power consumption, and increased cost.

Once the hardware setup is in place, it is crucial to choose an appropriate beamforming technique and acoustic data processing pipeline that match the physical constraints and performance objectives of the intended application. In real-time imaging applications, the number of reconstructed volumes per second is a critical parameter, directly influencing the frame rate. This rate is fundamentally constrained by the speed of sound in the medium – approximately 1540 m/s in soft tissue – and the imaging depth. Consequently, the number of transmission events is typically limited to a few thousand per second in clinical settings, restricting the acquisition of acoustic signals to a few hundred interrogation cycles. Moreover, conventional TD acoustic data processing based on the well-known delay-and-sum (DAS) beamforming (THOMENIUS, 1996) is notoriously computationally demanding. While real-time 2D ultrasound systems are now commonplace, in the case of volumetric TD DAS methods, the numerical complexity is of several orders of magnitude greater, even when employing channel-saving RCA arrays. Therefore, developing an efficient 3D acoustic data processing strategy is essential, one that minimizes the number of transmission events while employing time-efficient reconstruction algorithms to meet

the specific requirements of real-time imaging applications. This approach ensures that the system can achieve high frame rates and maintain image quality, both of which are critical for effective real-time imaging.

Several data acquisition and beamforming techniques have been reported recently to reduce the number of transmit events compared to traditional TD DAS methods. Among these, plane-wave imaging (PWI) approaches (CHENG, LU, 2006) using Fourier spectrum domain (SD) implementations allowed for the computational complexity of 3D acoustic data processing to be significantly reduced, especially for the fully sampled 2D array transducer for which they were developed. Specifically, real-time volumetric imaging using a 1024-element fully-sampled array transducer can reportedly be achieved using GPU acceleration capabilities (LEWANDOWSKI *et al.*, 2020). However, as reported in (TASINKIEWICZ, 2023), the imaging depth of the PWI method with volume reconstruction from a single transmission is limited due to the reduced SNR inherent in this acquisition mode.

The performance of medical ultrasound imaging is inherently constrained by tissue attenuation, which limits key parameters such as SNR, penetration depth, contrast, and contrast-to-noise ratio (CNR). Although higher transmit power can improve these metrics, it is strictly limited by safety regulations (International Electrotechnical Commission, 2015).

Conversely, axial and lateral resolution are influenced by factors including pulse duration, center frequency, and beam width, all of which are determined by the wavelength and aperture size of the transducer array (CHEN, PARKER, 2017). Increasing the center frequency improves resolution but also exacerbates attenuation (D’ASTOUS, FOSTER, 1986), thereby reducing the effective imaging depth. This trade-off between resolution and penetration depth presents a fundamental challenge in ultrasound imaging, necessitating the development of advanced techniques that balance these parameters to optimize image quality and diagnostic utility.

One promising solution to mitigate these limitations involves the use of wideband signal sequences, such as Golay complementary coded sequences (GCCS), with durations spanning several cycles (typically 16 to 32 periods) of the nominal frequency. Unlike conventional short pulse bursts comprising 1–2 sine cycles, these extended-duration sequences offer several advantages. They enhance the energy content of the transmitted signal, leading to improved SNR and penetration depth. Additionally, when appropriately matched with the transducer’s bandwidth, they help maintain or even improve axial resolution. Notably, simulations and experimental results have demonstrated that employing such wideband GCCS transmitted signals can significantly increase the amplitude of compressed echoes and preserve axial resolution, even when using transducers with narrower fractional bandwidths (NOWICKI *et al.*, 2003; TROTS *et al.*, 2011). More details on Golay coded sequences, their properties, and efficient compression techniques can be found in (TROTS *et al.*, 2004).

A primary limitation of GCCS in ultrasound imaging is the requirement for sequential transmission of complementary codes. This process inherently reduces the data acquisition rate, potentially halving the frame rate – a critical parameter for real-time applications. However, this constraint predominantly affects the imaging of dynamic structures. To mitigate this issue, a strategy utilizing two consecutive frames of acquired echoes, each corresponding to alternating sequences of the GCCS pair, can be employed. This approach allows for the reconstruction of a single ultrasound image from sequential data, thereby maintaining an adequate frame rate without compromising image quality.

The primary objective of this study was to experimentally investigate ultrasound echoes generated by the sequential transmission of GCCS pairs. This approach evaluates the feasibility of enhancing penetration depth in the 3D SD-STA framework for ultrasound imaging (TASINKIEWICZ, 2023). To complement the experimental work, numerical simulations were conducted using the Field II platform for MATLAB® to analyze acoustic field interactions and image formation. The integration of experimental data with simulation results enables a comprehensive evaluation of the benefits and limitations of employing GCCS in 3D SD-STA ultrasound imaging.

Although the method operates within a fully 3D reconstruction framework, the evaluation is presented using representative cross-sectional views and depth-resolved analyses extracted from the reconstructed volume for clarity.

The main contributions of this work are as follows:

- integration of GCCS excitation into the 3D SD-STA ultrasound imaging framework,
- development of an efficient SD compression strategy specifically designed for GCCS pairs,

- experimental validation demonstrating simultaneous improvements in image quality and penetration depth while maintaining high computational efficiency.

The rest of the paper is organized as follows. In the next section, the theoretical fundamentals of GCCS signal processing and the 3D SD-STA imaging method are presented. In [Sec. 3](#), the details of the experimental setup are discussed. [Section 4](#) and [Sec. 5](#) present the obtained experimental results and discussion, respectively. Finally, [Sec. 6](#) briefly summaries the work.

2. THEORY

2.1. GCCS

The primary advantage of GCCS lies in their ability to achieve zero side-lobe levels in their combined autocorrelation function. This characteristic effectively suppresses unwanted echoes, thereby enhancing the clarity and contrast of ultrasound images. Furthermore, when combined with matched filtering, GCCS facilitate pulse compression, resulting in compressed echoes that closely resemble conventional short pulses but with significantly higher amplitudes ([TROTS *et al.*, 2004](#)). This combination leads to an improved SNR and enhanced imaging performance. GCCS belong to a broader class of complementary algebraic codes. Specifically, GCCS require two consecutive transmissions to generate a single image scan line, in contrast to the single transmission required for short pulse ([TROTS *et al.*, 2011](#)). A GCCS consist of a pair of codes $A1(k)$ and $A2(k)$, each of length N (N -bit sequences, where $k = 0, N - 1$). For example, the 16-bit long GCCS ($N = 16$) pair used in this work is defined as follows:

$$\begin{aligned} A1 &= \{1, 1, 1, -1, 1, 1, -1, 1, 1, 1, -1, -1, -1, 1, -1\}, \\ A2 &= \{1, 1, 1, -1, 1, 1, -1, 1, -1, -1, -1, 1, 1, 1, -1\}. \end{aligned} \quad (1)$$

The GCCS pair $A1$, $A2$ exhibit autocorrelation functions (ACFs) with side lobes of equal magnitude but opposite signs. Consequently, when these sequence responses are combined, their ACFs sum to a composite function that is zero at all time shifts except at zero, where it attains a peak amplitude of $2N$. This property effectively cancels outside lobes, enhancing the clarity and contrast of ultrasound images ([TROTS *et al.*, 2004](#)):

$$\text{ACF}(A1(k)) + \text{ACF}(A2(k)) = \begin{cases} 2N, & k = 0, \\ 0, & \text{otherwise,} \end{cases} \quad (2)$$

where $2N = 32$ for the 16-bit sequence defined in Eq. (1). The actual transmitted GCCS signals corresponding to the complementary sequences defined in Eq. (1) are shown in [Fig. 1](#). These signals were obtained by modulating the phase of a harmonic signal of nominal frequency f_0 according to the pattern: $\varphi_0 = 0$ for $A_i = 1$, and $\varphi_0 = \pi$ for $A_i = -1$, where A_i is the i -th bit of the corresponding sequence ($A1$ or $A2$) and φ_0 is the phase shift.

