

Development and Commissioning of the KIT 3D Ultrasound Computed Tomography System III

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Zusammenfassung

Diese Dissertation beschreibt die Konzeption, Entwicklung und klinische Evaluierung des dreidimensionalen Ultraschall-Computertomographiesystems der dritten Generation (3D USCT III) zur frühen Brustkrebserkennung. Die Arbeiten wurden am Institut für Datenverarbeitung und Elektronik (IPE) des Karlsruher Instituts für Technologie (KIT) durchgeführt, wo ein vollständig volumetrischer USCT-Ansatz als weltweit erstes System dieser Art entwickelt wird.

Aufbauend auf früheren Demonstratoren, einschließlich des klinisch erprobten Prototyps 3D USCT II, wurde die dritte Generation entwickelt, um die in Patientenstudien identifizierten Einschränkungen zu überwinden. Dazu zählten ein unzureichend großes Bildfeld, Herausforderungen bei der Brustpositionierung, Patientenbewegungen während der mehrminütigen Aufnahme sowie prozedurale Ineffizienzen. Als Reaktion darauf wurde das Bildvolumen auf $20 \times 20 \times 20 \text{ cm}^3$ vergrößert und sämtliche mechanischen, elektronischen und prozeduralen Teilsysteme im Hinblick auf Robustheit, Reproduzierbarkeit und einen verbesserten Workflow überarbeitet.

Ein theoretischer Rahmen basierend auf Fourier-Optik und k -Raummethoden wurde in die akustische Domäne übertragen und ermöglichte die quantitative Bewertung von Aperturdesign, räumlicher Bandbreite, Auflösung und Abtastanforderungen. Diese Modelle leiteten die Entwicklung eines neuen Wandlerkonzepts mit PZT-Piezofasertechnologie, welches eine erhöhte Bandbreite, einen größeren Öffnungswinkel und eine homogene 3D-Feldantwort bietet. Die daraus resultierenden Amplitudenverluste wurden durch ein höheres Signal-Rausch-Verhältnis (SNR) kompensiert, realisiert durch schnellere Messsequenzen, codierte Anregung, Mittelung und deutliche SNR-Gewinne aus der SAFT-Rekonstruktion. Semi-zufällige dünne Abtaststrategien, gestützt durch Konzepte des Compressive Sensing, zeigten, dass nahezu vollständige Informationsgewinnung auch ohne Nyquist-dichte Aperturen möglich ist.

Zur Unterstützung der erforderlichen Datenraten und Anregungsverfahren wurden eine neue μ TCA-basierte Datenakquisitionsarchitektur sowie ein USCT-spezifisches ASIC entwickelt, welche die analoge und digitale Leistung erheblich verbesserten. Jede Messposition erfasst etwa fünf Millionen AScans, was ein ef-

fizientes Datenmanagement und eine enge Synchronisation aller Systemkomponenten erfordert.

Die Systemintegration erfolgte in Zusammenarbeit mit dem Industriepartner Realcan EQ9 und führte zu zwei funktional gleichwertigen Geräten. Die klinische Validierung wurde im Rahmen des EU-Horizon 2020-Projekts QUSTOM am Hospital Vall d'Hebron in Barcelona mit 59 Patientinnen durchgeführt. Die Studie zeigte eine verbesserte Beleuchtungshomogenität, eine Aufnahmezeit von etwa fünf Minuten pro Brust, zuverlässige Messabläufe sowie eine verbesserte Bandbreite, ein gesteigertes SNR und erweitertes Signal-Dynamikpotenzial.

Nach der ersten klinischen Bewertung befindet sich das 3D USCT III-System in der regulatorischen Vorbereitungsphase für eine geplante multizentrische Studie. Damit stellt 3D USCT III einen wesentlichen Schritt hin zur routinemäßigen klinischen Anwendung der volumetrischen Ultraschalltomographie zur frühen und zuverlässigen Brustkrebsdetektion dar.

Abstract

This dissertation presents the conception, development, and clinical evaluation of the third-generation three-dimensional ultrasound computed tomography system (3D USCT) for early breast cancer detection. The system was developed at the Institute for Data Processing and Electronics (IPE) at the Karlsruhe Institute of Technology (KIT) as the first fully volumetric USCT approach of its kind.

3D USCT III was designed to address limitations identified in clinical studies of its predecessor (3D USCT II), including limited imaging volume, breast-positioning variability, patient motion during acquisition, and workflow inefficiencies. The imaging region was expanded to $20 \times 20 \times 20 \text{ cm}^3$, and all major mechanical, electronic, and procedural subsystems were redesigned to improve robustness, reproducibility, and acquisition speed.

A theoretical framework based on Fourier optics and k -space methods was transferred to the acoustic domain, enabling quantitative evaluation of aperture geometry, spatial bandwidth, resolution, and sampling constraints. These considerations guided the development of a novel PZT piezo-fiber transducer with increased bandwidth, larger opening angle, and homogeneous 3D field characteristics. Semi-random sparse sampling strategies, supported by compressive-sensing principles, preserved imaging performance despite sub-Nyquist sampling densities. Enhanced data-acquisition procedures with higher AScan counts enabled substantial SNR gains in reflectivity reconstructions. To meet acquisition and excitation requirements, a new μ TCA-based data-acquisition system and a dedicated USCT ASIC were implemented, significantly improving analog and digital performance.

Two equivalent systems were realized in collaboration with the industrial partner Realcan EQ9. Clinical validation was conducted within the EU Horizon 2020 project QUSTOM at Hospital Vall d'Hebron (Barcelona) with 59 patients. The study demonstrated improved illumination homogeneity, approximately five-minute acquisition time per breast, enhanced bandwidth, and increased SNR and dynamic range. Following initial clinical evaluation, 3D USCT III has entered regulatory preparation for a multicenter trial, representing a substantial step toward routine clinical application of volumetric ultrasound tomography for early breast cancer detection.

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1 Introduction

Ultrasound as an imaging method has established a permanent position in modern medical diagnostics as well as in many industrial applications and domains [NL81]. This work describes the Karlsruhe Institute of Technology’s (KIT) novel 3D multi-modal approach to a medical US imaging device for early breast cancer detection.

1.1 Ultrasound-Based Imaging

Humans are accustomed to light or electromagnetic wave (EM)-based imaging and have an intuitive understanding of its properties. Ultrasound (US)-based imaging, on the other hand, behaves very differently — in some ways unintuitively, yet with fascinating and useful properties. Fig. 1.1 illustrates this contrast from a qualitative perspective. On the left, a common EM-based imaging scene: in front, a cup with pens, and to the left, an apple with buildings in the background. The right side of the figure shows an artificially transformed version of this image, roughly simulating what could be reconstructed under US imaging conditions. With as US as modality, the pens inside the cup become visible, and the grass behind the apple can be discerned owing to the semi-transparent propagation of US through many solids. However, the overall reach of US imaging is smaller because air as the surrounding medium strongly attenuates acoustic waves.

This difference can be explained by the fundamental propagation properties of these two types of waves, EM and mechanical US, as summarized in Tab. 1.1. For EM waves in the visible optical range, attenuation is extremely high in solids such as skin or tissue but almost negligible in gases. For mechanical US waves in the MHz range, the situation is the reverse: attenuation is high in gases, but comparatively low in liquids and soft solids such as biological tissue.



Figure 1.1: *Representing light-based vs US-based imaging. (Left) Light or EM-based imaging of several physical objects. (Right) Hypothetical US image of the same objects under conditions similar to those in US-based imaging systems with water as medium. Origin: photo taken in front of KIT IPE in 2012.*

Generalising these material dependencies, US imaging — unlike EM-based imaging — is characterised by “weak reflectivity” for many solids of interest. Whereas EM waves in the visible range primarily interact with surfaces, US waves often penetrate those same materials, revealing internal structures. Information carried by transmitted or scattered US thus illuminates the interior of the object and exposes its mechanical properties [Gri09, HC09]. In contrast to EM waves, which travel easily through air, US waves experience strong attenuation in air; however, once inside soft matter such as tissue, their attenuation is much lower, enabling volumetric imaging. The measurement medium (often water or tissue) is therefore itself a crucial part of the imaging chain.

A weak reflectivity or “weak scattering” at medium–object interfaces additionally facilitates semi-transparency and enables the formation of many secondary scattered waves. These secondary waves are the basis for multi-static acquisition setups, where multiple transmitter–receiver combinations observe the same region under different propagation paths — a core principle of advanced reconstruction techniques such as synthetic aperture focusing (SAFT) or full-wave tomography.

US as an imaging modality therefore provides a unique opportunity: its weakly scattering nature allows both transmission and reflection information to be exploited, making it naturally suited for imaging the interior of biological tissue. From this fundamental reasoning, US offers significant potential as a medical

Table 1.1: Comparison of intuitive light/EM-wave imaging properties with US-based imaging. For EM waves, attenuation in solids is strong and impedance contrasts are large, causing strong surface reflections [HC09][Gri09]. For US, attenuation in solids is weaker, and impedance contrasts are smaller, enabling weakly reflective, semi-transparent imaging that reveals interior structures.

EM wave ^a	Air	Water	Skin
Attenuation	< 0.001%/cm ^b	0.05%/cm	> 99%/cm
Refractive Index	1.000292	1.33	1.45
US ^c	Air	Water	Fat ^d
Attenuation	49.9%/cm ^e	0.13%/cm ^f	30.4%/cm ^g
Refractive Index			
(Acoustic Impedance) ^h	0.0004	1.49	1.42

^a Green light, $\lambda = 550$ nm (≈ 5.45 THz). ^b Approximately 50% attenuation at 120 km propagation distance. ^c Ultrasound at 2.5 MHz. ^d Fat used as approximation for skin tissue. ^e 1.2 dB/(cm · MHz). ^f 0.0022 dB/(cm · MHz). ^g 0.63 dB/(cm · MHz). ^h Acoustic impedance in MN · s/m³.

imaging tool — combining semi-transparent penetration, controllable focusing, and multi-path information acquisition in a way that complements and extends conventional EM-based modalities.

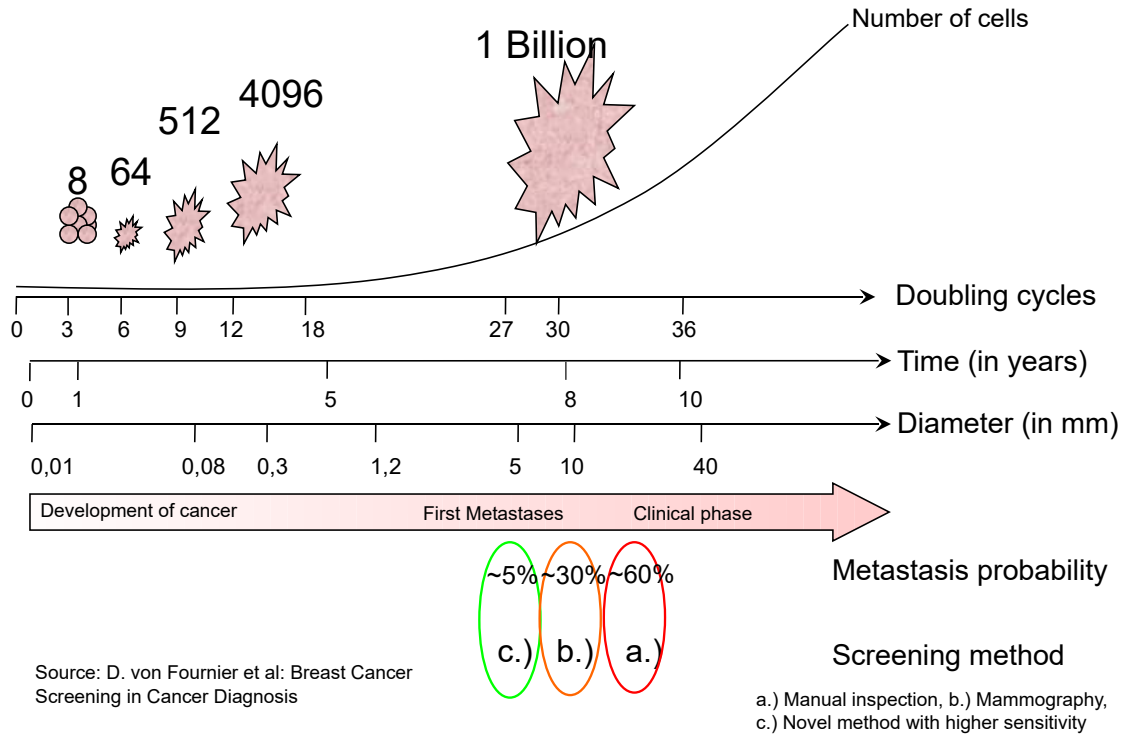


Figure 1.2: Medical motivation for a high-resolution imaging procedure: correlation between doubling cycles of cancer cells, tumor size, and metastasis probability [FAB92].

1.2 Medical Background: Breast Cancer Diagnostics

Breast cancer is among the most common and deadliest cancers in women in western industrialised nations. The survival probability is closely linked to the tumor size, as illustrated in Fig. 1.2. Above a certain threshold size, the growing tumor begins to release metastases into the bloodstream, which correlates with a drastic drop in survival probability [FAB92]. Therefore, the central medical goal of breast cancer diagnostics and screening is to detect tumors as early as possible in their development.

The current medical gold standard for early detection is regular mammography screening. In Germany, for example, such X-ray-based screenings are offered to women starting at the age of 50 years [Gem24]. However, X-ray mammography has several limitations: it relies on ionising radiation, requires painful compression of the breast, and often shows limited reproducibility between sessions. In routine screening practice, images from successive years are compared to monitor struc-

tural changes in breast tissue over long periods — often decades. If a suspicious lesion is identified, a tissue biopsy is performed to verify the diagnosis. Precise spatial localisation of such anomalies is essential for both accurate biopsy guidance and minimally invasive treatment. Conventional X-ray mammography, being a 2-dimensional (2D) projection method with strong tissue deformation, provides only limited spatial information for this purpose.

1.2.1 US and Breast Cancer Diagnostics

Although US offers fundamental advantages over ionising x-ray methods, such as non-ionising radiation and high soft-tissue contrast, conventional sonographic systems still exhibit significant weaknesses.

Standard handheld US scanners (sonography systems) often produce speckle-rich, noisy images (see Fig. 1.3, top-right) with spatially variable resolution, requiring significant interpretative experience from the physician. Moreover, manual sensor operation and pressure-induced tissue deformation make precise spatial localisation and reproducibility difficult. Meanwhile, reproducibility is particularly critical for screening applications and follow-up examinations of suspected cases.

Breast cancer and its precursor structures differ from surrounding media (fat, glandular, and muscle tissue) in several measurable acoustic properties — including speed of sound, attenuation, and acoustic impedance. An US modality that can accurately image or reconstruct these physical parameters holds promise for improved early detection compared to existing diagnostic techniques [GFI81]. Such a system would combine the safety and accessibility of US with the quantitative imaging power of tomographic methods.

1.3 Concept of Ultrasound Computed Tomography as Imaging Method

While US-based imaging has long been established across many domains — such as medicine and industry (non-destructive testing, NDT) — these devices are

1 Introduction

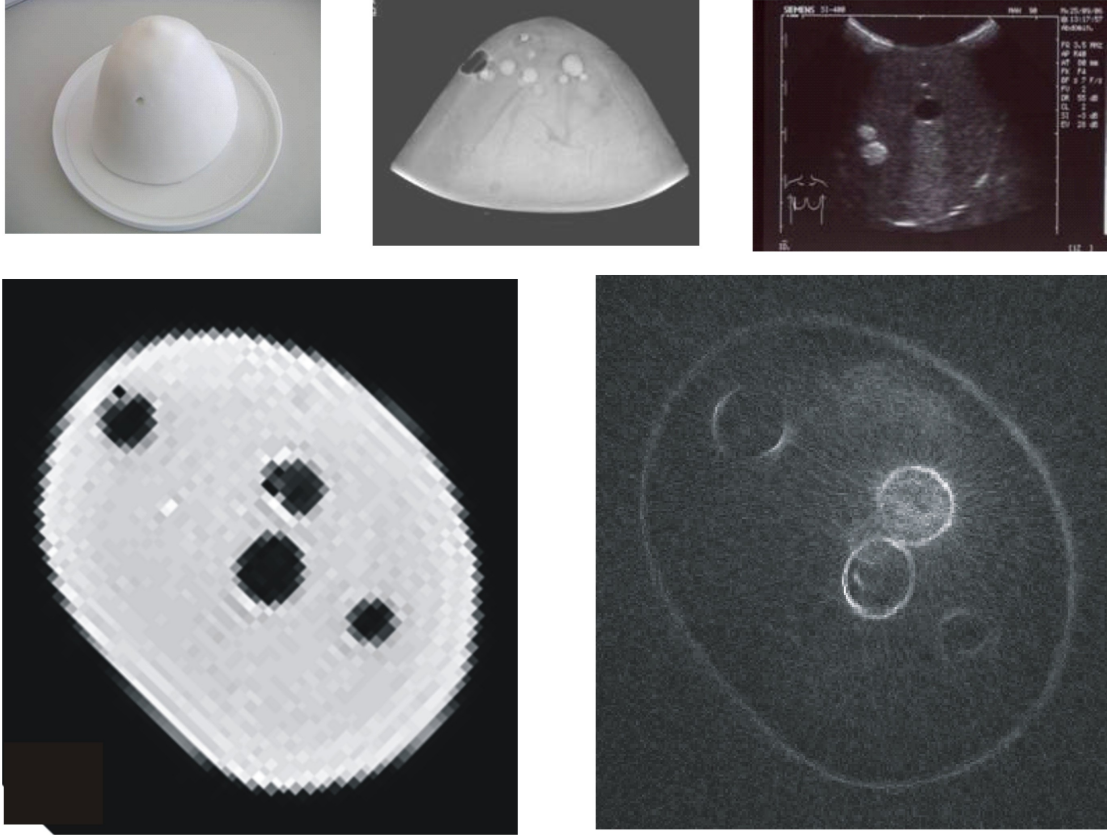


Figure 1.3: Imaging of a breast phantom using different medical methods: (top-left) medical biopsy breast phantom with cancer-like structures, (top-center) X-ray mammogram, (top-right) sonography image, (bottom-left) MRI image, (bottom-right) USCT reflection image.

typically implemented as *uni-modal* systems. An US-based device for early breast cancer detection, however, must go beyond that: it must be *multi-modal*, integrating several physical imaging principles within one system. This leads to the demand for an *Ultrasound Computed Tomograph (USCT)* that allows for both reflection and through-transmission measurements, whereas conventional US systems only conduct reflectivity imaging. To realise such a fundamentally different system, a re-conceptualisation is required — returning to the engineering foundations of wave imaging and redefining the system architecture from first principles.

1.3.1 Imaging Modality and Medium for a USCT

The basic principle of USCT is the use of a mechanical wave as the information carrier. The US waves interact with the human soft tissue, and imaging theory dictates that the object size d needs to be at least half the wavelength λ in order to become visible hence

$$d \geq \frac{\lambda}{2}. \quad (1.1)$$

From this relationship, a wavelengths corresponding to frequencies in the MHz range are required to resolve sub-millimetre structures.

A fundamental prerequisite for US measurements is the presence of a coupling medium. Since attenuation of MHz-range US waves in air is extremely high and the acoustic impedance contrast between air and human tissue is very high, another coupling and propagation medium is necessary. As human tissue consists of approximately 97% water, water serves as an excellent acoustic matching medium between the US source (transducer acting as emitter), object (human tissue), and US sink (transducer acting as receiver). Commercial sonography systems follow the same principle by using a gel, which is primarily water-based.

1.3.2 Medical US Frequencies for USCT

US frequencies between 1 and 10 MHz, as commonly used in medical imaging, are well transmitted in organic materials because human soft tissue consists largely of water. The theoretically achievable spatial resolution increases with frequency, but so does attenuation in both tissue and water. Hence, a balance must be found between resolution and penetration depth [WL53, JFD82]. This trade-off defines one of the key design parameters of a medical USCT system.

1.3.3 US Tomography for Multi-modality Imaging

The current gold standards for breast cancer screening — mammography and magnetic resonance tomography (MRI) — provide high sensitivity but suffer from

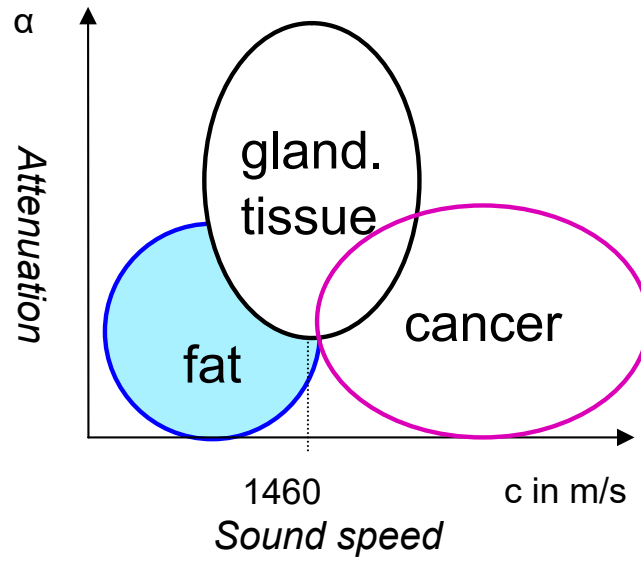


Figure 1.4: Two physical US parameters — speed of sound and attenuation — can be used to separate malignant breast cancer tissue from benign tissue such as fat and glandular regions [GFI81].

limited *specificity* due to their mono-modal nature. As early as the 1970s, Greenleaf proposed US as a means to improve diagnostic specificity. Two intrinsic physical properties of US propagation — *sound speed* and *attenuation* — offer complementary contrast mechanisms for distinguishing malignant from benign tissues [GFI81], as illustrated in Fig. 1.4.

A third US modality is *reflectivity*, which arises from changes in acoustic impedance at tissue interfaces. These reflections provide morphological information about structural boundaries and macro-geometry. Reflection-based imaging (e.g., real-time sonography) is a modality already familiar to clinicians.

Typically, soft tissues show only small variations in their acoustic parameters, resulting in weak scattering effects at their internal boundaries. To simultaneously capture attenuation, transmission, and reflection, the US frequency must be chosen carefully so that all effects are measurable within the same acquisition. The joint extraction of these properties — sound speed, attenuation, and morphology — enables a truly *multimodal* imaging approach.

Exploiting these physical object properties for increased sensitivity and specificity

— as envisioned by Greenleaf — forms the overarching goal of the Ultrasound Computed Tomograph for breast cancer detection.

1.3.4 Ultrasound Tomography for Complex Object Imaging

Breast anatomy presents significant structural complexity and variability — across age, ethnicity, hormonal status, and individual physiology [ELMH⁺13]. It includes a wide range of structures at multiple scales: homogeneous, millimetre-scale regions such as cysts or fat lobules; anisotropic, elongated features such as milk ducts (centimetres long but sub-millimetre in diameter); and diffuse, irregular structures such as malignant lesions or micro-calcifications.

Consequently, the breast cannot be adequately represented by a simple point scatterer, which is often used to characterise imaging system performance (point scatter response). Extending this analogy: if a point scatterer is the simplest possible object, and a non-destructive testing (NDT) task such as detecting delaminations in a pipe represents a medium complexity object, then the breast — with its branching ducts, layered tissues, and varying acoustic contrasts — represents a *highly complex* imaging object. Therefore, real-world 3D medical imaging imposes much higher requirements on imaging systems than 2D point-scatterer tests. These requirements are addressed later in the system design discussion.

1.3.5 Ultrasound Tomography Process and Setup

From the above considerations, a measurement principle can be derived. A USCT requires an aperture of many US transducers arranged around the object in 3D, capable of recording both transmission and reflection information. To extract all three modalities from a single dataset, the imaging step must be decoupled from the measurement process: the raw data (AScans) are recorded and serialised for later reconstruction. This is in contrast to conventional sonography, where image formation occurs concurrently with signal acquisition. The USCT therefore foregoes early analog data reduction in favour of full-wave digital storage, shifting complexity to the computational domain. This embodies the “computed” aspect of the system’s name — enabling flexible, software-based image formation

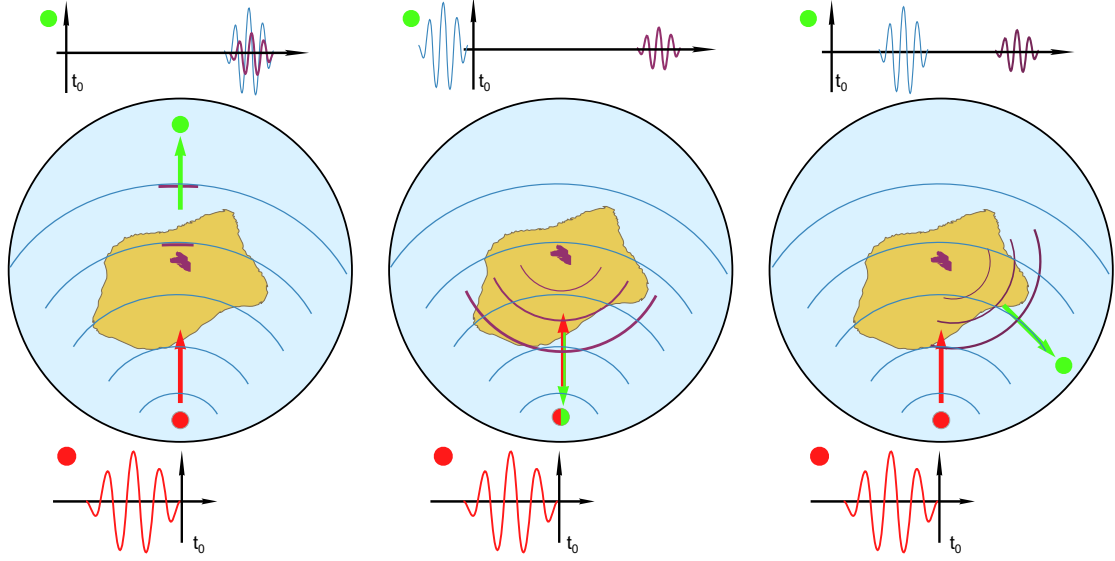


Figure 1.5: Data acquisition modes of a USCT measurement. Red dot: emitter, green dot: receiver. (Top) receiver pressure signal with blue transmission signal and purple the reflection signal. (Center) measurement scheme top-down view. (Bottom) emitter excitation signal. (Left) transmission configuration, information is obtained as amplitude attenuation and time delay; (center) pulse-echo configuration - emitter and receiver coincide, capturing morphological edge reflections; (right) bistatic reflection measurement — emitter and receiver separated, providing angular reflection data.

and algorithmic innovation. The principle of USCT can be illustrated in 2D; the concepts extend directly to the 3D case. Fig. 1.5 shows a schematic measurement setup: a circular transducer aperture surrounds a complex object with both weakly and strongly scattering regions. When US waves interact with the object, they undergo refraction and reflection depending on the local acoustic medium parameters impedance, speed of sound, and density. Where Z denotes acoustic impedance (Rayl), ρ the mass density (kg/m^3), and c the speed of sound which are linked with each other via the relation

$$Z = \rho \cdot c \quad (1.2)$$

[KFCS00].

The receivers measure time-dependent pressure signals starting from the emission of a short pulse, as shown in Fig. 1.5. The spatial distribution of transducers around the object enables two complementary imaging modalities: *transmission tomography* and *reflection tomography*. They can be distinguished by the angle α

between the transmitter and receiver directions.

Simplified, transmission tomography ($\alpha > 90^\circ$) provides measurements of *sound speed* and *attenuation*, as shown in Fig. 1.5 (left); reflection tomography ($\alpha < 90^\circ$) captures reflections from impedance discontinuities between tissue types, encoding morphological information, as illustrated in Fig. 1.5 (right).

1.3.6 Ultrasound Computed Tomography at KIT

At the Institute for Data Processing and Electronics (IPE) at KIT, an *Ultrasound Computed Tomography (USCT)* system is being developed. Its goal is to produce reproducible, spatially resolved images of organic tissue, specifically breasts. The diagnostic value should be comparable to MRI and it should offer specifically early detection of breast cancer and precancerous lesions. The operating frequency range of the US transducers is between 2 and 4 MHz [Wue00]. This range was chosen as an application-specific compromise between penetration depth, sound wave attenuation, and achievable resolution for soft-tissue imaging.

The USCT development at KIT followed a stepwise progression, from two-dimensional to three-dimensional systems, to systematically verify the measurement principle, reconstruction algorithms, and mechanical design.

First Demonstrator: 3D USCT I. With the first three-dimensional demonstrator (**3D USCT I**), the basic measurement and reconstruction principles were validated and optimised using static test objects (phantoms). The medical breast phantom used for calibration and validation was the model 051 from CIRS, Inc. [CIR10], which exhibits realistic breast-like acoustic, radiological, and magnetic properties. This phantom enables a direct comparison between US, X-ray, and MRI based imaging modalities (see Fig. 1.3), and therefore served as a quantitative reference object for the development of multimodal reconstruction methods.

Second Demonstrator: 3D USCT II. Building on these results, the next-generation system (**3D USCT II**) was developed to enable *in-vivo* measurements.

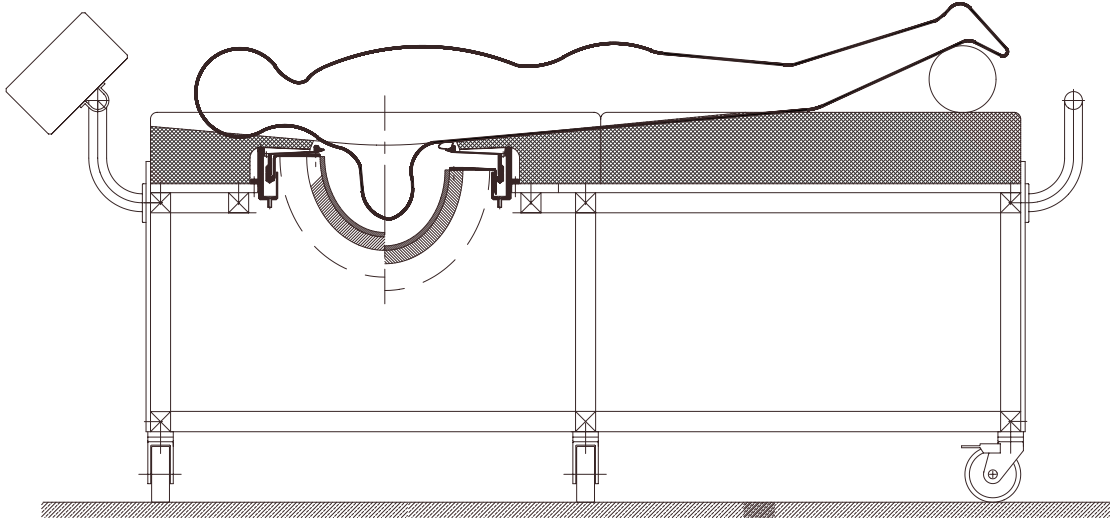


Figure 1.6: *3D USCT II schematic measurement process with participant. The patient lies in a prone position on a mattress; the breast is immersed in a water-filled container embedded in the system center. During the scan, the container can be rotated and lowered to acquire data from various aperture positions, while the patient remains stationary.*

This demonstrator implemented a redesigned mechanical setup, faster acquisition electronics, and refined reconstruction algorithms, making measurements on human participants possible within clinically acceptable time frames. A pilot study conducted at the University Hospital Jena demonstrated the comparability of multimodal diagnostic information from USCT with that of established MRI breast diagnostics. This study confirmed that the USCT approach can achieve clinically relevant image quality and repeatability, while being completely non-ionising.

A sketch of the measurement setup with a participant is shown in Fig. 1.6. The patient lies in a prone position on a mattress; a water-filled measurement container is positioned beneath the breast and contains the transducer aperture. During the measurement, the container can be rotated and vertically adjusted to acquire data from multiple view angles while the patient remains stationary — ensuring reproducibility and comfort during the scan.

Components of the USCT Systems

Measurement Container. The 3D USCT systems (see Fig. 1.7) are built around an approximately hemispherical measurement container where numerous US transducers are embedded within the wall. The entire container can be ro-



Figure 1.7: *3D USCT II overall structure: (center) the side-cover removed measurement setup; (right, integrated in patient bed) assembled measurement container with sockets for transducers and the rotation/lift mechanics below; (top-left) control PC; (top-right) data acquisition (DAQ) hardware.*

tated and vertically adjusted, allowing multiple geometric acquisition positions. Through combined rotation and translation, several million unique transmitter–receiver constellations can be measured. Degassed and temperature-stabilised water serves as the coupling medium between transducers and the centrally positioned object, ensuring acoustic transmission and minimising scattering at the interfaces.

US Transducers. The US transducers functionality is based on the piezoelectric effect and are optimised for a frequency range between 2 and 4 MHz [Gö2]. They are arranged as arrays whose individual elements can be independently controlled as transmitters or receivers. In contrast to conventional diagnostic US systems, the sensors used in the USCT are deliberately designed to be *unfocused*. This design choice enables the emission and reception of approximately spherical wave-

fronts, ensuring uniform angular coverage of the measurement volume and the included region-of-interest (RoI). A smaller active transducer area compared to classical sonography probes results in low emitted pressure levels or poor sensitivity in reception, which in turn necessitates an elaborate signal-processing chain to maintain a sufficient signal-to-noise ratio (SNR).

Digital Data Acquisition and Signal Processing. Data acquisition (DAQ) is carried out using parallelised DAQ hardware with up to 480 channels, each digitised at 10 MHz with 12-bit resolution [M. 06]. To optimise the SNR, several measures are employed: narrow analogue band filtering matched to the excitation frequency, temporal averaging of repeated measurements, and extensive digital signal processing in the subsequent reconstruction pipeline. These methods collectively improve the dynamic range and phase stability necessary for phase-sensitive reflectivity image reconstruction.

Measurement and Imaging Workflow. A complete measurement-to-image workflow is summarised in Fig. 1.8. In a typical USCT acquisition, one transducer is excited by a encoded pulse — either a damped sinusoid or a linear chirp tailored to the transducer’s bandwidth, with a centre frequency between 2 and 3 MHz. Due to the piezoelectric effect of the active element of the transducer, this excitation induces a mechanical oscillation and emission of an acoustic pulse into the water-filled imaging domain. All other transducers act as receivers, converting the resulting pressure fluctuations back into electrical signal via the inverse piezoelectric effect.

The measured electrical signals, representing the time-varying acoustic pressure at each receiver, are digitised and stored as *amplitude scans* (AScans). Each emitter–receiver pair thus forms one AScan; repeating the process for all emitter positions yields millions of AScans for a full 3D dataset. This exhaustive data acquisition is the foundation for subsequent numerical image reconstruction.

Imaging using the Synthetic Aperture Focusing Technique (SAFT). For reflection tomography, 3D image volumes are reconstructed using the *Synthetic Aperture Focusing Technique (SAFT)* — also known as the full-aperture sum-and-delay algorithm, ellipsoid backprojection, or synthetic aperture radar (SAR) reconstruction [AM95, DMS03]. Unlike conventional phased-array sonography,

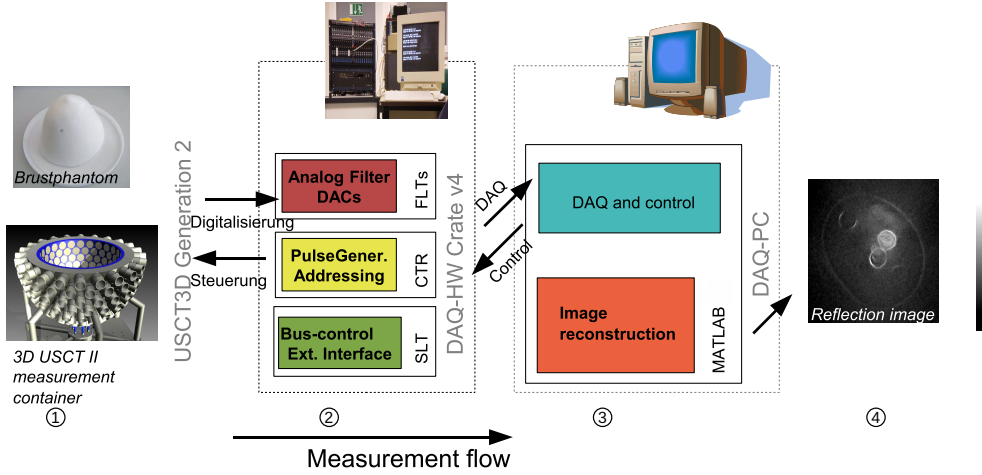


Figure 1.8: 3D USCT II system and measurement workflow showing the steps of a typical measurement process, from left to right: (1) Inserting a breast phantom into the measurement container; (2) excitation of a transmitter via the control board (CTR) while the receiver signals are digitised by the first-level trigger cards (FLT); (3) data transfer through the second-level trigger (SLT, master card) to the control PC via Ethernet; (4) reflectivity image reconstruction.

SAFT enables arbitrary spatial focusing during post-processing rather than during transmission, limited only by sampling accuracy and system calibration [SGSH03].

The fundamental idea of SAFT is to invert the measurement process. For each data point in an AScan (defined by a known emitter and receiver position and a measured time delay), all potential reflection points in the image space are determined. These points form a 3D ellipsoid whose foci are defined by the respective transducer positions. The measured signal value is then assigned to all points on that ellipsoid. This procedure is repeated for all AScans, such that signals from different emitter–receiver combinations constructively interfere only at the true reflection sites in the object. For all other spatial locations, incoherent summation leads to destructive interference and a reduction of background artifacts.

This process constitutes *coherent imaging*, as it requires the precise alignment of phase information and exact geometric knowledge of the measurement setup. In cases where phase coherence cannot be maintained — due to noise, model mismatch, or sampling inaccuracies — an alternative approach known as *incoherent imaging* can be used. Here, the positive signal envelope (obtained via Hilbert transformation) is summed, offering improved robustness at the expense of spatial resolution [SGSH03].

1.4 Motivation of this Work

This work aims to design and realise the *4th generation 3D USCT prototype (3D USCT III)* — the final system intended for deployment in medical institutions. It builds upon all prior developments of the USCT project and systematically addresses the technical and clinical requirements identified in preceding experimental and medical studies. In particular, this prototype will be constructed at two locations — one at KIT, Campus North (Karlsruhe), and one at Equilibrium 9 (EQ9) in Hangzhou — to enable parallel testing, development, and validation. The 3D USCT III prototype incorporates the lessons learned from the pre-study conducted at the Jena University Hospital (UKJ) and targets the following improvements:

- **Reduce measurement times** to below three minutes to meet clinical workflow requirements.
- **Increase robustness** of the mechanical setup, sensor coupling stability, and data acquisition reliability under long-term operation.
- **Extend applicability** to a broader range of patients by adapting physical parameters and system geometry, allowing more flexible positioning and alignment within the optimal RoI.
- **Ensure consistent RoI coverage** despite patient-specific anatomical variability, such as floating or non-centred breast positioning within water filled measurement container [RGHZ15].
- **Address signal-to-noise ratio (SNR) weaknesses** in the reflectivity imaging modality (SAFT reconstruction) caused by sparse, regular apertures and limited bandwidth through improved excitation and signal processing.
- **Resolve usability, safety, and regulatory challenges** by advancing the control architecture and establishing a compliant medical safety concept suitable for CE certification.

1.5 Scientific Contributions

This work contributes to a broad range of scientific and engineering domains, while ensuring that each theoretical and algorithmic advancement supports a practical medical imaging device.

Specifically, this work addresses the following research and engineering contributions:

- **Simulation and evaluation of transducer and aperture geometries** leading to quantitative design specifications for the mechanical and acoustic layout of the 3D USCT III prototype.
- **Simulation of frequency characteristics and bandwidth effects** for SAFT-based reflectivity imaging, advancing from traditional pressure- and sensitivity-optimised designs toward bandwidth- and divergence-driven transducer designs that enhance separability and image contrast.
- **Development of an optimisation framework for motion control** determining optimal movement patterns for the two motorised degrees of freedom in the aperture system, enabling efficient spatial coverage and reduced acquisition time.
- **Extension of temperature monitoring and compensation models** allowing dynamic modelling of acoustic and electronic drift over both space and time for increased measurement stability.
- **Design of the DAQ and measurement architecture** to handle large dynamic ranges, high data throughput, and time-synchronised signal capture from hundreds of parallel channels.
- **Alternative imaging modality: k-space reflectivity imaging** — a Fourier-based reconstruction approach providing time-efficient image formation for comparison with SAFT reconstructions.
- **Development of a lossless 2D and 3D image rotation algorithm** enabling iterative plane-wave reconstruction and orientation correction in multi-view imaging scenarios.

- **Algorithms for quantitative system evaluation and health monitoring** providing metrics for system performance, calibration stability, and automatic fault detection.
- **Advancement of an ISO-standard-based medical safety concept** ensuring conformity with CE certification requirements and defining the technical safety case for clinical operation.

Connection to Theoretical Modeling. The following chapters establish the theoretical and physical framework upon which these developments are based. Specifically, the next sections formalise the acoustic system concept, the linear wave model, and the sampling-theoretic interpretation of USCT imaging. These theoretical foundations link the practical design choices — aperture geometry, excitation bandwidth, and sampling density — to measurable image properties such as resolution, separability, and signal-to-noise ratio. This transition from system motivation to wave and k-space modeling ensures that later hardware and reconstruction design decisions remain grounded in the physical and mathematical principles described in the system design chapter.

2 Fundamentals of US Imaging

2.1 State of the Art

The topic of *Fourier-based simulation of imaging systems* in the context of US has been an active area of research for many years. In the original design of USCT, similar concepts were already used to assess the three-dimensional properties of the aperture ([Wue00], pp. 48–51, Chapter 3.1.2).

2.1.1 Existing Approaches for Simulating US Systems

Software frameworks for simulating US transducers and systems exist, such as Jensen’s *Field II* [Jen96a]. However, they are typically optimized for two-dimensional phased array configurations and planar transducer geometries. More general wave-based approaches, such as finite-element or finite-difference time-domain simulations implemented in *ANSYS* [ANS], *Wave3000* [Cyb], or *k-Wave* [TC10], provide higher physical accuracy but are computationally expensive. As demonstrated in previous studies [Web05], such methods are not capable of modeling the full three-dimensional USCT system with sufficient resolution or within reasonable computational limits.

The USCT project, as a 3D synthetic-aperture-focused (SAFT) imaging system, therefore requires a modeling approach that is both flexible and computationally efficient, yet still based on physically meaningful wave propagation.

A realistic system simulation and performance prediction was first established in the doctoral thesis of Schwarzenberg [Sch08a]. The central idea of that work was

to provide realistic image predictions and quantitative quality metrics that could serve as reference benchmarks for experimental systems by applying the same imaging algorithms used in reconstruction to the simulated data. This resulted in the need for a fast efficient forward solver.

2.1.2 Fourier-Based System Approaches for Simulating US Systems

This work extends the existing SAFT simulation framework by establishing a *Fourier-based modeling approach*. Such an approach can be succinctly characterized as: “...the use of spatial- and temporal-frequency domain representations of US system impulse responses and scattering functions to understand and analyze both conventional and hypothetical imaging methods.” (J. Goodman [Goo68]). This framework enables a simplified yet rigorous description of USCT as a linear, wave-based imaging system.

Consequently, the theoretical foundation of this work relies on concepts from Goodman’s *Fourier Optics* [Goo68], complemented by Anderson and Tahey’s *k-space Tutorial* [AT00] and by Kak and Slaney’s *Principles of Computerized Tomographic Imaging* [KS01].

The topics addressed in this work span multiple disciplines, medical imaging, US transducer design, system modeling, signal processing, and wave physics. The following sections introduce the relevant theoretical concepts and mathematical tools required to unify these fields into a consistent framework for USCT modeling. For detailed derivations and proofs, the reader is referred to the cited literature.

2.2 System Concept

A core concept in Fourier optics is the description of an imaging process as a *system* [Goo68]. This section introduces the general system concept and its relevant properties as applied to imaging.

The term *system* originates from electrical engineering, where it refers to networks

of components that describe the relationship between an input and an output quantity (e.g., voltage, current, or resistance) as a function of an independent variable, typically time. In an imaging context, these variables are extended to spatial coordinates (x, y) in 2D and (x, y, z) in 3D. Mathematically, a 2D system connects an input function g_1 with an output function g_2 via the system response operator S as

$$g_2(x_2, y_2) = S \{g_1(x_1, y_1)\}. \quad (2.1)$$

2.2.1 Linearity

A particularly important subclass of general systems is the *linear system*, which satisfies the properties of determinism and superposition (decomposability) for its input functions. This property is expressed as:

$$S\{a p(x_1, y_1) + b q(x_1, y_1)\} = a S\{p(x_1, y_1)\} + b S\{q(x_1, y_1)\}. \quad (2.2)$$

where p and q are arbitrary input functions, and a and b are scalar coefficients, and S denotes the system operator. Physically, this means that the system's response to a combined excitation is equal to the sum of the responses to each excitation individually. Such systems may respond to both real and complex (i.e., phase-carrying) input signals, which is particularly relevant in wave-based imaging. The property of linear superposition enables the decomposition of a complex input into elementary stimuli—such as spatial point sources, transducer sub-elements, or frequency components—whose responses can then be recombined to describe the total system behavior.

Superposition Integral

The concept of superposition can be formalized using the *superposition integral*, which equals for a 2D spatial domain:

$$g_2(x_2, y_2) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} g_1(x_i, y_i) H(x_i, y_i; x_2, y_2) dx_i dy_i, \quad (2.3)$$

where H denotes the *impulse response* or *point spread function* of the system, g_1 represents the input or object function, and g_2 the resulting system response. This formulation explicitly connects the local system response to each input point, forming the foundation for modeling optical and acoustic imaging systems as linear operators acting on spatial distributions.

2.2.2 Invariance

A further simplification of the system response H arises when it is assumed to be invariant with respect to absolute position—i.e., when the system exhibits *shift invariance*. In this case, the system response depends only on the relative displacement between input and output coordinates, not on their absolute positions. The superposition integral then takes the simpler form

$$g_2(x_2, y_2) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} g_1(x_i, y_i) H(x_2 - x_i, y_2 - y_i) dx_i dy_i. \quad (2.4)$$

This expression can be interpreted as the *convolution* of the input function with the system's impulse response, denoted as

$$g_2 = g_1 \otimes H. \quad (2.5)$$

Linear, shift-invariant systems are of particular interest because they can be fully characterized by their impulse response—or, equivalently, by their frequency response in the Fourier domain. This property forms the mathematical foundation for describing imaging performance through point spread functions (PSFs), modulation transfer functions (MTFs), and related frequency-domain representations.

In the next section, it will be discussed to what extent such linear and shift-invariant system considerations can be applied to US imaging systems, where wave propagation, medium heterogeneity, and nonlinear transducer behavior may challenge the underlying assumptions.

2.3 Ultrasound Physics

The applicability of system theory to US imaging can be motivated by considering US as a physical wave phenomenon. As succinctly described by Insana and Brown:

“Fluids have elasticity (compressibility κ) and inertia (mass density ρ), the two characteristics required for wave phenomena in a spatially distributed physical system whose elements are coupled. Elasticity implies that any deviation from the equilibrium state of the fluid will tend to be corrected; inertia implies that the correction will tend to overshoot, producing the need for a correction in the opposite direction and hence allowing for the possibility of propagating phenomena—acoustic pressure waves.” [IB93]

This intuitive description can be formalized by the scalar 1D wave equation, which models the oscillatory behavior of acoustic pressure p as a function of time t and spatial position $\vec{r} = (x, y, z)$:

$$\frac{\partial^2 p}{\partial z^2} = \rho_0 \kappa \frac{\partial^2 p(\vec{n}, t)}{\partial t^2}, \quad (2.6)$$

$$\nabla^2 p(\vec{r}, t) = \rho_0 \kappa \frac{\partial^2 p(\vec{n}, t)}{\partial t^2}, \quad (2.7)$$

where ∇^2 is the Laplacian, ρ_0 is the equilibrium mass density of the medium and κ its compressibility. The term $\sqrt{1/(\rho_0 \kappa)}$ corresponds to the propagation velocity c of the longitudinal acoustic wave in the medium.

This linear partial differential equation demonstrates that small acoustic pressure variations in a homogeneous medium behave as *linear waves*. Hence, to first order, acoustic wave propagation can be treated as a linear, time-invariant (LTI) system, where the superposition principle applies. This observation justifies the use of linear system theory—originally developed for optics and electrical networks—for US wave propagation and imaging.

For further modeling of US imaging systems, existing analytical frameworks from electromagnetic wave theory—known collectively as *optics*—can be directly leveraged, since both acoustic and optical waves obey analogous forms of the wave

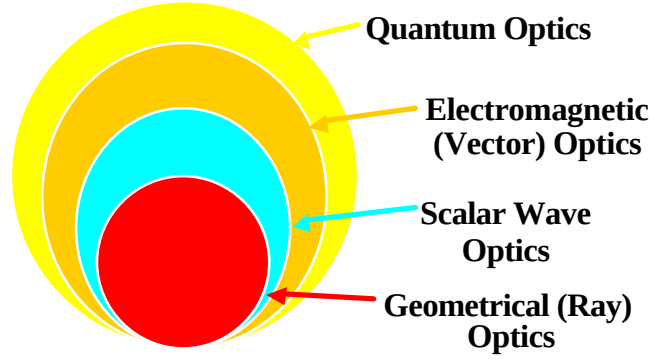


Figure 2.1: *Hierarchy of optics according to [Jag00], ranging from Snell’s geometric optics through wave equation-based diffraction optics to quantum optics.*

equation.

2.4 Optics and Ultrasound

The connection between US and electromagnetic optics arises from their shared wave nature. Both phenomena can be described by similar scalar or vector wave equations and therefore exhibit analogous propagation, diffraction, and interference behavior.

Wave-based optical theory offers a hierarchy of modeling approaches of increasing physical completeness, as illustrated in Figure 2.1. At the simplest level lies *geometric optics*, governed by Snell’s law and ray tracing. More comprehensive descriptions, such as *wave (or diffraction) optics*, explicitly account for interference and diffraction phenomena via the wave equation. At the highest level, *quantum optics* extends these models to the quantized nature of light and photon–matter interactions.

Since US imaging is governed by wave propagation at macroscopic scales, it corresponds most closely to the *diffraction optics* regime. In this regime, the scalar wave equation and Fourier-based methods—such as the angular spectrum and Fresnel or Fraunhofer approximations—can be applied. This analogy provides the theoretical foundation for the Fourier-based system description of USCT developed in the subsequent sections.

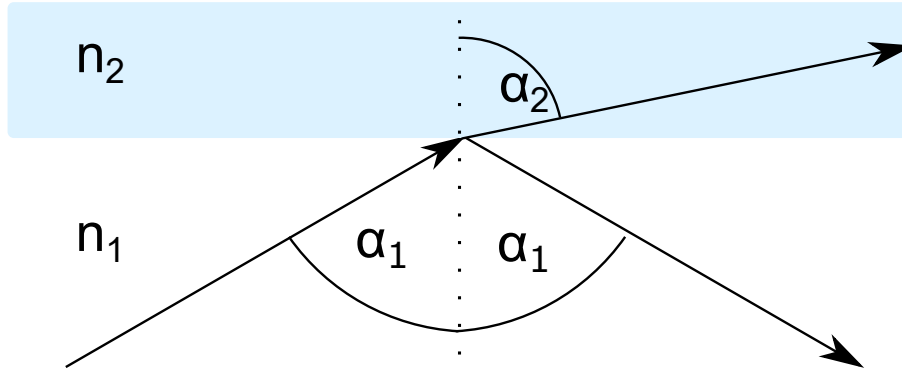


Figure 2.2: *Elementary effects of geometric optics: refraction at a boundary layer with incident angle α_1 and exit angle α_2 , reflection at a boundary layer with angle α_1 .*

2.4.1 Geometric Optics: Reflection and Refraction

The oldest and simplest description method is *geometric optics*, which approximates wave phenomena as rays. This allows the interaction of the US system with the measured object to be determined depending the wavelength and corresponding medium properties. This is formulated in Snell's law, i.e.

$$n_1 \cdot \sin \alpha_1 = n_2 \cdot \sin \alpha_2, \quad (2.8)$$

with n_1, n_2 the refractive indices (ratio of reference velocity c_0 to sound velocities c_1, c_2) and α_1, α_2 as angles relative to the boundary surface, see Fig. 2.2.