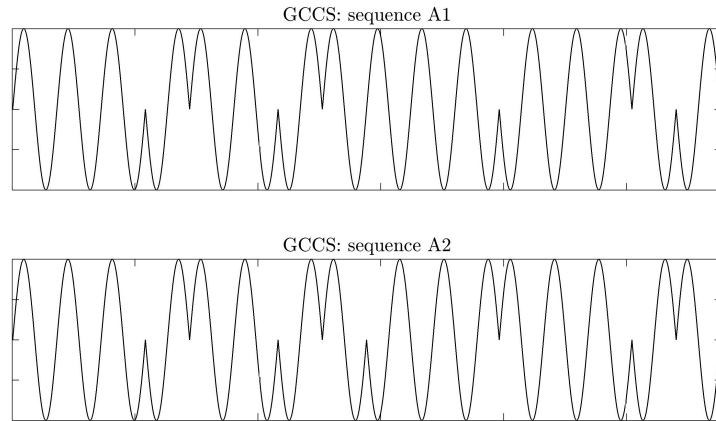


FIG. 1. Example of the 16-bit long GCCS pair used in this work; each bit of the code is encoded with one cycle of a sine wave at the nominal frequency (f_0).

The ACFs corresponding to the signals in Fig. 1 are displayed in the upper subplot of Fig. 2. The sum of these ACFs ($\text{ACF}(A1) + \text{ACF}(A2)$) yields a composite function that is zero at all time shifts except at zero, where it attains a peak amplitude of $2N$ (see Fig. 2, lower subplot).

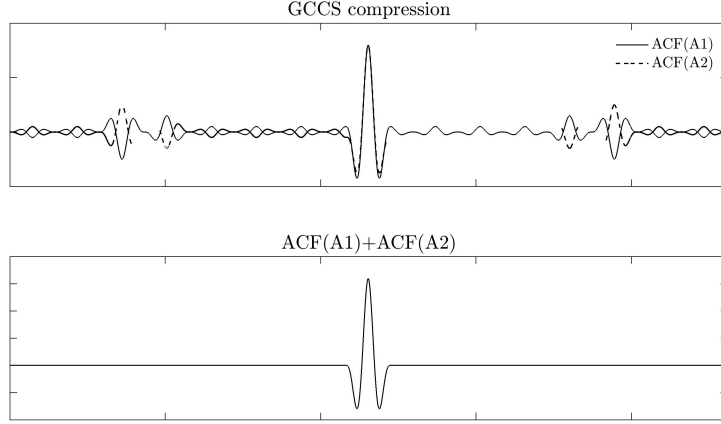


FIG. 2. ACFs (upper subplot) of the GCCS pair shown in Fig. 1, and their combined sum yielding the compressed GCCS signal (lower subplot).

In contrast to the idealized mathematical models presented in Eq. (1), achieving perfect side-lobe cancellation with physical signals is unattainable. As depicted in the lower subplot of Fig. 2, residual side lobes remain adjacent to the main lobe. However, this does not constitute a disadvantage; the magnitude of these side lobes and the duration of the compressed signal – both of which influence axial resolution – are comparable to those of a short pulse. More importantly, the use of GCCS results in a substantial increase in echo amplitude and SNR, theoretically offering a $2N$ -fold gain in signal amplitude.

2.2. SD 3D ULTRASOUND IMAGING

The primary advantage of SD data processing methods lies in their computational efficiency, making them well-suited for real-time 3D ultrasound imaging applications. This study focuses on the SD-STA method previously developed by Tasinkiewicz (2023). The SD-STA method integrates the benefits of plane-wave sub-aperture data acquisition with Fourier spectrum-based data processing to facilitate 3D image synthesis. A brief overview of the SD-STA methodology is provided below.

Without loss of generality, consider a square 2D $N \times N$ -element transducer array situated in the $z = 0$ plane and centered at the origin of a Cartesian coordinate system. The array's sides are aligned with the x - and y -axes. During a single transmit-receive (TX/RX) event, all elements of the array are simultaneously activated to emit a plane wave propagating in direction normal to the aperture into the examined volume (also referred to as the region of interest, or ROI). Subsequently, the transducer is switched to receive (RX) mode. Due to limitations in the number of available receiving channels, only a subset of the transducer elements is typically utilized for echo detection. These receiving elements are organized into so-called receiving sub-apertures, which can vary in size and configuration depending on specific imaging requirements and system constraints. The design and arrangement of these sub-apertures are crucial for optimizing image quality and system performance. In this study, we assume without loss of generality that the receive sub-apertures are square and disjoint, each comprising $M \times M$ elements, where $N/M = I$, I being a positive integer. Therefore, a full data acquisition cycle, which involves recording radio frequency (RF) signals from each of N^2 transducer elements, consists of I^2 TX/RX events in this case. Once a full acquisition cycle is completed, all echo signals collected by each 2D transducer element are processed and a single ultrasound image frame is synthesized.

The ultrasound echoes detected by the array represent a spatially sampled back-scattered acoustic field $s(\mathbf{r}_1, z = 0, t)$ within the aperture, where $\mathbf{r}_1 \equiv (\mathbf{e}_x x + \mathbf{e}_y y)$ – is the position vector in the $z = 0$ plane; here, $s(\mathbf{r}, t)$ is the back-scattered acoustic field, and $\mathbf{r} \equiv (\mathbf{e}_x x + \mathbf{e}_y y + \mathbf{e}_z z)$ is the position vector in the examined 3D ROI.