For objects larger than $\lambda/2$, Snell's law applies to reflection and scattering angles and depends on the scalar value of the sound speed. If the object to be imaged and the elements of the imaging system are of similar size or smaller than the wavelength, this description is insufficient, and a more complex modeling is needed.

2.4.2 Scalar Wave Optics: Interference and Diffraction

The basic idea in this section is to describe US as waves instead of rays. Elements for uniquely characterising a wave are its *amplitude*, i.e., the strength of its oscillation, and the direction of its propagation as a *direction vector*, and *phase* i.e. the state of the wave. The central point of wave optics is the assumption that

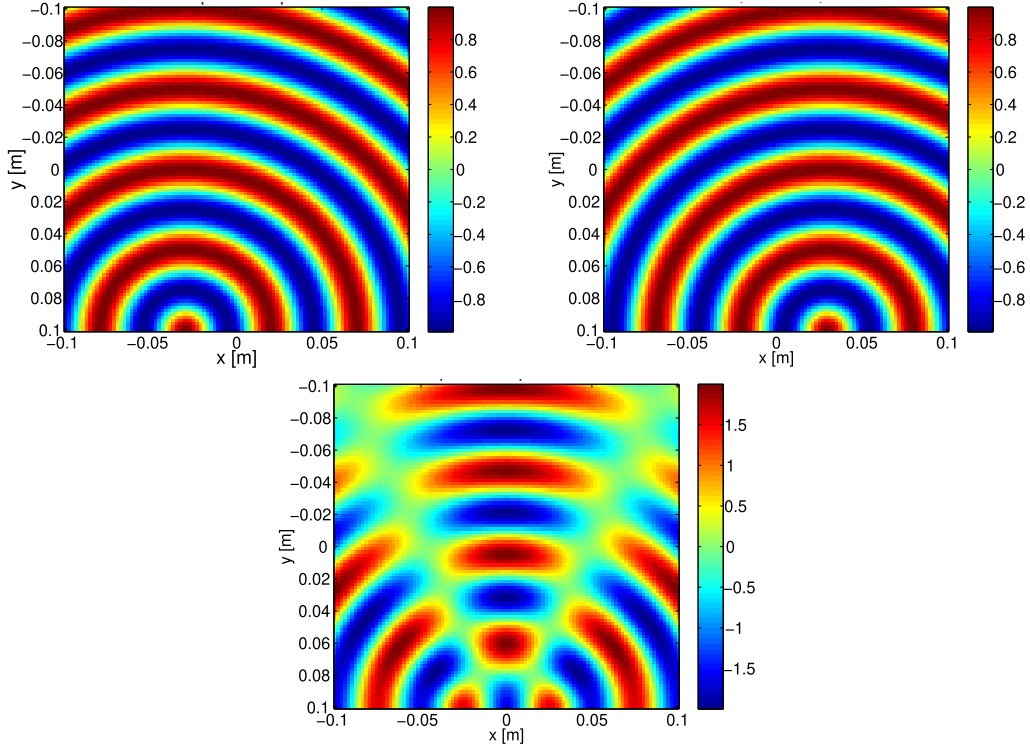


Figure 2.3: *Interference of two circular waves: (top left, top right) two time-stationary, monochromatic circular waves with a wavelength of 0.02 m and where, the wave centers are shifted by 0.06 m; (bottom) constructive and destructive interference of the two superimposed circular waves. Sound pressure is color-coded, red positive pressure and blue negative pressure.*

waves have a phase, which can overlap constructively or destructively with other waves, see Fig. 2.3. *Interference* is the effect that results from this, which can be modeled using the interference of Huygens' elementary waves.

This means for the US field that its interactions with point scatterers (an object with a size close to or below the wavelength) can no longer be described using Snell's ray geometry. The predicted behavior of a purely directional reflection does not occur in reality. In particular, the wave seems to spread around the point scatterer; while this scatterer exhibits omnidirectional radiation behavior. These effects are summarised under the term diffraction and are attributed to physical effects that can be explained by considering a wave with a phase.

This property allows interaction with objects close to and below the wavelength to be described. This is necessary for USCT, as both the objects to be imaged, as well as the functional elements of the USCT imaging system, are near or below

the wavelength used [Wue00].

2.4.3 Vectorial Wave Optics and Beyond

A vectorial description is necessary, for example, for electromagnetic waves, as these are driven by the two vectorial quantities – electric field and magnetic field. In the case of US imaging, the mechanical waves can typically be approximated by a scalar quantity the pressure field [Hui08] [vW06]. The scalar pressure field encapsulates the compressional mode as shear components are negligible in fluids.

But even for non-liquid media such as solids or structured objects where the underlying wave field is generally vectorial, since both longitudinal and shear waves can propagate, the scalar description is sufficient when two conditions are met [Goo68]:

- the characteristic dimensions are large compared to the wavelength, and
- the field is not evaluated too close to strongly diffracting boundaries,

Under such conditions a predominantly longitudinal response is obtained, and shear-wave effects become small. The field can be *reasonably well approximated* by a scalar description enabling even non-liquid ultrasound applications. These assumptions align well with typical USCT configurations, where water or soft tissue acts as an isotropic coupling medium and transducers operate in the far field relative to the region of interest.

Quantum optics are irrelevant for pressure waves, since quantum effects only occur at many orders of magnitude below the range relevant in the USCT demonstrator (which is approximately 1 to 1×10^{-4} m).

2.5 Diffractive US Systems as Linear Invariant Systems

The goal is to derive a scalar wave description for diffractive systems that meets the aforementioned conditions of invariance and linearity.

2.5.1 Spatial and Temporal Invariance in the US Context

Here we develop a physical interpretation of whether the system approach is *invariant* for USCT. Temporal invariance means that between consecutively sent or received pulses, the system does not change (e.g. degrade, etc.). This is approximately fulfilled as the piezoceramics used are chemically stable and operate well below the potential depolarisation voltage (which is several hundreds of Volts). In addition, it is valid to assume that the object to be scanned does not change during scanning.

Spatial invariance means that the system properties are identical for every point in space, which is generally not the case. Fortunately, for a limited region, in our case the *Region-of-Interest (RoI)*, this assumption is valid.

2.5.2 Scalar Wave Equation for Homogeneous Media

The first part begins with a general mathematical description of a wave propagation system and then introduces a physics interpretation.

We start with a purely monochromatic scalar pressure wave $U(\vec{r}, t)$ for homogeneous media with sound speed c_0 , with $\vec{R} = (x, y, z)$ for the spatial position and t for time

$$u(\vec{r}, t) = A(\vec{r}) \cdot \cos(2\pi f t + \phi(\vec{r})) \quad (2.9)$$

where $A(\vec{r})$ as the position-dependent amplitude function, f is the temporal frequency and $\phi(\vec{r})$ as the phase function. Here, $u(\vec{r}, t)$ represents the *acoustic pressure field* in the US case. The medium is assumed to be homogenous with speed of sound c_0 .

A more compact representation of the pressure field using the complex unit number $j = \sqrt{-1}$ equals

$$u(\vec{r}, t) = \Re\{U(\vec{r})e^{-j2\pi f t}\} \quad (2.10)$$

where $\Re$ denotes the “real part of”, and

$$U(\vec{r}) = A(\vec{r})e^{-j\phi(\vec{r})} \quad (2.11)$$

is the complex-valued amplitude distribution $U(\vec{r})$ or complex acoustic pressure amplitude. Clearly, the pressure field must satisfy the scalar wave equation for every point in *space* and *time*, hence

$$\nabla^2 u(\vec{r}, t) - \frac{1}{c_0^2} \frac{\partial^2 u(\vec{r}, t)}{\partial t^2} = 0. \quad (2.12)$$

A decoupling between space and time can be achieved by assuming that only a single *temporal frequency* f is considered (stationary state). Then $u(\vec{r}, t)$ is solely described by $U(\vec{r})$, and by substituting Eq. 2.11 into Eq. 2.12, the time-independent Helmholtz equation can be obtained

$$\nabla^2 U(\vec{r}) + k_0^2 U(\vec{r}) = 0 \quad (2.13)$$

with wave number $k_0 = \frac{2\pi}{\lambda_0} = \frac{2\pi f}{c_0}$, where λ_0 is the wavelength of the spatial wave in the homogenous medium.

Expanding the Laplacian operator ∇^2 results in the 2D form (for illustration):

$$\frac{\partial^2 U(\vec{r})}{\partial x^2} + \frac{\partial^2 U(\vec{r})}{\partial y^2} + k_0^2 U(\vec{r}) = 0 \quad (2.14)$$

Following the reasoning of Kak and Slaney [KS01] (p. 206, Eq. 9), we assume as a possible solution $U(\vec{r}) = e^{j\vec{k}\cdot\vec{r}}$, with $\vec{k} = (k_x, k_y)$ as a 2D propagation vector, and $U(\vec{r})$ then representing a 2D plane wave with the spatial frequency $|\vec{k}|$. This corresponds to the 2D Fourier transformation and allows arbitrary functions to be formed as sums of plane waves. Only planar waves that satisfy Eq. 2.15 fulfill the wave equation; hence

$$|\vec{k}|^2 = k_x^2 + k_y^2 = k_0^2 \quad (2.15)$$

Further results of this formulation can be found in [KS01], p. 206 ff.

2.5.3 Scalar Wave Equation for Heterogenous Media

In general, the medium is heterogenous and the speed of sound $c(\vec{r})$ is spatial dependent. Accordingly, the spatial wavenumber becomes $k(\vec{r}) = \frac{2\pi f}{c(\vec{r})} = |\vec{k}(\vec{r})|$. Therefore, the Helmholtz equation (Eq.2.13) for heterogenous media must include this spatial dependence:

$$\nabla^2 U(\vec{r}) + k^2(\vec{r}) U(\vec{r}) = 0. \quad (2.16)$$

Here, $U(\vec{r})$ again represents the complex amplitude of the acoustic pressure field, and $k(\vec{r})$ the local spatial frequency that varies with $c(\vec{r})$ and hence *vecr*.

Since the wavenumber depends on position, the Laplacian can no longer be separated from $k(\vec{r})$. To analyze the resulting wave field, the resulting wave number is approximately equal (up to first order)

$$k^2(\vec{r}) = k_0^2 + 2k_0 \Delta k(\vec{r}), \quad (2.17)$$

where $\Delta k(\vec{r}) = k(\vec{r}) - k_0$ describes the local deviation of the wave number. For small heterogenities ($|\Delta k(\vec{r})| \ll k_0$) or equivalently $|\Delta c(\vec{r})| \ll c_0$, the resulting problem can be treated as a weak scattering perturbation.

Substituting Eq. 2.17 into Eq. 2.16 gives

$$\nabla^2 U(\vec{r}) + k_0^2 U(\vec{r}) = -2k_0 \Delta k(\vec{r}) U(\vec{r}) \quad (2.18)$$

which separates the homogeneous part (left-hand side) from the heterogenous perturbation term (right-hand side). This expression serves as the starting point for the Born and Rytov approximations describing weak scattering in heterogenous acoustic or optical media.

When $\Delta k(\vec{r}) = 0$, Eq. 2.18 reduces again to the Helmholtz equation, see Eq. 2.13, for inhomogenous media, verifying consistency.

Reflections occur at material boundaries where the materials are characterised by different densities and compressibilities. For imaging that represents these, such

a formulation is necessary. Key elements here are the *Born and Rytov approximations*, which are not further elaborated but referred to in [Goo68] and [KS01] p. 212.

2.5.4 Angular Spectrum of Planar Waves

Among the aforementioned variants of introducing the scalar wave equation, the one used by [Goo68] (page 55, Chapter 3.10) is employed in this work, as it is closest to linear system theory. The basic idea is that for the wave-field $U(x, y, z = 0)$ at a planar aperture on the plane $z = 0$, the diffractive superpositions for any subsequent z -planes $U(x, y, z)$ result from the superposition of phase-shifted planar waves propagating from $U(x, y, z = 0)$, see Fig. 3.16.

In the following, the z -dimension denotes the propagation direction, which is also the normal vector to the planar aperture plane. The 2D Fourier description along $z = 0$ is then:

$$A(f_x, f_y; f_z = 0) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} U(x, y, z = 0) e^{-j2\pi(f_x x + f_y y)} dx dy \quad (2.19)$$

where $A(f_x, f_y; z = 0)$ is the spectrum.

Fourier transformation Eq. 2.19 can be considered as a linear decomposition of a complex function into simpler ones, planar waves in this case. To strengthen this perspective, we invert the equation, yielding the inverse spatial Fourier transformation of its spectrum

$$U(x, y, 0) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} A(f_x, f_y; 0) e^{j2\pi(f_x x + f_y y)} df_x df_y \quad (2.20)$$

In Eq. 2.20, the physical meaning can be directly interpreted, i.e. the inner part of the integral represents a *planar wave* with a direction vector.

The complex amplitude P of a time-independent, complex valued plane wave is

along the z -dimension is given by ([Goo68], Eq. 3-61):

$$P(x, y, z) = e^{j\vec{k}\cdot\vec{r}} = e^{j\frac{2\pi}{\lambda}(\alpha x + \beta y + \gamma z)}, \quad (2.21)$$

with $\vec{r} = x\hat{x} + y\hat{y} + z\hat{z}$ and $\vec{k} = \frac{2\pi}{\lambda}(\alpha\hat{x} + \beta\hat{y} + \gamma\hat{z})$, where $\hat{x}, \hat{y}, \hat{z}$ are the unit vectors.

The directional cosines are related by $\gamma = \sqrt{1 - \alpha^2 - \beta^2}$. Therefore, along the plane $z = 0$, the complex function $e^{j2\pi(f_x x + f_y y)}$ can be described as *wave propagation* with

$$\alpha = \lambda f_x, \quad \beta = \lambda f_y, \quad \gamma = \sqrt{1 - (\lambda f_x)^2 - (\lambda f_y)^2}. \quad (2.22)$$

In the Fourier decomposition of U , the planar complex-valued wave amplitude with spatial frequencies (f_x, f_y) becomes $A(f_x, f_y; f_z = 0) df_x df_y$ for $f_x = \alpha/\lambda$ and $f_y = \beta/\lambda$.

For this reason, the *wave propagation* is referred to as the angular spectrum of $U(x, y, 0)$:

$$A\left(\frac{\alpha}{\lambda}, \frac{\beta}{\lambda}; 0\right) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} U(x, y, 0) e^{-j2\pi\left(\frac{\alpha}{\lambda}x + \frac{\beta}{\lambda}y\right)} dx dy. \quad (2.23)$$

2.5.5 Propagation of the Angular Spectrum Along Z-axis

For an angular spectrum, let us now consider a distance z , corresponding to the propagation of the planar wave along z :

$$A\left(\frac{\alpha}{\lambda}, \frac{\beta}{\lambda}; z\right) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} U(x, y, z) e^{-j2\pi\left(\frac{\alpha}{\lambda}x + \frac{\beta}{\lambda}y\right)} dx dy. \quad (2.24)$$

If the relationship between $A\left(\frac{\alpha}{\lambda}, \frac{\beta}{\lambda}; 0\right)$ and $A\left(\frac{\alpha}{\lambda}, \frac{\beta}{\lambda}; z\right)$ can be found, the wave-propagation effect is completely described.

Using the Fourier integral theorem, Eq. 2.24 can be reformulated as:

$$U(x, y, z) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} A\left(\frac{\alpha}{\lambda}, \frac{\beta}{\lambda}; z\right) e^{j2\pi\left(\frac{\alpha}{\lambda}x + \frac{\beta}{\lambda}y\right)} d\left(\frac{\alpha}{\lambda}\right) d\left(\frac{\beta}{\lambda}\right). \quad (2.25)$$

Since U must also satisfy the Helmholtz equation, A must fulfill the corresponding

differential equation ([Goo68], page 57):

$$\frac{d^2}{dz^2} A\left(\frac{\alpha}{\lambda}, \frac{\beta}{\lambda}; z\right) + \left[\left(\frac{2\pi}{\lambda}\right)^2 (1 - \alpha^2 - \beta^2)\right] A\left(\frac{\alpha}{\lambda}, \frac{\beta}{\lambda}; z\right) = 0. \quad (2.26)$$

An elementary solution of this equation can be expressed as

$$A\left(\frac{\alpha}{\lambda}, \frac{\beta}{\lambda}; z\right) = A\left(\frac{\alpha}{\lambda}, \frac{\beta}{\lambda}; 0\right) e^{j \frac{2\pi}{\lambda} \sqrt{1 - \alpha^2 - \beta^2} z}. \quad (2.27)$$

The solution set $\alpha^2 + \beta^2 < 1$ corresponds to a pure phase shift for different propagation distances z . For $\alpha^2 + \beta^2 > 1$, an imaginary root arises, resulting in effective attenuation during propagation—a physical phenomenon known as *evanescent waves*. This explains why no angular-spectrum contribution is possible beyond the non-imaginary solution set: only components with real k_z physically propagate to the measurement plane. Additionally, this provides the crucial explanation of why no object resolution finer than the wavelength of the incident wave is physically achievable. Spatial frequencies representing sub-wavelength details lie in the evanescent region and consequently decay before they can be measured.

2.5.6 Amplitude Transfer Function

Following Goodman [Goo68] (page 58, Eqs. 3-70 and 3-71), the *Amplitude Transfer Function* (ATF) provides a linear description of how a thin, planar structure (e.g. an aperture, pupil, or amplitude/phase mask) modifies an incoming complex wave field. The ATF t_A is defined *in a single plane* as the ratio of the output field U_t to the input field U_i over the aperture plane:

$$t_A(x, y) = \frac{U_t(x, y, 0)}{U_i(x, y, 0)}. \quad (2.28)$$

We seek the resulting field as:

$$U_t(x, y, 0) = t_A(x, y) \cdot U_i(x, y, 0). \quad (2.29)$$

Using the Fourier convolution theorem, see Eq. 2.49, the same holds for the spatial-

frequency spectra:

$$A_t(f_x, f_y) = A_i(f_x, f_y) \otimes T(f_x, f_y), \quad (2.30)$$

where

$$T(f_x, f_y) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} t_A(x, y) e^{-j2\pi(f_x x + f_y y)} dx dy. \quad (2.31)$$

Thus, diffraction through an aperture or a planar transmission field is represented as a convolution between the input angular spectrum $A_i(f_x, f_y)$ and the transfer function $T(f_x, f_y)$ in the spatial-frequency domain.

2.5.7 Coherent and Incoherent Imaging Systems

If the considerations from Sec. 2.5.6 are applied to complex-valued (phase-sensitive) plane waves, a description of a *coherent* imaging system emerges. US-based imaging systems with phase-accurate beamforming typically belong to this class, since both amplitude and phase of the field $U(\vec{r})$ are preserved and contribute to image formation.

If, on the other hand, the same modeling is applied to the phase-independent *intensity*

$$I_i(x, y, z) = |U_i(x, y, z)|^2, \quad (2.32)$$

an *incoherent* imaging system is described — as is the case for most electromagnetic imaging modalities (e.g. X-rays or optical radiography), where only the intensity information contributes to image formation.

2.5.8 Fourier Diffraction Theorem

In contrast to the 2D, plane-anchored ATF of Sec. 2.5.6, the *Fourier Diffraction Theorem* (FDT) provides a 3D model for wave propagation and object-induced scattering under the weak-scattering (Born) approximation [KS01]. Here the goal is not to describe planar filtering, but the mapping between an object's 3D scattering potential and measured field data. In [KS01] (Chapter 6.3), the Fourier diffraction theorem is introduced in two ways and linearized using the “weakly

scattering” assumption. This assumption, also known as the *Born approximation*, requires that the imaged objects are only weakly scattering, since multiple scattering events are not considered in this formulation.

Assuming coherent imaging — as is typically the case for US — and following [Goo68] (page 58, Eqs. 3-70 to 3-71), a convolution-based description arises. The ATF of diffractive structures (e.g. the fields resulting from emitting and receiving apertures) can be derived as follows.

The amplitude transmittance t_A is defined as the relative strength of the input field to the output field over the three spatial dimensions:

$$t_A(x, y, z) = \frac{U_t(x, y, z)}{U_i(x, y, z)}. \quad (2.33)$$

We seek the resulting field as:

$$U_t(x, y, z) = t_A(x, y, z) \cdot U_i(x, y, z). \quad (2.34)$$

Using the Fourier convolution theorem (see Eq. 2.49), the same relationship applies to the spatial-frequency spectra:

$$A_t(f_x, f_y, f_z) = A_i(f_x, f_y, f_z) \otimes T(f_x, f_y, f_z), \quad (2.35)$$

where it is used that a multiplication in the spatial domain (x, y, z) yields spatial convolution in the frequency domain (f_x, f_y, f_z) .

$$T(f_x, f_y, f_z) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} t_A(x, y, z) e^{-j2\pi(f_x x + f_y y + f_z z)} dx dy dz, \quad (2.36)$$

For incoherent imaging, analogous methods can be applied using intensities instead of field amplitudes. This leads to convolution relations between intensity distributions and intensity transfer functions, rather than amplitude-based ones.

2.5.9 Diffraction Using the Fraunhofer Approximation

Another way to approximate diffraction is through the Fresnel approximation, and, as a further simplification, the Fraunhofer approximation. The basic idea is again the assumption of two (effectively) coplanar fields, $U_i(x_0, y_0, 0)$ and $U_t(x_1, y_1, z)$, defined on parallel planes.

We start from the Fresnel diffraction integral (see [Goo68], p. 66, Sec. 4.2, or [AT00], “k-space tutorial”, Eqs. (2.1)–(2.7)):

$$U_t(x_1, y_1, z) \approx \frac{e^{jkz}}{j\lambda z} e^{j\frac{k}{2z}(x_1^2+y_1^2)} \times \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} U_i(x_0, y_0, 0) e^{j\frac{k}{2z}(x_0^2+y_0^2)} e^{-j\frac{k}{z}(x_0x_1+y_0y_1)} dx_0 dy_0. \quad (2.37)$$

This represents the paraxial approximation, i.e. $\sin(\alpha) \approx \alpha$ for small propagation angles.

Using the Fraunhofer (far-field) approximation, the quadratic phase term $\frac{k}{2z}(x_0^2 + y_0^2)$ must satisfy $\frac{k}{2z}(x_0^2 + y_0^2)_{\max} \ll 1$, which implies, up to numerical factors of order unity,

$$z \gg \frac{(x_0^2 + y_0^2)_{\max}}{\lambda}. \quad (2.38)$$

With the Fraunhofer assumption, the quadratic phase term inside the integral in Eq. (2.37) can be approximated by unity. Using the Fourier convolution theorem, we obtain

$$U_t(x_1, y_1, z) \approx \frac{e^{jkz} e^{j\frac{k}{2z}(x_1^2+y_1^2)}}{j\lambda z} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} U_i(x_0, y_0, 0) e^{-j\frac{2\pi}{\lambda z}(x_0x_1+y_0y_1)} dx_0 dy_0. \quad (2.39)$$

Neglecting the quadratic phase factor in front of the integral, this expression is exactly the Fourier transform of the aperture function $U_i(x_0, y_0, 0)$.

Hence, the Fraunhofer diffraction approximation can be understood as a special case of the planar aperture description introduced in Sec. 2.5.4, with $U_t(x, y, z)$ corresponding to the far-field pattern at $z \rightarrow \infty$.

2.6 Properties of Imaging Systems in the Spatial Domain

Classical optics describe imaging systems using a set of quantitative metrics, which are introduced here following [Goo68].

2.6.1 Aperture (Aperture Opening)

In optics, the total spatial extent of the active imaging elements is referred to as the *aperture*. The spatial characteristics of this structure determine the imaging behavior of the system.

The radiation pattern of such a structure can be described as the interference of many elementary *Huygens' waves*. For example, the diffraction pattern of a 1D slit aperture is shown in Fig. 2.4. In US systems these components are the ultrasound emitters and receivers and their spatial extent and positioning. Further modeling and descriptions of the aperture influence will be discussed in later sections.

2.6.2 Resolution and Imaging Function

For imaging systems, the system concept of the *system function* transitions to the *imaging function*. For the special case of a Dirac delta function as the object, it becomes the *point spread function* (PSF). This also provides a rigorous definition of the intuitive concept of *resolution* [VGD95a].

The term *resolution* can be defined as the minimum distance at which two objects (point scatterers), imaged by a system, are just barely distinguishable [VGD95b], see Fig. 2.5. The limit of separability is often defined by the *full width at half maximum* (FWHM). Two point scatterers separated by a distance smaller than this measure cannot be distinguished and therefore it serves as a *resolution metric*.

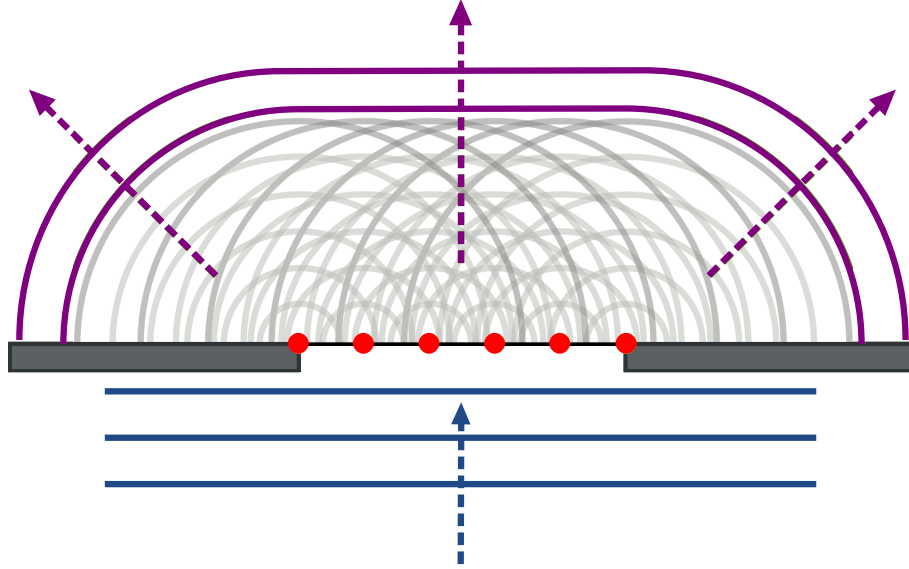


Figure 2.4: *Diffractive interaction of a plane wave (in blue) at a slit aperture, the directional vector given as blue, dashed arrow. This illustrates the discrete development over six Huygens' spherical waves (in gray) emitting from the the six point sources (red). The resulting wave front is given in purple, the direction vector as dashed purple arrow.[Wik10a].*

for the imaging system

$$r(x_1) = r(x_2) = \frac{1}{2} r(x_{\max}), \quad (2.40)$$

where $r(x_{\max})$ correspondes to the maximum values of the reconstructed image. $r(x_1)$ and $r(x_2)$ are the amplitudes of the reconstructed image at x_1 and x_2 respectively, where the response drops to 50% of the maximum. This measure is specific to each imaging system and its spatial characteristics.

Amplitude vs. Intensity PSF. In optical and acoustic imaging, two forms of the point spread function are commonly used:

- The **intensity PSF**, defined via $|h(\vec{r})|^2$, represents the measurable intensity or power distribution resulting from a point source. It is typically used in incoherent imaging systems such as X-ray or optical microscopy.
- The **amplitude PSF**, defined via the complex-valued field response $h(\vec{r})$ itself, preserves both amplitude and phase information. This definition applies to coherent imaging systems, including US and radar, where the superposi-

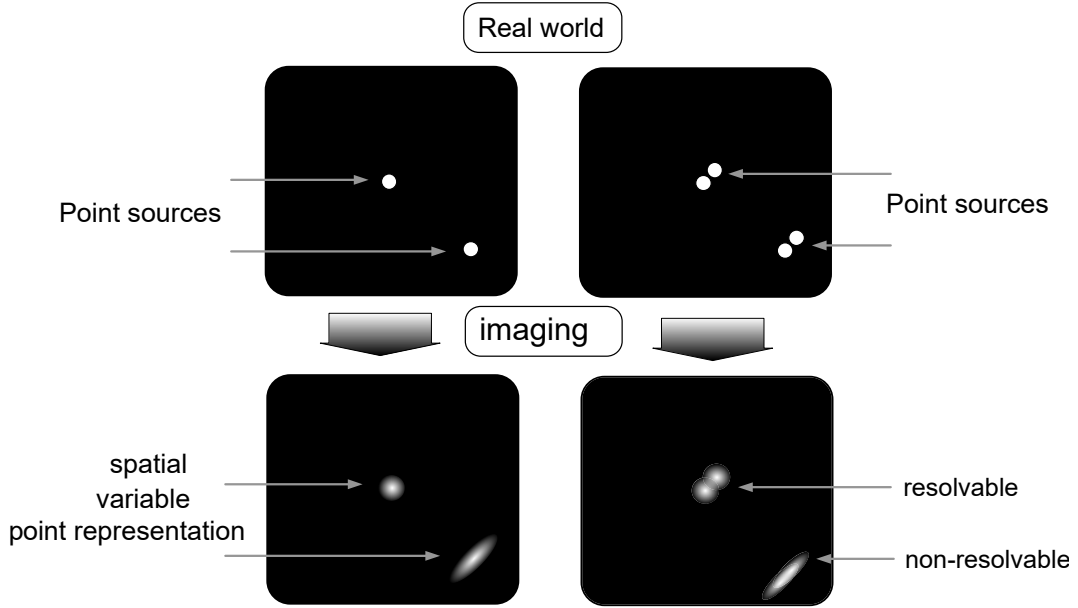


Figure 2.5: Resolution as the separability of objects in images. Real point sources (Top Left, Top Right) are imaged by a spatially variant imaging system. (Bottom Left) infinitely small point sources become spatially dependent point spread functions (PSFs). (Bottom Right) the central PSF suffices to distinguish the point sources at this specific distance, while the PSF in the lower-right corner no longer does [Sch08a].

tion of complex amplitudes leads to image formation.

Since US imaging and beamforming are inherently phase-coherent, all following derivations and simulations in this work will use the **amplitude PSF** definition.

2.6.3 Depth of Field

The term *depth of field* (DOF) originates from focused, planar optical systems. It describes the range along the optical axis that can still approximately be considered in focus, according to a specific resolution metric. For typical focusing imaging systems, there exists a direct relationship between the aperture size and the axial range that meets the given resolution criterion.

From a systems perspective, the DOF expresses the degree of *spatial variance* of the imaging function along depth (typically the z -dimension). A small DOF corresponds to strong spatial variance, while a large DOF indicates approximate spatial invariance along z .

Since the USCT imaging system as a whole does not possess a fixed planar aperture, this metric cannot be applied directly. Instead, the more general concept of the PSF is used to quantify focus behavior and resolution in depth. For USCT systems, synthetic aperture focusing in software effectively replaces the physical DOF limitation of conventional optics, enabling adaptive refocusing after acquisition.

2.6.4 Image Contrast

Several formulations of the concept contrast exist in optics and imaging theory. Contrast is of utmost importance and strongly connected to image uniqueness, system bandwidth, and separability of structures. It behaves differently depending on whether *coherent* or *incoherent* imaging is utilized and also depends on the system's bandwidth and the presence of side-lobes, which reduce effective contrast [Sza04a, Goo68].

For coherent systems, interference effects can either enhance or suppress contrast locally, while for incoherent systems contrast directly reflects the amplitude of the intensity distribution. In US imaging, side-lobes, limited dynamic range, and frequency-dependent attenuation are the main factors that degrade contrast.

In the next section contrast will be discussed from a Fourier-based perspective.

2.7 Imaging Systems in the Frequency Domain

After a frequency-independent consideration of imaging systems in the object space, we now turn to a Fourier-based analysis, analogous to the treatment in Goodman's *Fourier Optics*.

Here, we introduce the relevant ideas and terminology for both optical and ultrasound systems.

2.7.1 Fourier Transformation

The Fourier transformation occupies a central role in signal and image processing. One reason for this is the alternative perspective it provides for analyzing and solving problems from time- or space-dependent signals to their frequency-domain representations. In the temporal case, we use the standard engineering convention, where the forward transform carries a negative sign in the exponential:

$$\mathcal{F}_t\{g(t)\}(f) = G(f) = \int_{-\infty}^{\infty} g(t) e^{-j2\pi ft} dt, \quad (2.41)$$

$$\mathcal{F}_f^{-1}\{G(f)\}(t) = g(t) = \int_{-\infty}^{\infty} G(f) e^{+j2\pi ft} df. \quad (2.42)$$

For spatially varying object fields, however, we follow the convention commonly used in Fourier optics (Goodman), where the forward spatial transform uses a *positive* sign. This choice preserves the standard phasor form for wave propagation and simplifies the angular-spectrum representation:

$$\mathcal{F}_{x,y}\{U(x,y)\}(f_x, f_y) = A(f_x, f_y) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} U(x,y) e^{+j2\pi(f_x x + f_y y)} dx dy, \quad (2.43)$$

$$\mathcal{F}_{f_x, f_y}^{-1}\{A(f_x, f_y)\}(x, y) = U(x, y) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} A(f_x, f_y) e^{-j2\pi(f_x x + f_y y)} df_x df_y. \quad (2.44)$$

Using different sign conventions for time and space is standard practice in wave physics and Fourier optics: the temporal transform follows the signal-processing convention (useful for matched filtering and DAQ processing), while the spatial transform follows the optical convention to ensure a consistent propagation kernel in the angular-spectrum model. In the context of imaging, the goal is to shift from spatially dependent amplitude considerations to frequency-dependent magnitude and phase representations. This corresponds to the decomposition of a spatial image into spatially independent (only globally phase-shifted) frequency components,

n-D Fourier Transformation and Separability

In 2D or 3D, image functions (e.g., pixel or voxel representations) can be described by the 1D Fourier transformation introduced in Sec. 2.7.1, if they are separable. A function is *separable* with respect to its parameters if it can be expressed as a product of multiple functions, each depending only on one variable. For example, a function $g(x, y)$ is separable if there exist functions $g_x(x)$ and $g_y(y)$ such that:

$$g(x, y) = g_x(x) g_y(y). \quad (2.45)$$

For the 2D case, according to [Goo68] (page 10ff), the Fourier transform is defined as:

$$\mathcal{F}\{g(x, y)\} = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} g(x, y) e^{-j2\pi(f_x x + f_y y)} dx dy \quad (2.46)$$

If $g(x, y)$ is separable, the transform factorizes as

$$\mathcal{F}\{g(x, y)\} = \int_{-\infty}^{\infty} g_x(x) e^{-j2\pi f_x x} dx \int_{-\infty}^{\infty} g_y(y) e^{-j2\pi f_y y} dy = \mathcal{F}_x\{g_x\} \cdot \mathcal{F}_y\{g_y\}. \quad (2.47)$$

Finite Spatial Extent and Band Limitation

A fundamental characteristic of Fourier analysis is that each frequency component corresponds to an *unbounded* oscillation in the original temporal/spatial domain. Conversely, a time/spatially limited function requires infinitely many frequency components for its representation. This property is referred to as *temporal/spatial limitation*. Inversely, a function limited to a finite range in the frequency domain (*band limitation*) necessarily extends infinitely in the temporal/spatial domain. Therefore, a function cannot be both strictly time/spatially and band limited simultaneously [Goo68]. From a practical perspective, however, this issue can be mitigated by assuming that real signals, although not strictly band limited, have the majority of their energy contained within a certain finite bandwidth. Such functions can therefore be considered *approximately* both spatially and band limited, forming the basis of later concepts such as the *space–frequency product* and discrete sampling considerations.

2.7.2 Fourier Theorems

Here, we introduce a selection of mathematical properties of the Fourier domain that will be used in subsequent derivations. For detailed discussions and proofs, see [Goo68].

Linearity Theorem (Fourier Synthesis Theorem)

The linearity theorem states that the Fourier transform of a weighted sum of functions is equal to the weighted sum of the individual transforms:

$$\mathcal{F}\{\alpha g + \beta h\} = \alpha \mathcal{F}\{g\} + \beta \mathcal{F}\{h\} \quad (2.48)$$

with g and h as the functions to be transformed, and α and β as weighting coefficients.

Convolution Theorem

The convolution theorem states that the convolution of two functions in the spatial domain corresponds to the multiplication of their Fourier transforms, hence

$$\mathcal{F}\{f \otimes g\} = \mathcal{F}\{f\} \cdot \mathcal{F}\{g\} \quad (2.49)$$

with f and g as the functions to be Fourier transformed.

Fourier Integral Theorem

This theorem expresses the reversibility of the Fourier transform

$$\mathcal{F}\{\mathcal{F}^{-1}\{g(x, y)\}\} = \mathcal{F}^{-1}\{\mathcal{F}\{g(x, y)\}\} = g(x, y), \quad (2.50)$$

with $g(x, y)$ as the function to be transformed.

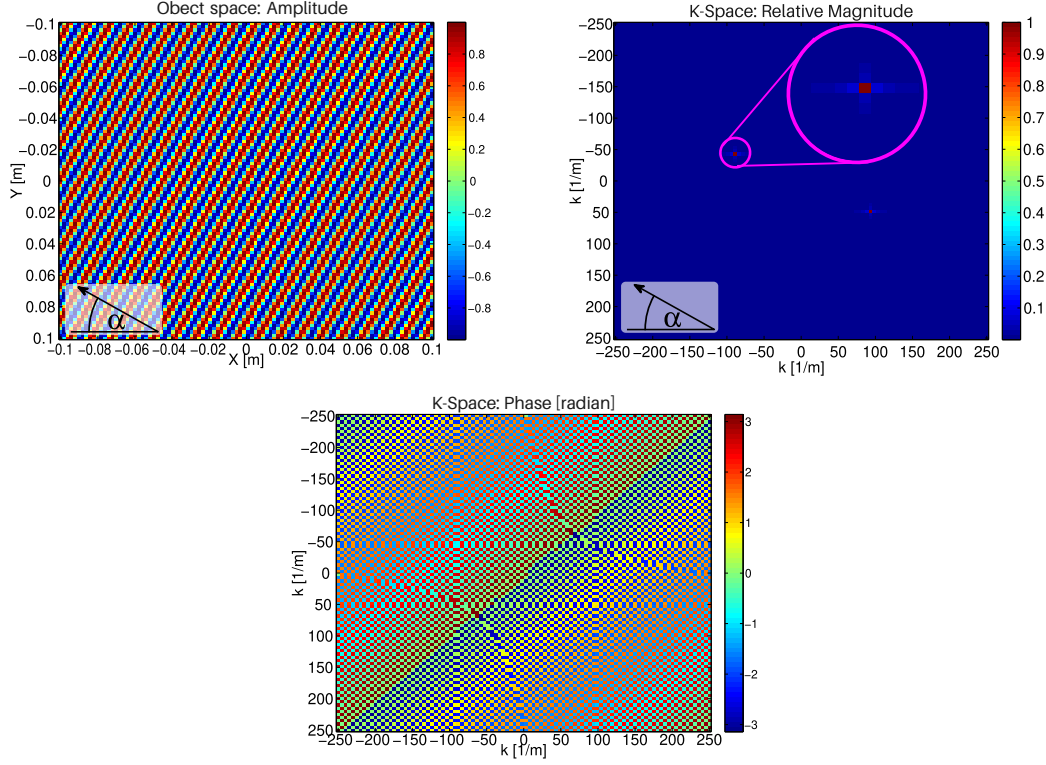


Figure 2.6: Plane wave in k -space: (top left) planar spatial monochromatic plane wave. (Top right) magnitude of the corresponding spatial frequency domain contribution for a monofrequent planar wave in k -space. The small pink circles zooms out to the big pink circle the very localized almost monofrequent contribution. (Bottom) phase of the corresponding frequency domain resulting from the plan wave. The propagation angle α is represented in both representations as the angle is fouriertransform invariant.

2.7.3 The Fourier Domain as a Spatial-Frequency Space (k-space)

Let us consider a 2D Fourier domain and interpret it as a space of *spatial frequencies*. Each point in this domain represents a plane wave component with a specific spatial frequency and phase. Figure 2.6 shows an example alongside its corresponding spatial representation.

We denote the Fourier transform as $\mathcal{F}\{g(x, y)\} = G(f_x, f_y)$, where the spatial coordinates (x, y) are measured in meters and their Fourier counterparts (f_x, f_y) in m^{-1} . When normalized to the wavenumber $k_0 = \frac{2\pi}{\lambda}$, the scaled spatial frequency domain is referred to as the k -space.

The characteristics of a multi-dimensional k-space are directly derived from those of the associated spatial domain. 2D or 3D image are typically characterized by the number of image points n_V (in pixel) along every spatial axis and their physical size s_V (in meters). In most imaging systems, the pixel or voxel spacing is identical along each spatial axis, so the periodic sampling of the continuous spatial domain can be expressed as a spatial frequency $f_S = \frac{1}{s_V}$, with all implications of discretization (see Sec. 2.7.4).

Through spatial sampling, the maximum representable spatial frequency is limited to half the sampling frequency, $f_S/2$, the *Nyquist frequency*. This defines the highest resolvable spatial frequency along each sampling axis.

In the spatial-frequency domain (k-space), the same limitation appears directly: the sampled image contains frequency components only within the interval $\pm f_S/2$. Because the Fourier transform of real-valued data is Hermitian-symmetric, k-space includes both positive and negative spatial frequencies.

The total image size, $n_V \cdot s_V$, in the spatial domain corresponds to the sampling interval in k-space, $s_{V,\text{k-space}} = \frac{1}{n_V \cdot s_V}$.

Superposition

According to the Fourier synthesis theorem, any spatial structure can be composed by superposing elementary waves. E.g. the spherical wave, important in US imaging, can be constructed out of a series of plane waves as it is shown in Fig. 2.7.

Spatial Translation

Shifting objects in the spatial domain appears in k-space as a phase shift of the frequency components, see Fig. 2.8.

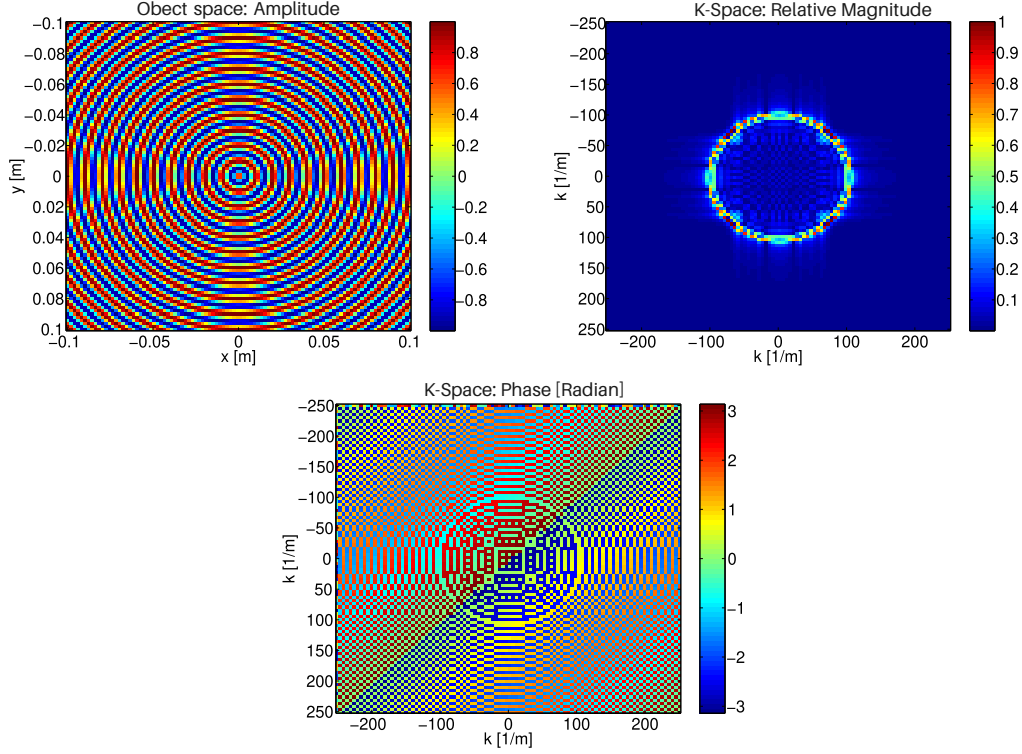


Figure 2.7: (left) 2D circular wave, (middle) magnitude of the corresponding frequency domain, (right) phase of the corresponding frequency domain; any spatial wave can be generated through the superposition of infinit planar elementary waves.

2.7.4 Whittaker–Shannon–Nyquist Sampling Theorem

The sampling theorem plays a central role in signal and image processing, and by extension, in all Fourier-based numerical systems [SAK06]. It strongly influences the assumptions made in this work, particularly regarding alias-free sampling of 1D signal data, 2D or 3D images, and k-space. Even the spatial distribution of discrete aperture positions in 3D space can be interpreted as *spatial sampling*, with all related effects such as spatial aliasing or grating lobes. From an implementation perspective, the sampling theorem also determines computational time and memory requirements, especially in cases of oversampling.

One way to introduce the sampling theorem is to consider the sampling of a continuous function $g(t)$ at equidistant time intervals. The sampling process can be modeled as the multiplication of $g(t)$ by an infinite train of Dirac delta functions

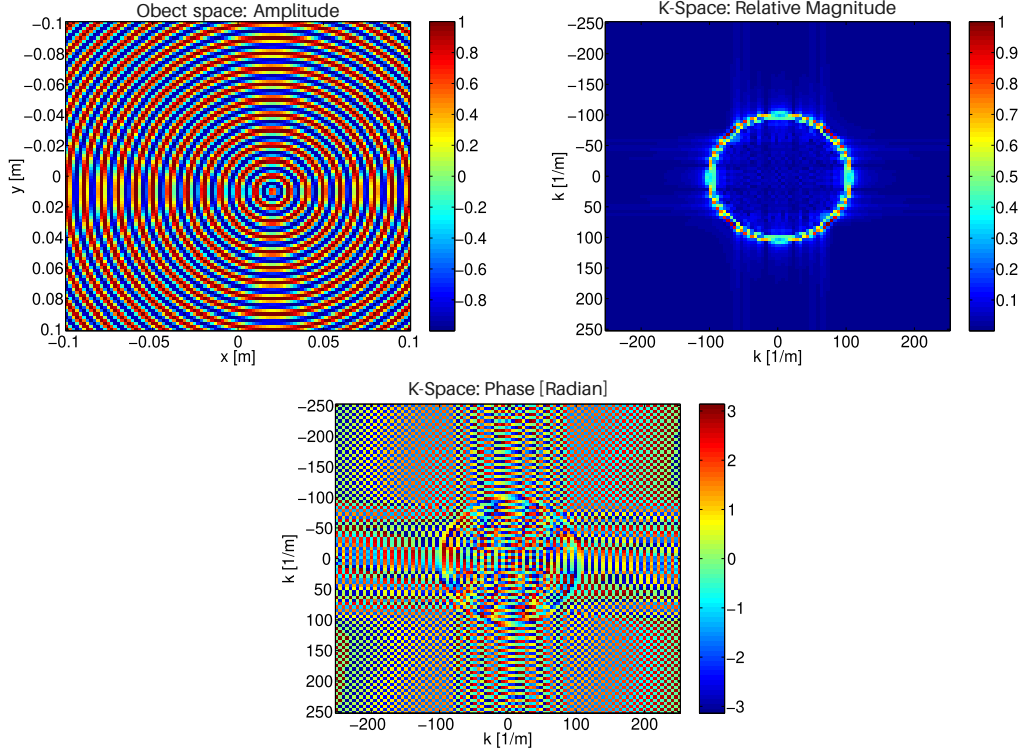


Figure 2.8: (left) shifted 2D circular wave, (middle) magnitude of the corresponding frequency domain, (right) phase of the corresponding frequency domain; a shift in the spatial domain is represented in the frequency domain as a pure phase shift, the magnitude remains unchanged. (Deviations in the frequency domain are due to the spatial limitation)

(Dirac comb), as described in [Goo68], (Chapter 2.4.1)

$$g_s(t) = g(t) \cdot \text{comb}\left(\frac{t}{T_s}\right) = \sum_{n=-\infty}^{\infty} \delta(t - nT_s) \cdot g(t), \quad (2.51)$$

where T_s is the sampling period and $f_s = 1/T_s$ the corresponding sampling frequency. The resulting function $g_s(t)$ represents the discretely sampled version of $g(t)$.

According to the Fourier convolution theorem, multiplication in the time domain corresponds to a convolution in the frequency domain, hence

$$G_s(f) = G(f) \otimes \text{comb}\left(\frac{f}{f_s}\right). \quad (2.52)$$

Hence, the spectrum of the sampled signal is a repetition of the original spectrum

$G(f)$, periodically shifted by integer multiples of f_s . If the original signal bandwidth B satisfies $B < f_s/2$, these shifted spectra do not overlap, and the original signal can be perfectly reconstructed. When $B \geq f_s/2$, spectral overlap occurs, leading to ambiguity between distinct frequency components — a phenomenon known as *aliasing*. In this case, the system no longer provides a unique or invertible mapping between the signal and its sampled representation. The lower frequency limit of the sampling range is determined by the duration of $g(t)$, while the upper limit is given by the sampling rate f_s . This concept generalizes directly to higher-dimensional domains. The same reasoning applies to spatial sampling in k -space (Section 2.7.3) and thus to 2D or 3D image sampling, as well as to discretized aperture distributions in US systems [Goo68].

2.8 Application of Fourier Analysis

As introduced in Sec. 2.7.3, the representation used in Fourier analysis of the spatial domain mapped by an imaging system is the Fourier transform of the spatial domain (also called the *reciprocal space*, due to its inverse unit m^{-1} , or *k-space*. By switching the perspective to a spatial frequency representation, new interpretation possibilities for the imaging system emerge. Some of these perspectives and applications are presented below.

2.8.1 System Function and Imaging Function

An imaging system described by linear system theory, assuming invariance in time and space, can be fully characterised by its spatial and temporal frequencies, including both phase and magnitude [Goo68]. The influence of such a linear system on an object to be imaged is referred to as the *imaging function*.

In the 2D spatial domain, the blurring of an object $f_O(x, y)$ by the system-specific PSF $f_A(x, y)$, see Sec. 2.6.2, can be mathematically described as a convolution. An object function $f_O(x, y)$ in the spatial domain is convolved with an *imaging*

function $f_A(x, y)$. Consequently, the resulting image $f_R(x, y)$ is given as

$$f_R(x, y) = f_O(x, y) \otimes f_A(x, y) \quad (2.53)$$

Using the convolution theorem, see Eq. 2.49, this convolution in the spatial domain becomes a simple multiplication of the Fourier-transformed object and imaging functions:

$$F_R(f_x, f_y) = F_O(f_x, f_y) \cdot F_A(f_x, f_y). \quad (2.54)$$

If the imaging function has only a real part and the imaginary part is zero, this filter function becomes a *bandpass function*. An example with the Shepp–Logan head phantom, commonly used as a simulation reference in medical imaging, is shown in Fig. 2.9 [SL74].