The spatial spectrum of the back-scattered acoustic field can be expressed as a linear combination of the harmonic waves resulting from randomly distributed scatterers in the ROI (CHENG, LU, 2006):

$$S(\mathbf{k}_\perp, t) = \int K(\omega) \left\{ \int_V f(\mathbf{r}) e^{i\mathbf{k}'\mathbf{r}} d\mathbf{r} \right\} e^{-j\omega t} d\omega, \quad (3)$$

where the function $f(\mathbf{r})$ is the object function describing the scattering intensity of the random distribution of scatterers in the ROI; $\mathbf{k}' = \mathbf{k} + \mathbf{k}^I$, where $\mathbf{k}^I = \mathbf{e}_x k_x^I + \mathbf{e}_y k_y^I + \mathbf{e}_z k_z^I$ is the wave-vector of the transmitted wave-field; $\mathbf{k} = \mathbf{e}_x k_x + \mathbf{e}_y k_y + \mathbf{e}_z k_z$ is the wave-vector of the back-scattered acoustic field; $\mathbf{k}_\perp = \mathbf{e}_x k_x^I + \mathbf{e}_y k_y^I$ is the reduced wave-vector of the back-scattered acoustic field corresponding to $\mathbf{r}_\perp$; and $K(\omega)$ is a transfer function that takes into account the bandwidth of the transmit and receive circuits. In the case of the SD-STA method considered in this study, the acoustic plane wave is transmitted in the normal direction (along the z -axis); therefore, $\mathbf{k}^I = \mathbf{e}_z k_z^I$. From Eq. (3), the relationship between the spatio-temporal spectrum of the back-scattered acoustic field $S(\mathbf{k}_\perp, \omega)$, and the spatial spectrum of the object function $F(\mathbf{k}')$ can be obtained as:

$$S(\mathbf{k}_\perp, \omega) = K(\omega) F(\mathbf{k}') \cong F'(\mathbf{k}'), \quad (4)$$

where $F'(\mathbf{k}')$ defines the ‘band-limited’ spatial spectrum of the object function $f(\mathbf{r})$. This term defines the 3D spatial spectrum of the ultrasound image reconstructed from a single data frame corresponding to a full data acquisition cycle: $F'(\mathbf{k}') \equiv I'(\mathbf{k}')$. The spatial spectrum of the ultrasound image $I(\mathbf{k}')$ can be obtained from the spatio-temporal spectrum of the back-scattered echoes detected by the elements of the matrix transducer:

$$I(\mathbf{k}'(\omega)) = S(\mathbf{k}'(\omega)), \quad (5)$$

where the spectrum $S(\mathbf{k}'(\omega))$ can be obtained from $S(\mathbf{k}_\perp, \omega)$ via the coordinate transformation:

$$k'_x = k_x, \quad k'_y = k_y, \quad k'_z(\omega) = \frac{\omega}{c} + \sqrt{\left(\frac{\omega}{c}\right)^2 - k_x^2 - k_y^2}. \quad (6)$$

The final reconstructed 3D ultrasound image can be obtained straightforwardly from $I(\mathbf{k}'(\omega))$. Specifically, applying the 3D inverse Fourier transform to $I(\mathbf{k}'(\omega))$ yields:

$$I(\mathbf{r}) = \frac{c}{4\pi^2} \iiint B(\mathbf{k}'(\omega)) I(\mathbf{k}'(\omega)) e^{j\mathbf{k}'(\omega)\mathbf{r}} d\mathbf{k}'(\omega), \quad (7)$$

where

$$B(\mathbf{k}'(\omega)) \equiv \frac{(\sqrt{k_z'^2(\omega) - k_x'^2 - k_y'^2})}{k'_z(\omega)} \quad (8)$$

is the Jacobian of the variable transformation in the spectrum domain defined in Eq. (6).

3. METHODS

3.1. MEASUREMENT SETUP

The 3D SD-STA method utilizing GCCS transmission, as proposed in this study, was validated through both synthetic data simulations conducted in the MATLAB[®] environment and experimental measurements. In the experimental setup, a Verasonics Vantage[™] 256 research ultrasound system (Verasonics Inc., Kirkland, WA) was employed to generate both a pair of 16-bit GCCS and a short pulse burst, defined here as a single sine cycle at the nominal transducer frequency. The system was configured with a square 1024-element matrix array transducer arranged in a 32×32 grid, operating at a center frequency of 3.47 MHz. Each element had a pitch of 0.3 mm and a width of 0.275 mm. The transducer was interfaced with the Vantage 256 system using a Universal Transducer Adapter (UTA) 1024-MUX, which enabled multiplexed acquisition of data from the matrix array. The sampling

frequency was 13.87 MHz, which corresponded to 4 time samples per acoustic wavelength of the recorded signal. A block diagram of the experimental setup is shown in Fig. 3.

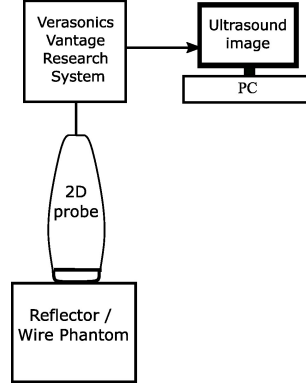


FIG. 3. Block diagram of the experimental setup illustrating the configuration and components used in the study.

For the experimental measurements, echoes from a reference reflector were first acquired using both GCCS and short-pulse transmissions. A 3 cm-thick brass plate, submerged in a water tank and exhibiting an acoustic reflection coefficient of approximately 0.86 in water, was employed as the reflector. The recorded reflected signals were then used to evaluate the performance and efficiency of GCCS transmission with the 1024-element matrix array transducer and the Verasonics ultrasound system. For this purpose the duration of the compressed GCCS signal was measured at both half-maximum and 0.1 of the maximum amplitude, and these values were compared to the corresponding durations of the short pulse. These measurement levels correspond to -6 dB and -20 dB relative to the signal peak, respectively. Additionally, the SLL was quantified for both transmission modalities defined here as the ratio of the amplitude of the side lobe with the maximum amplitude to that of the main lobe in the signal envelope. Furthermore, to evaluate the improvement in SNR, phantom measurements were conducted using different transmission signals. A commercial general-purpose phantom (Model 525, Dansk Fantom Service) with an acoustic attenuation of $0.5 \text{ dB}/(\text{cm} \cdot \text{MHz})$ (equivalent to $5.7 \text{ Np}/(\text{m} \cdot \text{MHz})$) was employed for these experiments. In this study, the SNR was defined as the ratio of the peak signal power to the root-mean-square (RMS) power of the noise. The SNR was evaluated as a function of depth using RF echoes acquired along the central axis of thin wires embedded in the general-purpose phantom, with the wires spaced at 10 mm intervals along the vertical axis. Comparisons were made between the compressed GCCS and short-pulse echoes. To calculate the average signal power, echo samples within a 2 mm rectangular window – corresponding to approximately five wavelengths ($\lambda \approx 0.44 \text{ mm}$) at the operating frequency – were selected, with the window centered on the echo peak at the depth of the corresponding wire. The RMS noise power was computed over a 10 mm rectangular window (approximately 22 wavelengths) located at depths between 62 mm and 72 mm. Under the experimental conditions used, this window selection ensured that the signal contributions from the phantom were negligible at these depths, thereby allowing the measurement to predominantly reflect noise.