2.8.2 Contrast in a Fourier Perspective

Picking up again from the previous spatial image domain perspective to a k-space domain perspective.

An imaging system’s contrast refers to the distinguishability of object features in the image. It can be formalised using Goodman’s *Modulation Transfer Function* (*MTF*), which describes how spatial frequency content is preserved through the system [Goo68]. The MTF can be computed as follows

$$\text{MTF}(f) = \frac{C_{\text{image}}(f)}{C_{\text{object}}(f)} = \frac{I_{\text{max}}(f) - I_{\text{min}}(f)}{I_{\text{max}}(f) + I_{\text{min}}(f)}, \quad (2.55)$$

where f is the spatial frequency, and $I_{\text{max}}(f)$ and $I_{\text{min}}(f)$ are the maximum and minimum image intensities at frequency f .

The imaging system’s capability to resolve and separate objects, composed of several spatial frequencies, depends on the range of spatial frequencies it can transmit. A broader bandwidth response—i.e. a wider k-space support—corresponds to better image fidelity and higher preserved contrast. While the highest transmitted frequency defines the theoretical resolution limit, the effective separability of nearby scatterers depends on the available bandwidth. A narrow bandwidth produces

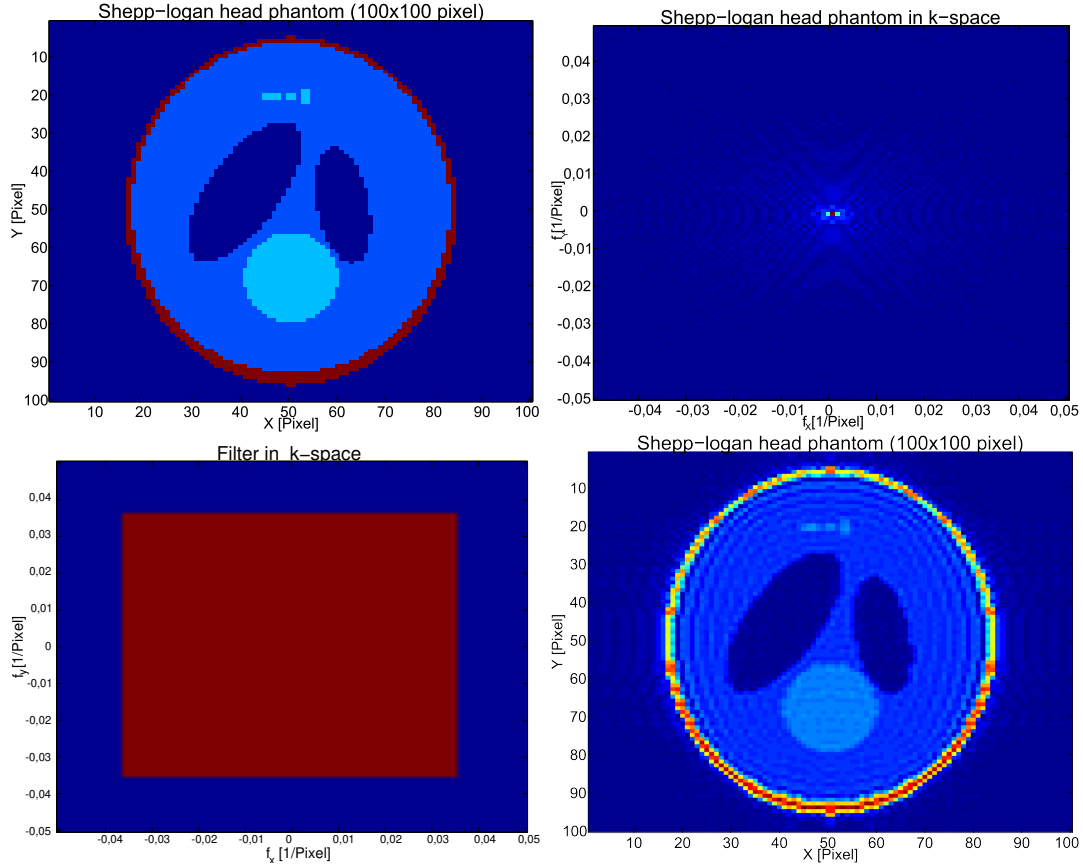


Figure 2.9: Example of k -space as a 2D filter for an object function: (top-left) Shepp–Logan head phantom (SL) normalized between zero (blue) and one (red), (top-right) SL in k -space representation, (bottom-left) a rectangular binary 2D low-pass filter with relative magnitude between zero and one (blue to red), (bottom-right) the resulting bandwidth limited SL image.

long, oscillatory point-spread functions whose sidelobes cause mutual interference between neighboring features, thereby reducing practical contrast. Conversely, a broad bandwidth shortens the effective pulse and suppresses sidelobes, improving both separability and visibility of fine structures. Hence, resolution and separability are distinct: the former relates to the system’s maximum spatial frequency, the latter to its spectral width and energy distribution.

2.8.3 Region-of-Support (Spatial Bandwidth)

A consequence of the spatial limitation of continuous objects is that they are typically unbound in the equivalent k -space, see Sec. 2.7.1.

By using a sampled and limited spatial domain with isotropic and constant pixel size, a limited k-space is obtained, with an additional assumption. This assumption is the *periodicity* (and thus the object's periodicity) of the described spatial domain. The pixel size (s_V [m] in each spatial dimension) and the total image size (s_B [m] in each spatial dimension) define the sampling properties of the wavenumber domain. Using the wavenumber definition for any of the spatial dimensions x, y, z as $k = 2\pi f$ (in radians per meter), the corresponding k-space limits are

$$k_0 = \frac{2\pi}{s_B}, \quad k_{\max} = \frac{\pi}{s_V}, \quad k_{\min} = -\frac{\pi}{s_V}, \quad (2.56)$$

for all spatial dimensions. These limits describe the minimal discrete sampling step (k_0) and the maximal representable wavenumbers ($k_{\min}, k_{\max}$) imposed by the spatial Nyquist criterion.

Ideally, the k-space should be fully covered by an imaging system, effectively functioning as an *all-pass system*, see Sec. 2.8.1. Real systems typically achieve this only partially. A quality metric for imaging systems can be derived, called the *spatial bandwidth or region-of-support* O_{BB} . To evaluate the imaging quality of an existing system, a quotient can be formed from the sum of contributions k_{system} to the sum of the theoretically ideal k-space k_{ideal}

$$O_{BB} = \frac{\sum k_{system}}{\sum k_{ideal}}. \quad (2.57)$$

2.8.4 Spatial Domain–Spatial Bandwidth Product

As a measure for the complexity of an object to be imaged denoted as $O(x, y, z)$, [Goo68], Chapter 2.4.2, proposes the product of its contribution in k-space $f_O(f_x, f_y, f_z)$ and its size in the spatial domain, denoted as $OOBB(O, f_O)$. The idea is that complex objects require extensive representation in both the spatial domain and k-space.

To identify whether an object contains non-trivial spatial content and non-trivial frequency content, we consider the binary support of the object in both domains.

Let

$$\text{bin}[U(x, y, z)] = \begin{cases} 1, & U(x, y, z) \neq 0, \\ 0, & U(x, y, z) = 0, \end{cases} \quad (2.58)$$

and analogously for the Fourier-domain object $F(f_x, f_y, f_z)$. The size of the spatial support is then

$$S_{\text{space}} = \sum_x \sum_y \sum_z \text{bin}(U(x, y, z)), \quad (2.59)$$

and the size of the frequency support is

$$S_{\text{freq}} = \sum_{f_x} \sum_{f_y} \sum_{f_z} \text{bin}(F(f_x, f_y, f_z)). \quad (2.60)$$

A complex object requires both supports to be non-zero:

$$O_{\text{OOBB}} = \begin{cases} 1, & S_{\text{space}} > 0 \wedge S_{\text{freq}} > 0, \\ 0, & \text{otherwise.} \end{cases} \quad (2.61)$$

An example of a non-complex object is an (infinite) plane with constant amplitude in the spatial domain. Its size in the spatial domain tends to infinity, but in k-space, with a vanishing bandwidth, it is represented only as a single frequency component, causing the spatial domain–spatial bandwidth product to approach zero. The maximum complexity is achieved when the contributions of both factors are of equal magnitude.

2.8.5 Emitter–Receiver–Object Interaction

An imaging system typically consists of two components: a source of waves represented by the emitter and a sink for waves represented by the receiver. The *Fourier Diffraction Theorem* describes the relationship between the transmitted

waves, object interaction, and received waves. This enables a modeling of the imaging system based on its components. The imaging capabilities of emitter–receiver combinations for an object represented in k -space can be derived intuitively using diffraction as interference with Huygens’ spherical waves. Fig. 2.10 illustrates an imaging scenario in the sense of Fourier analysis, where a plane-wave transducer continuously generates a monochromatic wave in the axial direction. Simultaneously, it acts as a potential receiver of a plane wave with the same frequency and a direction vector rotated by 180° (“pulse-echo case”). In the center of the image is a periodically repeating object that reflects the waves. The object is approximated by a discrete set of point scatterers. The physical spherical wave contributions (approximately) overlap to form a plane wave. Although the transducer is only sensitive to one specific plane wave with a single propagation direction but can detect *the phase and amplitude of this wave!* Thus, the transducer acts as a detector for a specific type of plane wave for which it measures the amplitude and phase information. Specifically in this case, the emitter propagated plane waves of λ wavelength are reflected at each discrete point scatterer. And after passing through $\lambda/2$ twice (once through the transmitted wave and once through the reflected wave), these reflected waves overlap without relative phase shift (a shift of λ is equivalent to no phase shift due to periodicity), with the newly forming wave at the next row of point scatterers. The reflection overlap is perfectly additive for this periodic structure. The equivalent k -space representation has a red point representing the imaging capability of the system and a black point representing the object function. Since they overlap, the objects imaging and detection is successful.

From a USCT design perspective, this is a central insight: each emitter-receiver transducer pair effectively probes *a single point in k -space* at the intersection between the emitted and received plane-wave vectors. Hence, the geometric arrangement and angular coverage of transducers directly determine how densely and the isotropically k -space is sampled. Systems with limited angular diversity or sparse aperture placement inevitably produce incomplete k -space coverage, leading to spatial anisotropy and reduced resolution in the reconstructed image. Dense ring geometries or multi-angle synthetic aperture setups aim to overcome

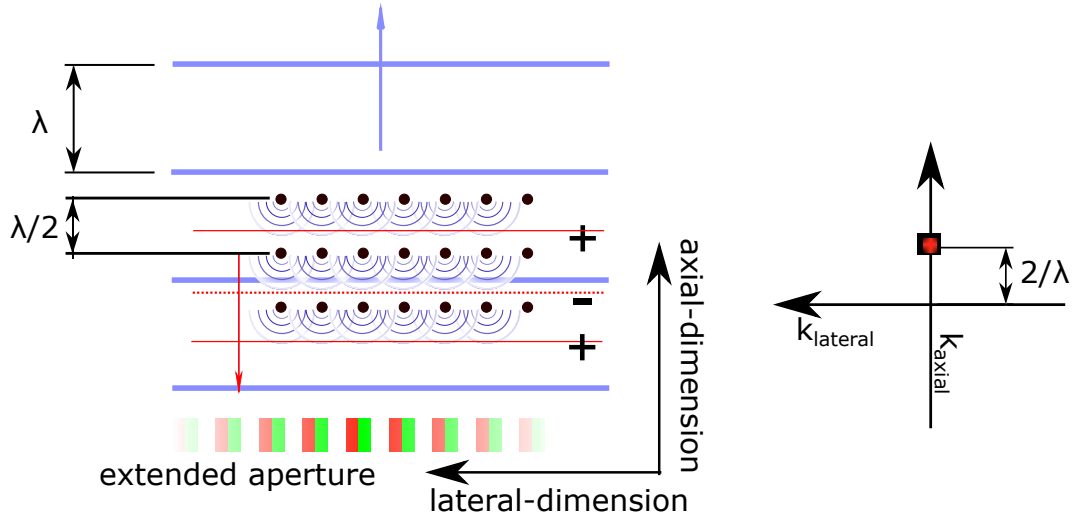


Figure 2.10: Emitter–receiver interaction with an object of characteristic size $\lambda/2$. **Left (spatial domain):** The red, spatially unbounded emitter aperture generates a plane wave of wavelength λ propagating in the direction indicated by the blue arrow. Simultaneously, the green reception aperture acts as a receiver for a plane wave of identical wavelength incident from the direction marked by the red arrow. The red waves represent reflections from the periodic structure. The plus and minus signs denote the polarity of two successive reflected contributions. Since the scatterers are separated by $\lambda/2$ along the propagation path, the accumulated phase shift corresponds to an integer multiple of λ , resulting in constructive interference. **Right (k-space):** The system transfer function (red point) and the object spectrum (black point) coincide; their overlap enables imaging and detection.

exactly this limitation.

Next, the question arises for which objects do the transmitted plane waves reflect and overlap to form exactly those required by the receiver?

An object/structure that can reflect an infinite plane wave must itself be spatially infinite large, hence the requirement for a periodic objects/structure. By geometrically constructive and destructive interference for potential periodicities of object structures, it can be shown that there is exactly one periodicity fulfilling the requirement for a continuous plane-wave reflection at this frequency. Surprisingly, in the “pulse-echo case” (emitter and receiver bistatic angle is 180°) see Fig. 2.13, an object with half the wavelength of the carrier frequency λ can be detected as the periodic structure width.

In the context of USCT, this defines a fundamental resolution limit: for backscattering (pulse–echo) configuration of emitter–receiver transducers, the smallest resolvable spatial period corresponds to $\lambda/2$, analogous to the Nyquist criterion. This limit is not imposed by the reconstruction algorithm but by the physics of coherent interference — a key constraint for the design of high-frequency US systems.

Figure 2.11 illustrates an example of an object structure width that cannot be detected. In this example, the spatially unlimited periodic object structure, discretely approximated as point scatterers, has a periodicity along the transmission direction of $\lambda/4$. The emitter propagated plane waves are reflected at each discrete point scatterer. After passing through $\lambda/4$ twice (once through transmitted waves and once through reflected waves), these reflected waves overlap with a phase shift of $\lambda/2$ with the newly forming reflection wave at the next row of point scatterers, resulting in perfect cancellation. This makes the object non-existent to the receiver aperture. The aperture acts as a receiver for a plane wave with wavelength λ in the reception direction shown by the red arrow. (right) Equivalent k-space: the red point represents the imaging capability of the system; the black point represents the object function. Since they do not overlap, no imaging/detection occurs. In a k-space representation this is represented by the lack of overlap between the red-marked support region frequency ($\lambda/2$) of the imaging system and the black-marked object function ($\lambda/4$).

The $\lambda/4$ example demonstrates another USCT design implication: if the system’s angular aperture or center frequency is too narrow, high-frequency structures (fine details) will fall outside the system’s accessible k-space region, leading to their complete suppression or aliasing. This emphasizes why both axial bandwidth and angular coverage (surrounding aperture) must be maximized to achieve isotropic resolution.

For the monochromatic case and different positions of the receiver and emitter, there exists exactly one object, defined by the k-space vector $\vec{v}_O$, with a periodicity/frequency (with arbitrary phase) at a specific angle that can be detected, see Fig. 2.12. The property that such a emitter–object–receiver combination must

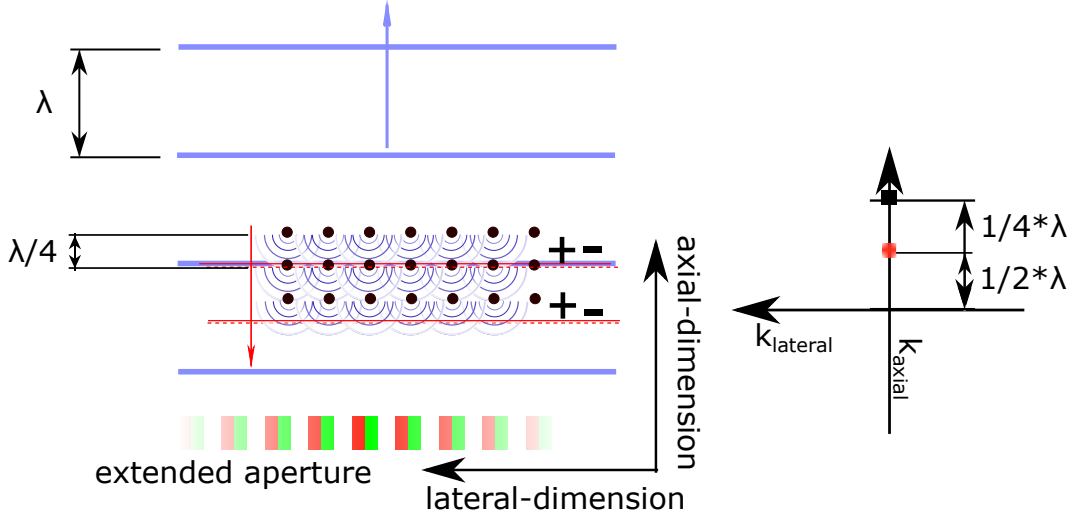


Figure 2.11: Emitter–receiver interaction with an object of characteristic size $\lambda/4$. **Left (spatial domain):** The red, spatially unbounded emitter aperture continuously generates a plane wave of wavelength λ propagating in the direction indicated by the blue arrow. Simultaneously, the green aperture acts as a receiver for a plane wave of the same wavelength incident from the direction marked by the red arrow. The emitted field is shown in blue, the reflected field in red. The plus and minus signs denote two reflected contributions that cancel due to a relative phase shift of $\lambda/2$ between successive scatterer rows. **Right (k-space):** The system transfer function (red point) and the object spectrum (black point) do not overlap; consequently, neither imaging nor detection is possible.

fulfill is an coherent, constructive wave interference with a phase shift of zero over time and space to form constructively a new plane wave.

This is fulfilled if the plane waves describing the emitter $\vec{v}_e$, the receiver $\vec{v}_r$, and the object $\vec{v}_O$ satisfy the following in k-space:

$$\vec{v}_O = -\vec{v}_e + \vec{v}_r, \quad (2.62)$$

$$\|\vec{v}_e\| = \|\vec{v}_r\| = k_e = k_r \quad (2.63)$$

The different signs of $\vec{v}_e$ and $\vec{v}_r$ result from the reflection at the object structure, see Fig. 2.14. All these potential contributions in k-space for a emitter, arbitrary receivers, and a monochromatic frequency are referred to as the *Ewald sphere*. In the context of X-ray crystallography this concept was first used to determine the periodicity of crystal lattices [Ewa69].

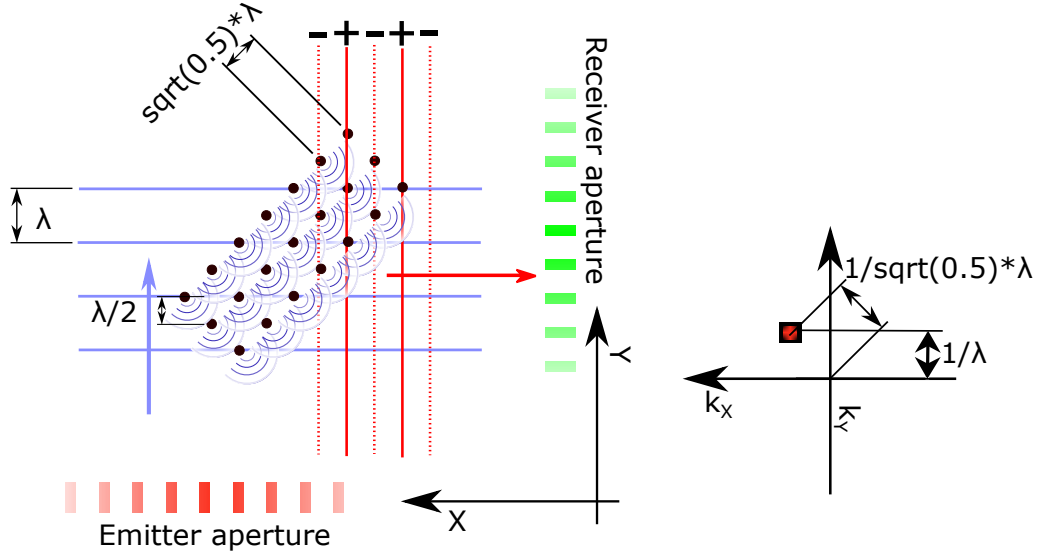


Figure 2.12: Emitter–receiver–object interaction for spatially separated emitter and receiver. **Left (spatial domain):** An unbounded red emitter generates a plane wave of wavelength λ propagating in the direction indicated by the blue arrow, while the green aperture receives a plane wave of identical wavelength from the direction marked by the red arrow. A periodic object (period $\lambda\sqrt{0.5}$), modeled as an array of point scatterers, reflects the incident field at 45° . Along the transducer axis, the cumulative optical path difference repeats every $\lambda/2$. After two successive contributions (incident and reflected), the total phase shift amounts to λ , leading to constructive interference. The plus and minus signs denote the polarity of two successive reflected waves. The resulting coherent superposition forms a plane wave propagating toward the receiver, rendering the object detectable. **Right (k-space):** The system transfer function (red point) coincides with the object spectrum (black point); their overlap enables imaging and detection.

From a USCT design viewpoint, the Ewald sphere defines the *k-space coverage surface* for a given frequency. The radius of this sphere ($k = 1/\lambda$) scales with frequency, meaning that broadband emitters sample multiple nested spheres, thereby filling the *k-space* more completely. Consequently, systems with multi-angle illumination and with transducers with broad bandwidth provide richer data, improving the imaging capacity of various algorithms.

2.9 Imaging system as Bistatic Ewald Sphere

Construction in the k-space

2.9.1 Propagation Modality in the k-space

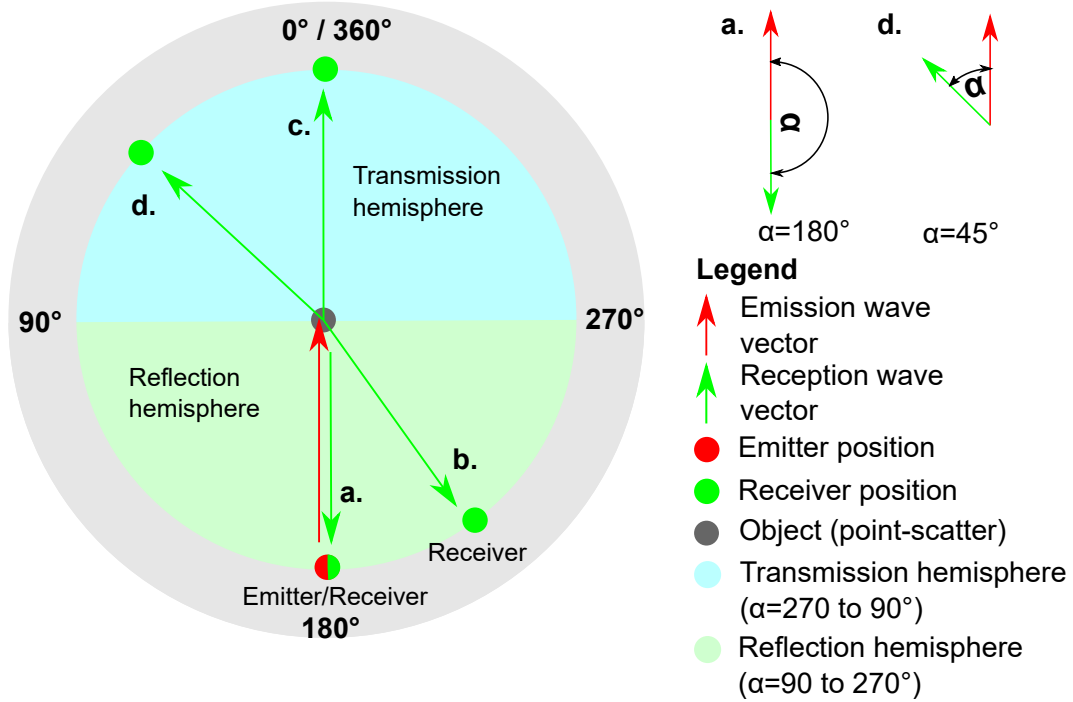


Figure 2.13: Bistatical transducer configurations in simplified 2D ring aperture. α denotes the bistatic angle, monostatic corresponds to $\alpha = 180^\circ$. Case a. denotes the monostatic case, emitter and receiver are the same and on the same position. Emitter wave vector and receiver wave vector are 180° to each other. Case b. denotes a case with separated emitter and receiver, a bistatical setup. Setup b. has the reflection wave directed back to the origin and is therefore in the reflection domain. Setup c. is the perfect transmission case under $\alpha = 0^\circ$. Case d. is also in the transmission domain due to forward reflection.

Most emitter-receiver configurations in USCT are bistatic, meaning to spatial separated transducer pairs. The case when they are identical is called monostatic. The possible combinations for an surrounding USCT system is shown in Fig. 2.13. Furthermore, this bistatic formulation can be directly interpreted geometrically in k-space using the Ewald sphere construction. Each emitter-object-receiver combination corresponds to a sampled spatial-frequency vector whose location in k-space is determined jointly by the propagation directions of the waves and by the

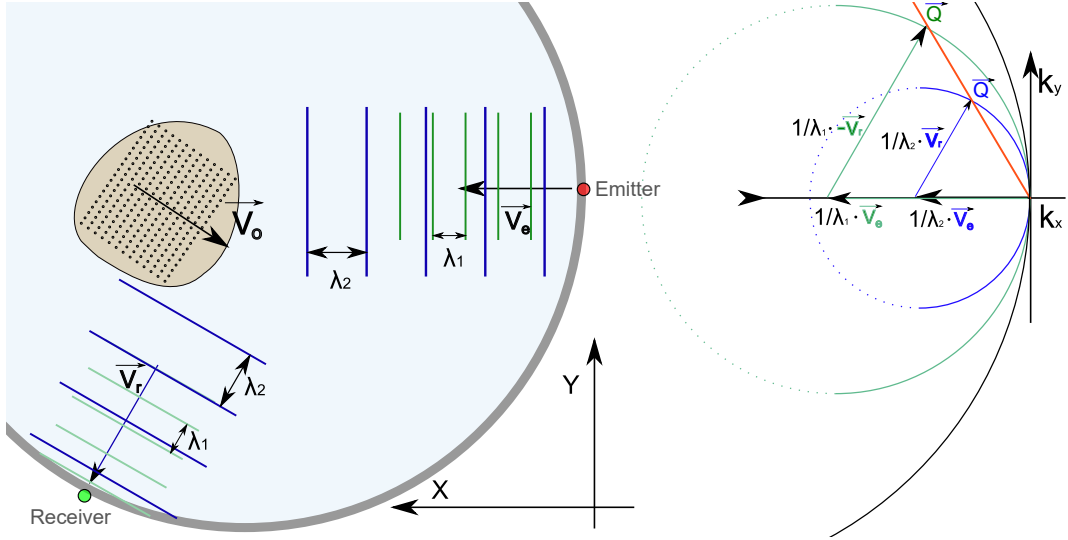


Figure 2.14: Ewald sphere construction for two monochromatic plane waves. (Left) 2D US setup, top down view. Separated emitter (red dot) and receiver (green dot) transducer configuration, in the center an object with periodical structures ($\vec{V}_o$). Emitter and receiver normal vectors in the spatial domain ($\vec{v}_e$ and $\vec{v}_r$) for two example wavelengths λ_1 (green) and λ_2 (blue). The brown object in the center is composed out of periodic structures respresented by the $\vec{V}_o$. (Right) k-space, the emitter and receiver normal vectors appear as $\vec{v}_e$ and $-\vec{v}_r$, respectively, with reciprocal length $k = 1/\lambda$. The red line shows the k-space trajectory and support ($\vec{Q}$) for constant $\vec{v}_e$ and $-\vec{v}_r$ but varying k , thus scaling the Ewald spheres with frequency. When $\vec{Q}$ and $\vec{V}_o$ overlap the structure is detected.

temporal frequency. Varying the frequency scales the Ewald sphere radially, while varying the bistatic angle rotates the sampled point on the sphere, see Fig. 2.14.

The *Fourier slice theorem*, see [KS01], Chap. 3, constitutes a special case of the Ewald sphere construction corresponding to the monostatic configuration $\vec{v}_e = \vec{v}_r$. In the general bistatic case, the Ewald sphere naturally splits into two hemispheres: bistatic angles greater than 90° form the *reflection* hemisphere, whereas angles smaller than 90° form the *transmission* hemisphere, see Fig. 2.13.

For USCT, these hemispheres correspond directly to the two propagation modalities:

- **Transmission tomography** samples spatial frequencies from

$$0 \ (\alpha = 0^\circ) \quad \text{to} \quad \sqrt{2} k \ (\alpha = 90^\circ) = \sqrt{2} \frac{2\pi}{\lambda}. \quad (2.64)$$

- **Reflection tomography** extends the accessible spatial-frequency radius from

$$\sqrt{2} k (\alpha = 90^\circ) = \sqrt{2} \frac{2\pi}{\lambda} \quad \text{to} \quad 2k (\alpha = 180^\circ) = \frac{4\pi}{\lambda}. \quad (2.65)$$

Thus, reflection tomography approximately doubles the accessible k-space maximum radius compared to pure transmission, at the cost of increased phase sensitivity and higher dynamic range requirements. These geometric relationships determine the achievable spatial bandwidth and are therefore fundamental constraints for USCT system design and simulation.

2.9.2 Temporal and Spatial Frequency in k-space

The spatial-frequency content of an object is described by its object vector $\vec{v}_O$ in k-space, representing the local spatial oscillation of the structure. For an object feature of characteristic size L , the magnitude of this object vector is

$$|\vec{v}_O| = \frac{2\pi}{L} \quad (2.66)$$

i.e. the wavenumber associated with that spatial scale. Thus, resolving an object feature requires that the bistatic measurement acquires a spatial-frequency contribution ($\vec{Q}$ in radian per meter) as large as $|\vec{v}_O|$ i.e.

$$|\vec{Q}| \geq |\vec{v}_O|. |\vec{Q}| = |\vec{v}_O|. \quad (2.67)$$

The measured spatial-frequency contribution follows as $\vec{Q} = \vec{k}_r - \vec{k}_e$. The emission wavevector $\vec{k}_e$ is constructed from the unit propagation direction $\vec{v}_e$ and the wavelength λ via $\vec{k}_e = (2\pi/\lambda) \vec{v}_e$. Analogously, the receive wavevector is given by $\vec{k}_r = (2\pi/\lambda) \vec{v}_r$.

We can achieve the maximum supported $\vec{Q}$ in the pulse-echo (monostatic) case from Eq. 2.65 this yields

$$|\vec{Q}_{\max}| = 2k = \frac{4\pi}{\lambda} \quad (2.68)$$

2.9 Imaging system as Bistatic Ewald Sphere Construction in the k -space

since $k = \frac{2\pi}{\lambda}$, we have an connection to λ .

Now, the question is which is the smallest structure L resolvable, with the condition

$$|\vec{Q}_{\max}| = |\vec{v}_O| \Rightarrow 2k = \frac{2\pi}{L}. \quad (2.69)$$

With inserting $k = \frac{2\pi}{\lambda}$

$$2 \cdot \frac{2\pi}{\lambda} = \frac{2\pi}{L} \Rightarrow L = \frac{\lambda}{2} \quad (2.70)$$

we end up in the familiar formulation connecting half-wavelength to structure size, an universal fourier imaging domain resolution criterion [Goo68].

L becomes the smallest resolvable feature (resolution) and with $|\vec{Q}_{\max}| = k_{\max} = 2k$ maximum spatial frequency supported by the system we get the following relationships

$$L = \frac{2\pi}{|\vec{Q}_{\max}|} = \frac{2\pi}{k_{\max}} = \frac{\pi}{k} = \frac{\lambda}{2} = \frac{c}{2f}. \quad (2.71)$$

This is a variant of the Rayleigh / Fourier resolution limit in its classical spatial-frequency representation e.g. also mentioned by Szabo [Sza04b] and Jensen [Jen96b] as

$$L \approx \frac{\pi}{k} \approx \frac{\lambda}{2}, \quad (2.72)$$

for resolution limits of US systems.

This corresponds to the pulse-echo configuration in US, e.g. the emitter and receiver are the same element, meaning the angle between emission and reception wave angle is 180° , the monostatic setup. When we go for more general configuration of spatial separated emitter and receiver transducers, angles smaller than 180° become possible in the bistatic configuration domain with α as bistatic angle (see Fig. 2.13). With bistatic angles from 270° to 180° to 90° we are in the reflection regime, with 90° to $0^\circ/360^\circ$ to 270° in the transmission regime.

In the full bistatic configuration for the surrounding USCT aperture, the temporal/axial frequency f of the emitted and received US determines the radius of the Ewald circle (or sphere in 3D), while the bistatic angle α determines the orientation of the sampled spatial-frequency vector. The accessible spatial frequency is

given by the magnitude of the scattering vector

$$|\vec{Q}| = 2k \sin\left(\frac{\alpha}{2}\right), \quad (2.73)$$

where $k = 2\pi f/c$ is the acoustic wavenumber and c the sound speed. To resolve an object feature of size L , the condition

$$|\vec{Q}| = |\vec{v}_O| = \frac{2\pi}{L} \quad (2.74)$$

must be satisfied. Solving this equality for the required temporal frequency yields the general bistatic resolution requirement:

$$f(\alpha) = \frac{c}{2L \sin(\alpha/2)} \quad (2.75)$$

The dependence on $\sin(\alpha/2)$ shows that small bistatic angles require a large temporal bandwidth in order to satisfy $|\vec{Q}| \geq |\vec{v}_O|$, whereas large bistatic angles (wide apertures) provide access to high spatial frequencies at comparatively moderate acoustic frequencies [Nat86, NW01]. In this way, the Ewald construction links the measurable temporal frequencies directly to the required object-vector contributions in k-space and spatial object feature sizes.

2.9.3 Fourier Diffraction Theorem and Sampling in 3D

The Ewald sphere formalism from Sec. 2.9.1 provides a geometric link between incident and scattered wave vectors in k -space. The *Fourier Diffraction Theorem* (FDT) extends this relation to a quantitative model: it connects the measured scattered field to the 3D Fourier transform of the object's scattering potential.

In practical systems, the FDT inversion is implemented in the time domain via *synthetic aperture focusing* (SAFT) or *delay-and-sum* (DAS) imaging. These methods perform coherent backprojection of received echoes along the travel-time surfaces:

$$O(\vec{r}) \approx \int_{\Omega} U_s \left(\vec{r}_R, \frac{|\vec{r}_R - \vec{r}| + |\vec{r}_T - \vec{r}|}{c} \right) d\Omega. \quad (2.76)$$

Each time delay corresponds to a constant-phase surface in k -space; SAFT thus reconstructs $O(\vec{r})$ by summing all consistent phase contributions—equivalent to the inverse Fourier mapping implied by the FDT.

3 US imaging system modeling and design

Building directly upon the theoretical and Fourier-domain system models established in Sec. 2.2, this chapter applies those formulations to the concrete design and modeling of 3D USCT hardware. While Sec. 2.2 developed the mathematical framework—linking the wave equation, k-space support, and sampling theory—this chapter translates those abstractions into system-level engineering models. In particular, it examines how the theoretical limits of resolution, separability, and bandwidth, introduced in Sec. 2.5.8–2.9.1, impose boundary conditions on mechanical aperture geometry, transducer bandwidth, and information-theoretic system performance.

3.1 Evaluation of Performance Boundary Conditions for Sonography Systems

The goals of the USCT project and the specific primary application of the 3D USCT system leads to the system modeling trade-offs: on one hand, the accuracy to be achieved, and on the other, the complexity of implementation and algorithm design must be within acceptable range.

Unlike conventional US approaches (e.g. sonography), 3D USCT is a true 3D method, which necessitates significantly more effort for 3D modeling and simulation. Additionally, USCT is a utilizing polychromatic imaging methods with a non-negligible bandwidth, meaning monochromatic approaches commonly used in optics may not be applicable.

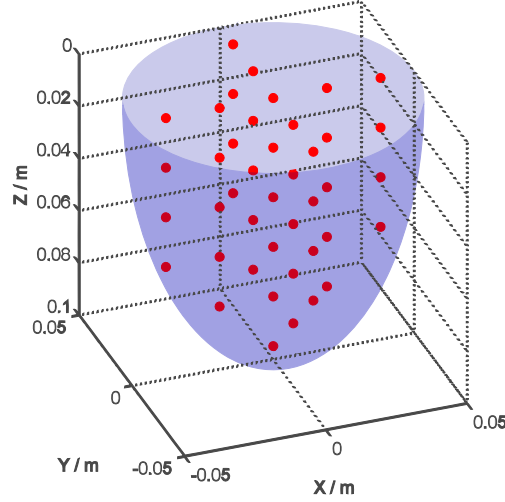


Figure 3.1: *Region-of-Interest for 3D USCT II: purple half-ellipsoid depicts the RoI, with the red dots forming the 42 evaluation points[Sch08a].*

Further properties related to the aperture, such as the distribution of aperture components in 3D, non-coplanar transducer orientation, and separate emitter and receiver positions, add complexity and are scarcely described in the literature for modeling and simulation.

3.2 Information Technology Boundary Conditions

The computational effort required for simulation is determined not only by algorithmic complexity but also by the «size» of the parameter space to be analysed. Here, size refers to the number of numerically discretized simulation support points, which is related to the sampling theorem, see Sec. 2.7.4. The target resolution of the USCT project is approximately $1 \cdot 10^{-4} \text{ m}$ [GR07] which is in line with the employed probing wavefield which has a center frequency of 2.0 MHz, and hence a wavelength of $\lambda = 5.3 \cdot 10^{-4} \text{ m}$ in water[Mar97].

The RoI covers a 3D cube of $0.1 \text{ m} \times 0.1 \text{ m} \times 0.1 \text{ m}$, motivated by typical breast of size B-C as aimed for patient population, see Fig. 3.1 [Sch08a].

A 3D voxel image calculated for the given RoI with $1 \cdot 10^{-4} \text{ m}$ resolution and double-precision floating-point data requires 7.45 GByte of memory. This memory

requirement clearly shows the need for efficient and memory-saving algorithms and corresponding implementations.

3.2.1 Approaches for US Systems: Apertures

Several apertures were selected for the imaging systems to be analysed, focusing on three systems: two directly related to the USCT project and one hypothetical system of theoretical interest.

The *3D USCT I* aperture, see Fig. 3.2, of the 3D USCT I tomograph was chosen because reference data is available for it. It also features a rotationally symmetric, layered cylindrical aperture optimized for uniform surface coverage of the aperture walls, see [Wue00]. The resolution of this system in the z-direction is significantly lower than along the xy-plane. The primary motivation for simulation studies was to compare it with the 3D USCT II demonstrator, at the time of this stage of research.

The geometry of the second-generation, the 3D USCT II system as shown in Fig. 3.3, was extensively optimized over several metrics. The goal was to reduce the z-resolution weakness and achieve a more homogeneous illumination of the target area [Sch08a]. The motivation here was to compare it with metrics obtained using more complex methods at the time.

The third imaging system assumes the geometry of a typical 2D-sonography aperture, see Fig. 3.4, but with frequency properties comparable to the USCT project. Such sonography apertures are typically operated in phased-array mode and are therefore designed under different premises than the two USCT apertures that serve SAFT imaging. Further motivation for a sonography geometry is to compare 3D and 2D apertures. While the aperture in the two USCT systems is distributed widely in 3D, the sonography aperture is spatially confined to a 2D plane, leading to significantly different imaging properties.

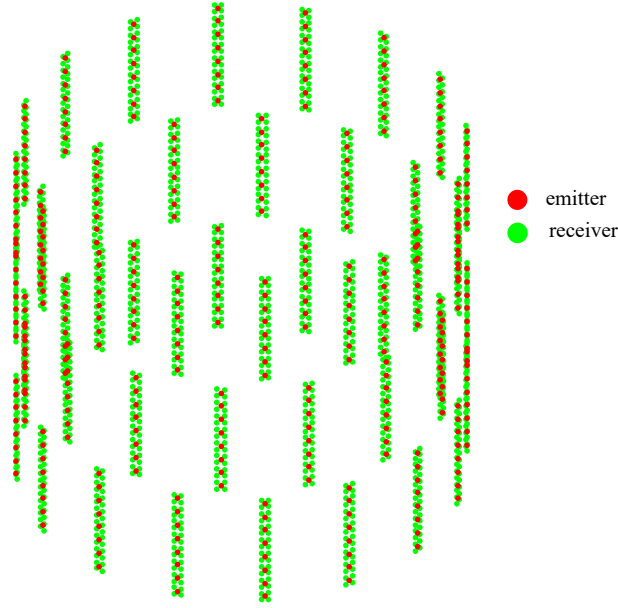


Figure 3.2: *3D USCT I transducer geometry with color-coded relative spatial positions for emitters (red) and receivers (green).*

3.2.2 Aperture 3D USCT I

The 3D USCT I demonstrator, completed in 2005, features a cylindrical measurement container with a diameter of 0.184 m. It has 24 evenly spaced layers along the z-axis, see Fig. 3.2. Each layer contains 16 emitters distributed equiangularly, totaling 384 emitters. Similarly, 48 layers of ring-shaped receivers are arranged equiangularly, each layer containing 16 receiver pairs, totaling 1536 receivers. The normals of the emitter and receiver emission surfaces are parallel to the xy-plane and directed toward the central rotational axis.

3.2.3 Aperture 3D USCT II

The aperture of the hardware for 3D USCT II is sparsely distributed on a semi-ellipsoidal 3D body, see Fig. 3.3. The parameters of this rotationally symmetric semi-ellipsoidal body are 0.18 m as the long elliptical axis in the z-direction (depth) and 0.12 m as the short elliptical axis in the xy-plane. The diameter at the upper edge is 0.24 m. On the 157 approximately evenly distributed transducer positions, nine receivers and four emitters are offset. Since these nine receivers and four emitters are on the same TAS, they share the same normal vector.

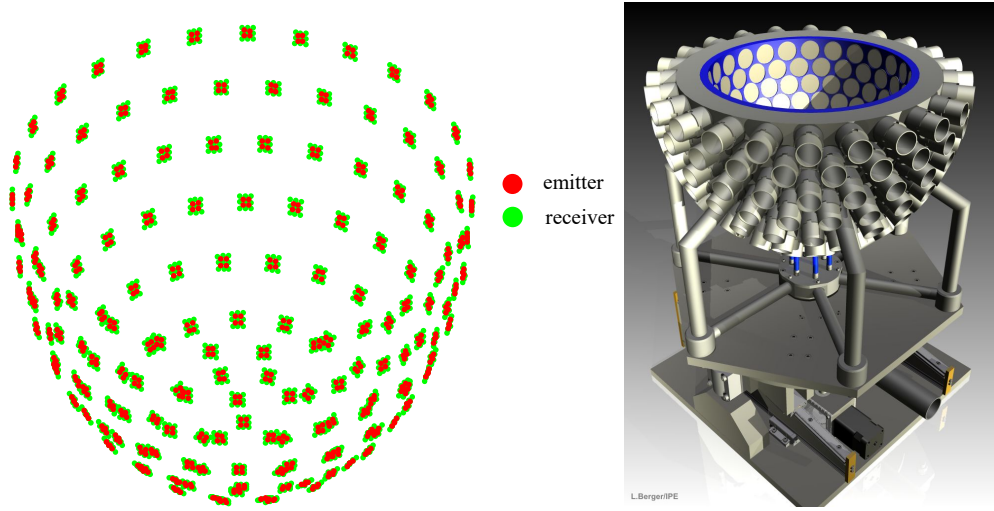


Figure 3.3: (left) 3D USCT II transducer geometry with color-coded emitters and receivers, red are the emitters, green are the receivers. (Right) scheme of the complete 3D USCT II system with mechanical base.

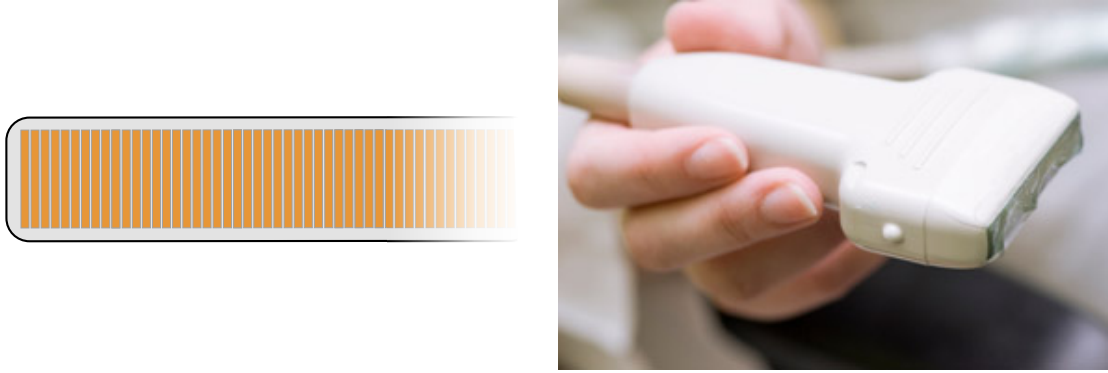


Figure 3.4: Structure of a typical sonography scanner: (left) 256 lamellae along the x -axis, brown elements represent transducer surfaces that can act as both emitters and receivers. (right) Photo of such a hand-held sonography probe.

The normals are directed towards approximately the centered RoI of $0.1\text{ m} \cdot 0.1\text{ m} \cdot 0.1\text{ m}$ size. This region will be the area containing the breast to be imaged during a measurement, as defined in Schwarzenberg's work[Sch08a].

3.2.4 Aperture 2D Sonography

The assumed sonography geometry also serves as an example of a phased-array architecture, see Fig. 3.6, consisting of many parallel transducer lamellae along the x -dimension. The dimensions of the individual elements along the x -axis

correspond to sub-elements of the 3D USCT I and 3D USCT II apertures, see Fig. 3.4. The elements are individually controllable and act as both emitters and receivers. The large emission area, achieved through significant z-axis extension of 4.9 mm compared to the x-axis dimension, generates a high sound intensity during emission and has a high sensitivity during reception. However, this large integration area also results in a poor resolution along z-axis, being orders of magnitude lower than in the xy-plane, effectively limiting the use of this aperture to 2D imaging of objects constant along their z-axis. The normal vectors of the elements are identical to each other and are parallel to the xz-plane.

The same RoI as for 3D USCT II is used, with a distance of 0.04 m from the aperture.

3.3 Simulation of Apertures as Plane-Wave Sources

Ewald Spheres as Monochromatic Emitter-Receiver Modeling

In the literature ([SHD08, AT00], etc.), a concept compatible with a linear system theory is used: a monochromatic emitter-receiver overall description using the *Ewald Sphere Construction* in the frequency domain see Sec. 2.14. Such a monochromatic frequency consideration can be transformed into a polychromatic approach through the linearity properties of k-space, see Chapter 2. Plane waves, as the simplest fundamental elements with their direction of propagation, serve as a modeling approach for apertures acting as wave sources and sinks.

However, the assumption that each emitter element emits a single, spatially unlimited plane wave implies that they have an infinitely wide aperture. Similarly, receivers must also be infinitely long to receive plane waves. Although such transducers cannot be constructed, they are a useful theoretical construct because they allow for simplifying considerations.

Such a plane wave can be mathematically characterized very simple and only two components, as finite dimensions (such as height, width, or structure) of the transducer are no longer required in this modeling. In a spatial mathematical

description as plane equation it is characterized by only its direction vector and position vector. In a k-space description, the position vector transitions into a periodic phase from 0 to 2π . This results in an aperture description through the direction vectors and position vectors of the transducers and their US frequency in the monochromatic case, or their bandwidth in the polychromatic case.

Spatial Information and Radiation Direction Vectors

The spatial positions and the radiation normals of the transducers can be derived from existing project data descriptions of the apertures. The spatial positions are represented as normalized position vectors from the origin, and the radiation normal vectors we denote as $\vec{v}_{nv}$. In the k-space representation, the position vector $\vec{v}_O$ transitions into a phase $P(\lambda)$:

$$P(\lambda) = 2\pi \cdot \text{mod}(\|\vec{v}_O\|, \lambda). \quad (3.1)$$

Where the Modulo is defined here continuously as $\text{mod}(\alpha, \beta) = (\alpha/\beta) - \text{floor}(\alpha/\beta)$ for arbitrary values of α and β .

3.4 Modeling Structured US Transducers

3.4.1 US Apertures

The above model described in Sec. 3.3 is limited by the assumption of infinitely extended transducers. In reality, transducer geometry is subject to geometric constraints that define thickness, length, and structure. Many factors, such as transducer efficiency, impedance matching, and reflection within the piezoelectric ceramic, must be considered to achieve transducers with sufficient signal intensity (sound pressure), bandwidth, and desired radiation characteristics, see [Gö2, KK90].

The spatial radiation characteristics can be understood as the superposition of many diffractively overlapping Huygens waves. In optics, these diffractive overlaps

are commonly divided into two distance-dependent regions: the near field and the far field [Goo68]. The near field can be approximated by Fresnel diffraction, which is not relevant here, as typical USCT application distances lie well within the far field (see Fig. 3.8). The far field relevant to USCT is described by Fraunhofer diffraction and is therefore predominantly angle-dependent; see Sec. 2.5.9.

The angle-dependent amplitude in the far field is a key factor in transducer design. The opening angle is typically inversely proportional to the aperture size, whereas the radiated amplitude (or receive sensitivity) scales with the aperture area. This introduces an inherent design trade-off: larger apertures provide higher acoustic output and hence a better SNR at the cost of a narrower opening angle. For SAFT-based USCT applications, a comparatively broad opening angle is required, whereas phased-array systems in sonography [KK90] benefit from narrower beams and larger apertures.

The fundamental geometric properties of the apertures can be seen as the results of a spatial sampling consideration: the pitch of two neighbouring elements should be smaller than $\frac{\lambda}{2}$, see Fig. 3.5. Neglecting this can lead to spatial aliasing, known as *grating lobes* in imaging system theory. For a detailed introduction to this topic, see Goodman[Goo68].

Complex radiation patterns with corresponding wavefield may be obtained by exciting each element simultaneously, but with varying amplitude and phase. This approach is typically used in phased-array applications where the wave field is focussed along a single line. However, the same focussing can be also achieved after the measurement during the processing of the data by using methods such as SAFT, see Fig. 3.6.

Exciting the elements with different gains, it is called apodization. The motivation here is to optimize the transducers's opening angle and reduce side and grating lobes [KK90].

3.4.2 Properties of US Transducers

The construction of US transducers is a vast and specialized field of research. Therefore, only a limited introduction is provided here; for detailed discussions,

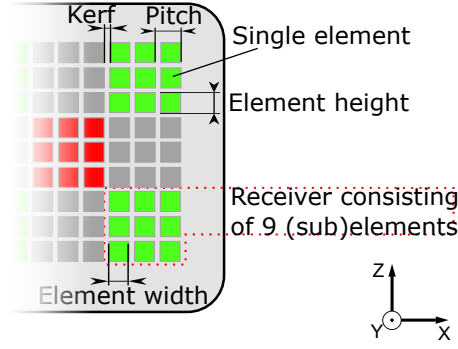


Figure 3.5: Transducer aperture terminology: Green elements act as receivers, red elements as emitters, and gray elements are inactive.

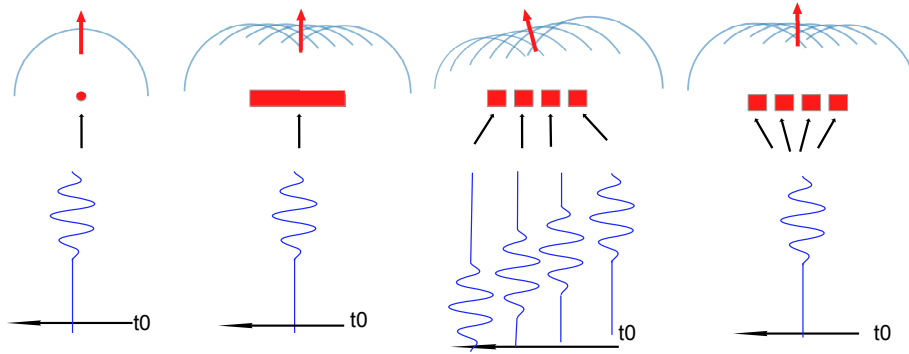


Figure 3.6: Interference of waves emitted at time t after excitation at t_0 : (far left) hypothetical point source, the ideal SAFT transducer; (left) real, flat rectangular source approximating a point source; (right) excitation of a substructured geometry (an “array”) through a time-phased pulse for a phased-array application. The phase shift enables targeted spatial overlap for focusing, both during emission and reception; (far right) substructured geometry without phase shift for a SAFT application, motivated by better radiation properties for SAFT use cases.

refer to [KK90].

3.4.3 Frequency Dependence and Radiation Characteristics

The efficiency of a real US transducers as a pressure-to-voltage converter (and vice versa) is frequency-dependent. The frequency at which the US receiver has its highest sensitivity is typically referred to as the *central or primary frequency*. Sensitivity decreases around this frequency, and the range where a 50 % sensitivity drop is reached is called the *bandwidth*. If a US transducer does not have an infinitely small aperture area (i.e., it is not a “point source”), it

exhibits radiation behavior depending on the aperture geometry and the distance. After compensating for the distance-related loss (geometric attenuation), the transducer behaves approximately only angle-dependent, the direction of the main lobe is described in 3D by azimuthal and elevation angles [KK90].

An angular sensitivity metric is well-known in many fields of transducer technology, referred to as *angular sensitivity* (or directivity, angular characteristics) for US transducers and hydrophones [Ond, KK90]. Specifically, in the case of transducers acting as emitters, it is called *radiation characteristics*; for receiving US waves, it is referred to as the *beamwidth*. Like the frequency bandwidth, it is usually defined by the half-power (-3 dB) point.

3.4.4 Lateral and Axial Bandwidth

In k-space, the bandwidth and angular characteristics of an US transducer map directly onto its *axial* and *lateral bandwidths*, respectively, following the terminology introduced by Anderson and Tahey [AT00].

The axial bandwidth results from the transducer bandwidth along the transducer's receiving or emitting axis in k-space, hence the term *axial bandwidth*. Lateral resolution („*side resolution*“) describes the width of the support along the radiation axis in k-space. For a monochromatic transducer emitting spherical waves, the lateral bandwidth forms along a ring or sphere concentric to the zero frequency in k-space, see Fig. 3.7. There is an inverse relationship between aperture size and lateral bandwidth. For example, a point aperture has a high bandwidth, while a large, spatially extended aperture has a low bandwidth [AT00]. Both bandwidths equally contribute to the k-space support, although typically not in equal proportions, forming a *ratio*. This ratio can serve as a metric for optimizing transducer designs, where different applications may demand varying proportions.

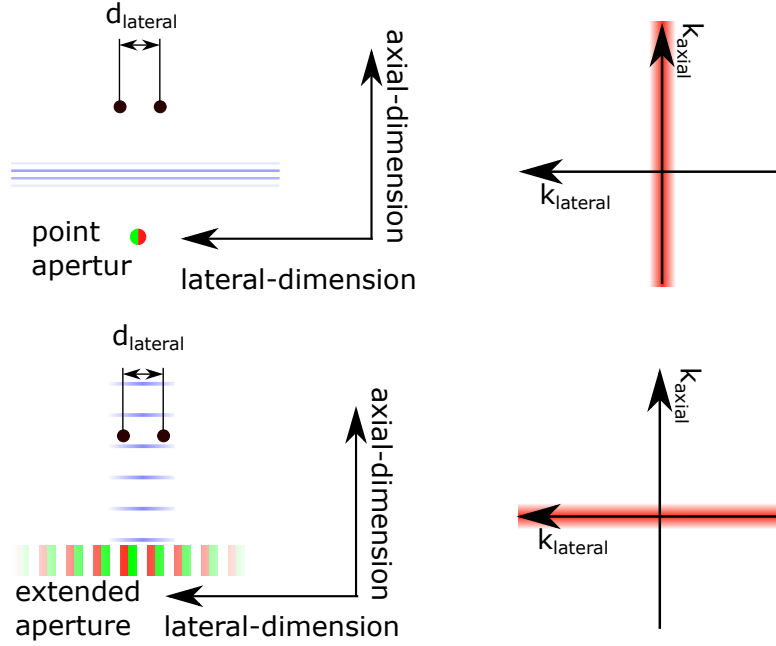


Figure 3.7: Resolution of two point objects separated only along the lateral dimension. (Top) point transducer with temporally unlimited bandwidth emitting a short temporal plane wave, represented in k -space as very high bandwidth. Acting as a receiver, the transducer can distinguish reflections from axially differentiated objects due to the ability to detect time delays. However, it cannot resolve laterally separated scatterers as no time delay exists. (Bottom) A spatially unlimited aperture with minimal temporal bandwidth can resolve the two point objects in phased array mode. By overlapping infinitely many spatially offset spherical waves along the aperture, a laterally narrow wave is generated that can “scan” the space for objects.

3.4.5 Simulation of Extended Transducer Apertures Using Fraunhofer Diffraction

The modeling of extended, planar transducer apertures can be achieved using diffraction via the Fraunhofer approximation. This approximation will be discussed in Sec. 3.4.9, as the far-field assumption for the 3D USCT II application is satisfied. The axial distance to the rectangular transducers from where on the far-field assumption is valid can be estimated using the “*antennas designer’s formula*” (see[Goo68], Eq. 4-27):

$$z > \frac{2 \cdot D^2}{\lambda} \quad (3.2)$$

where z is the minimum aperture-object distance, λ is the wavelength, and D is the aperture width, see Fig. 3.8.

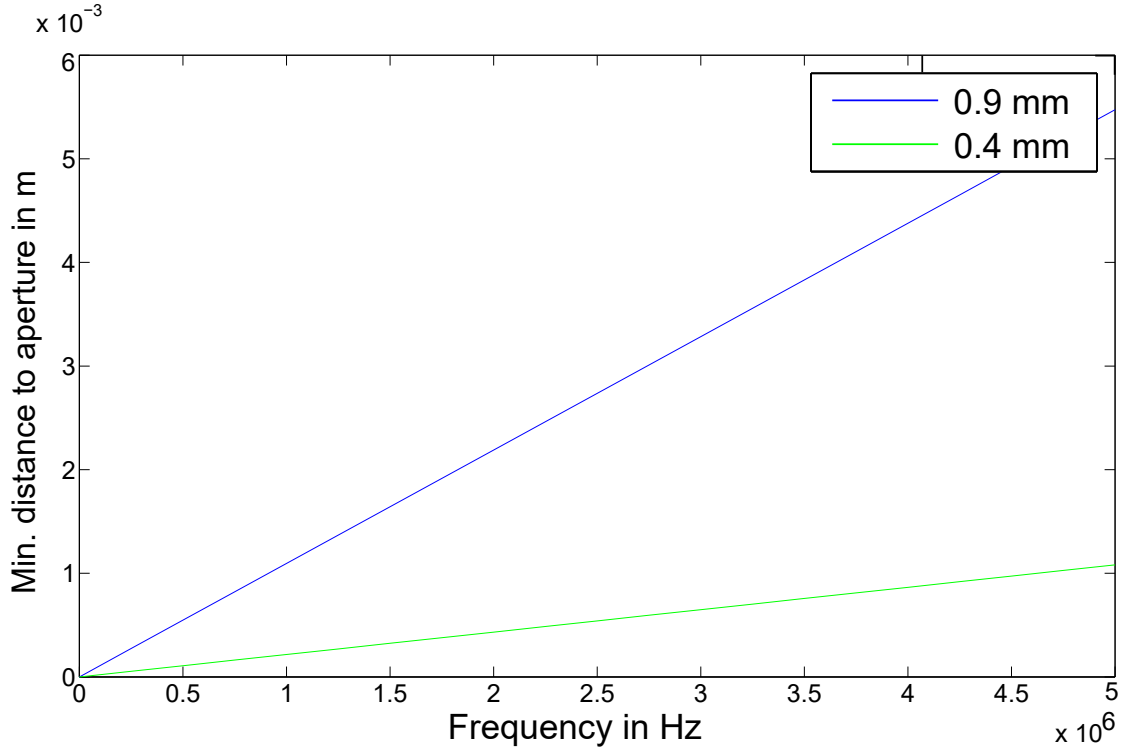


Figure 3.8: "Antennas designer's formula": Determination of minimum aperture-object distances required for (approximate) validity of the Fraunhofer approximation. (green) Smallest aperture width of the 3D USCT II, (blue) largest aperture width; x-axis: frequencies within the 3D USCT II system bandwidth (in Hz), y-axis: minimum distance (in m).

Given the spatial dimensions of the 3D USCT II measurement tank, with 0.12 m and 0.18 m along the two ellipsoid semi-axes and a centrally located RoI of 0.1 m, the minimum transducer-object distances required by Eq. 3.2 are exceeded by several orders of magnitude.

3.4.6 3D USCT I Transducer Structure

The 3D USCT I transducer system was manufactured at IPE and designed as part of a diploma thesis, see Fig. 3.9 [Gö2]. Structuring and sub-structuring were achieved by sub-dicing the piezoelectric core component, a PZT piezo plate [Gö2]. A saw blade thickness of $1 \cdot 10^{-4}$ m and a saw spacing of $4 \cdot 10^{-4}$ m resulted in the structure shown in Fig. 3.11. The substructured piezo plate was electrically connected to form larger emitter and receiver units. Each nine sub-elements, arranged

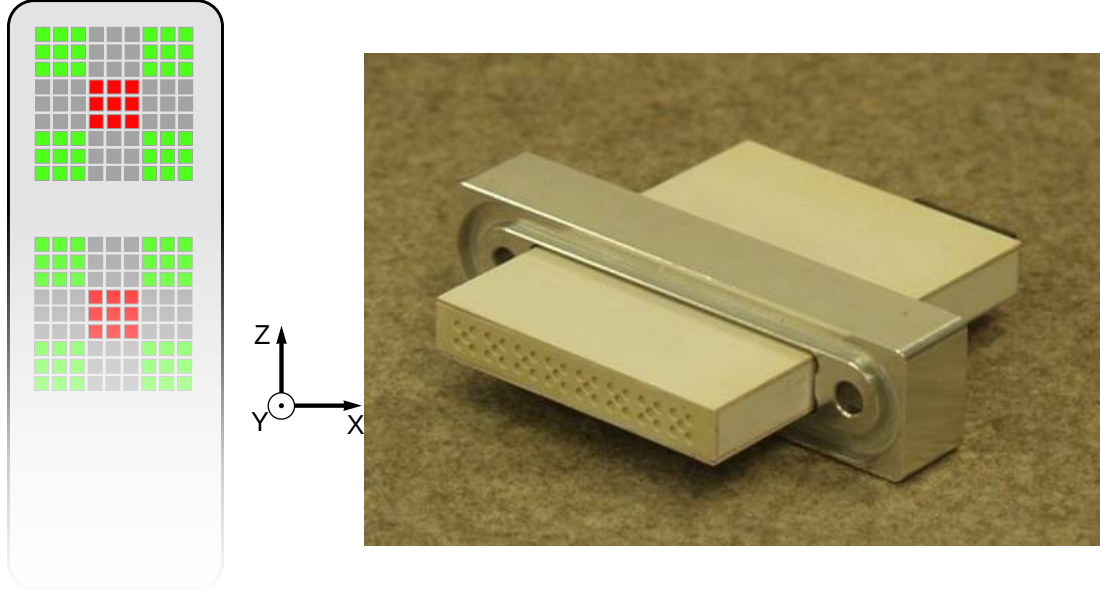


Figure 3.9: (Left) Structure and substructure of 3D USCT I transducers: green highlights the structure acting as a receiver, red highlights the structure acting as a emitter. Each consists of nine sub-structure elements electrically connected into a emitter or receiver structure. (Right) 3D USCT I transducer in housing.

in a square, form a controllable transducer by bonding together. Per piezo plate, four receivers and one emitter are formed. Together with amplifiers, multiplexers, and driver electronics, the piezo plate was sealed water-proof in a housing. Next, the radiating surface was polished to $\lambda/4$ to optimize radiation performance at the required center frequency. The resulting simulated and measured radiation patterns are shown in Fig. 3.11. These results show that the patterns are in excellent agreement with each other.

3.4.7 3D USCT II Transducer Structure

The 3D USCT II transducer system was also designed and manufactured at IPE. Structuring and sub-structuring were achieved by sawing a piezo plate. A saw blade width of $1 \cdot 10^{-4}$ m and a spacing of $4 \cdot 10^{-4}$ m resulted in the substructure shown in Fig. 3.12. Four quadratically arranged sub-elements are electrically bonded to form a controllable transducer. Each piezo plate contains nine receivers and four emitters, as shown in Fig. 3.12 left. This 3D USCT II transducer system is encapsulated in stainless steel housings with amplification and control electronics,

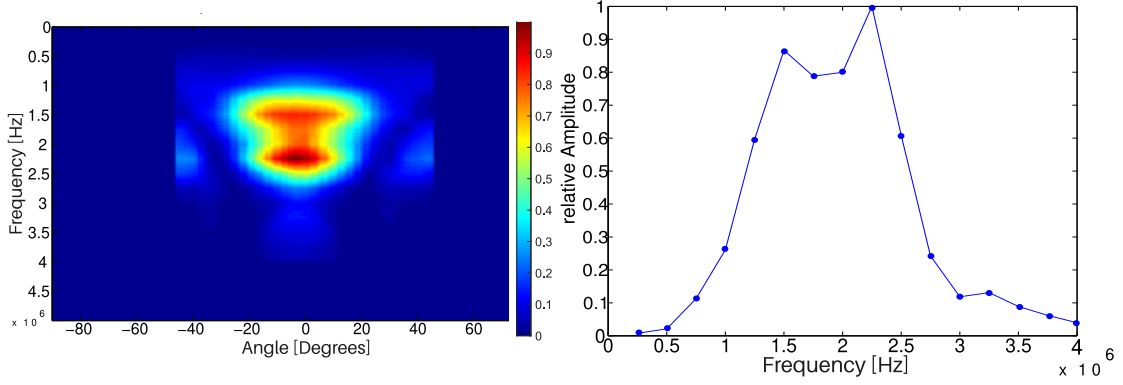


Figure 3.10: Measured characteristics of 3D USCT I: (left) Pressure measurement as frequency versus angle of the transducer structure I, (right) frequency-dependent transducer sensitivity for 3D USCT I transducer based on a 512-fold averaged measurement. From exp72-79-5-9.

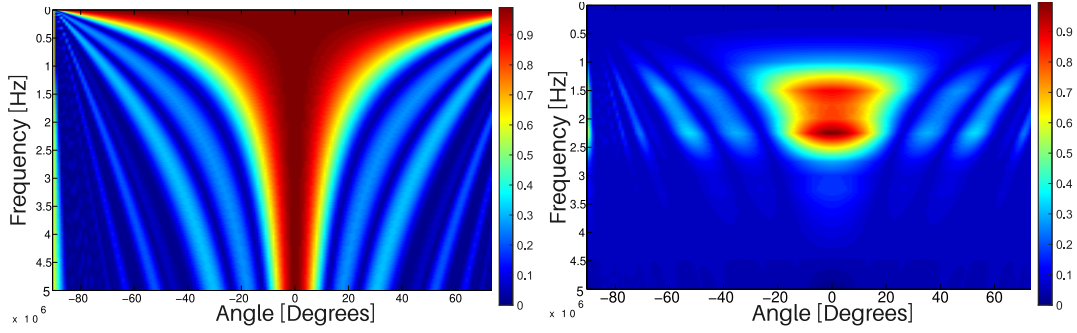


Figure 3.11: Simulated characteristics of 3D USCT I: (left) 1D simulation (Fraunhofer approximation) of relative pressure as frequency versus angle, (right) simulation weighted with the angular-independent axial frequency sensitivity obtained from the reference measurement, see Fig. 3.10.

as shown in Fig. 3.12. Additionally, scattering caps are attached around the active radiating surface to minimize potentially disruptive reflections from other emitters.

3.4.8 Sonography Aperture

The assumed element size along the x-axis matches the sub-element size of the 3D USCT I and 3D USCT II transducers, i.e., $4 \cdot 10^{-4}$ m. The extension along the z-axis corresponds to the total height of the 3D USCT II piezoceramic active area, i.e., $4.9 \cdot 10^{-3}$ m. Since no reference measurements are available, a Gaussian frequency sensitivity is assumed. The resulting radiation patterns are shown in

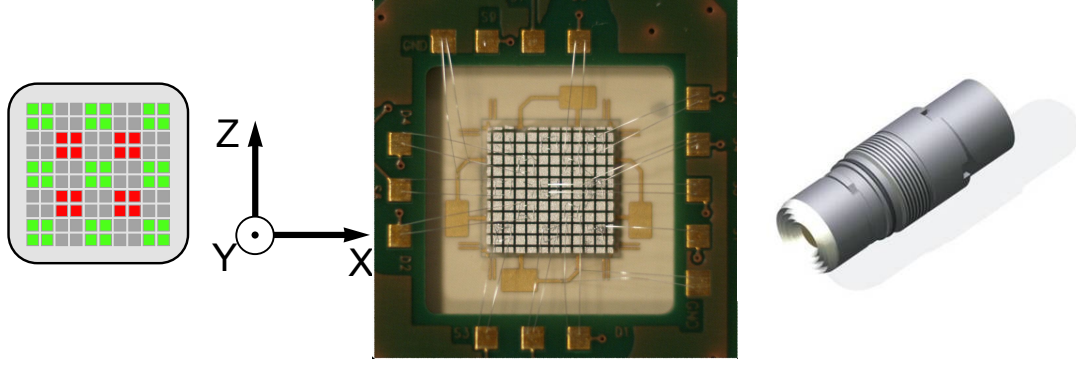


Figure 3.12: 3D USCT II transducer aperture: (left) Structure and substructure of 3D USCT II transducer System: green elements act as receivers, red elements act as emitters. Four sub-structure elements are electrically bonded into a single emitter or receiver structure, (middle) piezo structure and substructure, with an elemental sub-element cube side length of $4 \cdot 10^{-4}$ m and saw kerf width of $1 \cdot 10^{-4}$ m, (right) 3D USCT II in stainless steel housing with scattering caps on the radiating surface.

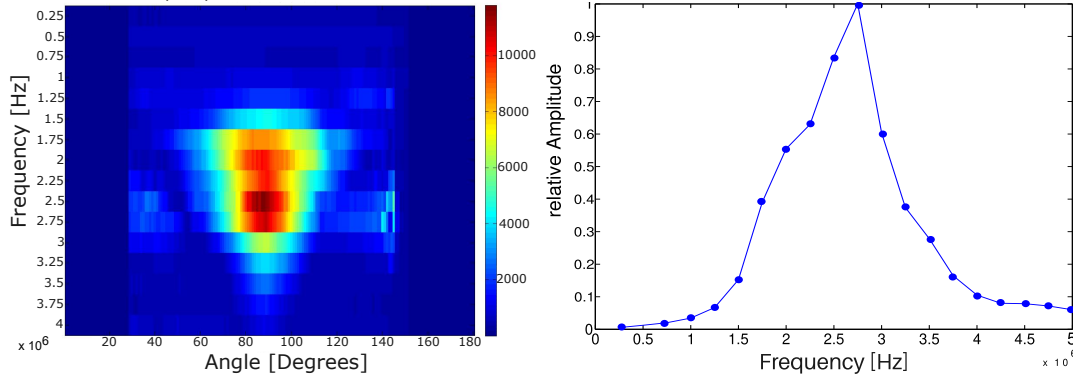


Figure 3.13: Measured characteristics of 3D USCT II: (left) Pressure measurement as frequency versus angle for the transducer structure II, (right) frequency-dependent sensitivity for 3D USCT II transducers based on a 512-fold averaged measurement. Data from experiment exp65_79-4-4.

Fig. 3.15. Note the natural focussing taking place in the angular direction due to elongation of the elements in the z-axis.

3.4.9 Lateral Bandwidth via Fraunhofer Diffraction Approximation

The 3D radiation of the transducer system over a range of frequencies is to be investigated. Fraunhofer diffraction is used, see Sec. 3.4.9, as the conditions for the geometries under consideration are met. For angular analysis, 1D definitions

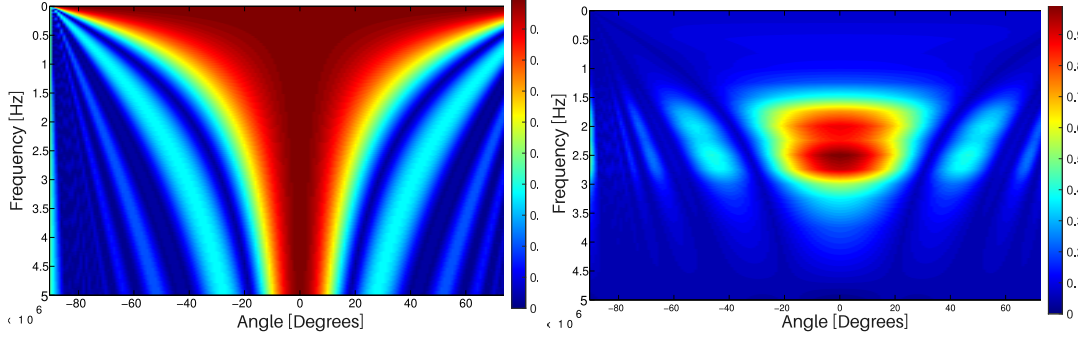


Figure 3.14: Simulated characteristics of 3D USCT II: (left) 1D simulation (Fraunhofer approximation) of relative pressure as frequency versus angle, (right) simulation weighted with the angular-independent frequency sensitivity derived from the reference measurement, see Fig. 3.13.

of the aperture are used for both the azimuth and elevation vectors (neglecting geometric extension).

The transducer aperture is modeled in 2D, analogous to optical single-slit and double-slit experiments (see also Goodman Chapter 3.4 “*Formulation of Diffraction as Planar Screen*”), as an aperture with properties of being either transparent or opaque. For this analysis, the thickness of the transducer system is not considered.

The introduced Fraunhofer diffraction can be used to describe the diffractive far-field of aperture structures, see Sec. 2.5.9, to determine the *lateral bandwidth*, see Fig. 3.16.

Afterward, a spatial 1D aperture $A(x)$, discretely sampled at s_f and transformed into the Fourier domain,

$$F_A(f_x) = \mathcal{F}\{A(x)\}, \quad (3.3)$$

can be used to approximate the angular radiation characteristics (shown here for the x -dimension).

In the Fraunhofer (focal) plane at distance d_f , the transverse coordinate x maps to the 1D spatial frequency axis

$$f_x = \frac{x}{\lambda d_f}, \quad (3.4)$$

where λ is the wavelength and d_f is the focal distance. Since the Fraunhofer

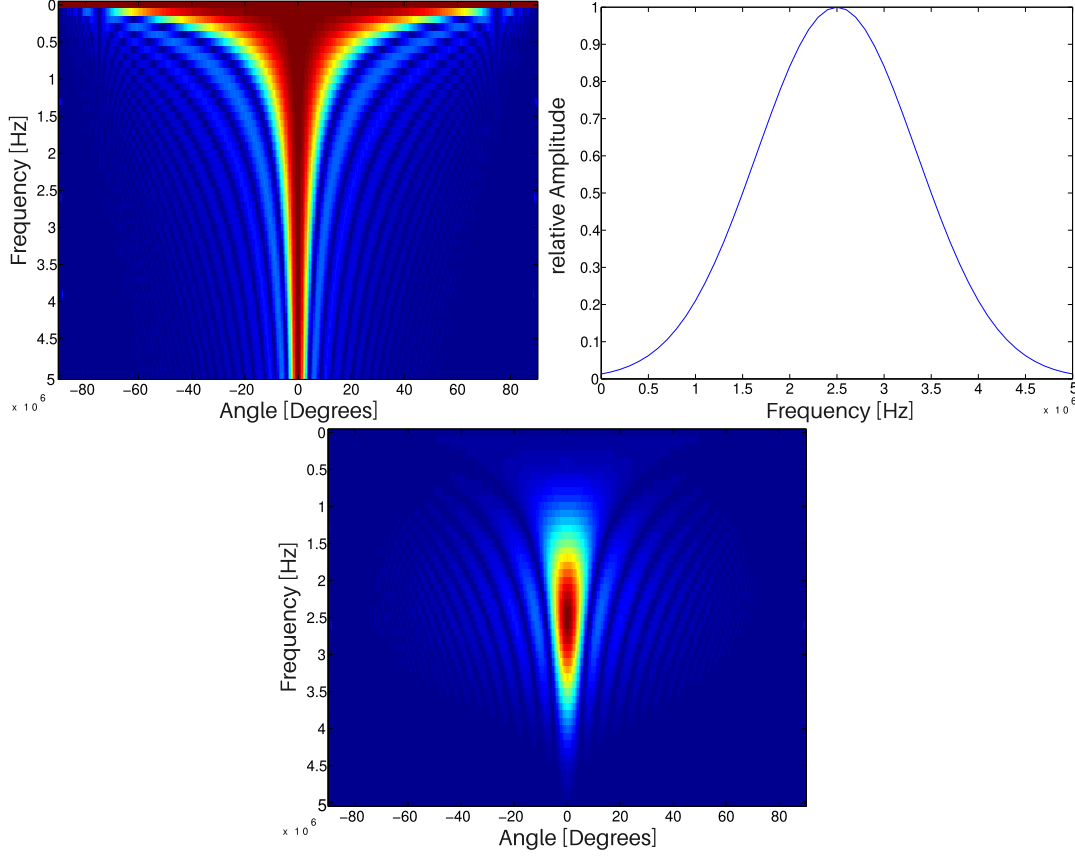


Figure 3.15: *Simulated characteristics of sonography aperture: (Left) Fraunhofer diffraction approximation for the radiation characteristics of the sonography aperture, (top-right) Gaussian frequency sensitivity assumption with a center frequency of 2.5 MHz and a bandwidth of 2 MHz, (bottom) simulation weighted with the angular-independent Gaussian frequency sensitivity.*

approximation provides a one-to-one mapping between transverse position x and diffraction angle β , it is often convenient to re-parameterize the same spectrum in angular coordinates (angular spectrum representation; see Section 2.5.4). Using the geometric relation

$$\tan(\beta) = \frac{x}{d_f}, \quad (3.5)$$

we obtain

$$f_x = \frac{x}{\lambda d_f} = \frac{\tan(\beta)}{\lambda} \equiv f_\alpha. \quad (3.6)$$

Here, f_α denotes the spatial frequency axis expressed in angular coordinates; the focal distance cancels identically [AT00, Goo68]. As consequence, complex 1D apertures can be constructed from simpler structures using Fourier theorems.

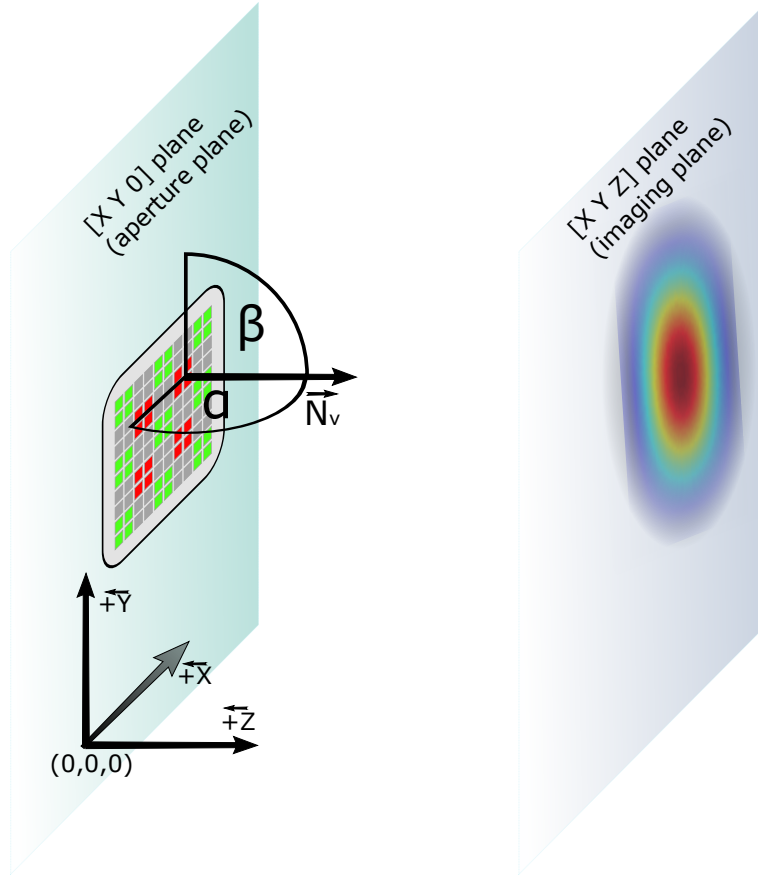


Figure 3.16: *Coordinate system for Fraunhofer diffraction description: (Left) arbitrary planar diffractive structure, showing the elevation angle β and azimuth angle α (the two diffraction angles) for the emitter at the top-right, with the radiation normal indicated as $\vec{N}_v$, (right) amplitude mapping of the diffractive wave phenomenon, showing the absolute value of the relative amplitude ranging from red (maximum pressure) to blue (minimum pressure) for an assumed image plane parallel to the aperture plane (“coplanar”).*

3.4.10 Axial Bandwidth from Reference Measurements

The axial bandwidth cannot be estimated without simulations incorporating material properties and the spatial depth structuring of the transducer [KK90]. Advanced simulation based on Finite element modelling is trying to address this [Koh10]. Alternatively, instead of assuming Gaussian axial frequency characteristics, reference data were obtained using the project’s transducer measurement setup, as shown in Fig. 6.6.

Geometrical considerations and initial evaluations against reference data revealed acceptable simulation accuracy, but certain effects are inadequately modeled. It

is plausible that the real US transducers operate with frequency and phase shifts due to damping, matching layers, and other properties.

3.5 Modeling Extended Transducers with Spatial Positions and Radiation Vectors

A logical evolution of the presented ideas and models is a simulation that combines the properties of the Ewald sphere construction with Fraunhofer aperture simulation. The goal is to improve simulation accuracy by integrating far-field modeling of transducer apertures and considering global transducer positions and radiation normals under more realistic geometric assumptions than those in the pure Ewald sphere construction.

The simulation of frequency characteristics extends beyond purely geometric aperture-based approaches, relying on measured data or assuming Gaussian characteristics, a common practice [AT00].

3.5.1 Amplitude via Fraunhofer Diffraction Approximation of the Far Field

The angular amplitude of an aperture structure is derived using Fraunhofer diffraction, as described in Sec. 3.4.9.

3.5.2 Focusing

In SAFT imaging systems, focusing on a point is achieved through time shifts in the spatial domain. A time shift, representing the passage of time, is linked to wave propagation but is not explicitly modeled. The simulation must be correct only for the moment when the wave interacts with the imaging point.

Since the propagation speed can be approximated as frequency-independent (i.e., no dispersion), the spatial propagation of US waves remains proportional to time,

independent of frequency [Hui08]. This is well-known and expressed as:

$$s(t) = v_{H20} \cdot t, \quad (3.7)$$

with $v_{H20} = 1480$ m/s. Frequency-dependent behavior, however, affects the phase $P(f, t)$, resulting in a phase shift over time due to propagation:

$$\begin{aligned} \lambda(f) &= \frac{v_{H20}}{f}, \\ P(f, t) &= 2\pi \cdot \text{mod}(s(t), \lambda(f)). \end{aligned} \quad (3.8)$$

3.5.3 Spatial Adjustment

Another phase shift arises from the spatial separation of diffractively interacting structures emitters, receivers, etc.). This is also modeled through the phase component. A translation along the position vector $\vec{V}_{dO}$, derived from the two position vectors $\vec{V}_{O1}$ and $\vec{V}_{O2}$, and the directional vector $\vec{V}_{PW}$ of the plane wave results in:

$$\begin{aligned} \lambda(f) &= \frac{v_{H20}}{f}, \\ \vec{V}_{dO} &= \vec{V}_{O2} - \vec{V}_{O1}, \\ S_f(t, f) &= \sum (\vec{V}_{dO} \cdot \vec{V}_{PW}) \cdot s(t), \\ P(f, s) &= 2\pi \cdot \text{mod}(S_f, \lambda). \end{aligned} \quad (3.9)$$

3.5.4 Diffractive Interaction in k-space as Convolution

In the *k-space Tutorial* and Goodman's *Fourier Optics* (Chapter „Angular Spectrum“, Chapter 3.10.3), a polychromatic k-space development is defined for complex emitter and receiver modeling, incorporating axial and lateral bandwidth, position vectors, and normal vectors. Initial steps have been taken to model diffraction as convolution in a 3D angular spectrum, though further theoretical groundwork is needed.

3.5.5 Transducer Characterization and Modelling

The transducer structures are assumed for the modelling to be separable into two one-dimensional functions due to their two-dimensional, mirror-symmetric piezo structure. The relative pressure P_m measured at the hydrophone comprises the frequency-dependent sensitivity $S(f)$ and two aperture-dependent components $A_x(x)$ and $A_y(y)$ along the symmetry axes x and y :

$$P_m(x, y, f) = A_x(x) \cdot A_y(y) \cdot S(f). \quad (3.10)$$

Reference values for the available 3D USCT I and 3D USCT II transducers were obtained using the project's transducer measurement setup [Pet05]. Since the transducer to be measured, 3D USCT I and 3D USCT II, are identical across their respective x and y axes, it suffices to measure along one axis. The other axis must be fixed at an angle of 0° , which here corresponds to $y = 0$ or $x = 0$, ensuring no angular characteristic influence from this dimension on the measurement. For example, $A_x(x)$ along $y = 0$ over x is represented as:

$$P_m(x, 0, f) = A_x(x) \cdot A_y(0) \cdot S(f). \quad (3.11)$$

The principal frequency dependency is determined by fixing both spatial parameters x and y and conducting a measurement run over f :

$$S(f) = P_m(0, 0, f), \quad (3.12)$$

with the aperture functions normalized such that $A_x(0) = A_y(0) = 1$.

The relative two-dimensional aperture function $P_m(x, y, f)$, derived from the measured values $P_m(x, 0, f)$, is calculated as:

$$P_m(x, y, f) = \frac{P_m(x, 0, f) \cdot P_m(y, 0, f)}{S(f)}, \quad (3.13)$$

with a single compensation of the twice-applied principal frequency dependency resulting from this multiplicative overlay.

Since the hydrophone cannot be rotated, the measured pressure from the emitting transducer is influenced by the angular sensitivity of the hydrophone. Using the calibration data of the hydrophone over frequency $S_H(f)$ and position $H_x(x)$, this effect in the measured data $P_{H,m}(x, 0, f)$ was compensated (“deconvolution”):

$$P_m(x, 0, f) = \frac{P_{H,m}(x, 0, f)}{H_x(x) \cdot S_H(f)}. \quad (3.14)$$

3.5.6 Parameter Space

The measurements serve as a reference for simulations based on the far-field assumption. Hence, measuring the entire radiation field across spatial coordinates (x, y, z) and additional frequencies, forming a 4D parameter space, is unnecessary. A 3D parameter space over elevation angle β , azimuth angle α , and frequency f at a fixed distance fulfilling the far-field condition suffices.

With the assumption introduced in Sec. 3.5.5 that the two angular characteristics are separable, and that the two transducer types measured are identical across their two mirror axes, the problem reduces further to a 2D problem.

Thus, the goal is to measure a 2D parameter space of relative angular sensitivity α along a mirror symmetry axis of the transducer and the relative frequency sensitivity f of the emitting transducer under test.

The measurement setup allows a step size of 0.001 m but is spatially constrained by side walls, water level, transducer port position, and hydrophone size. Due to the non-rotatable hydrophone and spatial constraints, the effective measurement range for angles is significantly smaller than the theoretically expected -90° to $+90^\circ$, approximately -50° to $+50^\circ$ across the YZ plane.

A coordinate list was defined with 180 positions along a semicircle for constant pressure without requiring additional compensation for geometric propagation loss. At each hydrophone measurement position, a narrowband pulse was emitted by the transducer. The sinusoidal pulse’s center frequency was incrementally varied during measurement across the known bandwidth of the transducer and

slightly beyond, between 250 kHz and 4 MHz in 250 kHz steps.

Since the transducers have a small radiation area (and the hydrophone also offers wide angular independence), the achieved signal-to-noise ratios (SNR) were low. Additionally, electromagnetic interference (EMI) from the measurement PC and other active electrical components (DAQ, AWG, etc.) overlapped the measurement signal with various frequency components. To minimize this influence, the measurement was repeated multiple times and averaged in software (512 measurements, resulting in $\sqrt{512}$ SNR gain). Furthermore, a bandpass filter covering the hydrophone's sensitivity range (1 to 10 MHz) was applied. The results of these measurements can be seen in Fig. 3.11 and Fig. 3.14 and were incorporated as axial bandwidth in the Fraunhofer simulations.

3.6 Evaluation of Simulated Transducer Characteristics

The quality of the simulated transducer characteristics across frequency and angle should be quantitatively compared to the measured reference data. For this purpose, both datasets were transformed into the 3D angle-frequency space, comprising elevation angle α , azimuth angle β , and temporal frequency f .

Quantifiable and comparable metrics are desired for evaluating simulation quality. The *mean-square-error* (MSE) is a standard metric, growing quadratically with deviations, making it sensitive to outliers and mean shifts. The MSE is computed as follows:

$$MSE(a, b) = \frac{1}{N_i} \sum_{i=1}^{N_i} (a(i) - b(i))^2. \quad (3.15)$$

Additionally, *coherence*, a standard measure from signal and system theory (distinct from the physical coherence concept) [Wik10b], is used. Coherence is a metric that provides similarity values between 0 (maximum dissimilarity) and 1 (identical). It can be considered a normalized correlation interpreted as a moving scalar-vector product forming the basis of a relative similarity measure.

For implementing coherence, the correlation was reformulated as a convolution

and computed in the Fourier domain using the Fourier convolution theorem, as convolution in the spatial domain is computationally intensive for large multidimensional problems. Consequently, the required correlation $\text{Corr}(f_1, f_2)$ is computed as:

$$\text{Corr}(f_1, f_2) = \mathcal{F}^{-1}(\mathcal{F}(f_1) \cdot \text{Conj}(\mathcal{F}(f_2))) \quad (3.16)$$

for mean-free, normalized functions f_1 and f_2 of arbitrary dimensions. The specific coherence $\text{Coer}(G_m, H_m)$ of the simulated and measured angle- and frequency-dependent 3D transducer characteristics $G_m(\alpha, \beta, f)$ and $H_m(\alpha, \beta, f)$ is then:

$$\text{Coer}(H_m, G_m, f) = \frac{\max(\text{Corr}(H_m, G_m))^2}{\max(\text{Corr}(G_m, G_m)) \cdot \max(\text{Corr}(H_m, H_m))}. \quad (3.17)$$

3.7 Deriving Ideal Apertures and Systems

3.7.1 Simulation of the Ideal Aperture Using the Ewald Sphere Construction

When considering k-Space contributions as an imaging function, the ideal filling corresponds to an equidistant and evenly weighted sampling of the entire space—an imaging function acting as an *all-pass filter* on an object function located in k-Space. This ideal k-Space support for all occurring k-Space frequencies is attempted to be generated using an inverse Ewald sphere construction of *plane-wave emitter-receiver combinations*, as described in Sec. 2.9.1.

Analyzing the potential filling through the Ewald sphere construction reveals that there is no uniqueness; instead, multiple pathways can achieve the same k-space point $\vec{k}(f_x, f_y, f_z)$:

$$\vec{k}(f_x, f_y, f_z) = k_e(-\vec{v}_e) + k_e \vec{v}_r = (\vec{v}_r - \vec{v}_e) \frac{\cos(\alpha)}{\vec{v}_e \cdot \vec{v}_r} \|\vec{k}(f_x, f_y, f_z)\|. \quad (3.18)$$

Here, $\vec{v}_e$ and $\vec{v}_r$ denote the unit direction vectors of the emitter and receiver, respectively. The corresponding wave-number magnitudes are identical in the

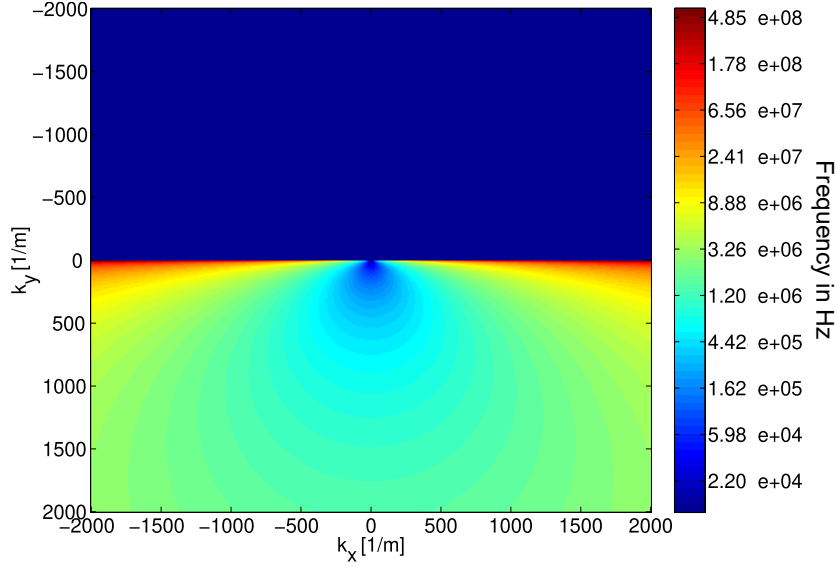


Figure 3.17: *k*-space construction with a given emitter normal vector $\vec{v}_s$. The plot logarithmically shows the required temporal frequency ($f = \frac{v_{H2O}}{k_e}$) over the 2D *k*-Space that a emitter-receiver combination must possess to contribute. With a single emitter, at most half of the *k*-Space can be filled, assuming that planar transducers realistically have a maximum radiation characteristic from $+90^\circ$ to -90° . RoI is 0.1 m^3 and pixel resolution is 0.5 mm .

elastic, monochromatic case and are given by

$$k_e = \frac{2\pi}{\lambda}. \quad (3.19)$$

The filling approach illustrated in Fig. 3.17 is obtained by assuming a fixed emitter normal vector

$$\vec{v}_{e,1} = [0, 1, 0]. \quad (3.20)$$

For reconstructing a spatial image characterized by a pixel count n_v and an isotropic pixel size s_v , the defining parameters of the corresponding *k*-space grid—namely the number of samples $n_{V_{\text{k-space}}}$ and the sampling interval $s_{V_{\text{k-space}}}$ —follow directly from n_v and s_v , as detailed in Sec. 2.7.3.

Using the inverse Ewald sphere construction, this *k*-space volume is subsequently populated. The receiver normal vectors $\vec{v}_r$ span a polychromatic result space, yielding a set of receiver directions whose cardinality is equivalent to the spatial pixel count.

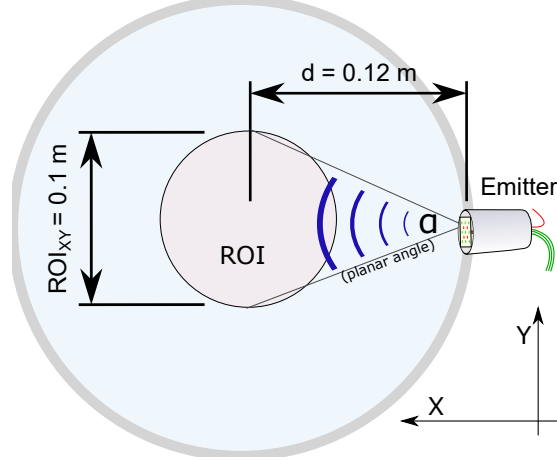


Figure 3.18: Estimation of the optimal radiation characteristics of a emitter for the 3D USCT II measurement setup: the optimal angle for an RoI of 0.1 m equals $\alpha = 45.23^\circ$.

The resulting temporal frequency bandwidth can be interpreted as a requirement for an imaging system capable of filling such a k-Space. Additionally, the angular differences resulting from the receiver normal vectors $\vec{v}_r$ and emitter normal vectors $\vec{v}_s$ can be interpreted as system requirements.

3.7.2 Simulation of the Ideal Aperture Using Fraunhofer Diffraction

The approach here involves the direct computation of a desired acoustic beam in k-Space. Through an inverse use of Fraunhofer diffraction, the absolute value of the k-Space representation of this acoustic beam, as a function of temporal frequency and spatial angles, can be interpreted as an aperture definition, as shown in Fig. 3.7.

Since the geometry scales differently for each temporal frequency, a physical aperture can only be optimal for one single frequency. For optimal 3D USCT transducers, we select the center frequency as the target frequency, though choosing the highest frequency would be also possible. The desired radiation characteristic is determined by the distance from the center of the RoI to the aperture. Here, the largest ring diameter of the 3D USCT II overall aperture is used as a *worst-case* scenario (as this presents the largest angle), with $d = 0.12$ m to the center of the RoI and an extension of the $RoI_{XY} = 0.10$ m across XY , as shown in Fig. 3.18.

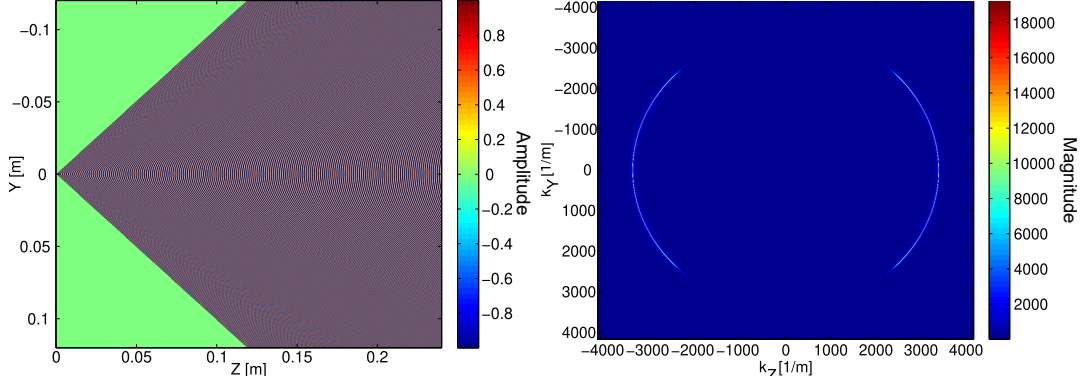


Figure 3.19: *Ideal radiation profile for 3D USCT II (2D): (left) Point emitter source at position $X = 0$ m, $Y = 0$ m, and $Z = 0$ m emits a monochromatic wave with a center frequency of $2.5 \cdot 10^6$ MHz and a radiation angle of 45.23° ; (right) representation in k -space.*

For this geometry the optimal radiation angle is $\alpha = 45.23^\circ$.

For the simulation, a wave with a monochromatic frequency of 2.5 MHz and a hard-limited opening angle of 45.23° is generated in the spatial domain, originating from an unknown sound source at $x = 0$ m and $y = 0$ m. In Fig. 3.19, both the spatial and spatial frequency domain representations are shown. The next step uses the coplanar configuration of Fraunhofer diffraction, with the aperture layer $U_i(x, y, 0)$ and the diffraction layer $U_t(x, y, z)$ at $Z = 0.096$ m. The choice of layer is arbitrary (aside from periodicity and sampling considerations), as only the angular dependency of Fraunhofer diffraction is used. The propagation direction of the wave is the z -dimension. This layer is inverse Fourier transformed to generate $U_i(x, y, 0)$, which is further scaled based on the distance and wavelength. In Fig. 3.20, the generated continuous aperture values are shown (top-right). These can be physically interpreted as “aperture transmissivity” or as various spatial excitation intensities. At the bottom-center of Fig. 3.20, the phase shifts required to produce the desired radiation characteristics are shown.

3.8 Summary

This chapter presents an optimal system design from an information theory perspective. The intended 3D image reconstruction algorithms of lower physical complexity require a linearly well-behaved system that provides complete spatial

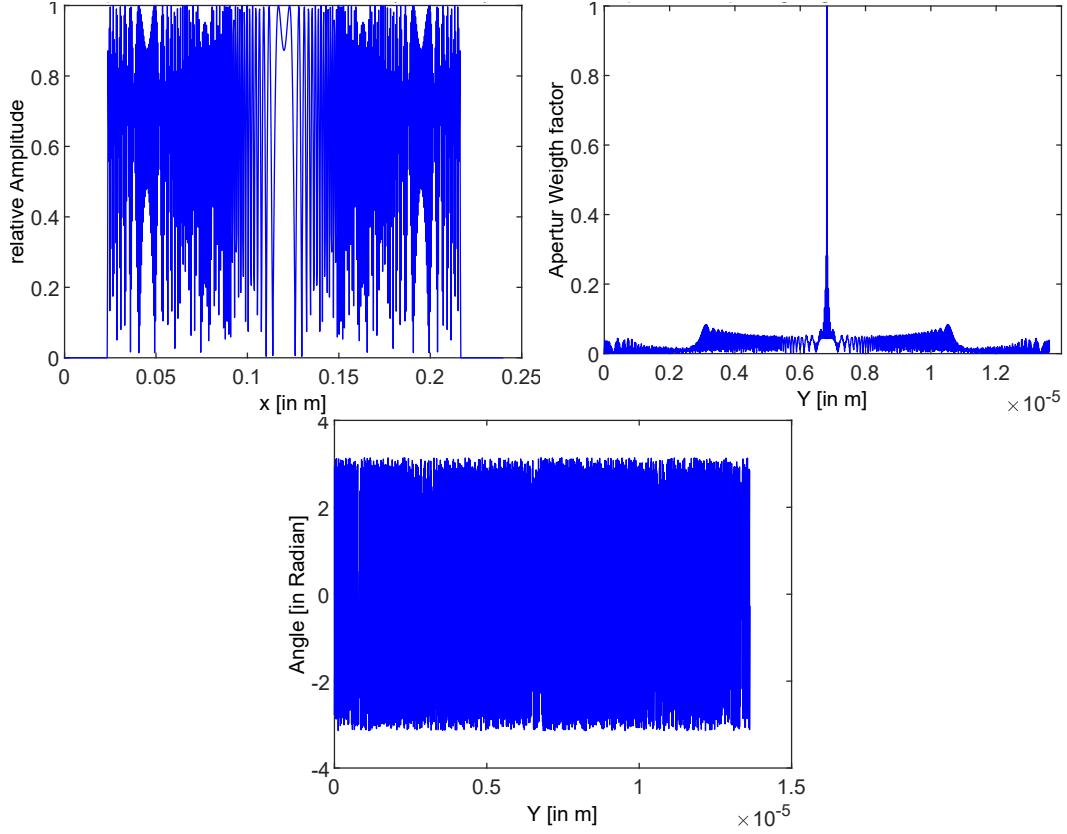


Figure 3.20: Inverse Fraunhofer diffraction for an opening angle of 45.2° : (top-left) x -dimension of a coplanar slice $U_i(x, y, z)$ for $Z = 0.096$ m (the values along y are identical to those along x); (top-right) the frequency- and distance-scaled inverse Fourier transform, interpretable as continuous aperture transmissivity for 2.5 MHz; (bottom-right) the phase shifts required to achieve the desired radiation characteristics for 2.5 MHz.

k-space coverage. This can be practically achieved through large axial and lateral bandwidths. Axial bandwidth corresponds to transducer bandwidth, whilst lateral bandwidth is achieved through a comprehensive set of multistatic emitter-receiver combinations covering the full angular spectrum.