3.2. NUMERICAL SIMULATIONS

For the numerical simulations, ultrasound echoes from a set of point scatterers were generated using the Field II platform (JENSEN, 1996; JENSEN, SVENDSEN, 1992) in the MATLAB[®] environment. The 1024-element matrix transducer was modeled using the same parameters as those used in the experimental measurements described above. A set of nine point scatterers, spaced at 10 mm intervals along the z -axis (perpendicular to the transducer face), was simulated within a volumetric domain of $10 \text{ mm} \times 10 \text{ mm} \times 90 \text{ mm}$. Each RF echo was simulated with $N_t = 1528$ time samples, corresponding to the assumed imaging depth of 90 mm. For the numerically simulated data, the amplitude gain resulting from the compression of the GCCS signals was calculated, and the echo amplitudes from the point scatterers were compared between the GCCS and short-pulse transmissions over depths ranging from 10 mm to 90 mm. Subsequently, the compression efficiency of the GCCS approach was evaluated

by analyzing both the amplitude gain of the coded sequences and the sum of their ACFs, and the results were compared against those obtained with the short pulse.

3.3. DATA ACQUISITION

In the numerical simulations, the full 32×32 element aperture was employed for both plane-wave transmission and RF echo detection. Acquiring a complete frame of acoustic data – which comprises 1024 RF signals, one for each element of the 2D array – required only a single TX/RX event for the short pulse and two sequential TX/RX events for the GCCS transmission. In the case of GCCS, the two complementary codes, A_1 and A_2 (see Fig. 1), were transmitted sequentially, and the corresponding echoes were stored for subsequent processing. For the experimental measurements, data acquisition needed to account for limitations in the number of available receiving channels on the scanner. Specifically, the Verasonics Vantage™ 256 research ultrasound system provides 256 independent receiving channels. Consequently, a complete data acquisition cycle involved four TX/RX events for the short pulse and eight TX/RX events for the GCCS transmissions, with the two codes transmitted sequentially. During each acquisition, the full 1024-element aperture was activated for plane-wave transmission. The resulting RF echoes were received by one of four dynamically selected RX sub-apertures, each containing 256 elements arranged in an 8×32 grid, as determined by the dynamic multiplexer (MUX) configuration. The digitized RF data were temporarily stored in the front-end electronics' local memory, and upon completion of a full acquisition cycle, the data were transferred in bulk via a high-speed PCI Express interface to the host computer for offline processing.

3.4. DATA PROCESSING STEPS OF THE MODIFIED 3D SD-STA METHOD

The steps involved in synthesizing a single 3D frame using the modified 3D SD-STA method under the experimental measurement setup (Verasonics scanner) are outlined below. A key feature of this approach, and a novel aspect of the present work, is implementation of GCCS processing and signal compression directly in the SD:

1. Transmit the codes A_1 and A_2 of the GCCS transmission signal sequentially using the full 32×32 -element TX aperture, and acquire ultrasound echoes using the i -th 8×32 RX sub-aperture, where $i = 1, \dots, 4$.
2. Record two matrices of the 1024 acquired RF echoes $S_j(\mathbf{r}_\perp, t)$, $j = 1, 2$ (one for each code A_j), and, after the full data acquisition cycle is completed, transfer the digitized data to the PC memory for further (offline) processing.
3. Compute separately the temporal spectrum functions $S_j(\mathbf{r}_\perp, \omega)$ with respect to the time variable t using the fast Fourier transform (FFT) algorithm: $S_j(\mathbf{r}_\perp, \omega) = \text{FFT}_{t \rightarrow \omega} S_j(\mathbf{r}_\perp, t)$.
4. Perform GCCS compression in SD:
 - 4.1. Compute the product of the spectrum functions $S_j(\mathbf{r}_\perp, t)$ with the temporal spectrum of its time-reversed code A_j : $\widehat{S}_j(\mathbf{r}_\perp, \omega) = S_j(\mathbf{r}_\perp, \omega) \cdot \widehat{A}_j(\omega)$, $\widehat{A}_j(\omega) = \text{FFT}_{t \rightarrow \omega} \{\text{rev}(A_j(t))\}$.
 - 4.2. Sum both spectrum function to obtain the temporal spectrum of the compressed GCCS signal: $S_{\text{GCCS}}(\mathbf{r}_\perp, \omega) = \sum_j \widehat{S}_j(\mathbf{r}_\perp, \omega)$, $j = 1, 2$.
5. Compute the spatial spectrum of $S_{\text{GCCS}}(\mathbf{r}_\perp, \omega)$ with respect to the spatial variables (x, y) using a 2D FFT algorithm: $S_{\text{GCCS}}(\mathbf{k}_\perp, \omega) = \text{FFT}_{\mathbf{r}_\perp \rightarrow \mathbf{k}_\perp} S_{\text{GCCS}}(\mathbf{r}_\perp, \omega)$.
6. Transform the spectral variables $(k_x, k_y, \omega) \rightarrow (k'_x, k'_y, k'_z(\omega))$ (see Eq. (6)) in the SD.
7. Compute the 3D spatial spectrum $I(\mathbf{k}'(\omega))$ of the 3D image (see Eq. (5)).
8. Compute the 3D inverse Fourier transform with respect to spatial spectrum variables (see Eq. (7)) to obtain the final reconstructed image itself: $I(\mathbf{r}) = \text{IFFT}_{\mathbf{k}'(\omega) \rightarrow \mathbf{r}} I(\mathbf{k}'(\omega))$.

For short-pulse transmission, step 4 of the acoustic data processing algorithm described above can be omitted, as in the original 3D SD-STA method. Additionally, the 3D spatio-temporal spectrum of the recorded echoes can be computed in a single step using a 3D FFT algorithm: $(\mathbf{k}_\perp, \omega) = \text{FFT}_{\mathbf{r}_\perp, t \rightarrow \mathbf{k}_\perp, \omega} S(\mathbf{r}_\perp, t)$, where $S(\mathbf{r}_\perp, t)$ denotes the matrix of 1024 RF echoes acquired using short-pulse transmission. Figure 4 illustrates the 3D SD-STA imaging framework, incorporating the SD-GCCS compression scheme described above.

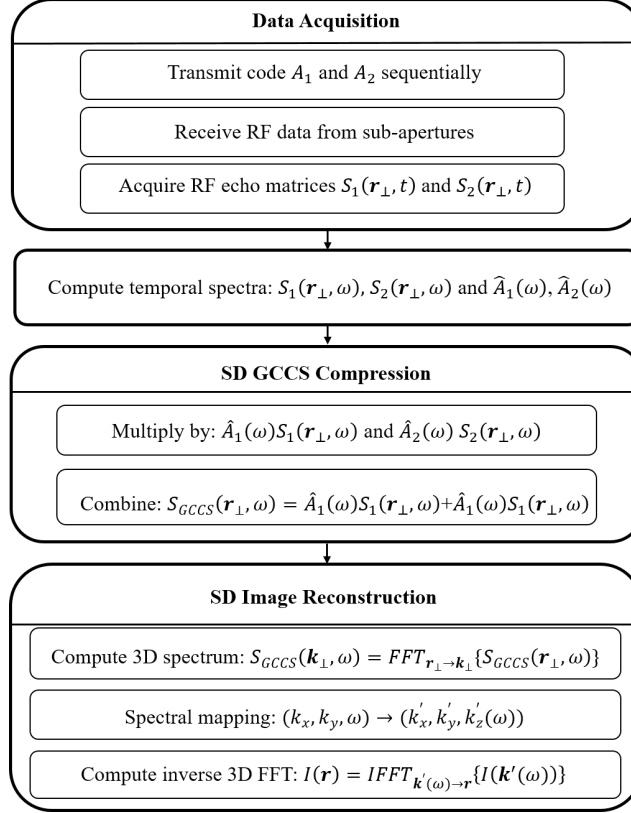


FIG. 4. Flowchart illustrating the successive data processing steps in the 3D SD-STA imaging framework, incorporating GCCS acquisition and SD compression.