Such ideal systems, whilst theoretically conceivable, are constrained by practical challenges and limitations. Practical trade-off options are explored in the following chapter.

4 Requirements for Next-Generation 3D USCT

After the information-theory based considerations presented in the previous chapters, we will now discuss application-based considerations. To this end we will compare the state of art in general as well as the status of the 3D USCT II device, marking the start of this thesis.

After the more general considerations, this chapter discusses the actual situation and vision of the KIT 3D USCT project at the begin of thesis.

The KIT USCT project went from its inception in 2001 through several significant device and prototype iterations, see Tab. 4.1. From a 2D SAFT concept evaluation device, over a 3D phantom evaluation device (3D USCT I) to a 3D device ready for testing humans (3D USCT II), see Fig. 4.1. The next step then envisioned was the 3D USCT III device, ready for a multicenter study with a statistical significant number of patients, paving the transition into the medical market.

4.1 3D USCT II Limits and Problems

From the pre-clinical study conducted at University Jena, several severe conclusions have been drawn [RGHZ15]. While ten patients were successfully scanned, several performance weaknesses were identified [RZD⁺13]. The assessment and review happend in collaboration with clinical Mammography specialists, see Fig. 4.1 right.

- The breast sizes of the patients were significantly larger and more variable than expected, and did not fit well into the assumed RoI, see Fig. 4.3.



Figure 4.1: (left) Jena Hospital study with 3D USCT II, system preparation and phantom test measurements at 2013-10-23. (right) Jena Hospital study with 3D USCT II: Review with Prof. Kaiser (Uni Jena), at KIT at 2013-12-04.

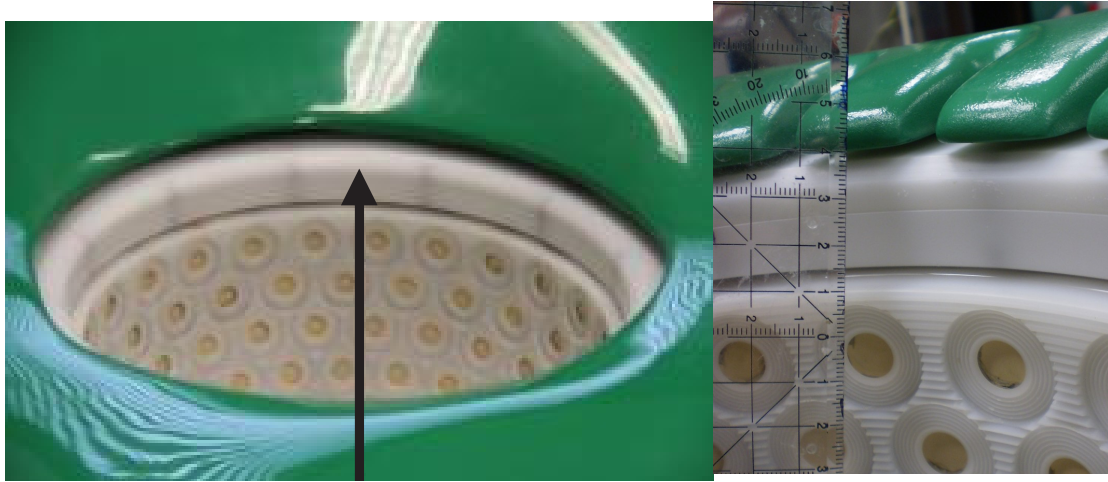
- Accurate positioning in the xy-plane as well as along the z-axis was substantially more challenging than anticipated.
- The measurement time was too long with *approx* 20–30 minutes, making that involuntary motions of the subjects were likely to happen. Moreover, it became a challenge and uncomfortable for the subject to lie still during the entire scan.

The solution to the first two issues is to increase the RoI from 0.1 m to 0.2 m in all three spatial directions. Especially the z-axis was already limited by and system inherent dead-space by 2.5 cm, see Fig.4.2. This enlarged RoI accommodates breast sizes up to approximately cup C while providing sufficient positional tolerance.

The last issue, measurement time and the associated risk of patient motion, must be addressed through technological improvements that enable faster acquisition. Possible strategies include increasing the number of parallel DAQ channels, accelerating the mechanical movement, or improving DAQ procedures to achieve higher SNR per measurement, thereby allowing a reduction in the required number of averages.

Table 4.1: Comparison of KIT 2D and 3D USCT systems. Performance and engineering metrics about the systems [Zap10, Sch08b].

Property	2D USCT	3D USCT I	3D USCT II
Measuring mode	2D	3D, 2D	3D, 2D, time resolved (4D)
Wave form	Cylindrical	Spherical	Spherical
Measurement operation	Semi automatic	Automatic	Automatic
Physical US transducers	16	32	157*13
Number of rotation positions	10^4	48	9
AScans	160,000	3,538,944	3,686,400
Parallel accumulated AScans	2	192	480
Measurement time (window 300 μ s)	768 s	11 s	2.3 s
Measuring time including averaging (16)	-	176 s	39 s
Readout speed	<1 MB/s	<3 MB/s	33 MB/s
Readout time including mech. adjustment	24 h	8 h	10 min
Data readout time (including compression)	-	1.5 – 4.5 h	2.2 min

**Figure 4.2:** (Left) the 3D USCT II aperture in the patient bed. Notable is the large dead space in Z from the ≈ 2.5 cm patient bed to the first transducer. (Right) detail view.

4.1.1 3D USCT II Evaluation: Measurement Performance

The overall measurement time per subject consists of (a) preparation time for proband and device, (b) the technical measurement time for left and right breast and (c) the after care for the subject and the device. The list of individual tasks

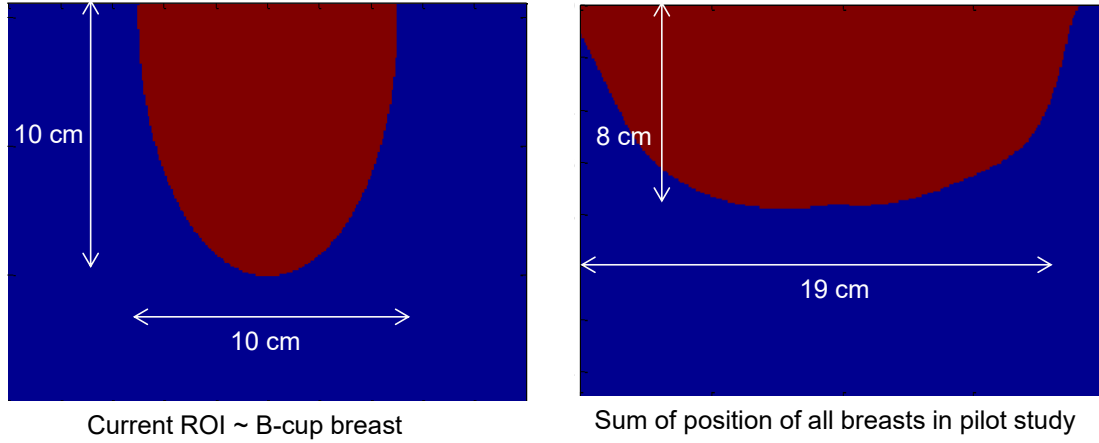


Figure 4.3: (Left) Assumed use case for 3D USCT II was a B cup breast fitting into an $0.1 \cdot 0.1 \cdot 0.1 \text{ m}^3$ sized RoI marked in red. (Right) averaged breast from the pilot study in Hospital Jena marked in red [RZD⁺13, RGHZ15].

Table 4.2: 3D USCT II: Overview of procedure steps and estimated time requirements

Step	Duration / Data
Device preparation and patient information talk	15–30 min
Breast positioning	2–5 min
Measurement time per breast (10 MPs · 51 s)	8.5 min
Proband after talk	5–10 min
Data serialisation and storage transfer data amount	10 MPs = 16.5 GB
Data transfer (16.5 GB@20 MB/s)	14.1 min
Technical device preparation for next proband (water / cleaning / heat-up)	15 min

in measurement procedure and their timings are given in Tab. 4.2 and the profiled measurement times per breast are given in Fig. 4.4

Overall this results with mean technical time measurement per breast of ≈ 8.5 minutes in an proband throughput of roughly one subject per hour.

The measurement time, where the subject had to lie down still, was described by many as bothersome, while being manageable. In the later image recreation step it was indeed found that many of the taken data sets are indeed noticable compromised by movement of the probands during the measurement procedure. This lead to significant blurring in the resulting reflectivity images, limiting resolution and constrast.

The effect on the image quality of the long measurement procedure is for exmaple

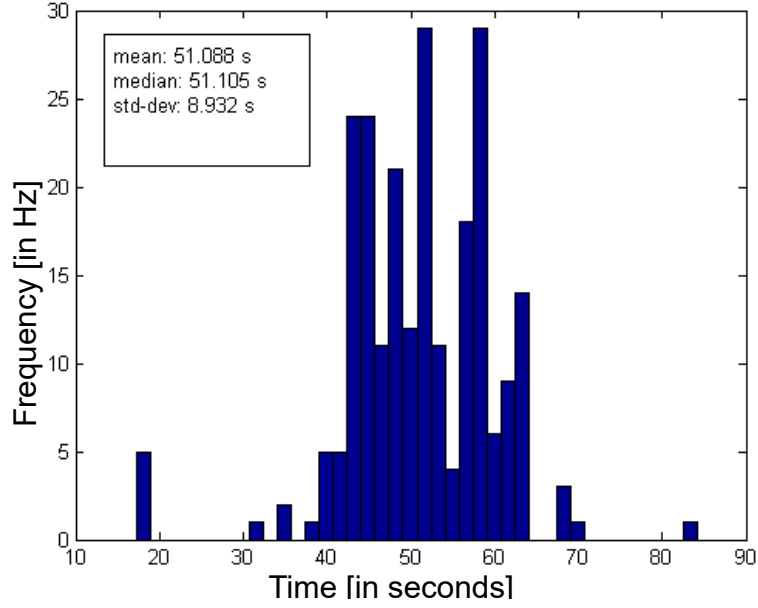


Figure 4.4: *3D USCT II measurement performance per motor position in Hospital Jena study as histogram from 236 measurement and motor positioning timings.*

visible in Fig. 4.5. This image shows a 2D reflectivity image slice of subject 16 (P16) from the Jena Hospital pilot study. The resulting image reconstruction shows artifacts due to inhomogenous illumination, grating lobes, and aliasing caused by sparse aperture and blurring because of subject movement. This will be discussed in more detail in Sec. 4.1.4.

Visible on top a sign the problem of the inhomogenous illumination can be shown as also the prominent visible grating lobes due to sparse aperture.

4.1.2 3D USCT II Evaluation: Accuracy and Reliability

In the built up of the mechanical system of 3D USCT II, the approach was to specify production tolerances tight enough to produce an accurate system which is capable of producing sharp images. The requirement is here the half wavelength criterium, which is challenging due to the nature of the system being a flexible scientific evaluation device with many degrees of freedom and therefore complexities and long error chains. Also, rapid prototyping, especially in the domain of transducer, led to unreliable components - notable the US transducer water proofing PU seals failing over time, see Fig. 4.6 left. Another source of failure was the

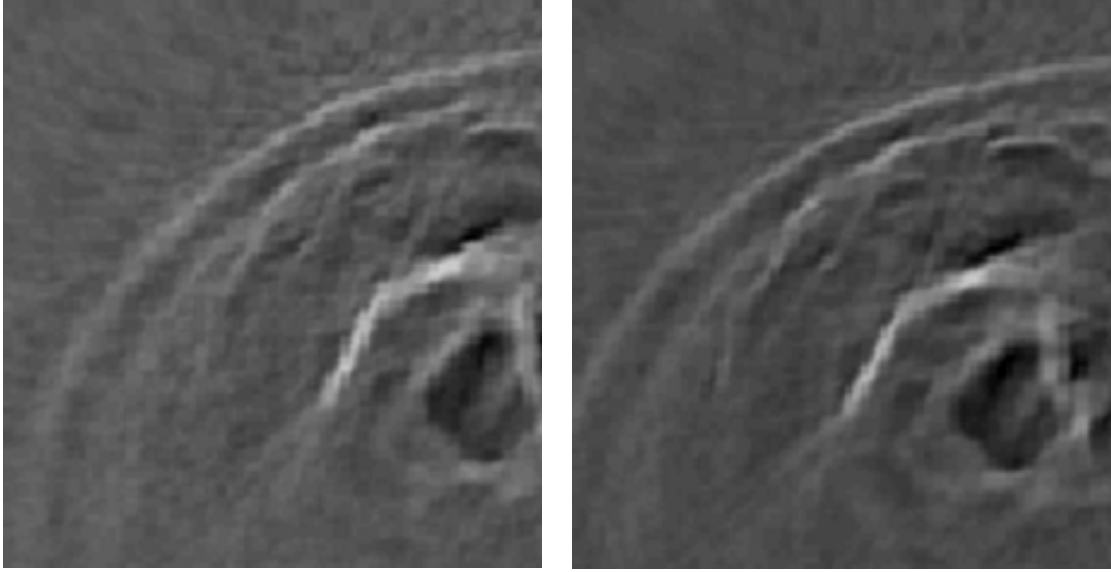


Figure 4.5: (Left) Jena reflectivity 2D slice image from proband 16 (P16) image is visible blurred due to subject movement. (Right) Jena reflectivity image P16 with a motion correction applied, the image becomes sharper and has increase contrast and SNR [MEG⁺ 16].

complex internal water system with many pipes, pumps, valves, seals, which failed on the start of the Jena study and needed to be repaired in an improvised way, see Fig. 4.6 right.

An overview of the reliability and spare situation of 3D USCT II is shown in Tab. 4.3.

For the next system, simplifications and in-system test and calibration capabilities instead of unrealistic production tolerance specifications should be put on the fore-front of the design priorities. For instance, as a result the water preparation system is planned to be completely externalised.

4.1.3 3D USCT II Evaluation: Bandwidth

The 3D USCT II system's bandwidth is defined by several frequency bandwidth limiting factors. The three main factors are: (a) the analogue response of the transducers, (b) the analog electronic chain and especially the electronical band-pass filter for anti-aliasing filtering (AAF), and (c) finally the digital stage AAF, This last stage digital lowpass filtering by a bandpass downsampling scheme, keeps

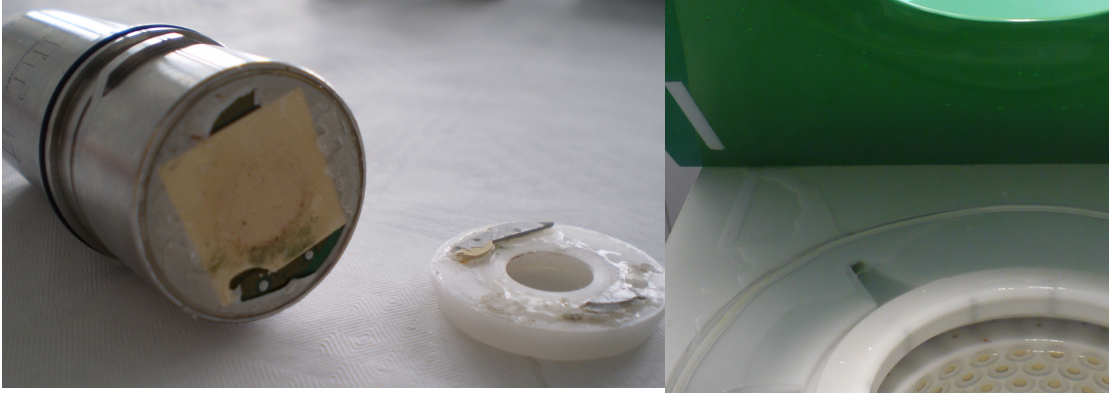


Figure 4.6: (Right) 3D USCT II TAS leaking due to sealing material failure. PU filling material failed to stick cohesive to the TMM4 ceramic. (Left) Jena Study water seal leak, water visible on top of the white patient bed beyond the seal.

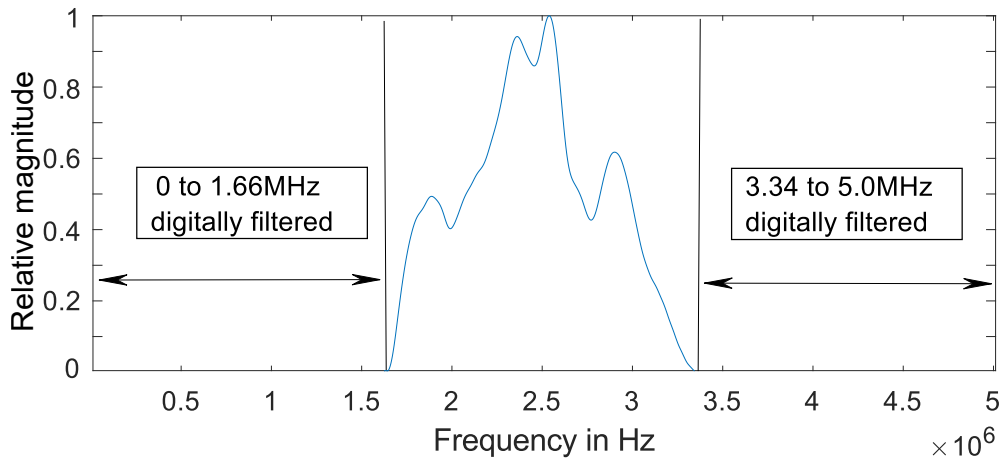


Figure 4.7: Extracted frequency response for the 3D USCT II from University Jena measurement for patient 13 from the transmission pulse.

only frequencies between 1.66 MHz and 3.3 MHz. This was motivated by the need for an additional factor three in data volume reduction and 3.3 MHz sampling rate data format, required for being able to store and transfer the data fast enough, see Fig. 4.7. For next-generation USCTs, the goals are to increase the analog signal chain bandwidth to be nearer to a flat-spectrum response as well as removing the transfer speed constraints. The latter goal will lead to a less restrictive data reduction process and storage format.

Table 4.3: *3D USCT II Component Inventory and Risk Assessment.*

Component	Quantity	Spares	Problem/Failure Risk (empirical)
Motion Mechanics	1	0	medium
Aperture	1	0	none
PrPMC Modules	1	2 (200%)	medium
Crate	1	0	low
SLT	1	1 (100%)	none
CtrlBoard	1	1 (100%)	none
FLTs	20	2–4 (10–20%)	low/medium
TAS	157	5–10 (3–6%)	high (≈ 20 failures)
Motor Control	2	1 (50%)	low
Motors	2	0	none
OS/HDD/Server	1	2 (200%)	medium

4.1.4 3D USCT II Evaluation: Imaging Quality

The reflectivity image displayed Fig. 4.5 is obtained during the pilot study at the Jena Hospital. The image from subject P16 reveals several of the encountered problems: inhomogenous illumination, prominent visible grating lobes due to the sparse aperture and the movement of the test subject. As result of these problems the image has become blurred.

Isotropy of the imaging illumination: surrounding aperture

A commonly voiced feedback by medical practioners about classical sonography are shadow effects in the image. This is the result of an inhomogenous illumination over the desired imaging domain, the RoI. The attenuating foreground with strongly reflecting objects might obscure background objects. The effect is especially strong for conventional sonography systems as they illuminate the object only from one side with US. Here significant improvement should be achieved with a next-generation device.

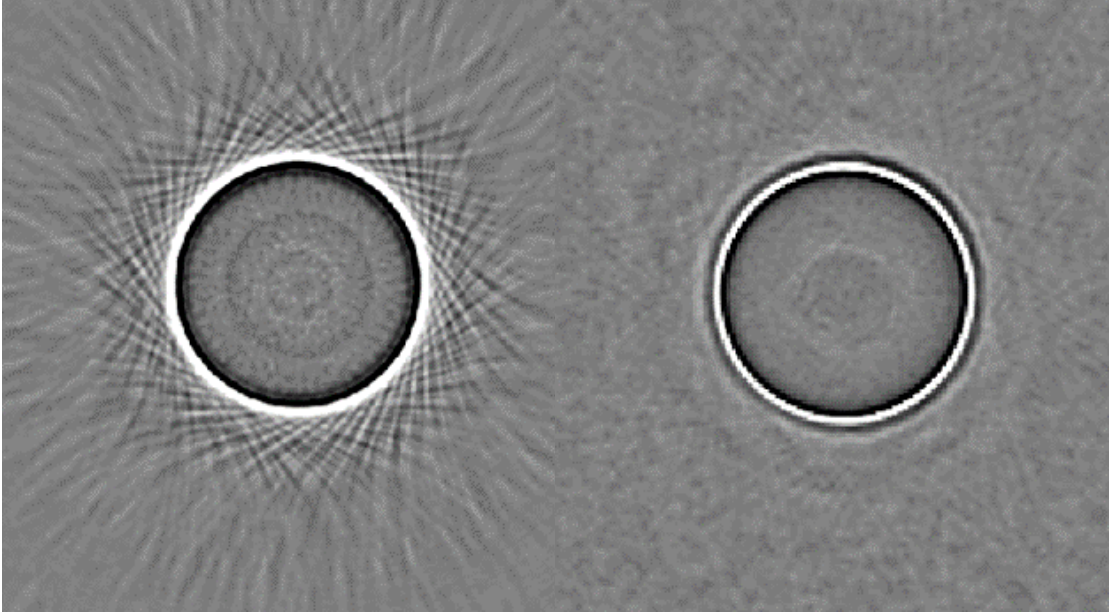


Figure 4.8: *Metal sphere imaging response: Comparison of irregular versus regular spatial sampling with sparse arrays based on 3D USCT II data, impact on reflectivity imaging with SAFT. (Left) 3D USCT II with regular aperture, (right) 3D USCT II with irregular aperture (emulated from regular data) [RZHG25, RZHG13].*

Sparse aperture and grating lobes

A sparse aperture inherently leads to grating-lobe artifacts, as illustrated in Fig. 4.8. The SAFT image was created from a subset of 3D USCT II data with to mimic a possible future improved aperture. For a fair comparison with the regular, but likewise sparse, 3D USCT II aperture, a comparable number of AScans was selected for the evaluation: 3D USCT II: 5,324,184 AScans (6 MP) regular, 3D USCT III emulation: 5,308,416 AScans (irregular selected subset from 12 MP), corresponding to 99.7% of the original data.

An encouraging reduction of grating-lobe-related artifacts can be observed in Fig. 4.8 (right). Smearing out the aliasing effect introduced by sparse apertures leads to a more diffuse background artifact pattern and leads to improved image quality. The image SNR, defined as the maximum background amplitude divided by the maximum sphere amplitude, i.e.

$$\text{SNR}_{\text{img}} = \frac{\max(B)}{\max(S)},$$

yielded 30.3% for 3D USCT II, while for the 3D USCT III the SNR value was reduced by half to 15.4%.

4.2 Design Principles for Next-Generation 3D USCT

The challenges observed in the second-generation 3D USCT II system were diverse and substantial, motivating a step back and a more fundamental reconsideration rather than an incremental refinement for next-generation USCT.

Building on the system-level perspective outlined in the previous chapters, the information-extraction process is first examined on an abstract level before addressing specific design decisions in the next chapter. Data acquisition in USCT can be regarded as the controlled excitation of a physical system comprising the medium, the hardware components, and the object. The resulting wave response encodes the relevant object information.

Image reconstruction is the inversion of the underlying wave-propagation process. Such inversions are computationally intensive and often ill-posed, and a perfect recovery of complex 3D structures is rarely achievable in practice. This has been experienced in earlier KIT 3D USCT systems, where the cubic scaling of computational effort with voxel resolution [Har12] quickly reached practical limits.

Although computational performance increased rapidly for several decades, Moore’s law is nearing its end, and memory-bandwidth limitations—commonly referred to as the “memory wall”—have become a central performance bottleneck. These constraints affect CPUs, GPUs, and distributed architectures alike. While quantum computing and AI-based inversion algorithms are promising future technologies, they are not yet available for large-scale USCT. Consequently, complexity reduction at the system-design level remains essential.

A well-established simplification is the use of ray-based or projection-based imaging methods, as adopted in the KIT USCT project [RZHG18]. These approaches require stable and reproducible system behaviour and benefit from approximate linearity, which greatly simplifies modelling and reconstruction.

This leads to a simplified representation of the physical wavefield (Fig. 2.1), linking

US propagation to the spatial-frequency content of the object (Fig. 3.7). The imaging chain can thus be described as a linear system that transports 3D spatial information through broadband 1D ultrasound data. Fourier optics, introduced in the previous chapter, provides the theoretical framework for this viewpoint.

From this perspective, several system properties become desirable: broadband linearity across all components, temporal and spatial stability, and well-defined spherical wavefields that reliably encode object information. The resulting multi-frequency paradigm extends and generalises earlier centre-frequency-oriented system designs, enabling more robust utilisation of the available information bandwidth.

4.2.1 Ideal Next-Generation 3D USCT

From previous considerations, the idealised wave-model assumptions can be summarised as follows:

- constant sound speed in a homogeneous, non-attenuating, non-dispersive medium,
- negligible multiple reflections under a weak-scattering assumption,
- far-field wave behaviour instead of near-field or higher-order effects,
- utilisation of a multi-frequency approach.

4.2.2 Multi Imaging-Modalities with US

We aim for a multi imaging-modalities created by a single US device. This in contrast to conventional sonography systems that process and integrate signals directly in hardware through phased-array beamforming resulting only in the possibility of one imaging modality - e.g. reflectivity or transmission imaging but not both at the same time.

Although phased-array approaches enable fast, real-time reflectivity imaging, the resulting image quality, and in particular the diagnostic specificity, is often limited.

A promising strategy is therefore to combine the advantages of reflectivity-based US with transmission-based parameters. This strategy is known to enhance the diagnostic specificity in breast cancer imaging as it allows for obtaining sound-speed and attenuation images simultaneously [GFI81].

This concept can be realised with an US device capable of providing multiple image modalities from a single acquisition. From a system-design perspective, this corresponds to the “computed” aspect of USCT: the system delivers all modalities in one device and one measurement procedure. This is achieved by decoupling the measurement phase from the imaging phase, doing the image reconstruction off-line and not in real-time, and avoiding premature data reduction so that the full raw dataset remains available for advanced reconstruction. With this approach, different reconstruction methods can be applied to the same raw data. The raw dataset serves as the common basis linking all imaging algorithms and enables simultaneous extraction of multiple modalities.

This vision should be implemented in a non-real-time imaging system using a synthetic-aperture focusing acquisition strategy.

4.2.3 Clinical Applicability

An additional key requirement for the next-generation 3D USCT system is clinical applicability. Beyond technical prototyping, the system must be sufficiently robust, flexible, and user-friendly for deployment in a clinical environment, considering the economic constraints of healthcare, established radiology workflows, patient needs, and the requirements of practising radiologists who must be able to rely on 3D USCT as a functional diagnostic tool.

With the envisioned 3D USCT III system, this objective should become attainable. A multicenter study will serve as the final cornerstone, demonstrating the technical and medical viability of the method as a clinical application and indicating its readiness for clinical adoption and eventual market adoption, see Fig. 4.9.

The clinical applicability of the system, defined as technology readiness level should be increased significantly. While TRL 6 to 7 was shown when the applicability was demonstrated with 3D USCT II in a relevant environment (University

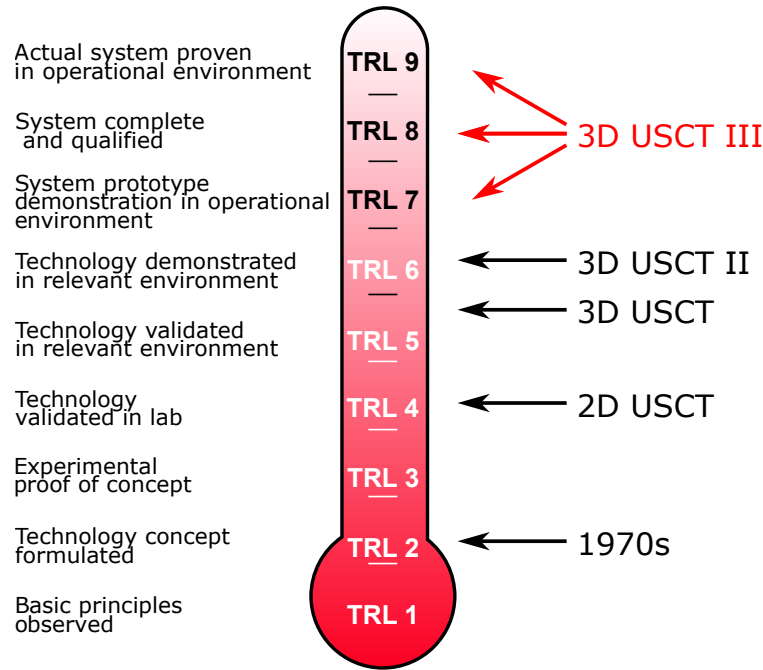


Figure 4.9: Technology readiness level adapted for 3D USCT (2015): according to the EU Horizon 2020 definition, from Wikipedia [Com14, Hé17, Har].

Jena) it should be increased TRL 8, a device with the potential for full medical regulatory approval and normative compliance.

For 10 Million AScans gathered for a sufficient image quality, the measurement time should be reduced from 20 minutes to below 3 minutes. Approach is the reduction of time-costly required mechanical repositions, compensating with improved DAQ electronics and measurement procedure.

The grating lobe artifacts prominent arising in reflectivity imaging due to the sparse and regular US aperture spatial sampling should be minimized. This should be achieved by innovating an irregular sampling approach for new US transducers.

The illumination homogeneity and subject placement reliability should be increased, leading to a bigger RoI of $20 \times 20 \times 20 \text{ cm}^3$. An adapted aperture and transducer design are the way forward here.

Table 4.4: Comparison of ultrasound imaging systems and KIT USCT systems (status 2014).

System	Resolution	Aperture geometry	Imaging principle	Bandwidth	Meas. time
Sonography	anisotropic (3D)	single-sided	integrated real-time beamforming	narrowband	real-time
Delphinus USCT [DLR ⁺ 14]	isotropic (2D)	2D surrounding ring	decoupled tomographic reconstruction	medium	minutes
KIT 3D USCT II	isotropic (3D)	regular sparse 3D surrounding array	decoupled synthetic aperture focusing	medium	> 10 min
KIT 3D USCT III	isotropic (3D)	irregular sparse 3D surrounding array	decoupled synthetic aperture focusing	broadband	< 3 min

4.3 Summary of Chapter

The summary of Chapter 4 can be formulated as a set of high-level system requirements, which can be compared with competing systems see Tab. 4.4. These requirements follow from the first “proper” hospital study conducted with the 3D USCT II device at University Jena [RGHZ15].

1. The technological readiness level (TRL) is too low. Currently, it is at TRL 6 and needs to be TRL 7-9 to become clinical viable.
2. The illumination of the RoI should become more homogeneous.
3. The size of the RoI should increase to 0.2 m in each direction and the breast should be positioned deeper and more centrally in this RoI.
4. The number of sensors should increase and distributed irregular along the surface to prevent the aliasing effects currently visible in the reflectivity images. As a result, the number of AScans will increase towards 10^7 , meanwhile the scanning time needs to be below 3 min.

5 System design for KIT 3D USCT

III

This chapter describes the core aspects considered during the design of the 3D USCT III device. One of the open questions is related to the spatial distribution of the transducer elements around the RoI, which was extensively analysed for the design of the 3D USCT II device [Sch08b]. In this chapter, a similar SAFT imaging simulation approach together with general design consideration based on fourier-optics are presented. Typical metrics for characterizing the image reconstructions are resolution, contrast, and homogeneity of illumination. Since resolution is already addressed in previous works and chapters, the focus here will be on the metrics contrast and illumination.

5.1 Homogeneity of Image Illumination

5.1.1 Isotropic Illumination Provided by a Surrounding Aperture

As discussed in previous Sec. 4.2.3 a key finding of the Hospital Jena study was that breasts' variability in size and placement needs to be addressed by an increased RoI. A RoI of $20 \times 20 \times 15 \text{ cm}^3$ ($\approx$ hemispherical) or $20 \times 20 \times 20 \text{ cm}^3$ (rectangular) is needed to accommodate the targeted breast cup size B, even under suboptimal positioning and with buoyancy-related deformation effects.

Additionally, a commonly voiced feedback by medical practitioners about classical sonography are shadowing effects in the resulting images. These effects are

amongst others due to inhomogenous illumination of the RoI. This effect is especially strong for conventional sonography systems which illuminate the object only from one side. Illuminating by all sides with an surrounding aperture will reduce this effect. A potential implementations of a surrounding aperture might be e.g. a circular array or a (hemi-)spherical aperture.

5.1.2 Aperture Dimensionality: 2D, 2.5D, or 3D Aperture

In general, as laid out in earlier chapters, there are 2D, 2.5D and 3D approaches to US imaging systems. 2.5D system may be obtained by moving a circular system along its central axis. The KIT 3D USCT project decided early on for a real 3D approach with the core argumentation that 2.5D systems are less capable of representing all kind of general structures. Out-of-plane reflections are undetectable with a 2.5D system but can be resolved by fully three-dimensional systems. An extreme example are “stealth structures”; these structures reflect energy exclusively out of plane and at large reflection angles away from the emitter direction, rendering them effectively invisible in a 2.5D system.

The second argument is that, with improvements in electronics and DAQ, early data reduction is no longer required, specifically the integrated phased-array step for emission and reception used in sonography. In a SAFT measurement principle device, this approach can be replaced by purely software-based focusing algorithms, such as ellipsoidal backprojection from the family of SAFT imaging algorithms. This means that part of the SNR gain can be shifted from the 1D signal domain to the 2D and 3D imaging domain. The downside is that the number of collected reflectivity waves per AScan increases, as they are now collected over a 3D volume instead of a 2D slice, leading to a much larger number of reflections (“information density”). For a complex object such as the human breast, this may lead to reflections overlapping with each other, which are nearly impossible to separate at the signal level with limited bandwidth. Due to this ambiguity, the increased AScan information content must be resolved and separated in the 3D image reconstruction domain.

Clearly, starting with planar transducer for 1D, followed by a moving circular array

for 2.5D, would mean a spherical transducer for 3D. However, as this is physically not feasible for the breast, a hemispherical array is the maximum achievable.

5.1.3 Size of the Aperture

The 3D USCT project aims for improved illumination over classical US systems by using a surrounding 3D hemispherical aperture. The optimal selection of the remaining degree of freedom of such hemispherical aperture, i.e. the diameter, is depending on the imaging approach. Far-field methods like ellipsoid sum-and-delay and backprojecting methods improve with growing diameter in homogeneity in their imaging representation [Sch08b]. A smaller diameter affects key imaging properties such as illumination, contrast and resolutions alike in an anisotropic way.

Other methods, not relying on far-field homogeneity like full-waveform inversion (FWI) methods can perform well with smaller apertures, putting the RoI inside or close-by the near-field.

As the 3D USCT project relies on the far-field ellipsoide SAFT imaging method for reflectivity, the diameter is a trade-off between imaging homogeneity and signal attenuation.

5.1.4 Optimal Aperture Diameter

One way of selecting the optimal diameter is by considering the extremes.

An aperture with a minimal diameter, just covering the RoI aperture in the far-field, would have an angular homogenous radiation pattern but not an isotropic response over an extended RoI. Increasing the diameter further increases the isotropy in the RoI, approaching for an infinite diameter full isotropy and homogeneity in the whole RoI as the spherical wave transforms to a plane wave. The spherical aperture would converge for an infinite diameter to a plane-wave imaging system with all its preferable properties. Yet, the resulting signal amplitude and SNR would approach zero, due to geometrical attenuation of growing distance.

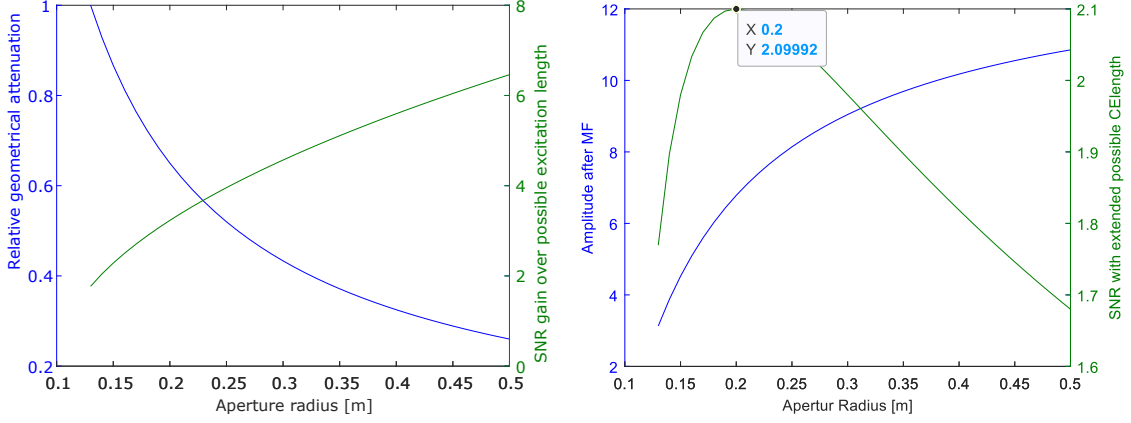


Figure 5.1: (Left), Blue curve represents the geometrical attenuation. Green curve depicts the gain in SNR due to the extension of the coded excitation length. Normalized for 3D USCT II radius 0.13 m. (Right), SNR gain comprised out of CE length extension combined with geometrical attenuation over the increase in radius of an aperture. The optimum for both together is at a radius of 0.2 m.

Meanwhile, a spherical aperture with minimal diameter, would show minimal geometrical and medium attenuation losses but would be maximally inhomogeneous and an-isotrop over the RoI with its imaging properties.

Therefore, the far-field condition demanding diameter is a necessary but insufficient criterion. Ideally, the aperture diameter is approaching the plane-wave as much condition as technically feasible given by the trade-off between RoI isotropy increase and sound-pressure drop.

Another way of thinking about this is considering the bistatic angles for every position in the RoI in dependence of the aperture diameter. The resolution-optimal angular sampling and the bistatic angle changes anisotrop over the RoI. For instance, for an 20 cm RoI, 3.5 MHz and 15° emitter-receiver in-between angle, the resolution drops factor five for a reflector on the RoI interface as object worst case and aperture best case, see Fig 5.2. Also this perspective motivates some aperture to RoI distance.

Another way to select a good trade-off is by considering the possible coded excitation (CE) length. This length is limited by the travel time before a sent excitation train or reflection might interfere with the next excitation. From Fig. 5.1 right, it can be seen that the optimum is at a radius of 0.2 m, which is considerable larger than the 0.13 m of the 3D USCT II device.

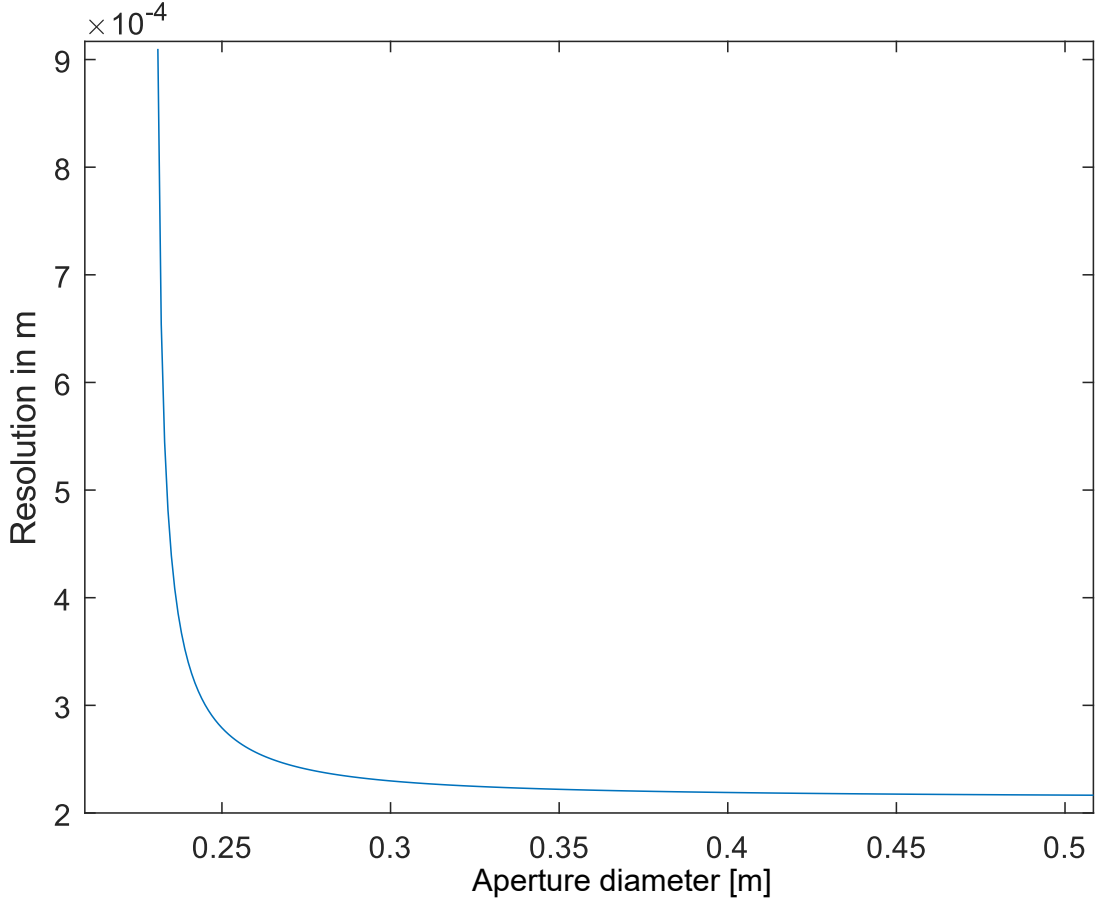


Figure 5.2: *Best-case resolution over aperture diameter, for a spherical RoI of 20 cm diameter, a 3.5 MHz frequency and an in-between transducer angle of 15°. Reflection point selected as worst-case (nearest distance) from nearest RoI interface to the emitter-receiver transducers.*

5.1.5 Aperture: Ratio of Emitting and Receiving Transducers

Two other design parameters for the 3D USCT III device are the number of elements and the emitter-receiver ratio. For the 3D USCT II, this ratio is four to nine. In addition, the emitters and receivers are fully separated from each other and wired via separate and specialised electrical excitation (TX) and reception (RX) lines. These lines are specifically designed for the US transducers and the electric requirements (load, impedance). For the TX and RX channel multiplexing, off-the-shelf components are used. The opposite design philosophy would be a unified transducer design where each element can act both as an emitter and a receiver. However, this choice may give problems related to the complexity of the

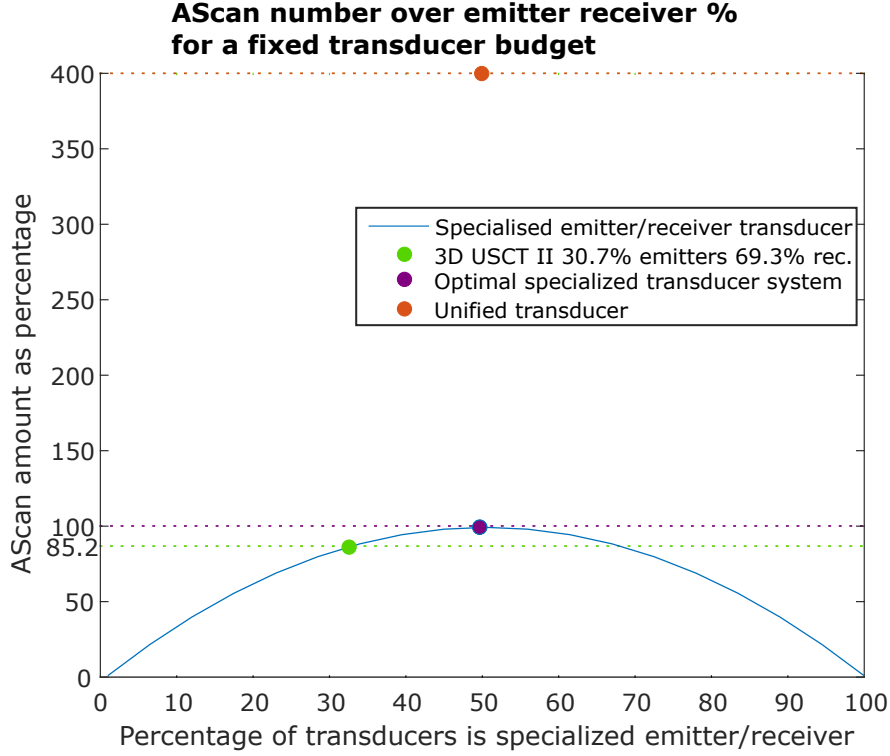


Figure 5.3: *Relative AScan amount per emitter-receiver percentage of a fixed transducer budget. X-axis percentage of the transducer budget is specialized as emitter/receiver, y-axis AScan amount as percentage, normalized to the optimal specialized transducer case with 50% emitter/receiver. (Blue graph) specialized emitter-receiver percentage design space. (Green dot) 3D USCT II case with 30.7% emitters. (Purple dot) optimal specialized transducer design with 50% emitter and 50% receiver. (Red dot) unified transducer design.*

electronics, such as matching electronics and electronic cross talk. On the other hand, the choice of a unified transducer design will lead to an increase in the number of resulting AScons for a given number of transducers n , i.e.

$$\begin{aligned} n\text{AScons}_{\max, \text{specialised}} &= \left(\frac{n}{2}\right)^2 = \frac{1}{4}n^2, \\ n\text{AScons}_{\max, \text{unified}} &= n^2. \end{aligned} \tag{5.1}$$

Also, if the ratio of emitter to receiver elements deviates from the balanced 50% to 50% percentage, the number of AScons generated per transducer decreases even further, see the left and right extremes in Fig. 5.3 of the blue curve. In green is the 3D USCT II case with an unbalanced emitter to receiver percentage of 30.7%

to 69.3% given, leading to only 85.2% of AScans. A balanced emitter-receiver percentage design for a 3D USCT III system would lead to an increase of AScans of 14.8%. But, given as red dot, a unified transducer design, would lead to an increase of available AScan amount of 400% for the fixed transducer budget.

Given that the 3D USCT III still operates in a sparse-aperture regime, additional AScans, and the information gain associated with them, are highly beneficial. Therefore, the unified transducer design has been selected despite the increase in system complexity. A key enabler for this choice is the next-generation front-end architecture, which goes beyond commercially available components. A custom IPE designed ASIC (Application-Specific Integrated Circuit) for a more integrated front-end design has been developed to support this change. This will be discussed in more detail in Sec. 6.5.1.

5.1.6 DAQ System Design: Maximising Number of AScans vs Minimising Measurement Time

The measurement performance of the 3D USCT III device is primarily determined by two factors: (i) the ability to produce a high number of (unique) AScans (n_{AScans}), and (ii) achieve this within the shortest possible measurement time (T_{total}). An optimum for these parameters arises partly from physical constraints such as the size of the RoI, and partly from technical limitations. The technical limitations may be related to the emitter-receiver ratio, number of TAS, number of DAQ channels, channel muxing, etc.

The measurement time (T_{total}) can be derived from geometrical considerations the size of the aperture, RoI sizes, and soundspeed considerations.

First, we have a measurement window time t_m which is given by the wave's travel time from the emitter through the measurement container and RoI to the receiver. This covers as maximum time length reflection pathes from the center of the RoI. To cover any potential reflection from the RoI, the travel-path extends to the back-side of the RoI and back to the receiver. But, there is a even longer reflection path inside the aperture. It is the path from the emitter to the furthest distanted aperture wall and back to the receiver. For a (hemi)spherical aperture this longest

travel path time is $t_f = t_{shot}$. While it carries no primary object information it needs to be considered in the overall measurement process as reflections from the back-wall might interfere with the next excitation and its measurement window t_m . With the diameter from the previous extracted optimum (see Fig. 5.1 right) $d = 2 \cdot 0.2 \text{ m}$ we get $t_f = \frac{2 \times d}{c}$.

We define now a extended measurement window time as shot time $t_{shot} = t_f$. Next, the soundspeed c is conservatively assumed as the slowest tissue soundspeed in breast tissue; the lower end is assumed with of fatty tissue of 1420 m/s [doi00]. Hence, the total time for t_{shot} is

$$t_{shot} = \frac{2 \times d}{c} = \frac{2 \times (2 \times 0.2 \text{ m})}{1420} \approx 600 \mu\text{s}. \quad (5.2)$$

with d the diameter of the aperture.

The overall time T_{total} scales with number of ASans and t_{shot} . In the employed SAFT imaging method, all emitter transducers are permuted with all receiver transducers, resulting in the total number of measured unique ASans

$$n_{ASans} = n_{Emitter} \times n_{Receiver}, \quad (5.3)$$

with $n_{Emitter}$ the number of emitters and $n_{Receiver}$ the number of receivers.

However, for the total measurement time, the number of emitters primarily determines the performance because US signals can be received in parallel without interference but cannot be sent in parallel.

Thus, when there are enough parallel DAQ channels the total measurement time for the complete iterative SAFT measurement process is given by:

$$T_{total} = n_{Emitter} \times t_{shot}, \quad (5.4)$$

where the number of $n_{Receiver}$ plays no role. Technically, this time optimum is limited by the number of available DAQ channels, often not sufficient for a 1:1 connection between each $n_{Receiver}$ and an unique DAQ channel.

Temporal multiplexing of DAQ channels can compensate for insufficient chan-

nels (DAQ channels $< n_{\text{Receiver}}$). However, this comes with a price as it requires $n_{\text{multiplex}}$, viz.

$$n_{\text{Multiplex}} = \lceil \frac{n_{\text{Receiver}}}{\text{DAQ Channels}} \rceil \quad (5.5)$$

$$T_{\text{total}} = n_{\text{Emitter}} \times n_{\text{Multiplex}} \times \text{Measurement time}. \quad (5.6)$$

Therefore the DAQ parameters which needs to be considered for optimal performance are emitter to receiver ratio, the number of Receiver DAQ channels, number of TAS, and $n_{\text{multiplex}}$. The number of physical receiver DAQ channels is crucial and also a design limitation. In the 3D USCT II device the 480 receiver DAQ channels are roughly contributing to 90% of the power consumption of the whole device and a large part of the hardware costs. Therefore, the reception channel multiplexing is important for reducing the number reception channels to a technical and economical feasible number. Finding a good balance between the desired minimal measurement time and optimal AScan number is crucial. For finding an optimal solution, an optimisation criterion is required. A metric as the deviation from the desired AScans number with the AScans per second measurement speed are equally weighted. Additional challenges and overhead arises from rounding up to an integer number of multiplexing cycles. Clustering transducer elements into TAS adds further overhead and will impact the total measurement time. The ratio of the number of emitters to receivers is also very import. In a configuration with unified transducers, i.e. elements that function both as emitter and receiver, the total number of AScans is always maximised. However, such an architecture also removes one degree of freedom that may enable alternative, potentially superior, optimisation strategies. Therefore we consider these options in the optimization as well.

To explore the resulting design space, we begin with the status quo: the simulation of the one million AScan DAQ system configuration of the 3D USCT II system, see Fig. 5.4. The relevant system parameters are an aperture diameter of 26 cm, 480 DAQ channels, and a target number of 1,041,420 AScans, which corresponds to the optimum at a TAS number of 157 and 13 elements per TAS. Along the diagonal of the optimisation map, the unified transducer configuration appears, where each element serves both as emitter and receiver. This arrangement inherently improves AScan efficiency by reducing the required number of TAS positions.

The quality metric used in Fig. 5.4 (e) equally weights the AScans per second and the total achievable number of AScans (c and d).

The hypothetical optimum for the 3D USCT III system lies at a 7:6 emitter-receiver ratio (MUX 2). At the same measurement speed per AScan, this allows for 15% more AScans compared to the current MUX 3 configuration with a 4:9 emitter-receiver ratio. The global optimum would be achieved with 17 TAS elements (9:8 emitter-receiver ratio, MUX 2) with a slightly higher measurement speed. The analysis shows that the maximum number of AScans occurs near the configuration in which the number of emitters equals the number of receivers, which is consistent with the optimum for the unified-transducer approach. At the same time, the measurement time per AScan varies significantly across the design space, by roughly 30%. Distinct “sweet-spot” configurations exist that provide better measurement-time efficiency than the unified-transducer setup. When evaluated using the combined quality metric, these sweet-spot configurations achieve improvements of up to 20% compared with the unified-transducer baseline. This shows that fine tuning of the parameters does matter for the design of a new system.

Interestingly, while having fewer sweet-spots than the specialised transducer solution, missing sweetspots is more expensive. In general the unified transducer design performs better, but in comparison, the singular global optimum is still with the specialised transducer approach (5% better).

As 3D USCT III device design moves to a new DAQ architecture from the 3D USCT II DAQ hardware, a table with optimal system design parameters over the available degree of freedom DAQ channels is presented in Tab. 5.1.

Conclusion for DAQ parameters for 3D USCT III

Based on the previous analysis, the following conclusions can be drawn.

- With the new system, the number of AScans should be increased by a factor of about 10, making the measurement time an even bigger challenge. Therefore, the optimal selection of design parameters becomes even more critical compared to the current design, where the measurement time is primarily

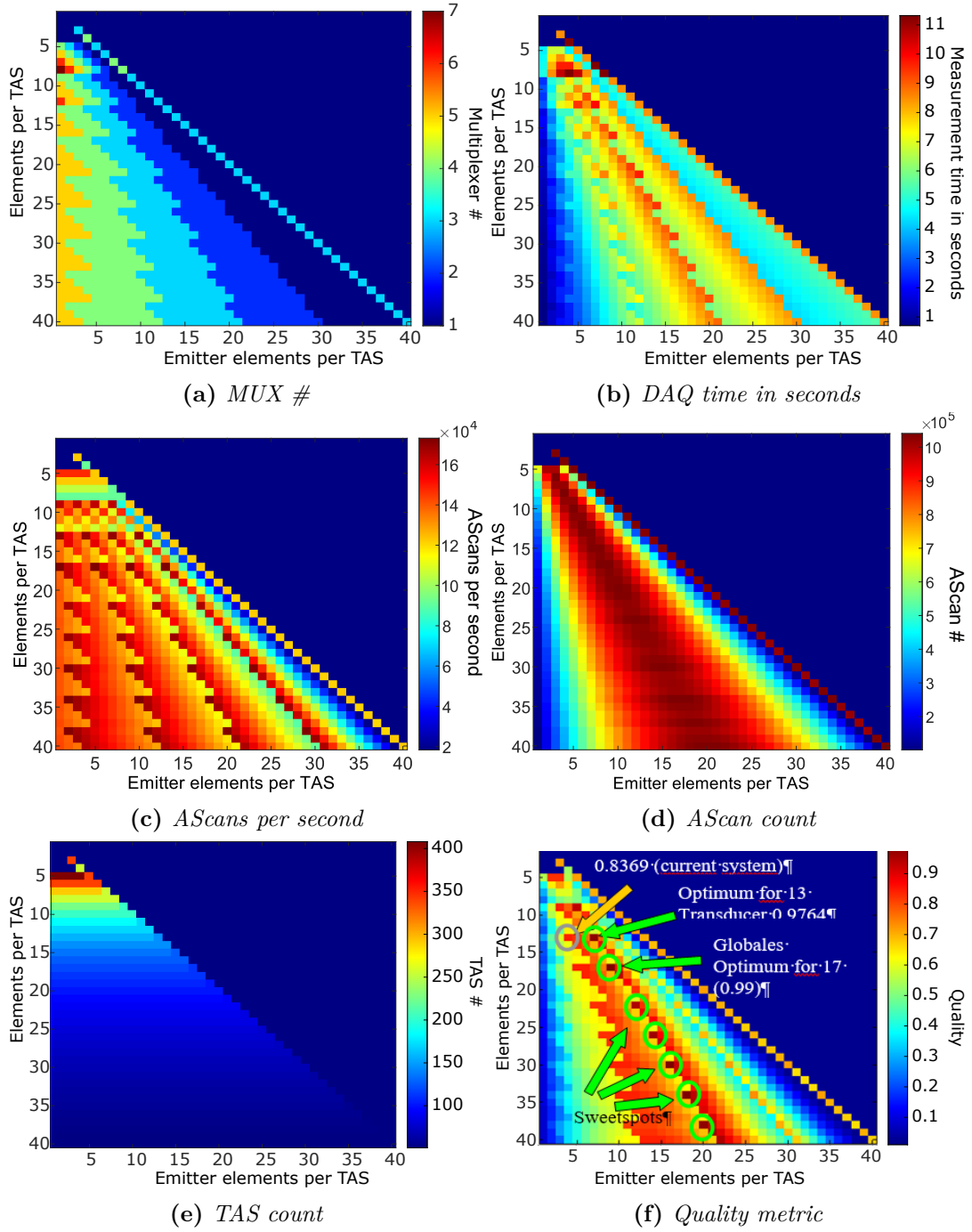


Figure 5.4: Optimisation specialised transducers for 480 DAQ channels (3D USCT II). Aim $10 \cdot 10^6$ ASans with 10 motor positions in transducer. Quality (graph e): equally weighted ASans/sec and number of ASans (graph c and d)

Table 5.1: *Unified transducer optimisation. Global sweetspots over the DAQ channel number, for $10 \cdot 10^6$ AScans, diameter 44 cm, filtered for reasonable TAS diameter, average 4, E-R transducer sweetspot ≥ 0.999 quality. Universal design sweetspots are a subset of E-R design sweetspots.*

DAQ Channel#	107	124	129	150	215	239	248	300	403	410	450
DAQ time [s]	452	393	378	332	227	211	197	166	121	121	121
AScans [10^6]	10.30	10.39	10.40	10.32	10.40	10.32	10.39	10.39	10.34	10.34	10.37
AScans/s [10^3]	22.8	26.4	27.5	31.1	45.8	48.9	52.8	62.4	85.7	85.7	85.8
MUX	30	26	25	22	15	14	13	11	8	8	8
DAQ channel/TAS	1	1	1	1	1	2	2	1	3	3	3
Shots [10^3]	96.3	83.8	80.6	70.7	48.4	45.0	41.9	35.5	25.7	25.7	25.8
TAS #	107	124	129	146	215	119	124	293	134	134	140
TAS d [cm]	5.21	4.84	4.74	4.46	3.68	4.94	4.84	3.15	4.66	4.66	4.55
E/R Ratio	1	1	1	1	1	1	1	1	1	1	1
Transducer elements	30	26	25	22	15	27	26	11	24	24	23
Quality [%]	100	100	100	96.55	100	95.94	99.94	97.55	99.94	99.94	89.58

limited by mechanical movement.

- A suboptimal choice of parameters could lead to a measurement time increase of up to 50%.
- A universal transducer approach allows for either the number of TAS or the number of transducers to be reduced to 50%. However, this usually results in slightly missing the optimum measurement time. The additional degree of freedom provided by a specialised emitter-receiver approach offers a small increase in efficiency.
- The optimum is usually near the point where the number of emitters equals the number of receivers but is not exactly on it.
- If there are no restrictions (e.g. production, costs) on the number of transducers per TAS, a specialised design approach or usage pattern approach could remain interesting to minimise measurement time. For specific optimal design suggestions, see Tab. 5.1.
- If the number of DAQ channels in the system is known, the number of transducer elements on the TAS should be adjusted accordingly.

Tab. 5.2 guided the DAQ system and analogue electronics design decisions, but could not be realized on a one-to-one basis. A more conservative design was

Table 5.2: *Comparison of Proposed and Realised USCT System Parameters*

Parameter	Optimal DAQ proposed	DAQ realised
DAQ channels	403	384
TAS	134	128
Unified transducer	yes	yes
Transducer elements	24	18
DAQ channels per TAS	3	3
MUX	8	6
Motor positions	1	2
n_{AScans}	$\approx 10^6$	$\approx 10^6$
T_{total}	121 s	129 s + ≈ 10 s (movement)

followed relying on a second motorposition relaxing the hardware requirements in DAQ and electronics hardware. Yet, the proposed combination in the ninth column of Tab. 5.1 with 403 channels is close to the practical realised solution with 384 channels and a comparable expected performance, see Tab. 5.2.

5.1.7 Summary of the Aperture based Considerations for 3D USCT III

Based on the presented work, the following summary can be made.

- An approximate hemispherical aperture of ≈ 0.2 m radius.
- Measurement window of $\approx 600 \mu\text{s}$.
- Unified transducer design with an emitter to receiver ratio of 1:1.
- Optimal DAQ system parameter proposal for $n_{\text{AScans}} = 10 \cdot 10^6$ ASscans in $T_{\text{total}} \approx 130$ s.

5.2 Imaging Contrast Optimisation for 3D USCT III

An important question to be answered is related to how to improve the contrast in reflectivity images. This is addressed in this section. This contrast can be

Table 5.3: Literature values of US attenuation coefficients for different breast tissue types, reported in dB/(MHz · cm).

Tissue type	Attenuation coefficient α [dB/(MHz · cm)]	Ref.
Fatty breast tissue	0.5 – 0.7	[IH93, Oel02]
Fibroglandular (dense) tissue	1.0 – 1.5	[Gre81, IH93]
Malignant breast tumour	1.0 – 2.0	[Oel02, Wea00]
Benign fibroadenoma	0.8 – 1.5	[Oel02]
Whole breast (mixed composition)	0.7 – 1.3	[Gre81, Wea00]

expressed as SNR in the imaging domain. Special attention will be given to the relation of imaging contrast with the features of US such as transducer spectral components and the influence on the imaging algorithms used in KIT 3D USCT reconstruction.

5.2.1 Object Complexity and Image Contrast

In the USCT application, the objects are complex, assuming many material interfaces acting as reflectors of various sizes. These sizes vary from 10 cm (breast size) down to 0.2 mm (e.g. micro-calcifications), all enclosed in a RoI of $20 \times 20 \times 15 \text{ cm}^3$.

To estimate the loss in signal strength, we consider a worst-case scenario. For this scenario, we assume a propagation path of 10 cm through highly attenuating, cancer-affected breast tissue at 3 MHz, combined with geometric amplitude loss due to spherical spreading of the wavefield. For malignant breast tissue, the upper-bound for the attenuation coefficient is $\alpha \approx -2 \text{ dB}/(\text{MHz} \cdot \text{cm})$ have been reported, see Tab. 5.3, corresponding to approximately $-6 \text{ dB}/\text{cm}$ at 3 MHz. Over a 10 cm path (from the boundary of the object to its center and back, see Fig. 5.5) this yields a tissue attenuation of about

$$A_{\text{tissue}} \approx -6 \text{ dB}/\text{cm} \times 10 \text{ cm} = -60 \text{ dB}. \quad (5.7)$$

In addition, spherical spreading causes a geometric amplitude decay proportional to $1/r$. A first case which utilises the surrounding aperture by exploring the object only until the center (r_{oneway}). r_{oneway} is considered as way from the transducer

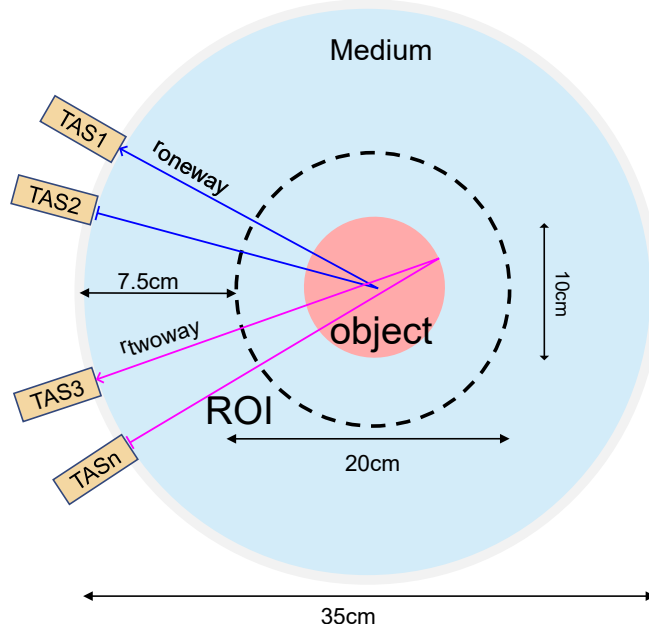


Figure 5.5: Aperture setup for a surrounding 3D USCT with its centred RoI and an object and the two attenuation considerations (top down view). Water medium distance between object and transducers is 7.5 cm. Object size is 10 cm. Blue the one-way-through-object travel path and pink the two way path. The dashed line demarks the RoI.

through the medium ($2 \cdot 7.5$ cm) to the object sitting centered in the 20 cm RoI (additional $2 \cdot 5$ cm) and back to the receiving transducer. Normalising to a reference distance r_0 close to the transducer ($r_0 = 3.8$ cm, $\approx$ begin of far-field), the corresponding geometric loss at a radius $r = r_0 + R_{oneway}$ can be expressed as

$$A_{\text{geom}} \approx 20 \log_{10} \left(\frac{r_{\text{oneway}}}{r_0} \right) \text{ dB} = \frac{35 + 3.8}{3.8} = -20.2 \text{ dB}, \quad (5.8)$$

which, for typical USCT geometries, is on the order of ≈ -20.2 dB. Taken together with the tissue attenuation, this leads to a total one-way worst-case loss of roughly

$$A_{\text{total}} \approx A_{\text{tissue}} + A_{\text{geom}} \approx -80 \text{ dB}. \quad (5.9)$$

This one-way wave travel assumption includes the normal case for transmission and reflections signals, fitting our surrounding aperture approach. However, the full reflectivity imaging potential covers also the wave which travels through the tissue completely and reflect at the back of the object to the front again. As worst case for USCT the application a two-way path (r_{twoway}) from the transducer to

the back of the object and backwards again can be defined, resulting in $r_{\text{tway}} \approx 45$ cm, see Fig. 5.5. With $r_{\text{tway}} = 45$ cm and $r = r_0 + r_{\text{tway}}$ a slightly bigger A_{geom} of -22.2 db results. With the significant increased tissue attenuation $A_{\text{tissue}} \approx -6 \text{ dB/cm} \times 20 \text{ cm} = -120 \text{ dB}$ resulting in a two-way worst-case attenuation of roughly $\approx -140 \text{ dB}$.

The minimum end of the dynamic range is simply assumed as a direct path between neighbouring transducers at only a few millimetres distance. In the water as coupling medium it can be assumed to be essentially non-attenuative and thus taken as a 0 dB reference.

This contrast between a near-lossless local path and a worst-case loss of up to approximately -140 dB clearly illustrates the need for an advanced DAQ and processing chain capable of reliably handling a dynamic range on the order of 0 to -140 dB. Formulated as technical goal, assuming a reasonable data type representation as 12-bit integer (typical ADC dynamic range, also in 3D USCT II), this dynamic range would come on top. As -140 dB corresponds to ≈ 23 -bit ($\log_2(10^{140/20})$), the overall ideal technical system design goal is therefore a 35-bit dynamic range resulting from the two-way worst case for the reflectivity signals. A reduced case is with the one-way assumption reducing the need to 25-bit (12-bit+13-bit) for transmission signals and reflectivity signals. This also underlines the importance of the surrounding aperture approach allowing the relaxed second case. The dynamic range consideration is evaluated later under Sec. 8.2.5.

5.2.2 Required Bandwidth for Object Representation

From a theoretical perspective, it would be ideal to have a full flat spectrum starting at 0 MHz and going towards infinity. However, from a technical perspective this is not realisable.

Fortunately, using the k-space synthesis we can derive the bandwidth subset which we need for our specific imaging requirements, see Fig. 2.13 in Sec. 2.9.2. For an object size of 0.2 mm (e.g. a micro-calcification) and a best-case pulse-echo configuration with a bistatic angle $\alpha = 180^\circ$ for a emitter-receiver combination

(see Eq. 2.75) the required frequency is

$$f = \frac{c}{2 \cdot L \cdot \sin(\alpha/2)} = \frac{1520 \text{ m/s}}{2 \cdot 0.0002 \text{ m} \cdot 1} = 3.8 \text{ MHz.} \quad (5.10)$$

For the biggest feasible object in the RoI, e.g. $\approx 0.10 \text{ m}$, using the pulse echo configuration a frequency of 7.6 kHz would be required. This would lead to a very broad bandwidth, with also very low frequencies. This is unrealistic to achieve with common US transducers and analogue electronic chains.

Fortunately, with a device which provides both the transmission and backscatter sensitive emitter-receiver configurations by the bistatic α angle, these frequency requirements can be significantly relaxed; as is the case with the surrounding 3D USCT aperture. For such a surrounding aperture with (almost) full angular coverage the needed frequencies for relevant object sizes are given in Tab. 5.4. For instance, for big objects (e.g. 10 cm) the effective temporal frequency requirement shrink significantly with the availability of emitter-receiver combinations with smaller bistatic angles α , e.g. for $\alpha = 1^\circ$ the frequency is 870 kHz, see Eq. 2.75.

Moreover, if a fully covered bistatic angular space is available, as in the case with the surrounding 3D USCT III aperture, it becomes easier to realise bistatic combinations with corresponding frequencies are available, see green marked fields in the Tab. 5.4. This substantially reduces the required temporal (axial) frequency support. In our case, the effective requirement can be reduced from a fractional bandwidth of approximately 200% (from 7 kHz to 430 MHz) to about 125% (from 870.91 kHz to 3.8 MHz), which is more realistic bandwidth for the 3D USCT III system.

In contrast, a purely transmission-based system with limited angular coverage (only high α) would for the same target structure sizes of 10 cm, 5 mm, and 0.2 mm, require a frequency span in the order of 7 kHz to 3.8 MHz to fulfil the half-wavelength criterion, while a purely reflectivity-based system constrained to small bistatic angles would need to extend its usable frequency range up to several hundreds of MHz ($\approx 430 \text{ MHz}$ for the 0.2 mm structures). Such bandwidth requirements are difficult to realise in a practical technical system, while for the

Table 5.4: Required temporal/axial frequency for different bistatic angles α for resolving structures of 10 cm, 5 mm, and 0.2 mm at $c = 1520$ m/s. Green marked the lowest, a center example and the highest frequency, spanning the required frequency range. Derived from Eq. 2.75.

α (deg)	$\sin(\alpha/2)$	$f_{10\text{cm}}$ (kHz)	$f_{5\text{mm}}$ (kHz)	$f_{0.2\text{mm}}$ (MHz)	Scattering regime
0	0.0000	∞	∞	∞	Transmission
1	0.0087	870.91	17418.14	435.45	Near-transmission
2	0.0175	435.47	8709.40	217.74	Near-transmission
3	0.0262	290.33	5806.64	145.17	Near-transmission
4	0.0349	217.77	4355.36	108.88	Near-transmission
5	0.0436	174.23	3484.69	87.12	Near-transmission
15	0.1305	58.25	1164.92	29.12	Forward scatter
30	0.2588	29.36	587.28	14.68	Forward scatter
60	0.5000	15.20	304.00	7.60	Strong forward scatter
90	0.7071	10.75	214.96	5.37	Side scatter
120	0.8660	8.78	175.51	4.39	Partial backscatter
150	0.9659	7.87	157.36	3.93	Strong backscatter
165	0.9904	7.68	153.56	3.84	Strong backscatter
180	1.0000	7.60	152.00	3.80	Backscatter / pulse-echo

combined transmission–reflection system 3D USCT III, this is technical feasible.

Consequently, given these considerations, we aim for a bandwidth of approximately 0.8 to 3.8 MHz for the 3D USCT III in order to cover structure sizes between 10 cm and 0.2 mm.

5.2.3 Required Bandwidth for Object Separation on Signal Level

Another more practical approach with predictive power for the system design is considering the USCT signal measurement process and considering the separability of individual point scatterers. With a Huygens wave synthesising argument it can be argued that with enough of such simple scatterers a complex object like a breast’s structure can be composed and represented. Ideally, with a full-bandwidth system, a reflection would be represented by a sharp Dirac response.

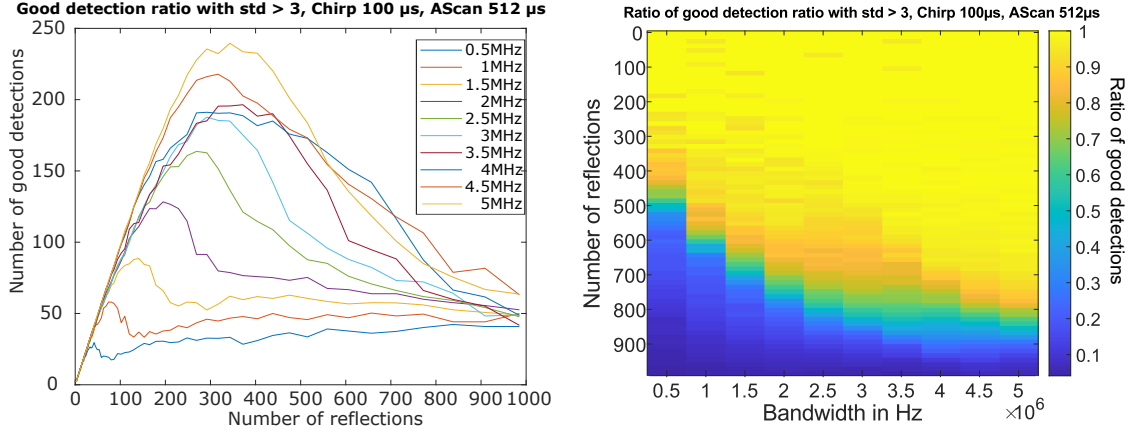


Figure 5.6: *Simulation of the separability of reflection signals over bandwidth and reflection. Parameters chosen for 3D USCT III. (Left) absolute number of separable reflections in the signal for curves of growing bandwidth, x-axis is growing number of reflections. (Right) ratio of properly detected reflections, y-axis growing number of reflections, x-axis is bandwidth in Hertz.*

As signal processing theory tells us that with a reduced bandwidth this Dirac pulse degrades to a sinc pulse whose number and amplitude of sidelobes grows with falling bandwidth.

The design decision of increasing the opening angle and using 3D transducers allows us to receive the reflections from all objects at once. So, every AScan contains more information from more objects than in the case of using conventional 2D sonographic transducers or the transducers of the previous 3D USCT II system which have a smaller opening angle. As a result, the information density in the 1D AScan increases with the new 3D USCT III device. Therefore, separability of information and pulses in the 1D ASAns becomes a larger challenge.

In the following simulative analysis we consider a USCT measurement window of 512 μs and an excitation pulse length of 100 μs with a centre frequency of 2.5 MHz. Next, we model synthetically created artificial reflections to analyse the impact of many overlapping reflections in clustered time-window. Here a complex object is represented by a collection of point-scatterer, placed in the center of the RoI. Next, we map the spatial position to time windows. The degrees of freedom analysed are the bandwidth and the number of reflectors, as emulation of object complexity. The bandwidth was defined between 0.5 and 5 MHz as parameter for linear chirps as excitation signal. The number of reflectors analysed are between 1

and 100. As criterion for successful separation of an individual scatterer from the background, cluttered by many side-lobes, a $> 3 \cdot \sigma$ criterion is used. The results can be seen in Fig. 5.6. The signal processing and metric extraction is defined as follows.

The matched-filter output $A_{\text{mf}}(n)$ is obtained by correlating the complex-envelope AScan $A_{\text{ce}}(n)$ with the complex conjugate of the reference excitation $C_E(n)$ in the frequency domain

$$A_{\text{mf}}(n) = \mathcal{F}^{-1}\{\mathcal{F}[A_{\text{ce}}(n)] \cdot \mathcal{F}[C_E(n)]^*\}. \quad (5.11)$$

A detection set idx_1 is then defined by thresholding the matched-filtered signal at three times its standard deviation

$$\text{idx}_1 = \{n \mid A_{\text{mf}}(n) > 3 \sigma_{A_{\text{mf}}}\}. \quad (5.12)$$

The number of correctly detected reflections D corresponds to the intersection between the expected reflection positions and the detected indices

$$D = \left| \text{round}(\mathbf{p}_1) \cap \mathbf{i}_{\text{idx}_1} \right|. \quad (5.13)$$

Finally, the effective SNR gain G_{SNR} is defined as the ratio of correctly detected reflections to the total number of expected reflections

$$G_{\text{SNR}} = \frac{D}{R}. \quad (5.14)$$

This ratio serves as a quantitative measure of the detection efficiency achieved through matched filtering.

Ultimately it seems that on signal level for separability of the complex objects a full bandwidth would be required. This gives us a strong argument for image domain level object separation.

5.2.4 Imaging-based Object Separation Considerations for the Aperture

The impact of the aperture on the separability of point scatterers is analysed using simulated data for three different setups.

The first setup is a 2D sonography aperture resulting in one-sided illumination. The two apertures have a lateral length of 2.5 cm and 12.5 cm, see Fig. 5.7. Here, the left column shows the two setups where the red dots form the aperture and the green dots are the point scatterers. On the right column the resulting SAFT image is given. Top row is an aperture with 300 elements at 2.5 MHz center frequency and 1 MHz bandwidth, and a lateral length of 2.5 cm. Bottom represents the same setup, but now the lateral length of the aperture is increased to 12.5 cm. As a result, the separability of the point sources improves noticeable along the lateral dimension while the axial resolution is hardly affected.

The second setup is a 2D surrounding aperture setup, positioned circular around the point scatterers. As a result, this setup results in an all-sided illumination, see Fig. 5.8. Again, the left column shows the setup and the right column the resulting SAFT image. The aperture has 300 elements at 2.5 MHz center frequency and 1 MHz bandwidth, and a diameter of 26 cm. The separability of the point scatterers improves noticeable along all directions compared to the 2D sonography aperture. In addition, the resolution is homogenous in all directions.

Next the center frequency is varied, see Fig. 5.9. From top left to bottom down the center frequency is increased, from 1.66 MHz to 10 MHz. The separability of the point scatterers increases with rising frequency as expected, the blurred representation of the point-spread-function becomes smaller.

Finally, the bandwidth is varied while keeping the center frequency fixed at 2.5 MHz, see Fig. 5.10. From top left to bottom down, the bandwidth increases from 15 kHz to 2.5 MHz. The separability of the point scatterers increases significantly with rising bandwidth. As demonstrated, a high center frequency is not sufficient for achieving a unique representation. Hence, a broad bandwidth is required. For a very small bandwidth (16 kHz), an almost monochromatic case, the resulting point-spread-function becomes periodic. Such visible periodic ripples

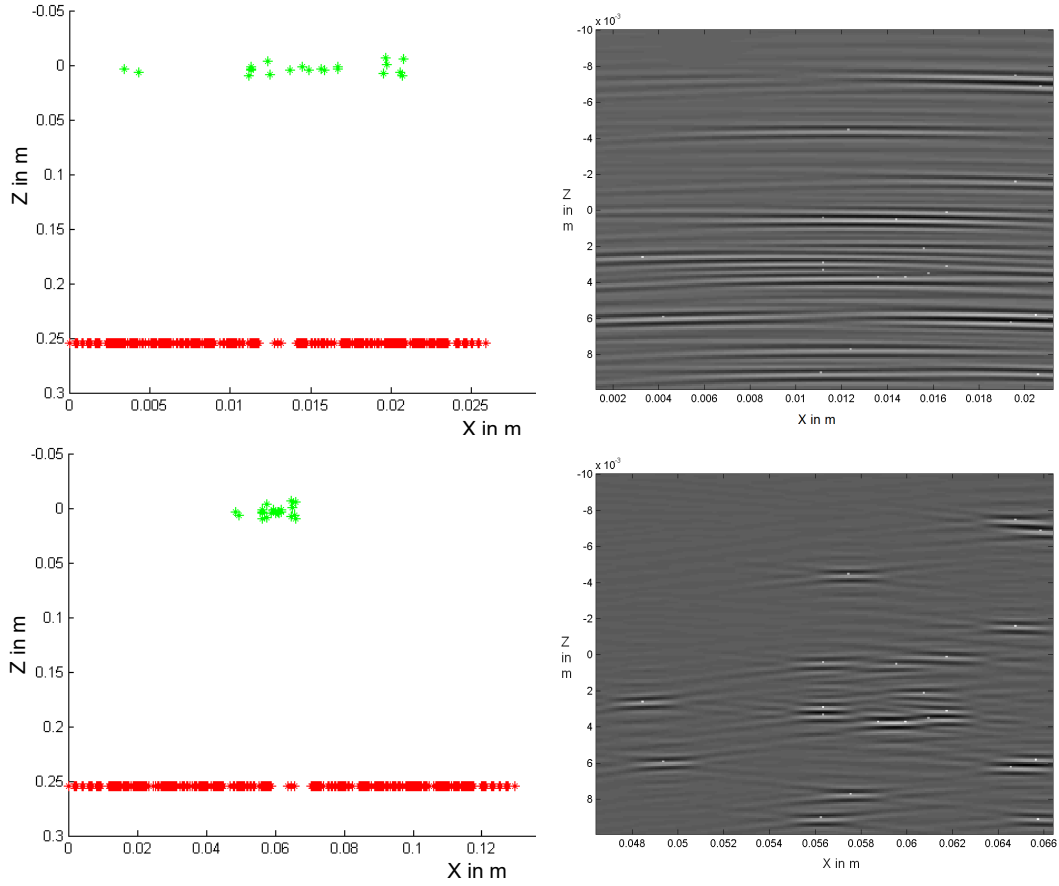


Figure 5.7: Planar systems: impact of lateral aperture lateral length. (Left column) Sonography aperture in red, green point sources. X-axis lateral dimension, z-axis axial dimension. Axis units in meter. (Right column) SAFT response of this aperture, overlaid with ground truth (as white dots). (Top row) Sonography aperture with lateral length of 2.5 cm, (Second row) 12.5 cm lateral length.

in the resulting SAFT images are also called grating lobes.

5.2.5 SAFT Imaging based SNR Considerations

In classical sonography, image level SNR (which corresponds to contrast) benefits from two strong mechanisms: temporal averaging ($\sqrt{n}$) and beam-forming based SNR gain in the image creation step. Because many transducer elements emit and receiver US in parallel, their echoes add coherently up due to beamforming and provide substantial instantaneous SNR. In a SAFT-based system, each measurement is performed with only a single emitter and every resulting AScan is formed only by a single emit–receive transducer pair, resulting in a very low

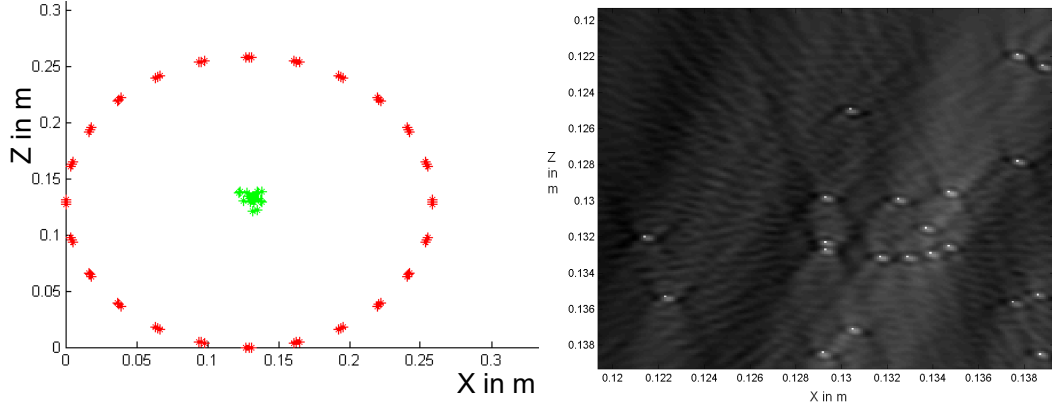


Figure 5.8: Surrounding aperture 12.5 cm diameter: impact on lateral resolution. (Left column) aperture in red, green point sources. X-axis lateral dimension, z-axis axial dimension. Axis units in meter. (Right column) SAFT response of this aperture, overlaid with ground truth (as white dots).

acoustic pressure and therefore SNR. This effect is even more pronounced for 3D systems, with small omni-directional transducers whose emission and reception sensitivity is significantly smaller than in typical larger 2D sonography arrays. While in SAFT-based system temporal SNR gain is possible, practical constraints such as patient motion or measurement time limit this possibility.

Luckily the SNR gain potential of sonography system by beamforming is not lost for the 3D USCT system, but just shifted to another domain, i.e. from the hardware implemented signal beamforming domain to the digital imaging method itself, and hence the SAFT image reconstruction. The signal contributions from each weak 3D transducer interfere constructively at the correct voxel locations and grow with n , whereas the predominantly uncorrelated noise components grow only with $\sqrt{n}$. In this way, the $\sqrt{n}$ SNR improvement that conventional sonography obtain through parallel beamforming is, in 3D USCT, produced algorithmically by the SAFT imaging method.

To illustrate this image based SNR gain with SAFT, a 3D USCT setup including point scatterers is simulated. The hardware setup is a 3D surrounding aperture with fully omnidirectional (large opening angle) transducers. The low-pressure on emission and reception of 3D transducers is emulated by adding a high noise level, resulting in a low or negative SNR level. The noise distribution is Gaussian and non-coloured. Two forms of SAFT imaging algorithms are analysed: perfectly

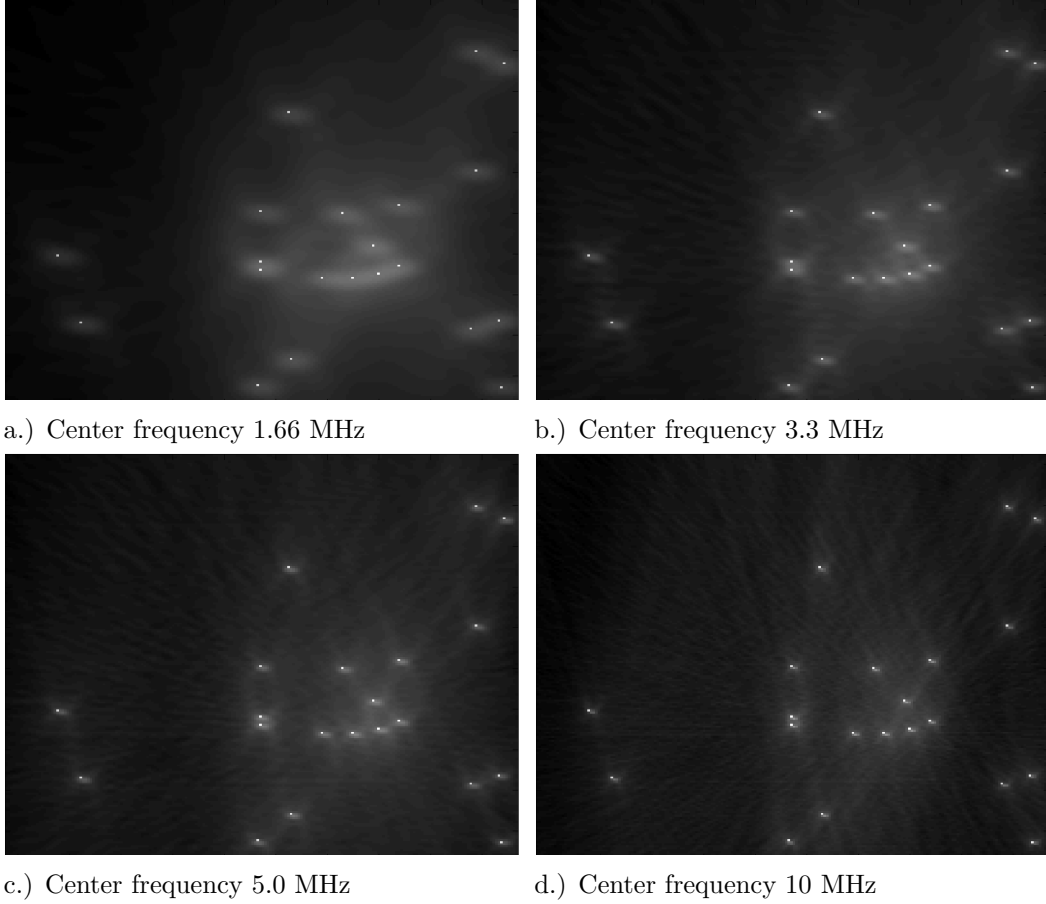


Figure 5.9: *Impact of varying axial center frequencies on reflectivity SAFT imaging. A surrounding circular aperture with a radius of 12.5 cm (see Fig. 5.8), 300 AScans, and a ground truth consisting of 20 point scatterers (as white dots overlaid). All axes are in meter.*

focussing requiring coherent SAFT, and an error-tolerant pulse-detection variant of SAFT, called optimal-pulse SAFT[RSZG08]. The simulation was done with a CPU-implemented SAFT in MATLAB and the following parameters: $\approx 58,000$ AScans in a 35 cm surrounding 3D aperture with realistic parameters; transducer frequency sensitivity of 0.8 MHz to 3.8 MHz; 20 point scatterers with random positions inside the RoI and with equal omnidirectional reflectivity. Angular characteristic and geometrical attenuation is added in the SAFT imaging step. In Fig. 5.11 (top left) the distribution of the scatterers in a $0.02 \times 0.02 \text{ m}^2$ 2D image plane can be seen. In the top right graph the AScan representation without noise is shown, in the bottom right the case with 0.2 noise strength and in the bottom left the case with a noise strength of 1.0. The noise strength factor was tuned

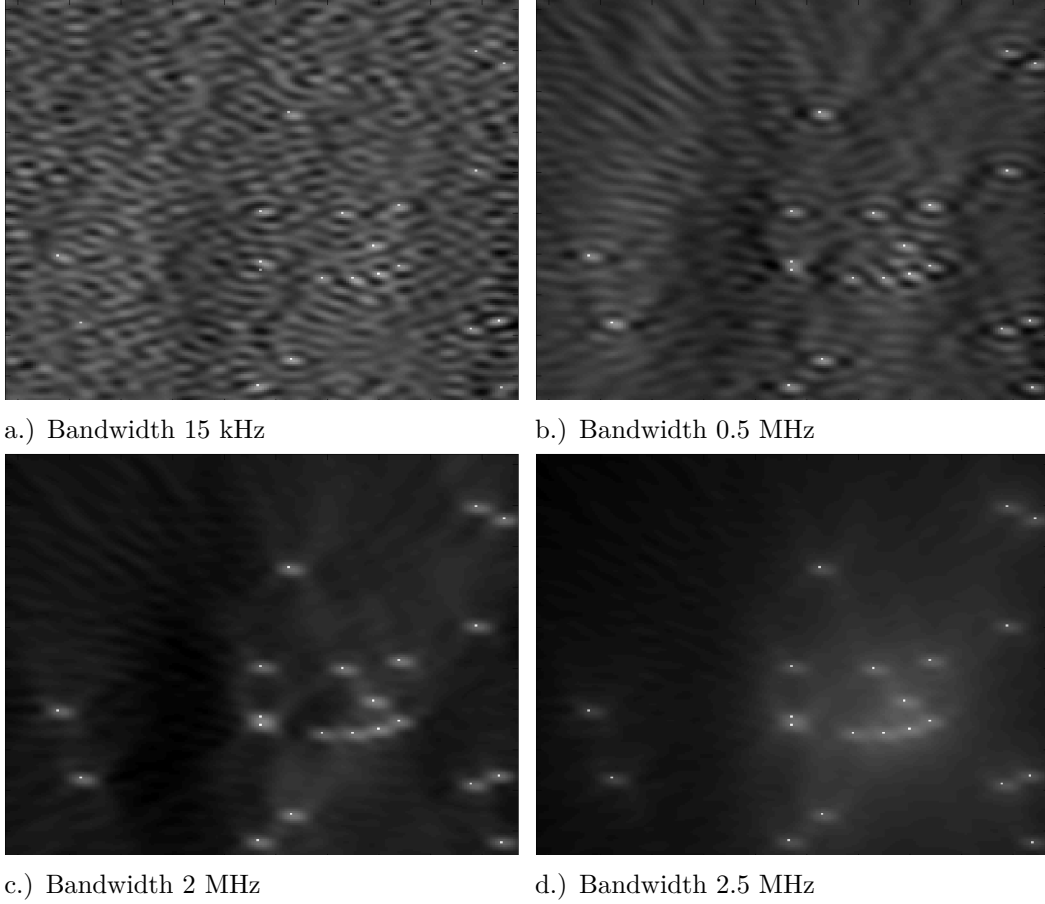


Figure 5.10: *Impact of bandwidth on SAFT imaging. Surrounding circular aperture 12.5 cm radius, 300 AScans, ground truth of 20 point scatterers as white dots overlaid. All axes are in m.*

such that the signals amplitude corresponds to the σ of the added noise in RoI distance, hence the amplitudes are similar.

In Fig. 5.12 the resulting SAFT images are shown, with in the top row the coherent SAFT and in the bottom row the non-coherent SAFT image with optimal pulse. The left column is the noiseless case, the right column is the case with a noise level of $\sigma = 4 \cdot A_{max}$, with A_{max} the maximum amplitude present in all recorded AScans. For the noiseless case, both SAFT methods perform well and can separate the point scatterers. For the noise case, the coherent SAFT has some background noise but still shows most scatterers. The optimal-pulse SAFT suffers significantly and several scatters are undistinguishable from the noise floor. In this example the SNR ratio is below 1. Also notable, in the optimal-pulse case the absolute amplitude has degraded from being defined ideally only by the physical

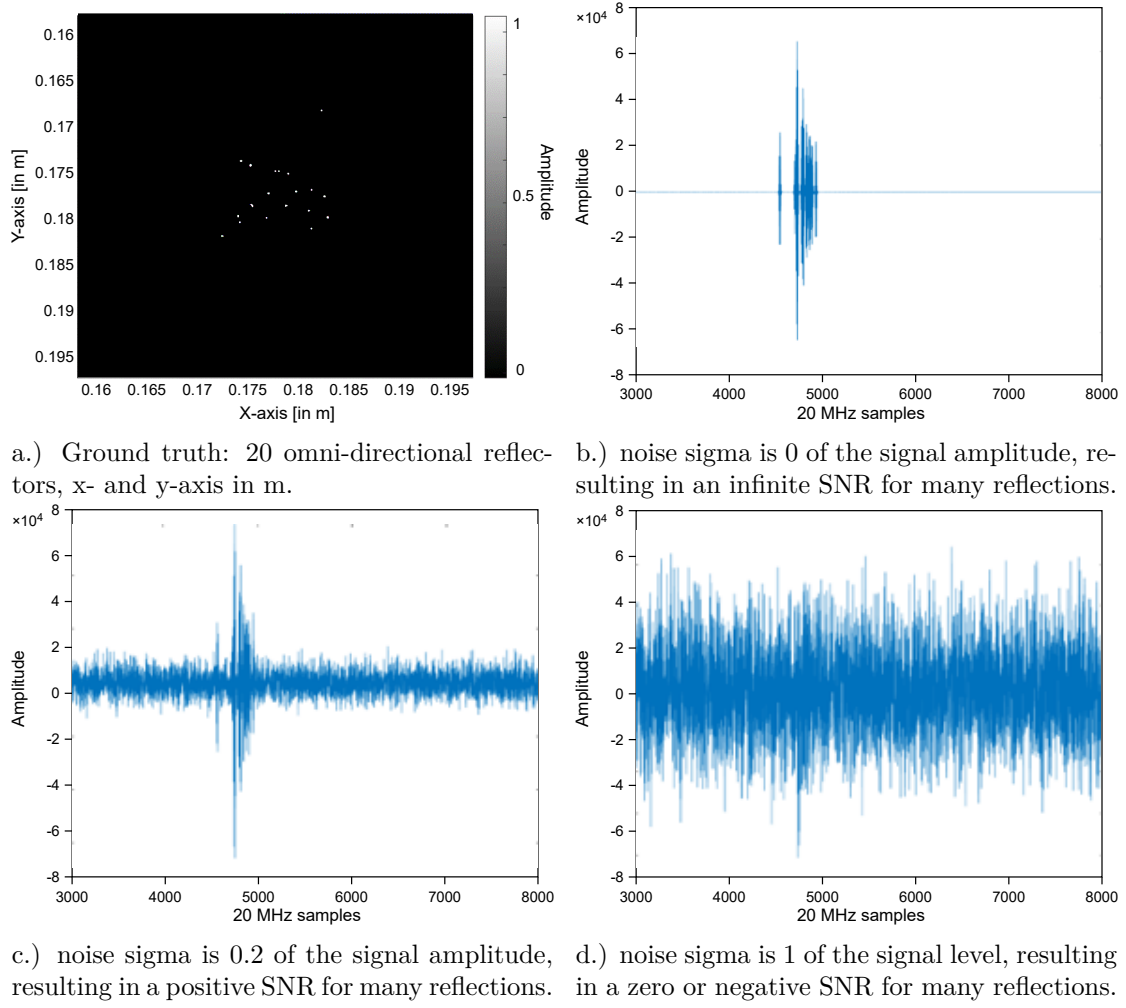


Figure 5.11: *Simulation with realistic USCT parameters: 3D positions, transducer characteristic, reconstruction bandwidth etc. Objects are 20 omnidirectional point scatterers in RoI: a) 2D ground truth image, b–d) simulated AScans for a exemplary emitter-receiver combination for various noise levels.*

property reflectivity to a lower value determined by the probabilistic properties of the optimal pulse inherent signal-level detection step. In contrast, in coherent SAFT image the absolute amplitude level of the scatterers remain the same between the noise and noiseless case, achieving a proper proportional representation of the reflectivity modality of these scatter objects.

In Fig. 5.13 the SNR behaviour over the AScans is shown for the four cases displayed in Fig. 5.12, evaluated for point scatterer #6 as example. The noise strength is 2, resulting in a noise standard deviation of $\sigma \approx 2$. The typical signal-level SNR is therefore below one. The image SNR is defined as the ratio between

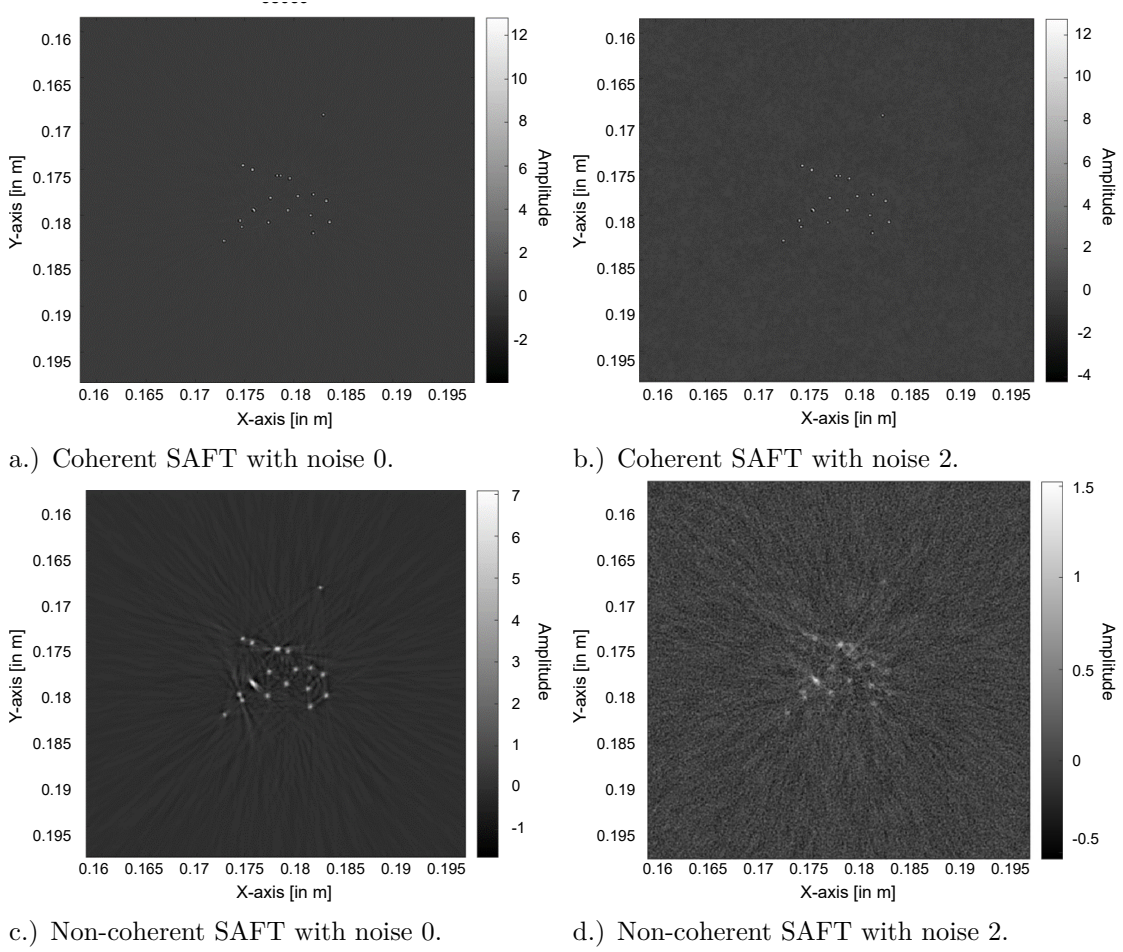


Figure 5.12: *Simulation with realistic 3D USCT parameters with 58,059 AScans: 3D transducer positions, transducer characteristic, bandwidth etc. All axes are in m. Objects are 20 omnidirectional point scatterers in RoI. (Left column) no noise; (Right column) noise strength 2. a,b: coherent SAFT reflectivity image; c,d: non-coherent optimal pulse SAFT reflectivity image.*

the reconstructed amplitude of the point scatterer and the standard deviation of the complete image as noise approximation:

$$\text{SNR}_{\text{image}} = \frac{A_{\text{ps}}}{\sigma_{\text{image}}}, \quad (5.15)$$

where A_{ps} is the peak reconstructed amplitude and σ_{image} the standard deviation of the background noise.