4. RESULTS

4.1. NUMERICAL SIMULATIONS

The simulated signals acquired from a set of point scatterers positioned along the normal axis at depths ranging from 10 mm to 90 mm, spaced every 10 mm, are illustrated in Fig. 5 and Fig. 6 for both the GCCS and short-pulse transmissions. Plane-wave insonification was performed using all 1024 elements of the 2D transducer, and the echoes were recorded independently by each element. The signals presented in Fig. 5 and Fig. 6 correspond

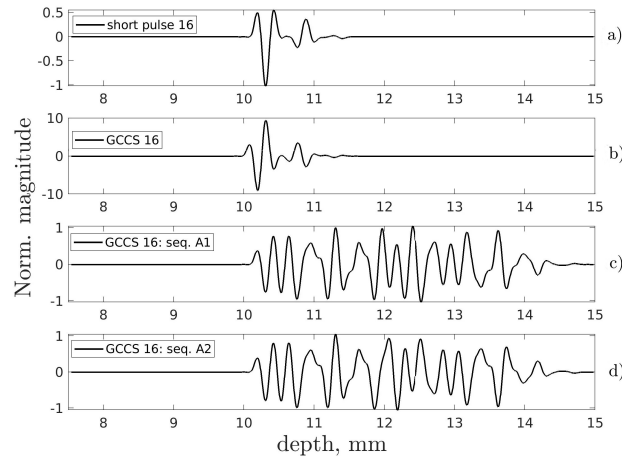


FIG. 5. Normalized RF echoes from the point scatterer closest to the transducer, simulated using the Field II program in MATLAB[®]. The signals were recorded by the central element with indices $\#(16, 16)$ under plane-wave insonification: a) short-pulse transmission, b) and c) GCCS sequences A1 and A2, respectively, d) compressed GCCS echo corresponding to the sequences shown in b) and c).

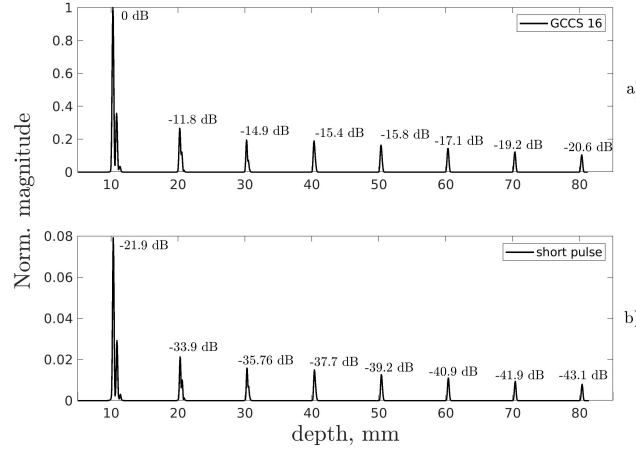


FIG. 6. Envelopes of the normalized echoes from a set of point scatterers simulated using the Field II program in MATLAB[®]. Signals were recorded by the central element with indices $\#(16, 16)$ under plane-wave insonification: a) GCCS transmitted signals, b) short-pulse transmitted signals. Echo amplitudes at different depths are shown relative to the maximum GCCS signal at a depth of 10 mm.

to the time traces detected by a single central element with indices $\#(16, 16)$. Figure 5 shows the normalized RF signals originating from the nearest point scatterer at a depth of 10 mm.

Specifically, the echo for the short-pulse transmission is shown in Fig. 5a. The GCCS compressed echo is shown in Fig. 5b and the corresponding transmitted GCCS sequences A1 and A2 are depicted in Fig. 5c and Fig. 5d, respectively. It should be noted that the magnitude of the compressed GCCS signal (Fig. 5b) was normalized to the maximum value of the RF echo corresponding to the transmitted sequence A1, in order to highlight the amplitude gain resulting from code compression. In Fig. 6, the envelopes of the compressed GCCS echoes (Fig. 6a) and the short-pulse echoes (Fig. 6b) are presented, respectively. All signals were normalized to the maximum amplitude of the GCCS echo. For comparison, the echo amplitudes at different depths – corresponding to the positions of the point scatterers – are shown relative to the maximum GCCS signal at a depth of 10 mm, which corresponds to the scatterer nearest to the transducer aperture.

4.2. MEASUREMENTS

The RF echoes from reference reflector detected for the different transmitted signals are shown in Fig. 7. It should be noted that the magnitude of the compressed GCCS (Fig. 7b) was normalized to the maximum value

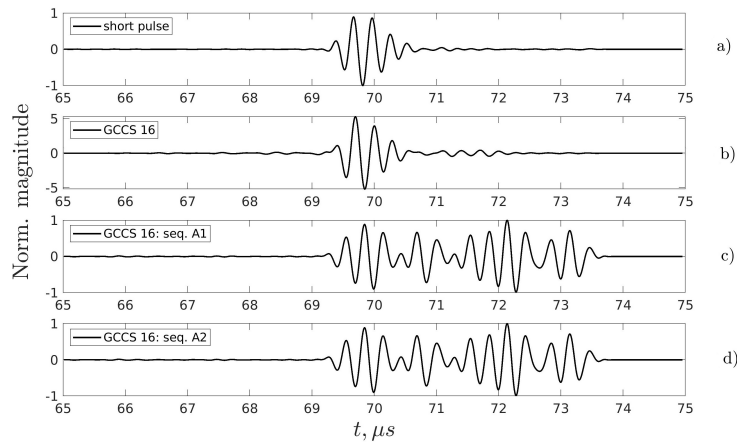


FIG. 7. Normalized RF echoes from a reference reflector, recorded by a single central element with indices $\#(16, 16)$ of the 1024-element matrix transducer connected to the Verasonics Vantage[™] 256 research ultrasound system during plane-wave insonification: a) short-pulse echo, b) compressed GCCS echo, c) and d) echoes resulting from the sequential transmission of the complementary coded sequences A1 and A2.

of the RF echo corresponding to the transmitted sequence A1 to emphasize the compression gain in amplitude, whereas the remaining signals were normalized to their maximum values.

In Fig. 8, the envelopes for the short-pulse transmission and the compressed echo for the GCCS transmission, corresponding to the RF echoes in Fig. 7a and Fig. 7b, are shown.

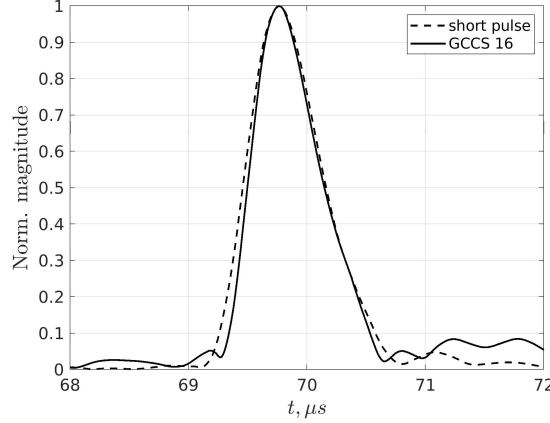


FIG. 8. Envelopes of the RF echoes from a reference reflector obtained using short-pulse and GCCS transmitted signals. The echoes were recorded by the central element with indices $\#(16, 16)$ of the 1024-element matrix transducer connected to the Verasonics VantageTM 256 research ultrasound system during plane-wave insonification.

The half-maximum time durations (which corresponds to the -6 dB amplitude level) of the compressed GCCS and short-pulse envelopes were $0.65 \mu\text{s}$ and $0.69 \mu\text{s}$, respectively. In terms of axial distances, given a speed of sound in water at room temperature is about 1482 m/s , these durations correspond to approximately 0.96 mm and 1.02 mm , respectively. The corresponding time duration values measured at 0.1 of the maximum envelope amplitude (-20 dB amplitude level) were $1.19 \mu\text{s}$ and $1.26 \mu\text{s}$ for the GCCS and the short pulse, respectively. In terms of axial distances, these are 1.76 mm and 1.87 mm , respectively. The SLL, determined as the peak value of the maximum side-lobe of the envelopes presented in Fig. 8, was -20.9 dB for the compressed GCCS (measured at time $t \approx 71.25 \mu\text{s}$) and -24.5 dB for the short pulse (measured at time $t \approx 69.25 \mu\text{s}$).

B -mode images of the general-purpose phantom (Model 525), acquired using the 3D SD-STA imaging method with GCCS and short-pulse transmissions (see Subsec. 2.2 and the data processing algorithm in Subsec. 3.4), are presented in Fig. 9 and Fig. 10 for two orthogonal cross-sectional views in the $x = 0$ and $y = 0$ planes,

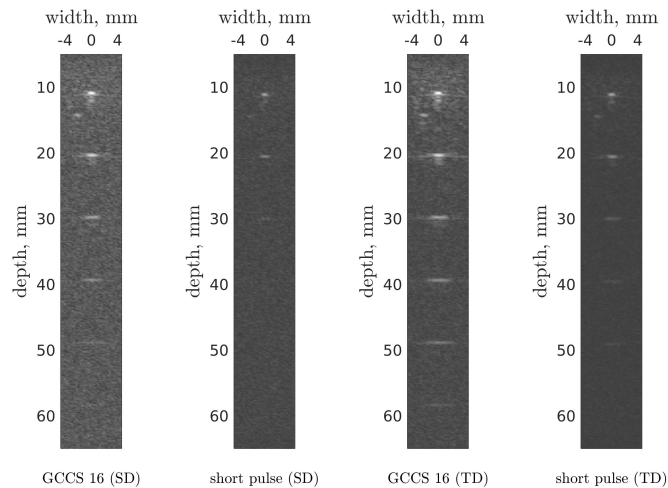


FIG. 9. B -mode cross sections ($x = 0$ plane) of the general-purpose phantom (Model 525) reconstructed using the 3D SD-STA method with GCCS and short-pulse transmissions (labelled ‘GCCS 16 (SD)’ and ‘short pulse (SD)’), respectively). For comparison, images obtained via the conventional TD-PWI method with GCCS and short-pulse excitation are shown (labelled ‘GCCS 16 (TD)’ and ‘short pulse (TD)’). All images are displayed on a 40 dB dynamic range logarithmic scale

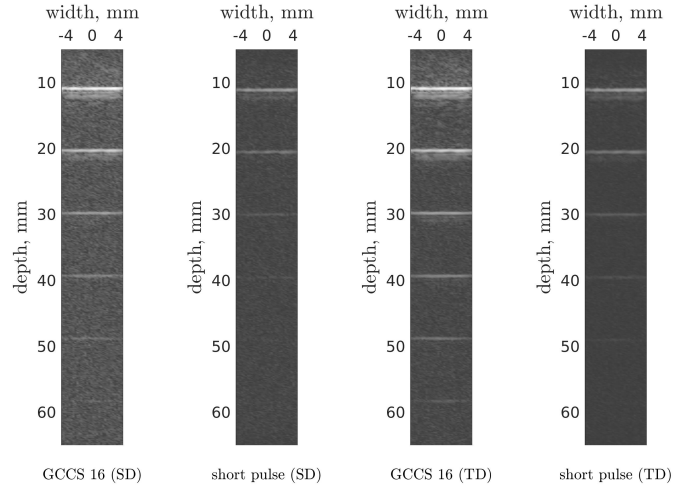


FIG. 10. *B*-mode cross sections ($y = 0$ plane) of the general-purpose phantom (Model 525) reconstructed using the 3D SD-ST method with GCCS and short-pulse transmissions (labelled ‘GCCS 16 (SD)’ and ‘short pulse (SD)’), respectively). For comparison, images obtained via the conventional TD-PWI method with GCCS and short-pulse excitation are shown (labelled ‘GCCS 16 (TD)’ and ‘short pulse (TD)’). All images are displayed on a 40 dB dynamic range logarithmic scale.

respectively, which represent the symmetry planes of the 2D transducer aperture perpendicular to its surface. For comparison, images reconstructed using the conventional time-domain plane wave imaging (TD-PWI) method are also provided for both GCCS excitation and short-pulse insonification. The peak-to-peak driving voltage on the Verasonics Vantage™ 256 scanner was set to 10 V to demonstrate the better visualization of deeper wires when using GCCS transmission compared to the short-pulse signal. Prior to visualization, the reconstructed images were normalized to their respective maximum values.

Figure 11 shows the normalized RF echo magnitudes from the general-purpose phantom (Model 525) recorded by transducer element # (16, 16) for various transmitted signals. These signals correspond to the central scanline of the *B*-mode images in Fig. 9 (specifically the subplots labelled ‘short pulse (SD)’ and ‘GCCS 16 (SD)’), respectively), which intersects the vertical wire array. Prior to visualization, all signals were normalized to their respective peak amplitudes.

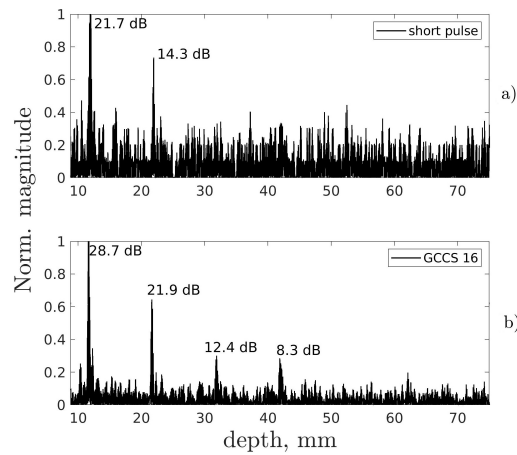


FIG. 11. Magnitude of the normalized RF echoes from the general-purpose phantom (Model 525), recorded by transducer element # (16, 16) under plane-wave insonification for: a) short-pulse transmission and b) GCCS transmission. SNR values at selected depths are indicated.