In the left graph, i.e. the noiseless case, both SAFT methods perform well but both seem to saturate in SNR already. On the right side, i.e. the strong-noise

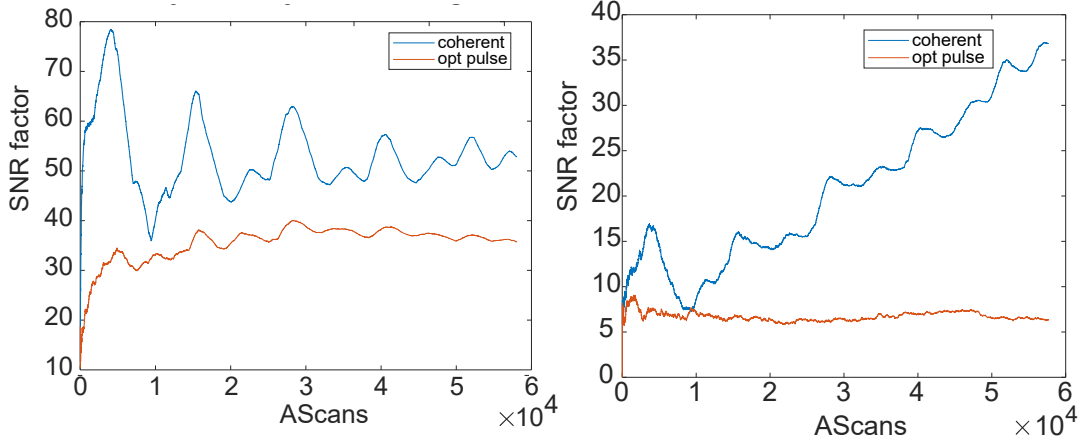


Figure 5.13: Reflectivity SAFT imaging SNR simulation with realistic 3D USCT parameters and 20 point-scatters. Point scatterer #6 SNR ratio analysed over 58,059 AScan AScan addition. X-axis is AScan count; y-axis is the SNR gain ratio. Blue curve is the coherent SAFT SNR; red curve is the non-coherent SAFT (optimal pulse) SNR. (Left) noiseless case. (Right) noise strength 2, corresponding to a signal SNR ratio of 0.5, the noise amplitude is higher than the signal.

case, both methods start with a similar SNR, but optimal pulse SAFT is already at its saturation point due to the below one SNR, while coherent SAFT gains SNR when additional AScan are added.

Consequently, in the high signal-level SNR regime both methods might work well if enough scatterers are separable on signal level due to enough bandwidth (see Sec. 5.2.3). Reaching a signal SNR ratio of 1, either due to weaker reflectivity of an object or general system-inherent low SNR, the optimal pulse SAFT starts having problems resolving objects. While methods like optimal-pulse-based SAFT imaging methods have the benefit of being more phase-error tolerant and do not require perfect focussing of an US system. This benefit comes with the downside of a reduced image SNR and potentially disproportional reflectivity.

For the 3D USCT III system design this means a well understood and well calibrated system is key for realising the full SNR potential with SAFT reflectivity imaging. Some evaluation is provided in Sec. 7.9.

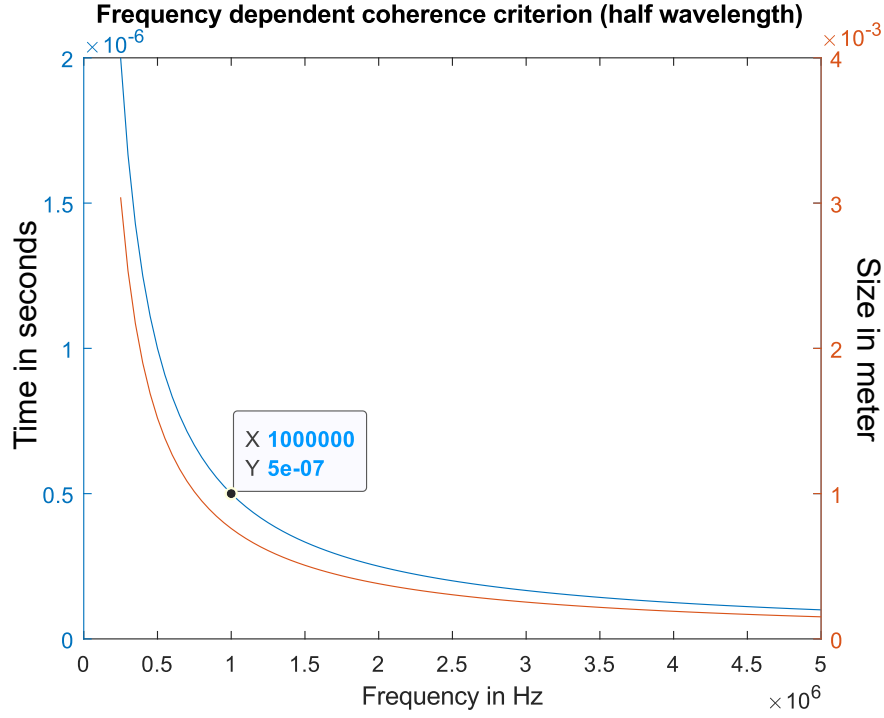


Figure 5.14: Tolerance over frequency on the x-axis. Blue graph given as time criterion ($0.5 \cdot \lambda/f$) on the left y-axis, red as spatial criterion for $c=1520$ m/s in meter on the right y-axis ($0.5 \cdot c/f$) For instance for 1 MHz, in time tolerance this is $0.5 \mu\text{s}$ (correlate to five 10 MHz sample in the USCT) or as spatial tolerances of 0.76 mm.

5.3 Accuracy and Tolerance Requirements

A fundamental requirement for coherent wave overlapping is the half-wavelength coherence criterion, which governs the required precision of all system components in hardware, signal processing, and imaging. The permissible path-length deviation ΔL , must remain below half the acoustic wavelength to avoid destructive phase errors. Using $\lambda = c/f$, the condition becomes

$$\Delta L < \frac{\lambda}{2} = \frac{c}{2f}. \quad (5.16)$$

For the operating frequency range of 0.5 to 4.5 MHz, this yields a frequency-dependent coherence tolerance from several millimetres at low frequencies down to sub-millimetre precision at the higher frequencies, see Fig. 5.14.

5.3.1 Considerations for the 3D USCT

For the 3D USCT III, and assuming a maximum utilised frequency of $f = 4.5$ MHz, the half-wavelength criterion in time is $0.1111 \mu\text{s}$, which can be interpreted as the allowable time tolerance. This corresponds to approximately 1.1 samples at the targeted sampling rate of 10 MHz, which represents a challenging requirement. For the spatial requirement, using a representative sound speed of $c = 1520$ m/s, the spatial half-wavelength criterion yields at $f = 4.5$ MHz

$$\frac{\lambda}{2} = \frac{c}{2f} \approx 0.169 \text{ mm}, \quad (5.17)$$

which defines the spatial tolerance. Similarly, the angular precision necessary can be derived for the outer boundary of the RoI. Given the spatial tolerance of $\Delta x = 0.169$ mm and a RoI radius of $R = 10$ cm, the corresponding angular tolerance is

$$\Delta\theta = \arctan\left(\frac{\Delta x}{R}\right) = \arctan\left(\frac{0.169}{100}\right) \approx 0.097^\circ. \quad (5.18)$$

These tolerances apply to all components in the system that can introduce time or phase shifts onto the US signal and impede the phase coherence.

This question of correct and coherent signal extraction inside the given tolerances is crucial for optimal imaging quality. Three principal approaches are possible: error reduction (design, process), error tracking (calibration, measurements and modelling in) and increasing the tolerance (algorithmic robustness).

5.3.2 Uncertainty and Error Sources

There are many sources of error and uncertainties known from the experiments and calibration of the 3D USCT II system: transducer positioning uncertainties, electrical system delays, transducer phase-shifts, medium sound-speed errors, and physical effects.

Spatial errors come from manufacturing accuracies on the transducer positions and mechanical fixtures. The planned approach of handling these errors for the 3D USCT III is geometrical calibration afterwards using extracted data. This

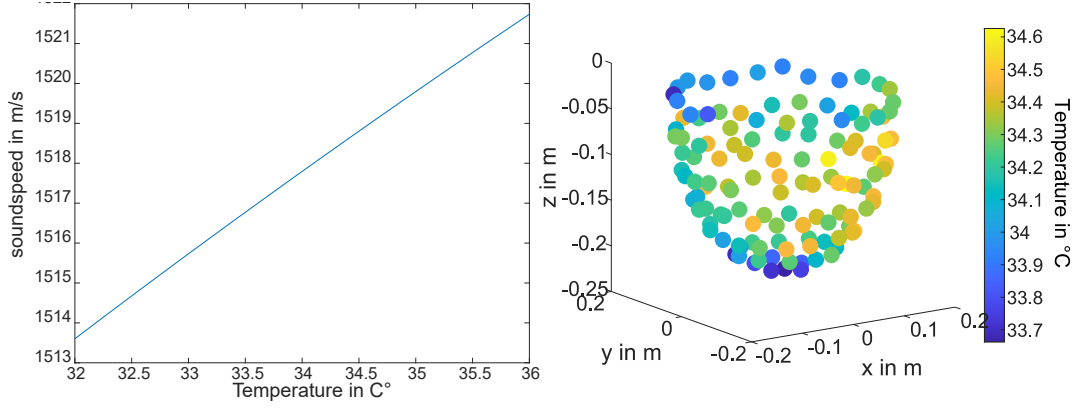


Figure 5.15: (Left) water medium's sound-speed dependence over temperature in the relevant temperature range for 3D USCT. (Right) temperature variation inside the 3D USCT II, measured values from the temperature sensors in each TAS.

approach is successfully used with the 3D USCT II system, relaxing the need for excessive required mechanical production tolerances significantly.

Additionally, time delays coming from the electronics and signal chain should be characterized using self-calibration approach included in the hardware system and measurable on signal level. This feature is missing in 3D USCT II.

With respect to the physical effects, the medium sound speed is a big source of uncertainty. Unfortunately, for the typical medium for US devices, i.e. water, it is known that the sound speed is temperature dependent[Mar97], see Fig. 5.15 left.

In a large 3D USCT system with around 20 l water as surrounding medium we have a high potential for temperature variability over space and time. This unlike conventional sonography systems where there is no significant amount of surrounding medium or for ring-based 2D US systems, which have typically shorter travel paths and less water. Therefore this topic is of greater importance for 3D USCT and similar 3D systems.

From the 3D USCT II device experience, variations of 1°C over space occur due to water layering, heating or cooling effect and are common despite many efforts of keeping the temperature stable, see Fig. 5.15 right. To partly control this problem within the 3D USCT II and III devices, temperature sensors are included in the surface of the devices and transducers.

To show the effect of these temperature variations on the allowable tolerance we

calculate the error in the travel time for a wave which travels twice over a distance of $r = 7.5$ cm, i.e. the distance between the transducer surface and the skin of the breast see Fig. 5.5. For a temperature variation of 0.5°C around 35°C a timing error of

$$t_{\text{err}} = \frac{2r}{c_{35^\circ\text{C}}} - \frac{2r}{c_{34.5^\circ\text{C}}} = 0.645 \cdot 10^{-7} \text{ s} \quad (5.19)$$

is observed. This accounts already for 59% ($0.65 \cdot 10$ MHz samples) of the allowable temporal tolerance budget for coherent focussing, see Fig. 5.14.

5.3.3 System Uncertainty Mitigation Approaches

Challenges related to the uncertainties and error calibration were addressed repeatedly throughout the history of the USCT project. Several theses and projects are related to finding solutions to calibrate the system [Zap03, Hop03], achieving focussed reflectivity images. Among these approaches to realise the full imaging potential are two PhDs: Adam Filipik (2007) and Weiyap Tan (2017) [ZRF⁺07, Tan17]. In these works variations in the transducer position, the sound speed, and the time delay were modelled and corrected for. As a result, both resolution and contrast could be significantly improved.

Yet, there are indications that additional uncertainties and error sources still exist, and the overall system error remains too large to realise the full potential of coherent SAFT and compounding [Nei23]. Potential additional system error sources may include frequency-dependent phase shifts from system components such as the transducers or dispersion effects originating from the tissue. These kinds of phase shifts are currently not included by the calibration or the imaging model.

From the error-mitigation perspective, one already implemented method is a variation of SAFT, the so-called optimal-pulse approach, which increases robustness against such errors [RSZG08]. Newer iterative, optimisation-based imaging algorithms may also compensate for these system uncertainties and tend to be less susceptible. Another mitigation mechanism is to exploit the available bandwidth by using only the lower-frequency components, since they exhibit a higher tolerance to system errors and uncertainties, see Fig. 5.14.

For the 3D USCT III system, it can be concluded that a precise characterisation and reduction of intrinsic uncertainties, the implementation of an in-system calibration strategy, or the use of robust model-based and iterative imaging algorithms will be essential to exploit the theoretically achievable resolution and coherence of the system.

5.4 Transducer Design

In addition to the already existing considerations from previous Chapters, additional degrees of freedom left for the transducer design are answered with a Huygens based wave field projection approach. For this approach, a spatial sampling is utilised at ≈ 11 sample-points per wavelength, the maximum pressure curves are taken in the far-field.

The 3D USCT III transducer design is guided by the system-level decision to build a true 3D US system and by the operational experience gained with the 3D USCT II. Several design questions must be addressed:

- Trade-off between transducer diameter, RoI illumination, beam divergence/opening angle.
- Trade-off opening angle versus emitted pressure.
- Phase divergence minimization over frequencies.
- Axial versus lateral bandwidth contribution optimization in a surrounding aperture.
- 3D radiation characteristics for the transducer, square versus circular aperture geometry.
- Focused transducers versus planar-aperture transducers, including the implications for phase alignment.

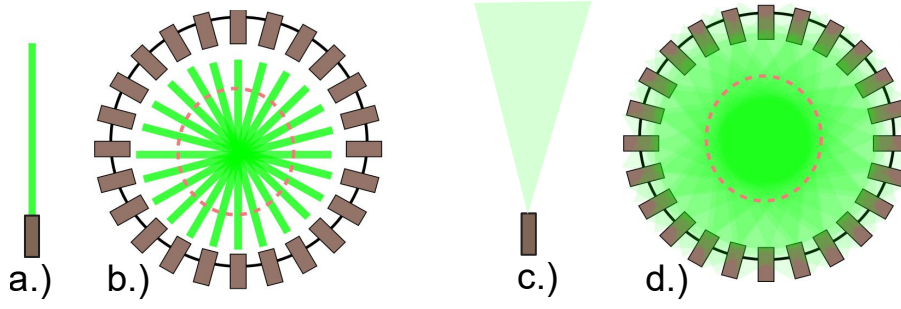


Figure 5.16: *Illustration of the effect of an increased opening angle on RoI illumination in a top-down view. Brown shapes represent transducers, green regions indicate emitted wave fields, and the dashed red circle marks the RoI centrally inside the aperture. (a) Transducer with a narrow opening angle (low divergence). (b) Resulting composite wave field when using such transducers around the aperture, producing areas of over- and under-illumination. (c) Transducer with an increased opening angle. (d) Resulting composite field when using wide-angle transducers, producing a much more homogeneous illumination of the RoI.*

5.4.1 RoI Illumination Homogenisation

The primary motivation for increasing the opening angle of the US transducers is to achieve a more homogeneous illumination in the RoI.

When a transducer's opening angle is too small, only the center region will be illuminated from all sides, while the outer regions are only illuminated from one side. This effect is illustrated in Fig. 5.16 a.) and b.), Figure c.) illustrates a hypothetical transducer with a wider wave field, i.e., a larger opening angle. The composite radiation pattern in d.) demonstrates a that this approach results in a significantly more homogeneous illumination of the RoI.

5.4.2 Required Opening Angle in Dependence on Aperture Diameter

The question of the required opening angle for the new RoI of $20 \times 20 \times 15 \text{ m}^3$ will be addressed here. First we will assume the aperture as perfect (hemi-)spherical and we will design for symmetry and simplification reason the RoI as rotational symmetric (hemi-)spherical with a 20 cm diameter.

In Fig. 5.17 a.), the situation for the current 3D USCT II aperture is illustrated,

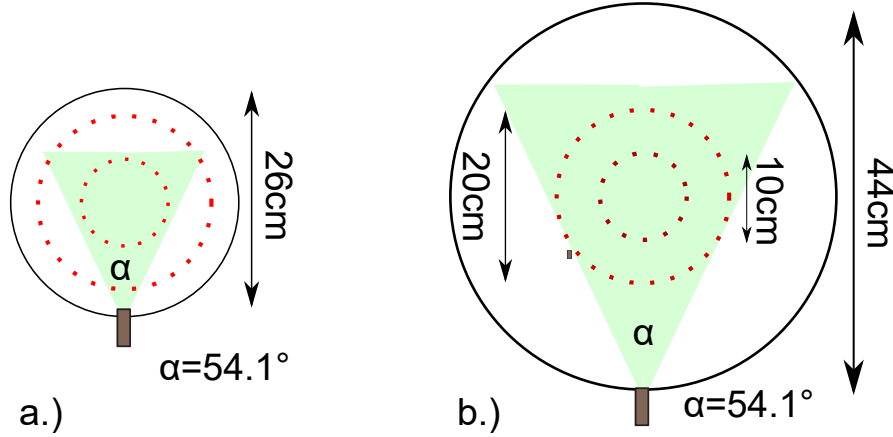


Figure 5.17: Wave field of an emitting transducer in different surrounding circular aperture configurations, inner red dotted circle is the 3D USCT II spherical RoI 10 cm, outer red dotted circle is the new spherical RoI of 20 cm. (a) 3D USCT II setup with a diameter of 26 cm, where the opening angle covers only the 3D USCT II RoI. (b) Potential 3D USCT III setup with 44 cm diameter and an opening angle of 54.1° .

the opening angle covers the smaller inner spherical RoI of 10 cm but is insufficient to cover the larger 20 cm spherical RoI. The required opening angle for the 3D USCT II aperture of 26 cm diameter which covers the new RoI completely can be derived by a triangular calculation

$$2 \cdot \arcsin\left(\frac{0.10}{0.13}\right) = 105.7^\circ, \quad (5.20)$$

resulting in $\alpha = 106^\circ$, which is a quite large opening angle from manufacturing point of view. Therefore, the consideration is taken that increasing the diameter of the the new 3D USCT III aperture might reduce the opening angle requirement substantially. For instance, if we increase the aperture diameter to 44 cm for the new RoI, i.e.

$$2 \cdot \arcsin\left(\frac{0.10}{0.22}\right) = 54.1^\circ, \quad (5.21)$$

we can relax the opening angle requirement to almost the half with an $\alpha = 54.1^\circ$, see Fig. 5.17 (b).

The general relationship between aperture diameter and required opening angle is given in Fig. 5.20 as blue curve and can help to find a the optimal value. Two further side-constraints are considered and included as color-code. The clinical application with patients lying on the device should measure only one breast. If the aperture opening gets too big, proper placement becomes a bigger challenge.

Table 5.5: *Transducer designs sound pressure for f_{max} (3.5 MHz)*

Design #	Transducer	Sound pressure
1	0.900 mm square (3D USCT II)	100%
2	0.400 mm square	25%
3	0.902 mm circular	100%
4	0.456 mm circular	25%

Also, hospital standards for door openings limit the diameter and with increasing diameter the water amount increases significantly, with challenges the logistics and timing. Therefore, a maximum diameter of 0.5 m is considered to be the end of the possible design range (red encoded). Second constraint is the opening angle; when larger than 90° , point source like transducers are required which are hardly feasible (yellow area). Consequently, the optimal design space is the green encoded area in Fig. 5.20.

5.4.3 Transducer Aperture Size Relationship to Opening Angle

In the previous section it was shown that there is the need to optimize the transducer aperture size in view of the resulting opening angle and amplitude of the pressure field, where the amplitude is linearly related to the SNR. In general, an increasing aperture size leads to a reduction of the opening angle, but to increase of the amplitude of the resulting pressure field. Based on experiences with the 3D USCT II device, it is preferred to keep the amplitude the same.

To compare and select appropriate designs, four transducer apertures are simulated, see Tab. 5.5. First, the original, reference 3D USCT II transducer aperture created by dice-and-fill technology with a side-length of $0.9 \times 0.9 \text{ mm}^2$, composed out of four $0.4 \times 0.4 \text{ mm}^2$ square sub-structures (kerf 0.1 mm). Second, as natural progression from the square production technology and realisable with current technology, a single square transducer of $0.4 \times 0.4 \text{ mm}^2$ is selected. Third, as ideal reference, a 0.902 mm diameter circular transducer, with a pressure and surface equivalent to the reference design. Fourth and last, a circular equivalent to the second square aperture with a diameter of 0.456 mm.

The transducers are simulated in the far-field using a Huygens wave interference simulation model with a sampling satisfying Nyquist criterion. This is chosen over

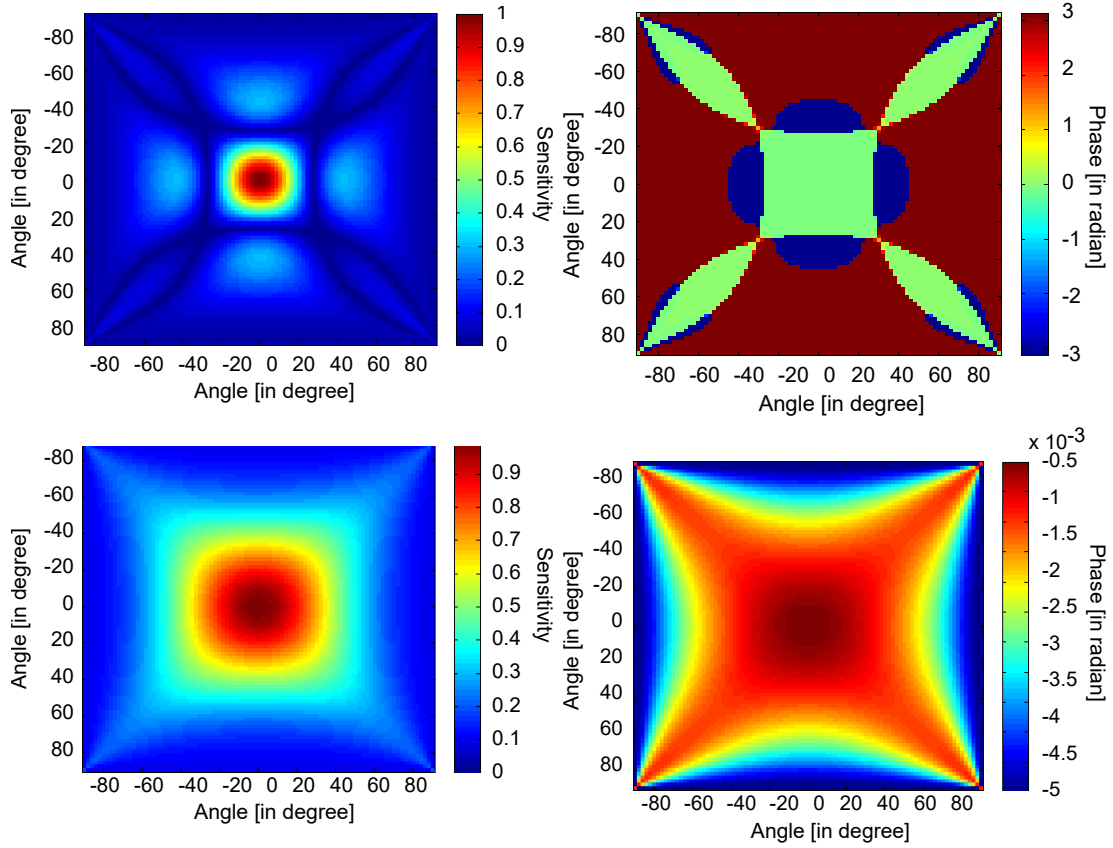


Figure 5.18: Simulation results for transducer apertures with Huygens wave synthesis for 3.5 MHz. (Left) Relative pressure over elevation angle (y -axis) and horizontal angle (x -axis). (Right) phase response in radian over elevation and horizontal angle. (**First row**) Square transducer (4 elements, 0.9×0.9 mm), 3D USCT II transducer design. (**Second row**) Square transducer (1 element, 0.4×0.4 mm), 25% sound pressure.

the Fraunhofer approximation models due to increased flexibility in the projection planes and projection spaces. Phase and pressure are extracted equispaced in the far field and associated with directivity angles. The resulting amplitudes and phases in the far-field are shown in Fig. 5.18 and Fig. 5.19.

Notable is the central phase inversion (phase value jumps by $\pm\pi$) for the two large transducers (design 1 and 3), happening around $\pm 20^\circ$. This phase inversion coincides with the minima in the pressure amplitudes. The large square design features a second phase jump along the diagonals, showing a non-homogeneous behaviour along elevation and horizontal angles. The circular transducer does not show this behaviour and is therefore preferred, and should be notably more robust for tasks requiring stable phase fronts, such as coherent wave field synthesis, like

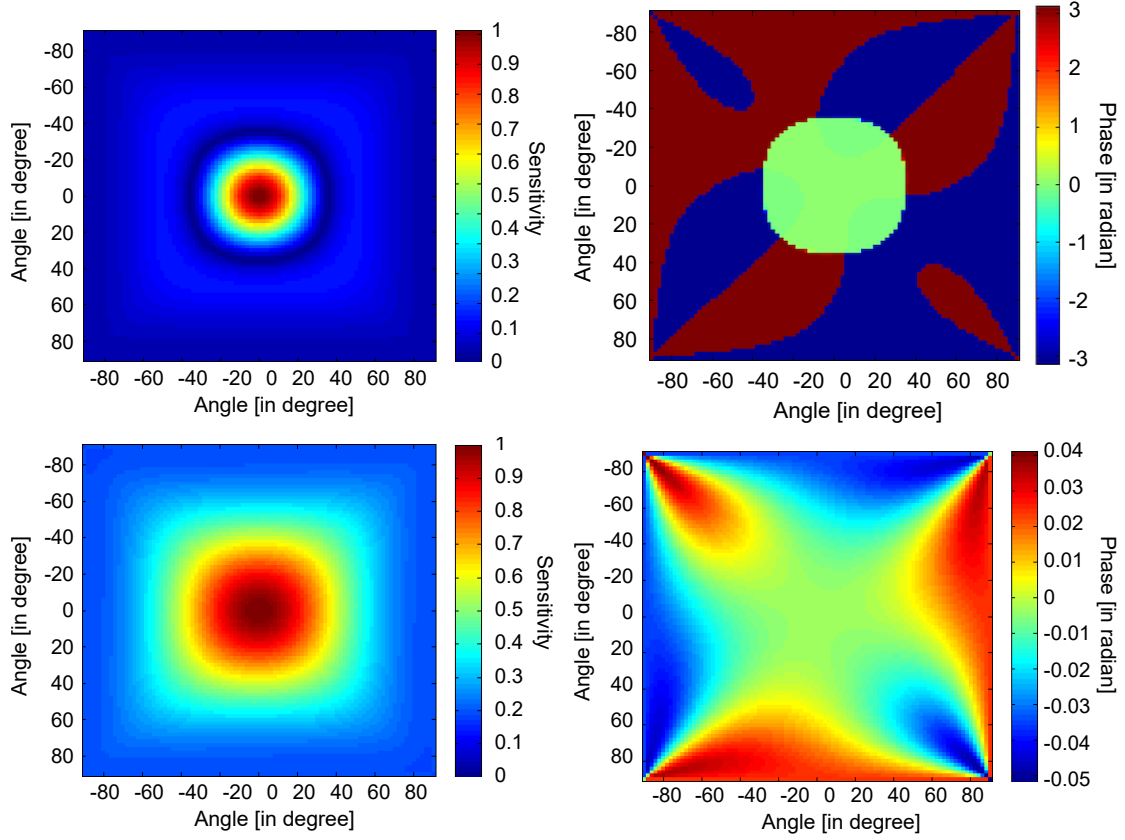


Figure 5.19: Simulation results for transducer apertures with Huygens wave synthesis for 3.5 MHz. (Left) Relative pressure over elevation angle (y -axis) and horizontal angle (x -axis). (Right) phase response in radian over elevation and horizontal angle. (First row) Circular transducer (diameter 0.902 mm), sound pressure equivalent to the 3D USCT II transducer (100%). (Second row) Circular transducer (diameter 0.456 mm), sound pressure equivalent to the 3D USCT II 0.4×0.4 mm sub-element (25% sound pressure).

SAFT.

The last step is considering the impact of emitter and receiver transducer sensitivity, which are assumed to behave symmetrically. Therefore, the final relative sensitivity results is the product of the emitter and receiver contributions. The extracted pressures are given in Tab. 5.6.

We define the practically usable opening angle as the angle at which the pressure drops to 50% at 3.5 MHz, the upper end of the centre-frequency range, in order to retain a safety margin. When we plot these 50% pressure opening angles for the four transducers in the “Opening angle over radius” diagram in Fig. 5.20, we

Table 5.6: *Opening angle for $f = f_{max} = 3.5$ MHz, product of receiver and emitter contribution for the Face-to-face configuration (bi-static angle 180°).*

Transducer	20% drop	50% drop	Sound pressure
0.900 mm square (3D USCT II)	14°	22°	100%
0.400 mm square	30°	54°	25%
0.902 mm circular	16°	28°	100%
0.4561 mm circular	33°	58°	25%

see that the only realistically realisable options are the two smaller transducer variants.

For the 0.4 mm^2 square transducer, the opening-angle performance implies an aperture diameter of approximately 0.44 m for optimal operation. For the 0.456 mm^2 circular transducer, the optimal aperture diameter would be ideally $\approx 0.412\text{ m}$.

These dimensions guided the later design decisions. Ultimately, however, the optimal diameter could only be approximated. Due to practical limitations, for example, the availability of material (rod production material availability of maximum diameter 0.5 m) and general device size constraints for practical handling, the realised aperture diameter was about 15% smaller, at approximately 0.35 m .

5.4.4 Transducer Aperture Frequency Behaviour in the RoI

After analysing the influence of opening angle, while taking into account the amplitude of the resulting pressure field in a monofrequent way, we now analyse in more detail the relation between the apertures and the frequencies in the relevant bandwidth. We expand the monofrequent analysis by an analysis covering the full bandwidth. As excitation signal we utilise the 3D USCT II linear chirp signal with a center frequency of $f_0 = 2.5\text{ MHz}$, a bandwidth (BW) of 1 MHz , and a $5\text{ }\mu\text{s}$ cosine attenuation window. Representative AScans are taken from specific positions in the RoI.

First setup is the 3D USCT II transducer (square $0.4 \times 0.4\text{ mm}^2$ design) with its RoI of $10 \times 10 \times 10\text{ cm}^3$. This configuration acts as reference setup and the relevant images are shown in Fig. 5.21. An AScan is taken from the centre of the

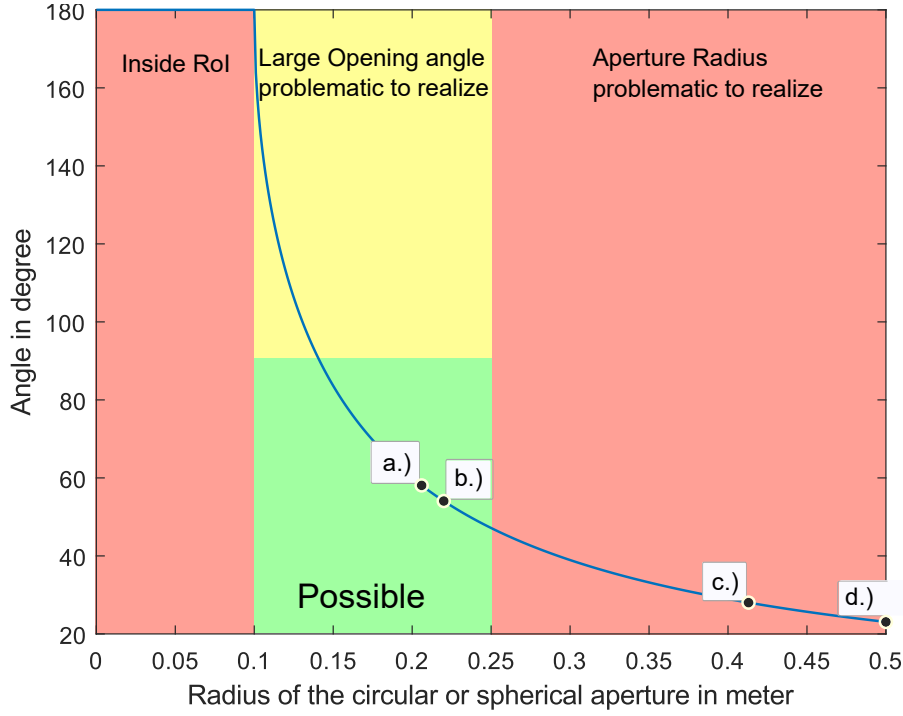


Figure 5.20: Required opening angle for a spherical RoI of 20 cm diameter: Blue curve, transducer opening angle over aperture radius for covering a spherical 0.2 m RoI. Red area, aperture radius > 0.25 m is unreasonable for clinical usage, ≥ 0.1 m is too small for the required RoI. Yellow area, opening angles beyond 90° are challenging for emerging as point-source with low pressure and problematic aspects ratios. Case a.) circular transducer of 0.456 mm diameter having a aperture radius of 0.21 m with an opening angle of 58.08° . B.) is a rectangular transducer of 0.400 mm side-length having a aperture radius of 0.220 m with an opening angle of 54.07° . C.) is a circular transducer of 0.902 mm diameter having a aperture radius of 0.413 m with an opening angle of 28.02° . D.) is a rectangular transducer of 0.900 mm side-length having a aperture radius of 0.500 m with an opening angle of 23.07° .

RoI, the ideal case. Top right shows in blue the reference chirp as excitation signal with overlaid in green the output response of the transducer. Bottom left shows the corresponding Fourier-domain response, i.e. the magnitude of the chirp input (blue) and the transducer influence (green). Bottom right shows the normalised full-bandwidth response with phase difference. There is little deviation between input and output, as expected.

The second test corresponds to same the 3D USCT II setup again with its RoI of $10 \times 10 \times 10$ cm³, see Fig. 5.22. This time the AScan is taken from the worst-case assumption. As worst-case assumption, motivated by the previous angular results assuming that the maximum deviation from the ideal, representative AS-

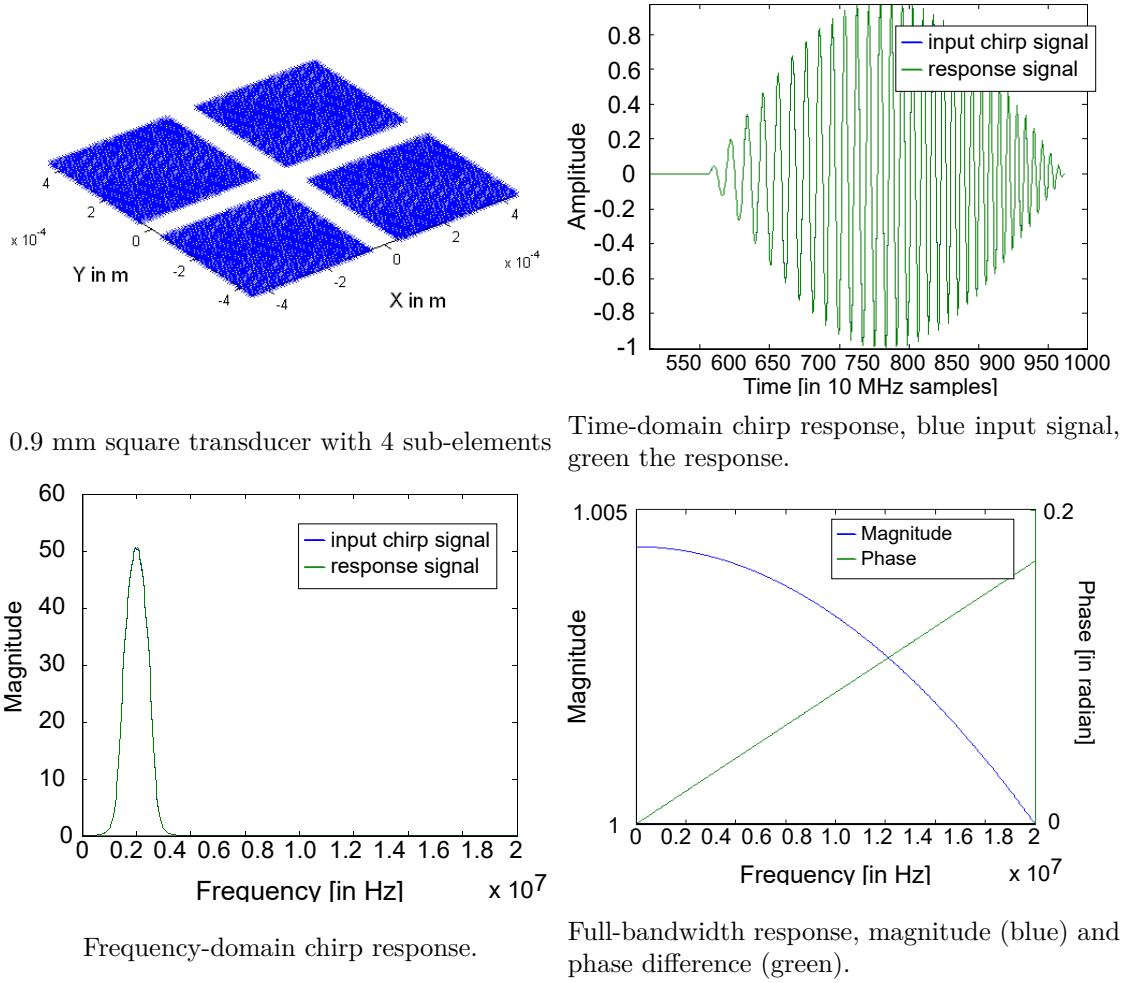


Figure 5.21: Frequency response for transducer 0.9 mm square, RoI $10 \times 10 \times 10$ cm, characteristic centred (best case). Chirp: $f_0 = 2.5$ MHz, bandwidth 1 MHz, cosinusoidal damped. Blue curves input signal, green curves response with transducer influence.

cans are extracted on the border of the RoI. We see in the time-domain and frequency-domain representations a significant deviation between the input and output signal. In the bottom right the full normalised spectrum response with phase is shown. Here we see significant phase inversion, starting at approximately 2.5 MHz. This will already significantly impede the performance of the signal-processing chain for imaging by potentially breaking the phase coherence criterion for SAFT.

Third setup is the 3D USCT II setup again but this time with the 3D USCT III aimed for a RoI of $20 \times 20 \times 15$ cm³, see Fig. 5.23. The AScan is taken again from the worst-case assumption, which means in case of this RoI even further out, under

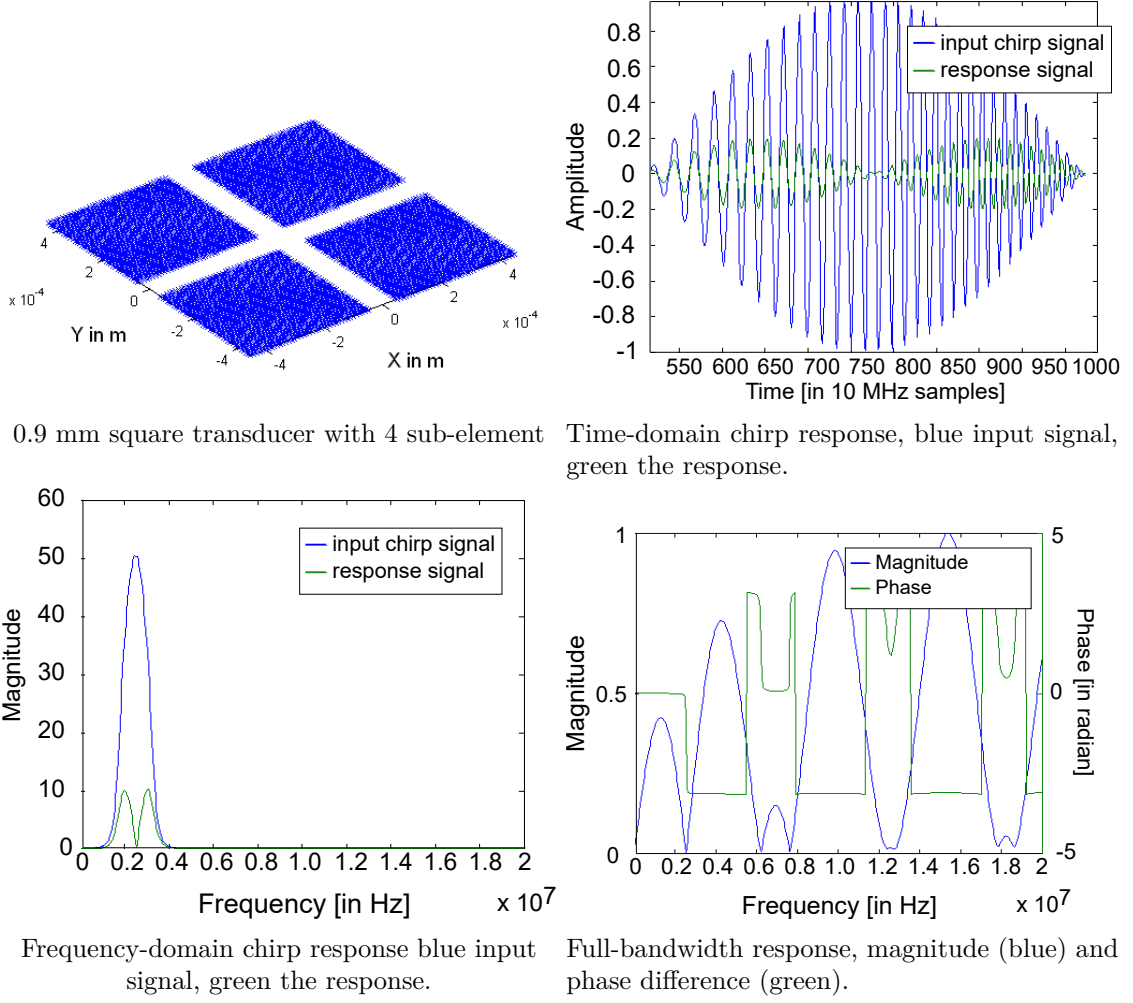
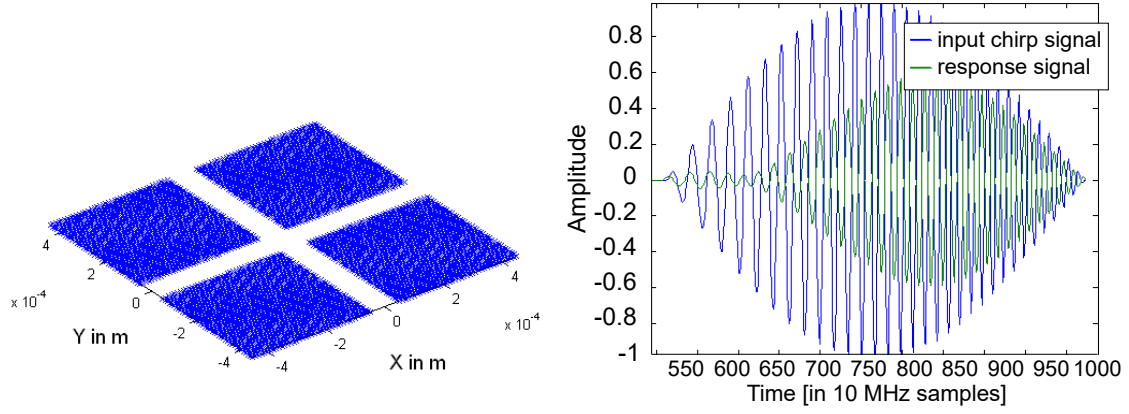


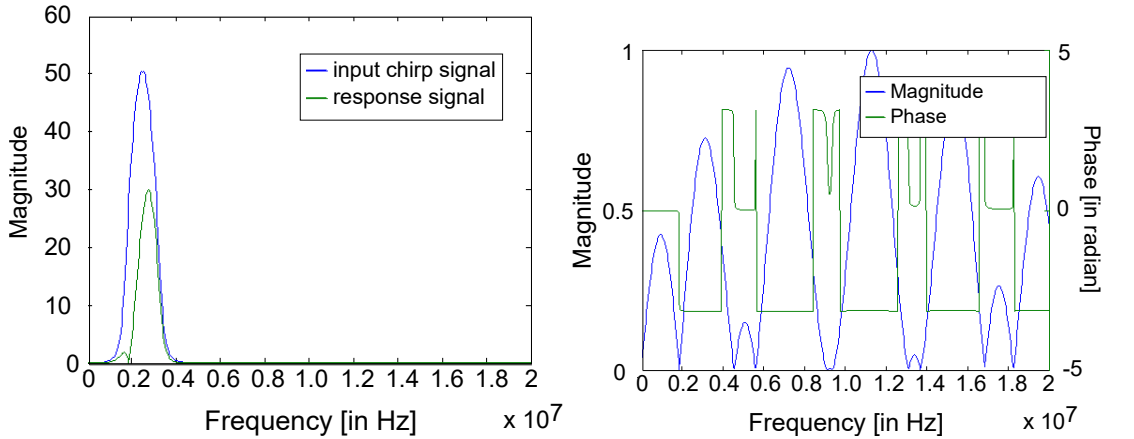
Figure 5.22: Frequency response for transducer 0.9 mm rect, RoI $10 \times 10 \times 10$ cm, characteristic at $[0.057 \text{ m}, 0 \text{ m}, 0.13\text{--}0.056 \text{ m}]$ (RoI border, worst case). Chirp: $f_0 = 2.5 \text{ MHz}$, bandwidth 1 MHz , cosinusoidal damped. Blue curves input signal, green curves response with transducer influence.

a higher angle. We see in the time-domain and frequency-domain representation a significant deviation between input and output chirp. In the bottom right the full normalised spectrum response with phase we see significant phase inversion, starting even earlier with approximately 2.0 MHz . This case is even more impeded in its performance of the signal-processing chain for later imaging.

As fourth setup we consider now the 0.4 mm^2 square transducer as potential 3D USCT III design, with the aimed RoI of $20 \times 20 \times 15 \text{ cm}^3$, see Fig. 5.24. The AScan is taken again from the worst-case assumption, which means in case of this RoI even further away, under a higher angle. We see in the time-domain



0.9 mm square transducer with 4 sub-elements Time-domain chirp response blue input signal, green the response.



Frequency-domain limited bandwidth chirp response blue input signal, green the response. Full-bandwidth response, magnitude (blue) and phase difference (green).

Figure 5.23: Frequency response for transducer 0.9 mm square, RoI $20 \times 20 \times 15$ cm, characteristic at $[0.113 \text{ m}, 0 \text{ m}, 0.13\text{--}0.056 \text{ m}]$ (RoI border, worst case). Chirp: $f_0 = 2.5 \text{ MHz}$, bandwidth 1 MHz , cosinusoidal damped. Blue curves input signal, green curves response with transducer influence.

and frequency-domain representations only minor deviation between input and output chirp. In the bottom right we have the full normalised spectrum response with phase there we see the first phase inversion at 4.5 MHz right outside of our aimed bandwidth. This setup should have, due to the very little extra amplitude attenuation and non-existing phase shift inside the desired bandwidth, very little degrading effect on the signal quality and later imaging.

As fifth setup we take a look at the 0.456 mm circular transducer as potential 3D USCT III design, with the aimed RoI of $20 \times 20 \times 15 \text{ cm}^3$, see Fig. 5.25. The AScan is taken again from the worst-case assumption, which means in case of

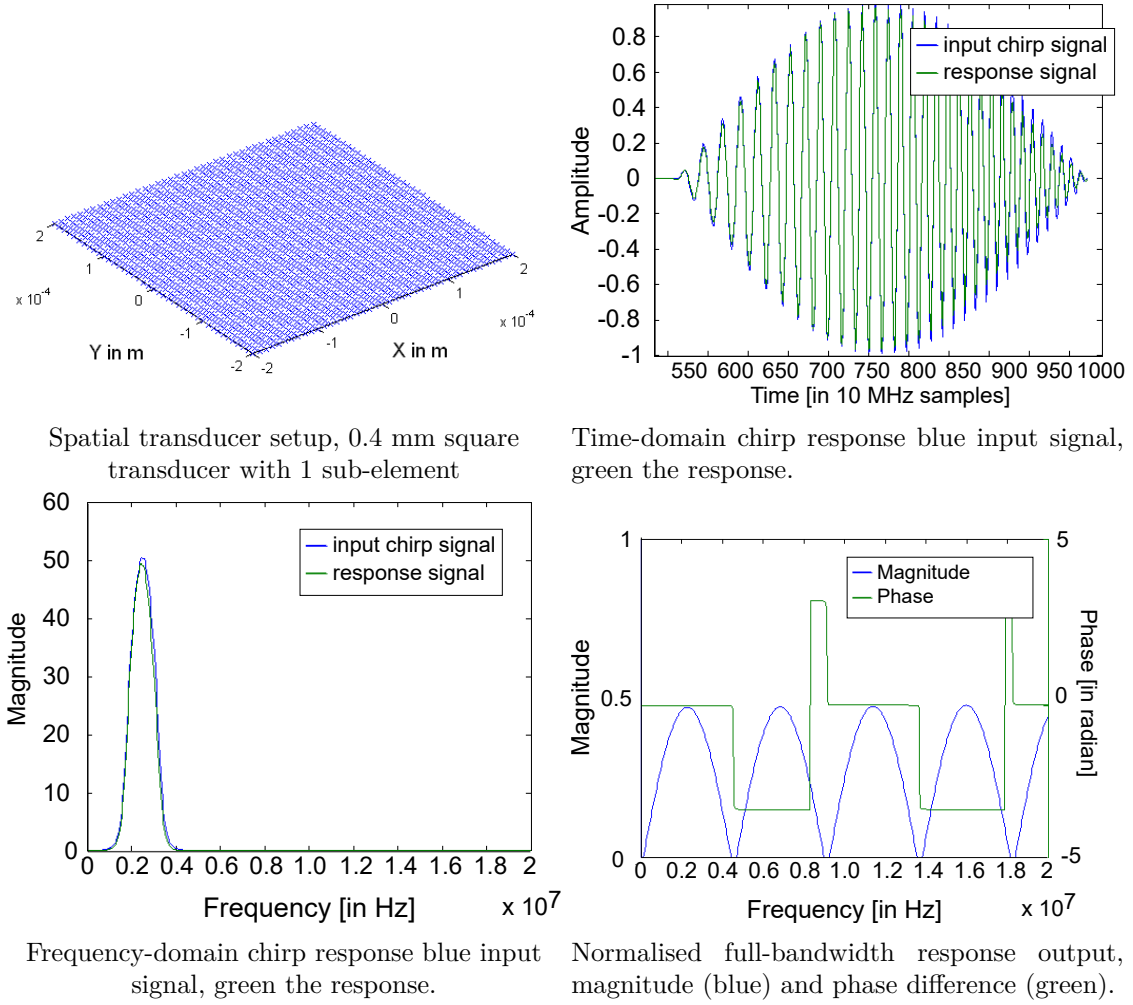


Figure 5.24: Frequency response for transducer $0.4 \text{ mm}^2 \text{ rect}$, RoI $20 \times 20 \times 15 \text{ cm}$, characteristic at $[0.113 \text{ m}, 0 \text{ m}, 0.13\text{--}0.056 \text{ m}]$ (RoI border, worst case). Chirp: $f_0 = 2.5 \text{ MHz}$, bandwidth 1 MHz , sinusoidal damped. Blue curves are the input signal, green curves is the response with transducer influence.

this RoI even further out, under a higher angle. We see in the time-domain and frequency-domain representation only minor deviation between input and output chirp. In the bottom right full normalised spectrum response with phase we see the first phase inversion at approximately 4.6 MHz (slightly higher than the square transducer), right outside of our aimed bandwidth. This setup should have, due to the little amplitude attenuation and non-existing phase shift inside the desired bandwidth, very little degrading effect on the signal quality and later imaging.