Figure 12 presents the SNR values as a function of depth, calculated from the normalized beam-formed RF echoes at scanline # (16, 16) of the reconstructed volume. For comparison, SNR curves are provided for both the proposed SD-ST method and the conventional TD-PWI method.

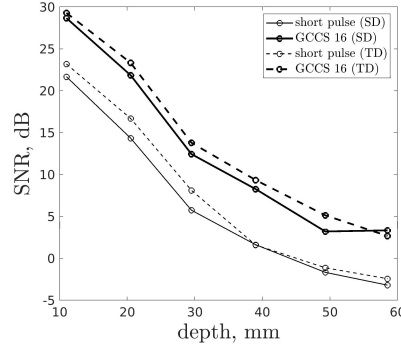


FIG. 12. SNR comparison calculated from normalized beamformed RF echoes of the general-purpose phantom (Model 525) at scanline #(16, 16) of the reconstructed volume. Results are shown for the proposed SD-STa method (denoted as ‘SD’ in the legend) and the conventional TD-PWI method (denoted as ‘TD’), comparing both short-pulse and GCCS-coded transmissions.

As shown in Fig. 12, the SNR improvement for the compressed GCCS signal compared to the short-pulse signal ranged from approximately 6.5 dB to 7.5 dB across the entire depth range, except at 50 mm, where a 4 dB gain was observed. For the conventional TD-PWI method, the corresponding SNR improvement ranged from 5.1 dB to 7.7 dB over the same depth interval.

5. DISCUSSION

The results of both numerical simulations and experimental measurements presented in the previous section clearly demonstrate the effectiveness of the proposed approach, which enables a substantial enhancement of 3D ultrasound imaging performance in the spectral domain through the transmission of complementary Golay code pairs. Key parameters – including signal duration, which directly influences the axial resolution, SLL, and SNR – were evaluated and compared for GCCS and short-pulse transmissions.

Specifically, the axial resolution of the imaging system was estimated from the time duration of the RF echo envelopes obtained from the reference reflector, measured at two amplitude levels: -6 dB (0.5 of the maximum amplitude) and -20 dB (0.1 of the maximum amplitude), as shown in Fig. 8. For GCCS transmission, the axial resolution improved by approximately 6 % at the -6 dB level compared to the short-pulse case. At the -20 dB level, the corresponding improvement was about 5.6 %.

The SLL was also evaluated from the echo envelopes shown in Fig. 8. The measurements indicated that the maximum SLL for the GCCS signal did not exceed -20.9 dB, whereas for the short pulse it was -24.5 dB. These results confirm the effectiveness of side-lobe suppression achieved by matched filtering and compression of the coded GCCS signals (see Fig. 7b and Fig. 7c). Importantly, despite a moderate increase in SLL (approximately 3.6 dB), the GCCS transmission provided a substantial gain in signal amplitude – about 13.9 dB higher than the short-pulse case, as illustrated in Fig. 7a and Fig. 7b. A similar trend was observed in numerical simulations with point scatterers (Fig. 5), where the peak-to-peak amplitude of the compressed GCCS signal increased by roughly 22 dB compared to the short pulse, evaluated for the scatterer located at a depth of 10 mm. Consistently, the envelope amplitudes of echoes across the full depth range (10 mm to 90 mm) demonstrated an enhancement of 20.9 dB to 23.8 dB with GCCS transmission (Fig. 6).

SNR values for GCCS and short-pulse transmissions were compared using beamformed RF echoes at scanline #(16, 16) of the reconstructed volume (Fig. 9 and Fig. 10). The resulting SNR curves as a function of depth are presented in Fig. 12. As illustrated, the compressed GCCS signal consistently outperformed the short pulse, yielding an SNR improvement of approximately 6.5 dB to 7.5 dB across the entire depth range, with the exception of a 4-dB gain at 50 mm. For comparison, Fig. 12 also depicts the SNR curves for the conventional TD-PWI method using both short-pulse and GCCS transmissions, where SNR improvements ranged from 5.1 dB to 7.7 dB over the investigated depth range. Furthermore, a comparison of SNR values for the SD-STa and TD-PWI methods using GCCS transmission shows a slight SNR increase across most depths, with gains ranging from

0.6 dB to 1.7 dB. An exception occurs at 60 mm, where the SD-STA method outperforms the TD-PWI by 0.6 dB. In the case of short-pulse transmission, the corresponding SNR increase for the TD-PWI method over SD-STA ranged from 0.5 dB to 2.5 dB. Finally, the proposed approach for improving image quality in 3D SD-STA ultrasound imaging via GCCS transmission was validated using experimental phantom data. Figure 9 and Fig. 10 present *B*-mode cross sections of the reconstructed volume in the $x = 0$ and $y = 0$ planes, respectively, which correspond to the symmetry planes of the 2D transducer aperture. For comparison, reconstruction results for both GCCS and short-pulse transmissions are illustrated. It should be emphasized that while the proposed approach is inherently 3D, results are reported using selected 2D slices extracted from the reconstructed volume for clarity of presentation and evaluation. A visual assessment of the results of ultrasound image reconstruction indicates that the imaging quality achieved with GCCS transmission surpasses that obtained using a conventional short-pulse transmission. This effect is observed for both SD-STA and TD-PWI reconstruction approaches and arises from the intrinsic properties of coded excitation, which improve penetration depth by increasing the SNR (see Fig. 12). Specifically, Fig. 9 demonstrates a significant improvement in imaging depth: target wires remain visible down to approximately 50 mm when using coded transmissions in both TD and SD methods. In contrast, the short-pulse approach limits visibility to around 30 mm (assuming a 40 dB dynamic range on a logarithmic scale and a 10 V transducer driving voltage).

6. CONCLUSIONS

The results presented in this paper demonstrate the effectiveness of using Golay complementary coded transmission to improve the quality of 3D SD-STA ultrasound imaging. We compared the properties of the GCCS transmit signals and the imaging parameters associated with them with the parameters of the standard short-pulse approach (one period of the fundamental frequency of the transducer). The comparative analysis conducted in this study has demonstrated that a significant enhancement of image quality obtained using 3D SD-STA ultrasound imaging method is achievable via coded transmission, particularly through the use of GCCS. Crucial parameters, such as axial resolution, SNR, SLL, and imaging depth, were improved, as evidenced by both numerical simulations and experimental measurements.

Critically, the application of GCCS transmitted signals permits an enhancement of the SNR via pulse compression – without increasing the transmitted amplitude or requiring the high-peak-power driving signals – which in turn significantly extends the visualization depth. This capability is especially pertinent in SD imaging modalities, where the depth of penetration often remains a limiting factor. Therefore, this approach represents a viable strategy to enhance the performance of ultrasound imaging systems within the constraints of existing regulatory limits.

Equally significant is the SD-GCCS signal compression method proposed in this study, which naturally integrates with the 3D SD-STA ultrasound imaging method, enhancing its efficiency and performance.

FUNDINGS

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

CONFLICT OF INTEREST

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

AUTHORS' CONTRIBUTIONS

Jurij Tasinkiewicz conceptualized the study, performed the analysis and wrote the original draft. Ihor Trots conceptualized the study and contributed to data interpretation. All authors reviewed and approved the final manuscript.

USE OF AI TOOLS

During the preparation of this manuscript, the authors used ChatGPT (OpenAI) solely for English-language proofreading and correction. The authors reviewed and verified all AI-assisted outputs and take full responsibility for the integrity and content of the published article.

REFERENCES

1. BÄUMLER M. *et al.* (2011), Accuracy of prenatal three-dimensional ultrasound in the diagnosis of cleft hard palate when cleft lip is present, *Ultrasound Obstet Gynecology*, **38**(4): 440–444, <https://doi.org/10.1002/uog.8933>.
2. BERNAL M., CUNITZ B., ROHRBACH D., DAIGLE R. (2020), High-frame-rate volume imaging using sparse-random-aperture compounding, *Physics in Medicine & Biology*, **65**: 175002, <https://doi.org/10.1088/1361-6560/ab9372>.
3. CAMPBELL S., LEES C., MOSCOSO G., HALL P. (2005), Ultrasound antenatal diagnosis of cleft palate by a new technique: the 3D reverse face view, *Ultrasound in Obstetrics and Gynecology*, **25**(1): 12–18, <https://doi.org/10.1002/uog.1819>.
4. CHEN S., PARKER K.J. (2017), Enhanced axial and lateral resolution using stabilized pulses, *Journal of Medical Imaging*, **4**(2): 027001, <https://doi.org/10.1117/1.jmi.4.2.027001>.
5. CHENG J., LU J. (2006), Extended high-frame rate imaging method with limited-diffraction beams, *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, **53**(5): 880–899, <https://doi.org/10.1109/tuffc.2006.1632680>.
6. D’ASTOUS F.T., FOSTER F.S. (1986), Frequency dependence of ultrasound attenuation and backscatter in breast tissue, *Ultrasound in Medicine & Biology*, **12**(10): 795–808, [https://doi.org/10.1016/0301-5629\(86\)90077-3](https://doi.org/10.1016/0301-5629(86)90077-3).
7. HANSEN-SHEARER J., LERENDEGUI M., TOULEMONDE M., TANG M.-X. (2022), Ultrafast 3-D ultrasound imaging using row-column array-specific frame-multiply-and-sum beamforming, *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, **69**(2): 480–488, <https://doi.org/10.1109/TUFFC.2021.3122094>.
8. International Electrotechnical Commission (2015), *Medical electrical equipment – Part 2-37: particular requirements for the basic safety and essential performance of ultrasonic medical diagnostic and monitoring equipment* (IEC 60601-2-37), Geneva, Switzerland.
9. JENSEN J.A. (1996), Field: a program for simulating ultrasound systems, [in:] *Proceedings of 10th Nordic-Baltic Conference on Biomedical Imaging Published in Medical & Biological Engineering & Computing*, **34**(Sup.1): 351–353.
10. JENSEN J.A., SVENDSEN N.B. (1992), Calculation of pressure fields from arbitrarily shaped, apodized, and excited ultrasound transducers, *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, **39**(2): 262–267, <https://doi.org/10.1109/58.139123>.
11. KIM Y. *et al.* (2024), Improving prenatal diagnosis precision for congenital clubfoot by using three-dimensional ultrasonography, *Diagnostics*, **14**(1): 117, <https://doi.org/10.3390/diagnostics14010117>.
12. KOTSIANOS-HERMLE D., HILTAWSKY K.M., WIRTH S., FISCHER T., FRIESE K., REISER M. (2009), Analysis of 107 breast lesions with automated 3D ultrasound and comparison with mammography and manual ultrasound, *European Journal of Radiology*, **71**(1): 109–115, <https://doi.org/10.1016/j.ejrad.2008.04.001>.
13. LEWANDOWSKI M., JAROSIK P., TASINKEVYCH Y., WALCZAK M. (2020), Efficient GPU implementation of 3D spectral domain synthetic aperture imaging, [in:] *2020 IEEE International Ultrasonics Symposium (IUS)*, pp. 1–3, <https://doi.org/10.1109/IUS46767.2020.9251552>.
14. LOK U.-W., LI P.-C. (2018), Microbeamforming with error compensation, *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, **65**(7): 1153–1165, <https://doi.org/10.1109/TUFFC.2018.2834411>.
15. MORTON C.E., LOCKWOOD G.R. (2003), Theoretical assessment of a crossed electrode 2-D array for 3-D imaging, [in:] *IEEE Symposium on Ultrasonics, 2003*, **1**: 968–971, <https://doi.org/10.1109/ULTSYM.2003.1293560>.
16. NOWICKI A., SECOMSKI W., LITNIEWSKI J., TROTS I. (2003), On the application of signal compression using Golay’s codes sequences in ultrasound diagnostic, *Archives of Acoustics*, **28**(4): 313–324.
17. PADILLA F. *et al.* (2013), Breast mass characterization using 3-dimensional automated ultrasound as an adjunct to digital breast tomosynthesis, *Journal of Ultrasound in Medicine*, **32**(1): 93–104, <https://doi.org/10.7863/jum.2013.32.1.93>.

18. SALARI A., AUDOIN M., TOMOV B.G., YIU B.Y.S., VILAIN THOMSEN E., JENSEN J.A. (2025), Beamformer for a lensed row-column array in 3-D ultrasound imaging, *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, **72**(2): 238–250, <https://doi.org/10.1109/TUFFC.2025.3526523>.
19. TASINKIEWICZ J. (2023), 3D synthetic aperture imaging method in spectrum domain for low-cost portable ultrasound systems, *Archives of Acoustics*, **48**(4): 559–572, <https://doi.org/10.24425/aoa.2023.146820>.
20. THOMENIUS K.E. (1996), Evolution of ultrasound beamformers, [in:] *1996 IEEE Ultrasonics Symposium. Proceedings*, **2**: 1615–1622, <https://doi.org/10.1109/ULTSYM.1996.584398>.
21. TROTS I., NOWICKI A., SECOMSKI W., LITNIEWSKI J. (2004), Golay sequences – side-lobe canceling codes for ultrasonography, *Archives of Acoustics*, **29**(1): 87–97.
22. TROTS I., TASINKEVYCH Y., NOWICKI A., LEWANDOWSKI M. (2011), Golay coded sequences in synthetic aperture imaging systems, *Archives of Acoustics*, **36**(4): 913–926.
23. WISKIN J., MALIK B., NATESAN R., LENOX M. (2019), Quantitative assessment of breast density using transmission ultrasound tomography, *Medical Physics*, **46**(6): 2610–2620, <https://doi.org/10.1002/mp.13503>.
24. YEN J.T., STEINBERG J.P., SMITH S.W. (2000), Sparse 2-D array design for real time rectilinear volumetric imaging, *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, **47**(1): 93–110, <https://doi.org/10.1109/58.818752>.