In general we conclude from this, that these two base transducer apertures should cover the required bandwidth and should provide from their base aperture design

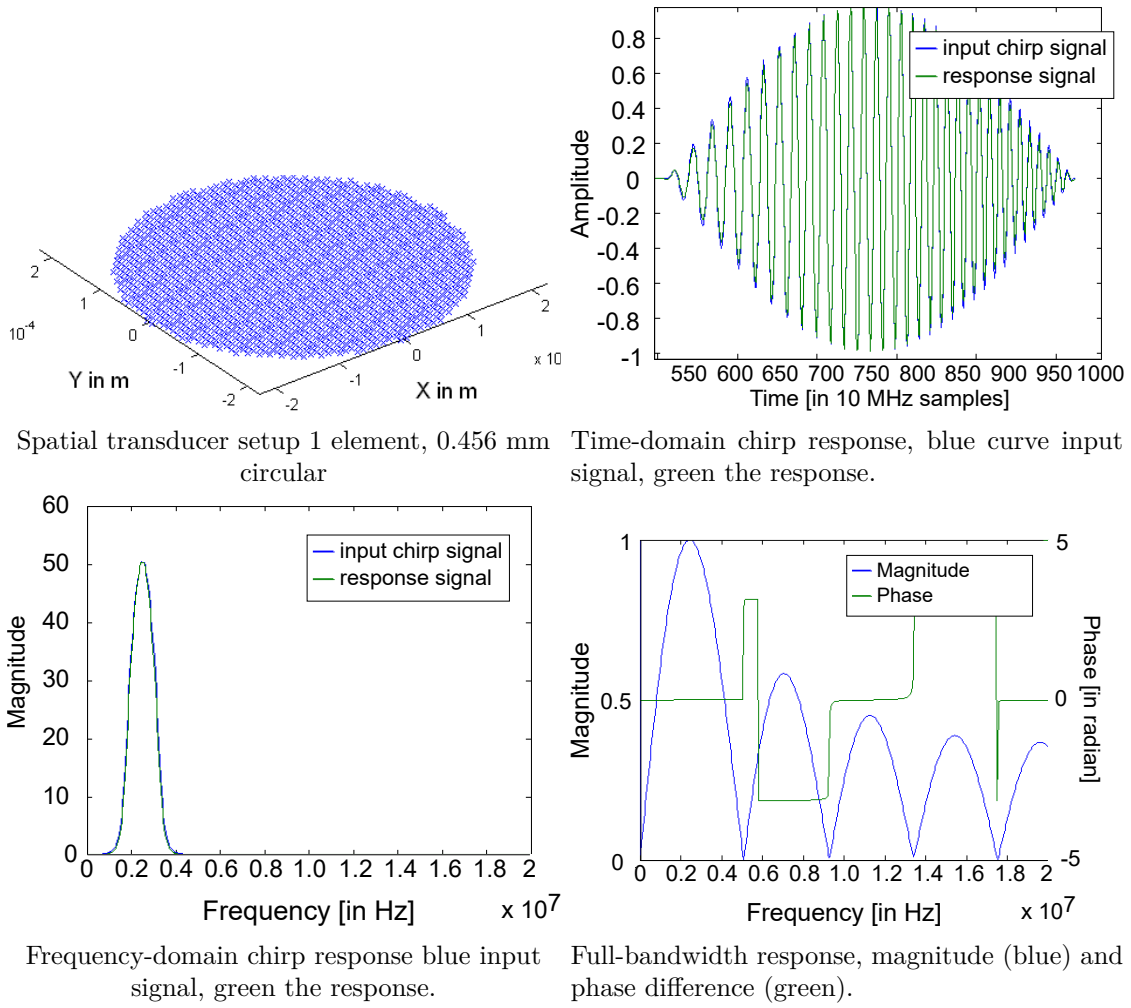


Figure 5.25: Frequency response for transducer 0.456 mm circular, RoI $20 \times 20 \times 15$ cm, characteristic at $[0.113 \text{ m}, 0 \text{ m}, 0.13\text{--}0.056 \text{ m}]$ (RoI border, worst case). Chirp: $f_0 = 2.5 \text{ MHz}$, bandwidth 1 MHz, cosinusoidal damped. Blue curves input signal, green curves response with transducer influence.

a phase-deviation-free response over the full required bandwidth.

5.4.5 Curved Surface Transducers

One last degree of freedom considered is planar versus curved surface transducers. The idea is that a curved transducer can have an increased opening angle without reducing the surface too much, finding another sound-pressure to opening-angle trade-off optimum.

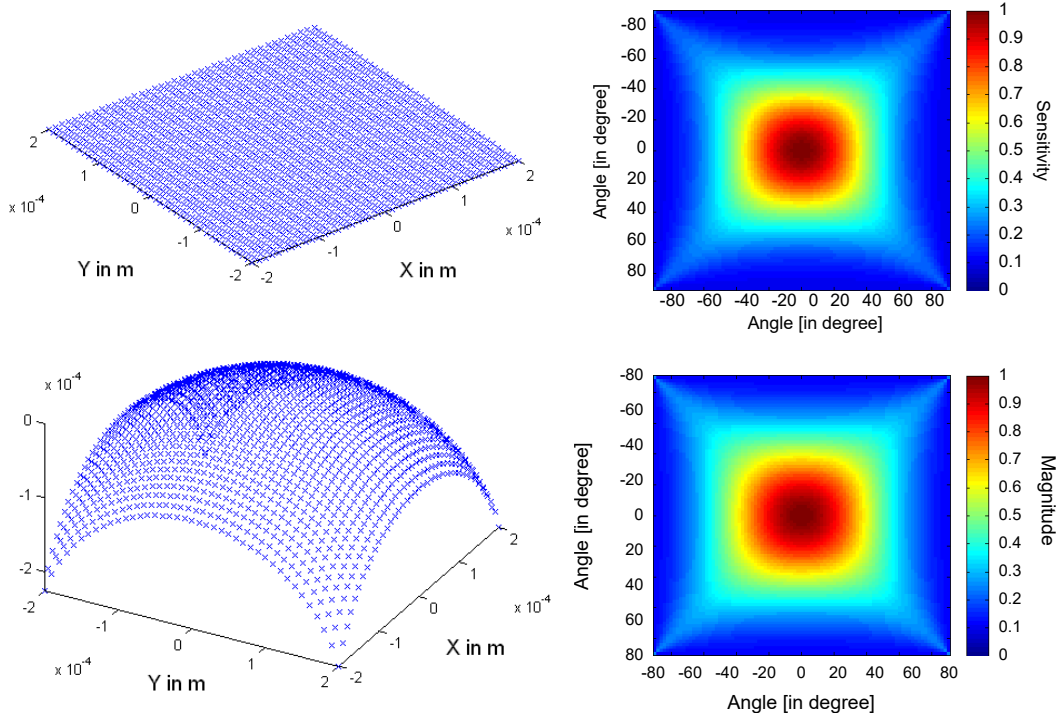


Figure 5.26: Far-field for $f_{max}=3.5$ MHz over 2D projection angles for curved square transducers of $0.4\text{ mm} \times 0.4\text{ mm}$. (Top row), curvature inf (flat surface). (Bottom row), curvature 0.29 mm (max. spherical). (Left column), spatial representation of the transducer. (Right column), far-field for $f_{max}=3.5$ MHz over 2D projection angles (Note: elevation and rotation, NOT a spherical projection)

The curvature is introduced as a design variation of transducer design two (square, 0.4 mm^2); see Fig. 5.26. The planar reference transducer is shown at the top, while the maximally spherically curved transducer (radius 0.29 mm) is shown at the bottom. The left column depicts the spatial aperture, whereas the right column shows the far-field amplitude response, represented in terms of elevation and horizontal angles.

Results for several other curvature variations, i.e. fresnel lenses, are given in Tab. 5.7. Overall, there is very limited improvement of the opening angle. Probably the transducer surface is too small, too similar to a point source for being capable to act as a sufficient lens.

Therefore a bigger transducer design was evaluated, regarding its performance. The third transducer design, the circular transducer aperture with a diameter of 0.902 mm was applied with a curvature. Fig. 5.27 shows as reference the perfor-

Table 5.7: *Beamwidth opening angle along x-axis for curved square transducer of 0.4 mm^2 in the far-field for 3.5 MHz. Defined as pressure drop given in the columns.*

Curvature radius	20%	3 dB (29.7%)	50%
∞ (flat)	46°	58°	84°
0.8 mm	46°	58°	84°
0.5 mm	48°	60°	84°
0.29 mm	52°	64°	92°

mance of the planar transducer in magnitude over elevation and horizontal angle (left) and on the right the phase response, again for 3.5 MHz. Fig. 5.28 shows the performance of the curved (radius 0.632 mm) transducer in magnitude over elevation and horizontal angle (left) and on the right the phase response for 3.5 MHz.

Fig. 5.29 shows the side by side performance of the two variations, green the curved transducer, blue the planar. The distinctive drop to zero magnitude at 38° for the planar transducer is smoothed out for the curved transducer but also the opening angle was not significantly improved. In the phase curve on the right, the planar transducer kept its zero phase error until 38° , while the curved transducer shows some smooth phase deviation over the angle.

In Fig. 5.30 we see the frequency dependent analysis for the curved transducer. Left image is the magnitude, right image the phase; x-axis the angle, y-axis the frequency between 0 and 3.5 MHz. Colour coded is the relative response, red is one, blue is zero. The general trend for broader opening angle for lower frequencies can be seen, which is not a specific feature for the curved transducer. On the right, the phase response is given. X-axis is angle, y-axis is frequency in Hertz. Colour coded is the phase in radian. Over the frequencies the trend is similar like for the 3.5 MHz frequency, the phase deviates slowly over the angle, for all frequencies.

Conclusion for the curved transducers, while some opening angle improvement is possible, there is always a phase shift which is additionally frequency dependent which is an undesired feature. This is unlike the planar transducer which has the beneficial property that there is zero phase shift until some specific angle. Hence, when designed and selected properly for a specific bandwidth and required opening angle an ideal planar transducer would impose zero phase shift over a frequency range, which is a very beneficial property. Also, the behaviour to impose only

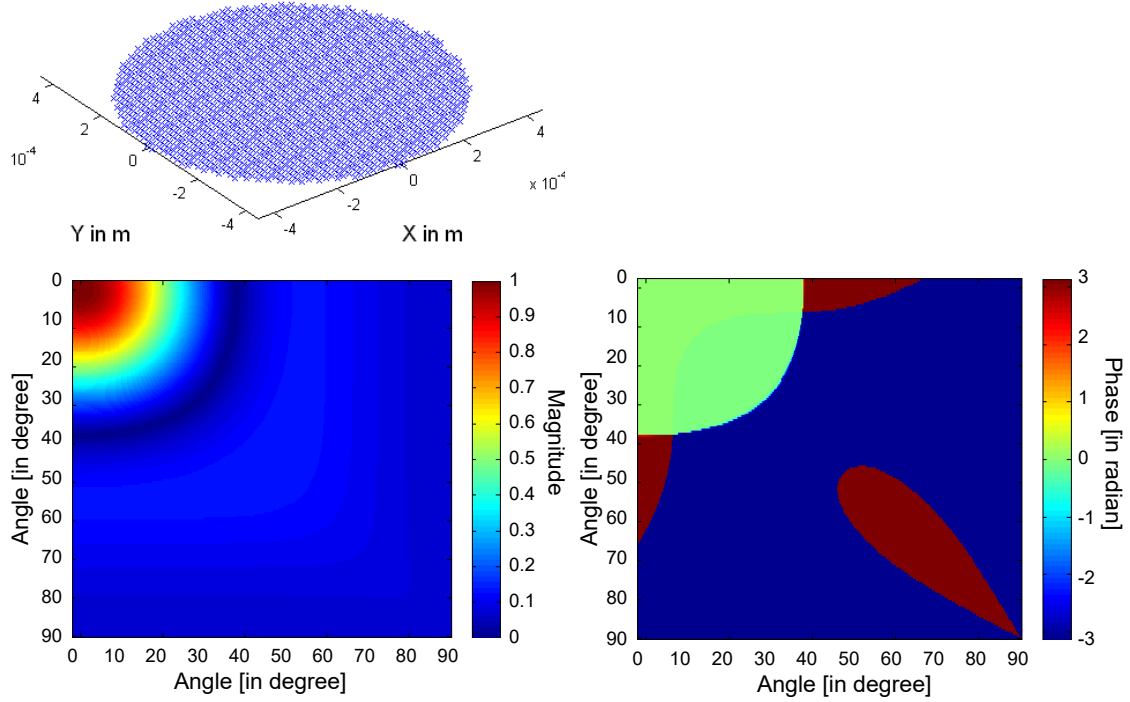


Figure 5.27: *Transducer Setup 3 element circular 0.902 mm (no curvature): Far-field for $f_{max}=3.5$ MHz over 2D projection angles. (Top left) Transducer aperture: 1 element circular (0.902 mm), curvature radius infinite (no curvature). (Bottom left) Magnitude over angles in far-field (for 3.5 MHz). (Bottom right) Phase in radian over angles in far-field (at 3.5 MHz).*

phase inversions (180° degree phase shifts) is a distinctive feature of the planar transducer which might be utilised as an easy to detect feature characteristic over the angle, most likely much easier to separate than a continuous varying phase shift over the angle.

As the goal for 3D USCT III is to utilise a broad bandwidth phase-correct for coherent focussed SAFT imaging, a curved transducer aperture design is determined to be a too risky design choice. It also imposes an additional source of phase uncertainty. Therefore, curved transducers are no longer being considered.

5.4.6 Transducer Array Aperture

The transducers forms an array in an housing, this unit is called a TAS. As the distribution of the transducer elements will be significantly larger than $\lambda/2$, the resulting array will be sparse. Their performance together regarding their side-

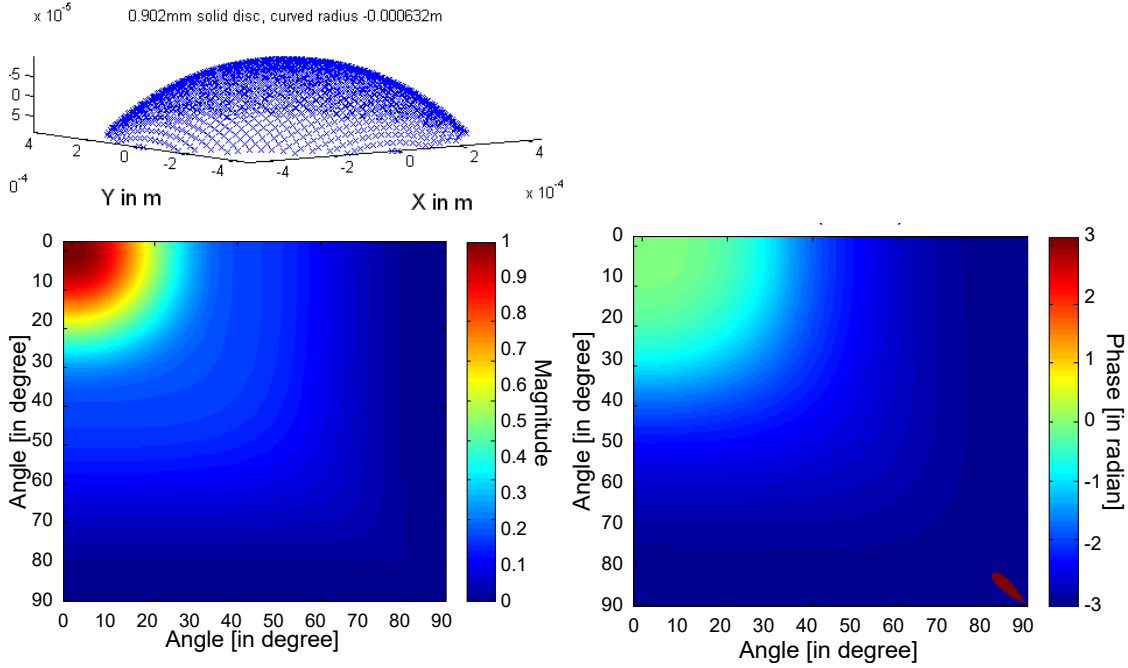


Figure 5.28: *Transducer Setup 3 element circular diameter 0.902 mm but with curvature radius 0.632 mm: Far-field for $f_{max}=3.5$ MHz over 2D projection angles for curved transducers. (Top left) Spatial transducer aperture, 1 element circular (0.902 mm), curvature radius 0.632 mm. (Bottom left) Magnitude over angles in far-field (at 3.5 MHz). (Bottom right) Phase in radian over angles in far-field (at 3.5 MHz).*

lobes is an important performance factor for SAFT and beam-forming imaging.

In Fig. 5.31 left we see a transducer array with 13 elements, like 3D USCT II, with an sparse but regular distribution. Around the center element, there are equi-angular and equidistant six elements in two circles placed. On the right image we see all transducers elements excited by a broad-band signal at the same time and their far-field response in a 2D angular space. Visible are significant side-lobes despite the broadband signal, due to the arrays sparseness.

In Fig. 5.32 we see an irregular transducer array aperture with 21 elements covering the same bandwidth. This design improves the radiation characteristics for the 3D USCT III. Around a center element are again two rings established, but this time prime numbered, seven and thirteen. The resulting irregular angular sampling leads to a much better suppression of side-maxima compared to the first case.

The zero phase shift situation between the elements' phase response of special

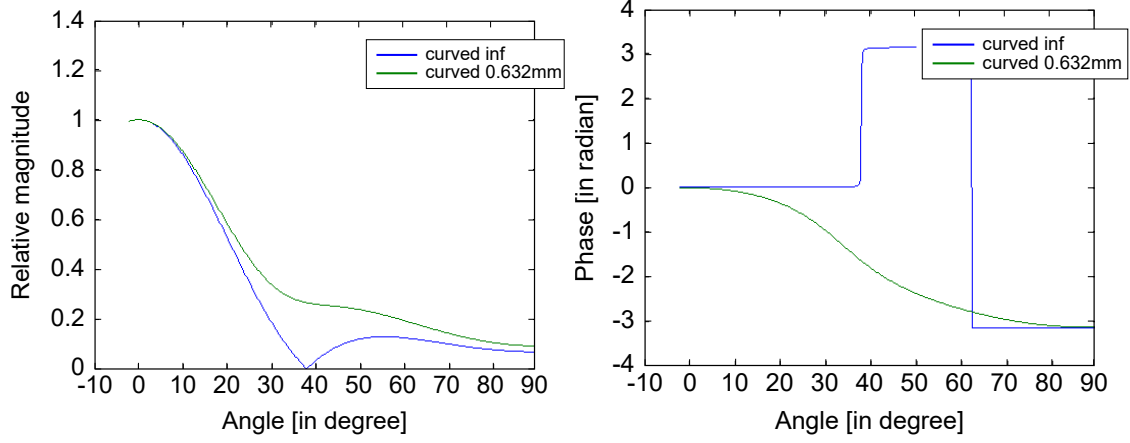


Figure 5.29: Comparison transducer Setup 3 with a transducer element between circular curved (green curves) and a uncurved (flat surface) transducer (blue curves). Far-field for $f_{max} = 3.5 \text{ MHz}$ over 2D projection angles for curved transducers curvature radius 0.632 mm , uncurved transducer. (Left) magnitude analysis, x-axis angle, y-axis relative magnitude. (Right) phase analysis x-axis angle, y-axis phase in radian.

importance among all beam-form combinations. The case zero phase is also the case which happens as cross-talk between the channels, therefore this case serves exemplary for all other cases. The second case has $\approx -8.5 \text{ dB}$ suppression between main-lobe to maximum sidelobe while the first case has $\approx 0 \text{ dB}$. The final array distribution selection is described in Sec. 5.6.

5.4.7 Axial Transducer Definition

After the core design of the surface aperture is defined for the transducer, another degree of freedom is left to decide: the axial dimension of the transducer, i.e., the thickness of the piezoelectric material.

While there are known recipes for achieving small-bandwidth transducer designs with maximum pressure output, the case for USCT aiming for a broadband solution is more complex.

Transfer in general from the piezo to the medium can be considered an impedance-matching problem guided by the acoustic impedance of the components, defined as

$$Z_{\text{MRayl}} = \frac{\rho \cdot c}{10^6} \quad (5.22)$$

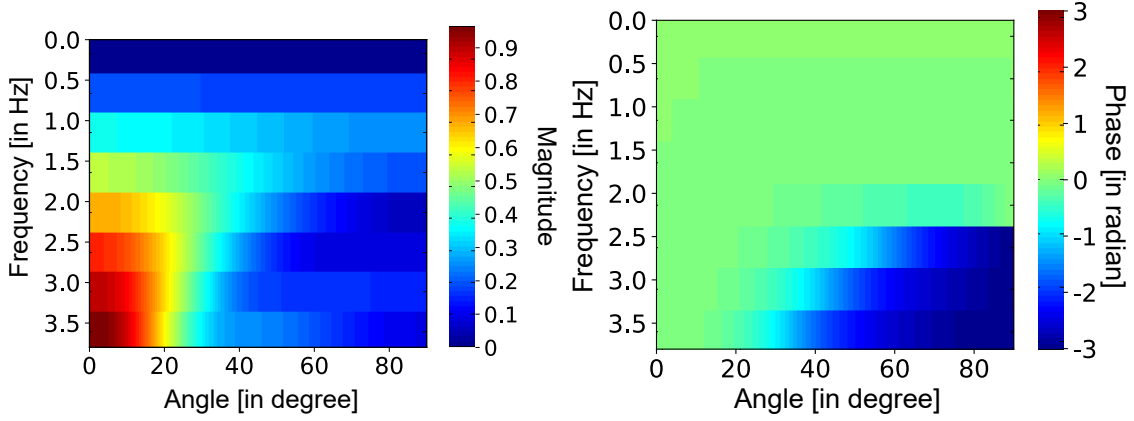


Figure 5.30: *Transducer Setup 3 element circular, curved 0.632 mm: Far-field for f_{max} 0 to 3.5 MHz over 2D projection angles for curved transducers curvature radius 0.632 mm. (Left) magnitude analysis x-axis angle, y-axis frequency in Hz. (Right) phase analysis in radian, x-axis angle, y-axis frequency.*

Table 5.8: *Acoustic impedances of materials relevant for 3D USCT III.*

Material	Density ρ (kg/m ³)	Sound speed c (m/s)	Acoustic impedance $Z = \rho c$ (MRayl)
PZT (PIC255)	7410	4000–4500	29.6–33.3
Water (20–25°C)	1000	1480	1.48
Polyurethane (PU)	1600	≈ 1500 (typ.)	2.40
TMM4 (Rogers)	2105	≈ 2500 (est.)	5.26

where ρ is the material density and c the sound speed.

Typically, one or two matching layers made of materials with an intermediate impedance are applied to improve the transfer of acoustic energy from the piezo to the water, together with a piezo backside damping layer for increasing bandwidth by attenuating residual oscillations. Proper selection of materials and dimensions for the matching layer(s) and the backing of the transducer has a significant impact on either improving or impeding the pressure efficiency and broadband performance of the device.

Previous USCT transducer designs employed a single matching layer made of Rogers TMM4 and a Flexovoss PU backing, which served as a starting point [Gö2]. To compute the thickness of the different layers (backing, PZT, matching) a 1D KLM model[KLM70], implemented in MATLAB is employed.

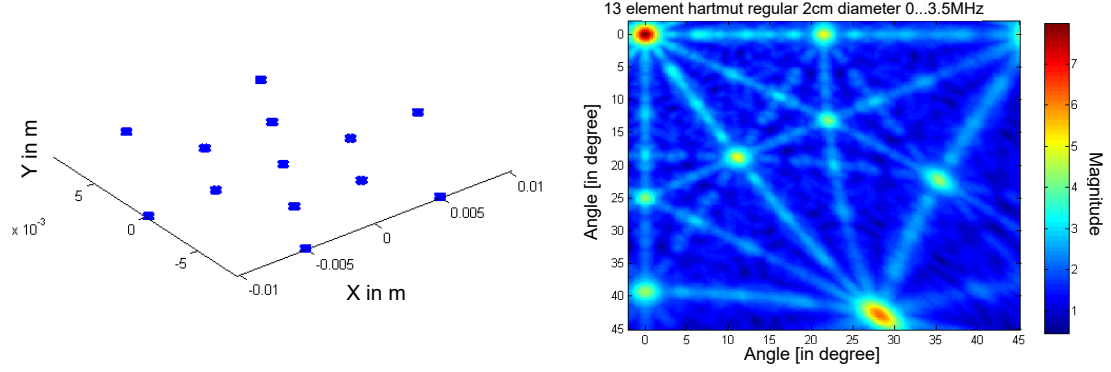


Figure 5.31: Transducer array 13 elements angle-regular, 2 cm array diameter, transducer element 0.456 mm circular. (Left), spatial setup, x-axis in m, y-axis in m. (Right), Far-field wave response under excitation all elements. Crosstalk case and 0-phase beam-form case. Integrated Amplitude for Frequency 0 to 3.5 MHz (red big, dark blue small). Over emission angle x and y in degree.

First, the 3D USCT II transducer setup was reproduced and tuned until it fitted the measured metrics reasonable, see Fig. 5.33.

Then, potential options for multilayer matching layer solutions like with Araldit, POM and aluminium were considered (and partly tested) for potential energy transfer efficiency benefits. This experiment was dropped early on for complexity concerns and the focussed was put on the established single TMM4 matching solution, which still seems a reasonable solution compromise between performance and complexity.

Finally, TMM4 single layer options for a more broadband 3D USCT III transducer were considered with 2D KLM modeling, see Fig. 5.34. Top graph is again the 3D USCT II performance as reference, visible are the two strong resonances around 3 MHz, the (primary) resonance under 1 MHz is very weak. For the 3D USCT III transducer proposal, a classical $\lambda/4$ design shifts the resonance originally below 1 MHz to higher frequencies and broadens the bandwidth. With a 100 μm thicker TMM4 layer, the resonance could shifted somewhat lower without losing as much energy as with the 3D USCT II design, offering another interesting bandwidth compromise in design.

As a result, the following two options are considered as good starting points by a 2D KLM optimum search for a Piezo of 550 μm thickness

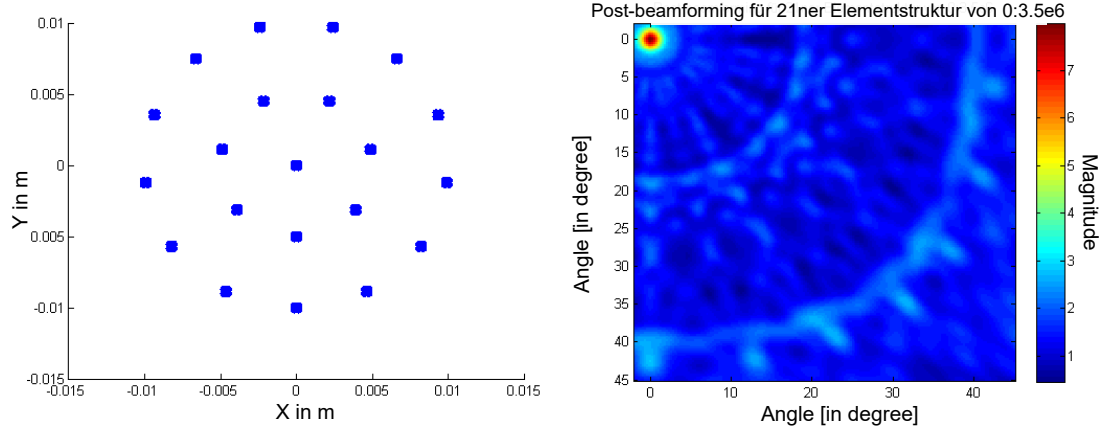


Figure 5.32: *Transducer array angle irregular setups: 21 elements angle irregular, 2 cm array diameter, transducer element 0.456 mm circular. (Left), spatial setup, x-axis in m, y-axis in m. (Right), Far-field wave response under excitation all elements. Crosstalk case and zero-phase beamform case. Integrated amplitude for Frequency 0 to 3.5 MHz (red big, dark blue small). Over emission angle X and Y in degree*

- KLM: TMM4 414 -200 μm ($\frac{\lambda}{4}$ for 2.5 MHz).
- KLM: TMM4 414 -100 μm .

for later implementation, see Sec. 6.4.4.

5.5 Aperture: Increasing Spatial Transducer Sampling Density

With the 3D USCT II system, the transducer surface sampling is very sparse but regular, see Fig. 5.35 left. To compensate for the large gaps, it is essential that the aperture could be moved mechanically. This strategy allows the system to emulate a much denser transducer sampling pattern and, in turn, reduces the hardware costs significantly: the number of TAS elements can be synthetically increased by a factor of 12, corresponding to the number of motor positions, see Fig. 5.35 right. The downside of this approach is a substantially increase in overall measurement time.

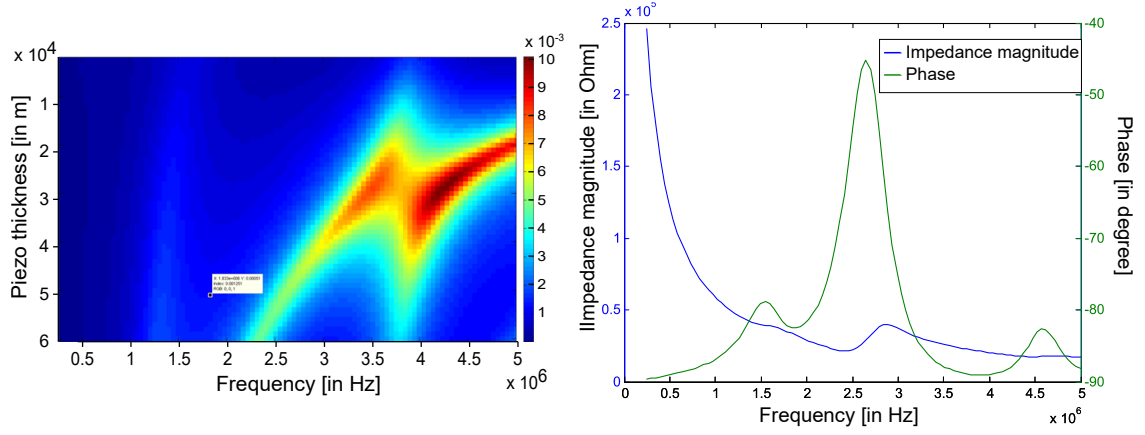


Figure 5.33: KLM modelling for the 3D USCT II fitted with Piezo thickness $550\text{ }\mu\text{m}+49\text{ }\mu\text{m}$, TMM₄ $414\text{ }\mu\text{m}-100\text{ }\mu\text{m}$. (Left) 2D response over frequency (x -axis) and TMM₄ matching layer thickness (y -axis). (Right) impedance derived with magnitude in Ohm (blue) and phase in degree (green).

Rotation and lift requirements for 3D USCT III aperture

With the 3D USCT III it is intended to overcome the sparseness of the spatial sampling present with the 3D USCT II device, while a fully densely sampled aperture is out of technical reach. To overcome this problem, the question arises of how many degrees of movement freedoms are required for an optimal sampled 3D USCT III pseudo irregular aperture. The following degree of freedom options are considered as technical realisable options: no mechanical movement at all, rotation around the z -axis, lift along the z -axis and the combination of both.

To select the optimal movement degree of freedom, an aperture correlation function based on self-similarity is minimised. The function covers the rotational space from -180° to $+180^\circ$ and a lift along the z -axis from 0 up to 3 cm.

The optimisation is implemented as a 2D correlation in the Fourier domain. Along the rotational dimension, the periodicity of the Fourier-space is utilised over its border. Along the z -axis (lift), periodicity was deliberately suppressed using zero-padding.

As expected, adding more mechanical degrees of freedom reduces self-similarity for both 3D USCT II and III (Fig. 5.36 top row). In both systems lift reduces self-similarity slightly better than rotation. The 3D USCT III is overall “tighter” because it does not have the periodic gaps that are present in 3D USCT II.

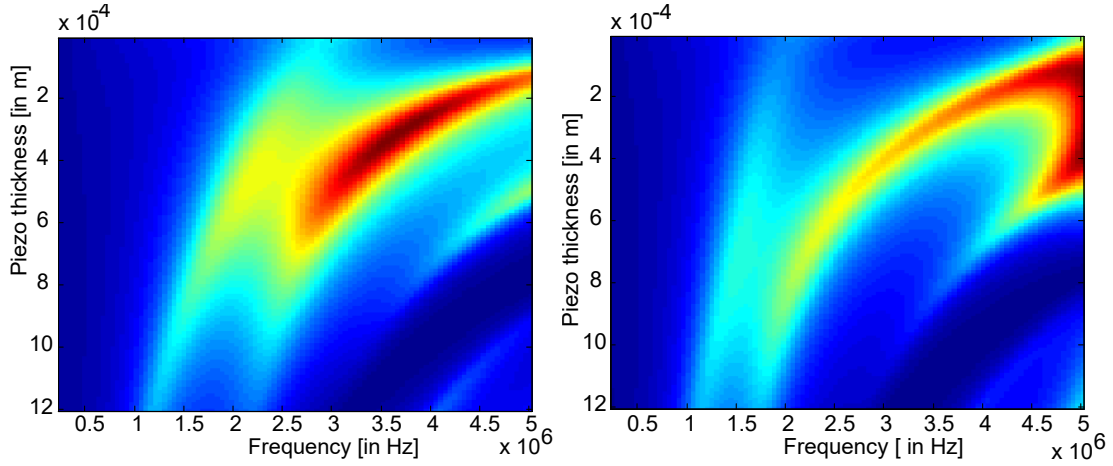


Figure 5.34: 2D KLM based TMM₄ matchinglayer and transducer selection, x-axis is frequency (0 to 5 MHz), y-axis is PZT thickness (0 to 1200 μm). 3D USCT III options: (Left) $\lambda/2$ (for 2.5 MHz) matching layer with TMM₄ (414 -200 μm $\lambda/4$, 2.5 MHz). (Right) manually tuned solution with 313 μm TMM₄.

The relative performance results (Fig. 5.36 bottom row) show that for 3D USCT III with 10 motor positions that uses only lift is just 3.4 % worse than using both degrees of freedom, while using only rotation is 20.1 % worse. For 3D USCT II, the penalties for losing a degree of freedom are significantly larger.

In the end, the decision was made to keep both rotation and lift in 3D USCT III, just like in 3D USCT II, to retain the option of filling the sampling aperture as optimally and densely as possible, especially for research and phantom studies.

5.6 Aperture: 3D USCT III Transducer Distribution

After defining the core design principles, such as aperture diameter, transducer shape and size, and the available mechanical degrees of freedom, a global optimisation was conducted to determine both the final layout of the overall aperture and the individual TAS elements.

The argument for using *irregular apertures* to minimise grating lobes (i.e., spatial aliasing caused by sparsely sampled apertures) follows directly from compres-

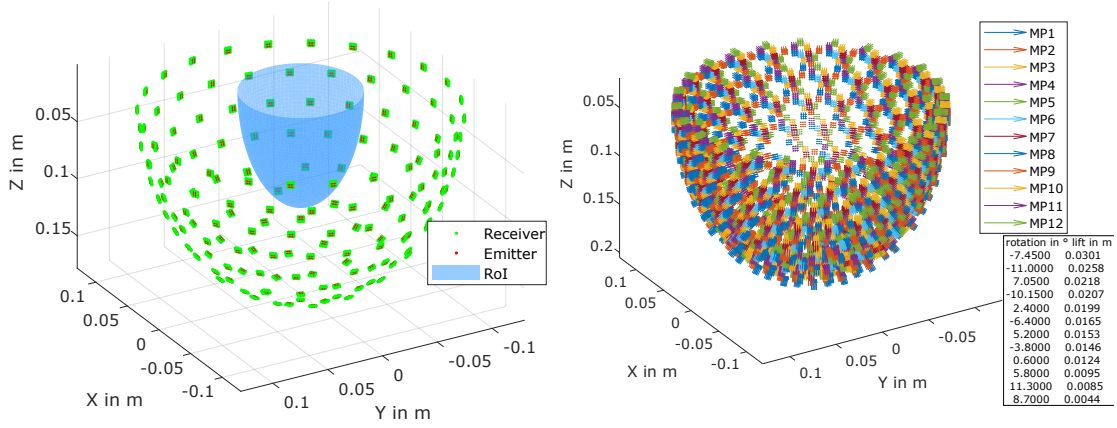


Figure 5.35: 3D USCT II aperture: (Left) one motor position, resulting in 880k AS-cans. (Right) synthesised denser 3D USCT II aperture with 12 motor positions (11 movements), color-coded in the graph, for 10 million resulting AS-cans. Compare with the 3D USCT III aperture shown in Fig. 5.39.

sive sampling theory. Irregular, non-periodic sampling patterns suppress coherent aliasing structures and therefore reduce strong, periodic grating lobes that arise in regularly spaced sensor distributions [SHDR07]. This is particularly relevant for USCT, where the aperture is necessarily sparse due to construction constraints.

To investigate this design space, the large number of possible transducer distributions is explored using Monte Carlo simulations. The evaluation criterion included image contrast, coverage of the RoI and the minimisation of characteristic reconstruction artifacts [GBH⁺18]. Since the transducer set must be manufacturable at industrial scale, the TAS design was constrained to be identical for all array elements.

Compared to the 3D USCT II system, the coverage of the exposed aperture surface has been significantly improved. The earlier system exhibited very low effective surface utilisation, whereas the optimised 3D USCT III configuration achieves a far more efficient spatial coverage of the available aperture. The resulting distribution of TAS elements for 3D USCT III is shown in Fig. 5.37 left.

Despite being identical produced, the 3D USCT III TAS have another possibility for further spatial randomisation following the compressive sensing concept. In addition to the global uniform random distribution of the transducers an irregular pattern with varying angle differences between the transducers should be achieved.

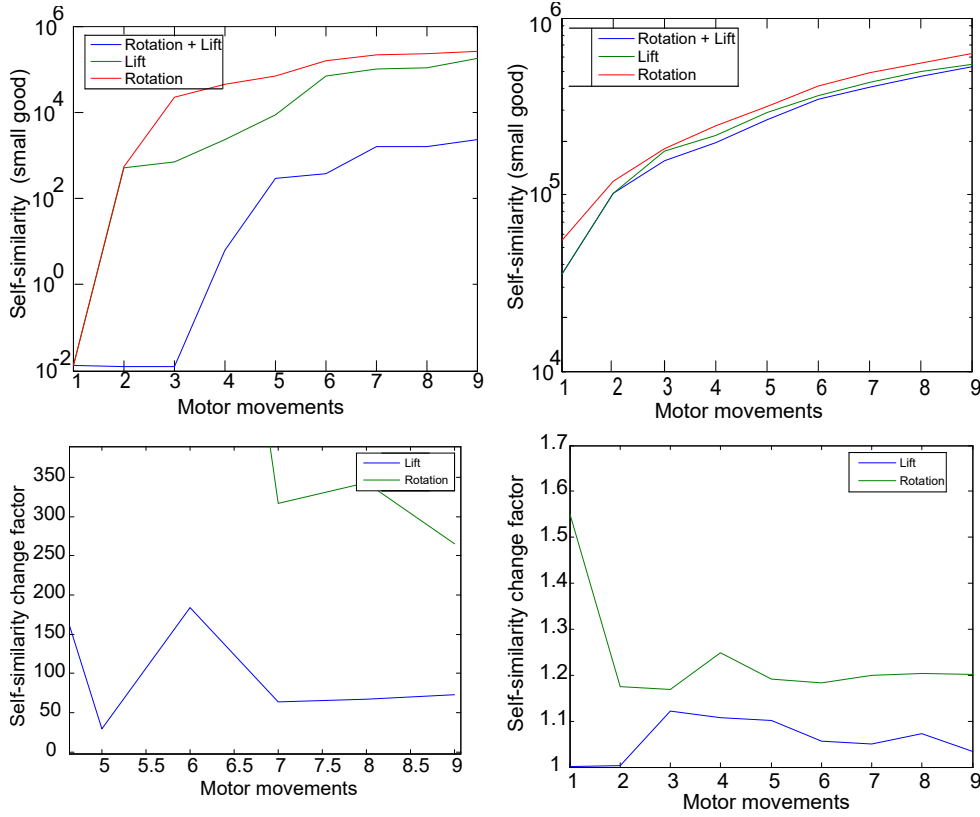


Figure 5.36: Self-similarity analysis for the 3D USCT apertures and their degrees of mechanical movements. (Top row) absolute comparison, red curve rotational degree of freedom, green curve is lift degree of freedom, blue curve are both, y-axis self-similarity (small is good), x-axis motor positions. (Top left) 3D USCT II performance over 10 motor positions, (Top right) 3D USCT III over 10 motor positions. (Bottom row) relative performance of only lift or rotational degree of freedom over the combined one. Blue only lift vs combined, green only rotation vs combined. (Bottom left), 3D USCT II; (Bottom right), 3D USCT III.

As solution, the TAS allow for an individual rotation in a pseudo-random pattern: the mechanical mounting design provides a bayonet like locking mechanism offering accurate rotational mounting with three discrete mounting positions of 120° degrees of freedom. The outer stainless steel ring where the bayonet connection snaps in offers seven additional rotational positions. Overall, each TAS can be mounted in 21 different rotational configurations (see Tab. 5.9), enough variety for approximating arbitrary positions, see Fig. 5.38.

Furthermore, to achieve the desired uniform randomness on the aperture surface, the TAS's need to be distributed optimally. For this, the aperture surface is subdivided into equal areas and the frequency distribution of the hits in these

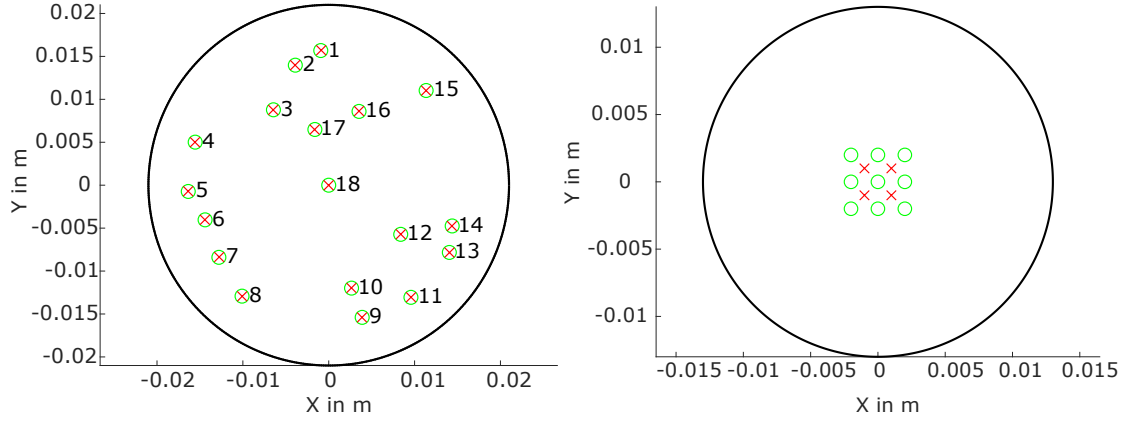


Figure 5.37: Transducer distribution on the TAS. Green circles are receiver positions and red are emitter positions. (Left) 3D USCT III TAS transducer distribution, almost homogeneously covering the exposed surface. (Right) 3D USCT II TAS transducer surface, the transducers are tightly and regular, clustered in the center.

areas by the position of the transducers is analysed by a chi-square test similarly to the test of uniform random number generators. For a globally uniform transducer distribution the TAS's have to be spaced as tightly as possible, i.e. the area of the surface not covered by the TAS's should be minimal. The density of aperture surface coverage with transducer arrays was increased from 75 % in 3D USCT II to 80 % in 3D USCT III. The result for 3D USCT III is shown in Fig. 5.39, the reference 3D USCT II aperture is depicted at Fig. 5.35.

5.7 Excitation Chain for 3D USCT III

Resulting from above aperture and transducer decision, requirements for the signal chain, see Fig. 5.40, composed out of the excitation (emission) and DAQ (reception) until imaging components result.

5.7.1 Signal-level SNR Gain

When considering repetition rates, acoustic fade-out times, and the goal of minimising the overall measurement duration, a fundamental trade-off emerges: excitation length must be balanced against the risk of overlapping with the next temporal measurement window. This trade-off is governed by the size of RoI, and

Table 5.9: Bayonet rotation configurations of a single TAS. Angles are defined mathematically positive (counter-clockwise) with respect to the water container reference; due to external mounting, all effective rotations appear negative.

Configuration level	Rotation angles (deg)
Relative bayonet positions (A, B, C)	120°, 120°, 120°
Absolute bayonet offsets	0°, -120°, -240°
Pin-hole increments	0°, -17°, -20°, -16°, -19°, -18°, -17°
Cumulative pin-hole angles	0°, -17°, -37°, -53°, -72°, -90°, -107°
Group A (base pos.)	0°, -17°, -37°, -53°, -72°, -90°, -107°
Group B (offset -120°)	-120°, -137°, -157°, -173°, -192°, -210°, -227°
Group C (offset -240°)	-240°, -257°, -277°, -293°, -312°, -330°, -347°

the device’s aperture geometry.

The optimisation aims at determining the excitation length that maximises the SNR while ensuring that the coded excitation and its acoustic tail decay before the arrival of the first reflection in the next time window. Increasing the coded-excitation (CE) length improves the signal level proportionally, whereas the SNR increases only with the *square root* of the extension factor.

For the 3D USCT III design, several geometric and hardware-related changes compared to 3D USCT II affect the available SNR:

- system diameter increased from 26 cm to 35 cm,
- effective transducer surface reduced by a factor of 4.

The combined SNR scaling factor is therefore:

$$\left(\frac{26}{35}\right) \cdot \left(\frac{1}{4}\right)^2 = 0.046. \quad (5.23)$$

Thus, an SNR factor of approximately 21.54 must be recovered to achieve SNR neutrality relative to 3D USCT II.

Assuming that an excitation-voltage increase from 20 V_{pp} (3D USCT II) to around 100 V_{pp} (3D USCT III) provides an additional SNR factor of about 5, a remaining

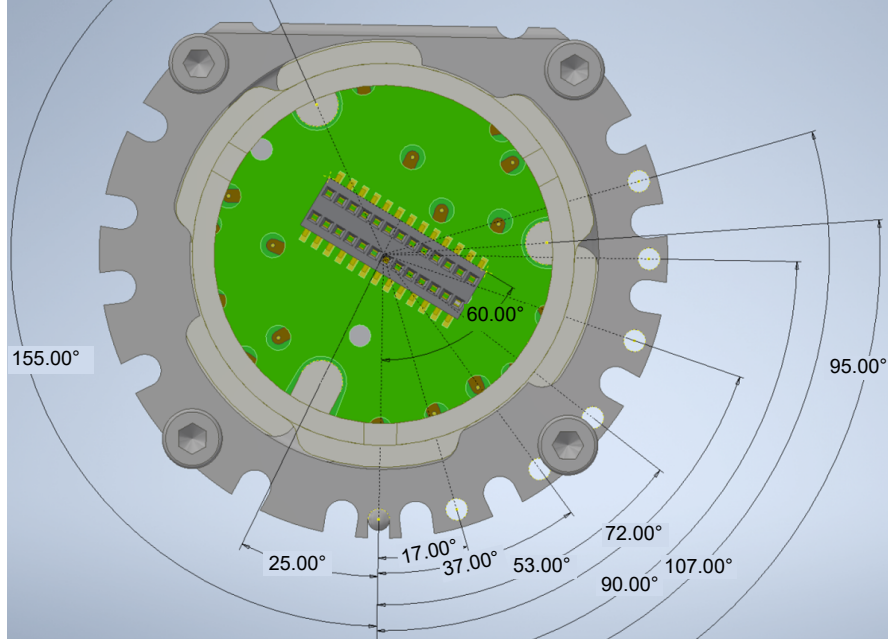


Figure 5.38: Rotational degree of freedom provided by a stainless-steel bayonet ring. The rotations are defined counter clockwise/mathematical positive from perspective of the water container. As mounting of the bayonet rings happens from the outside view, the angles are negative (going clockwise direction).

factor of roughly 4.3 must be compensated through coded-excitation extension. Since SNR grows with the square root of the CE extension, its length must be extended by

$$4.3^2 \approx 18.5. \quad (5.24)$$

So the CE must be extended by a factor of about 18.5. With the standard CE length of $12.7 \mu\text{s}$, the extended excitation becomes:

$$12.7 \mu\text{s} \times 18.5 \approx 235 \mu\text{s}. \quad (5.25)$$

The measurement window of $400 \mu\text{s}$ must therefore be enlarged by:

$$235 \mu\text{s} - 12.7 \mu\text{s} = 222.3 \mu\text{s}, \quad (5.26)$$

which corresponds to:

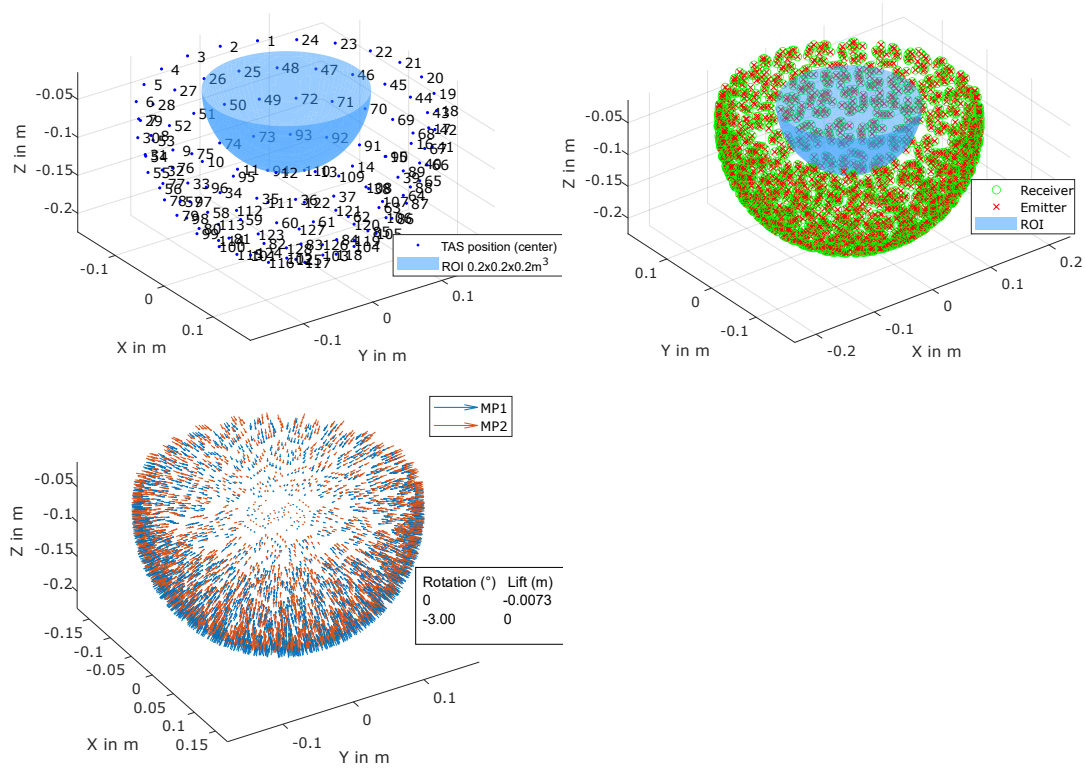


Figure 5.39: 3D USCT III aperture: (Top left) TAS positions of one motor position with RoI. (Top right) all transducer elements with RoI, $0.2 \times 0.2 \times 0.2 \text{ m}^3$. With this configuration 5.3 million ASans are obtained. (Bottom left) direction vectors of the transducers for two motor positions, together 10.6 million ASans. Compare with the 3D USCT II aperture shown in Fig. 5.35.

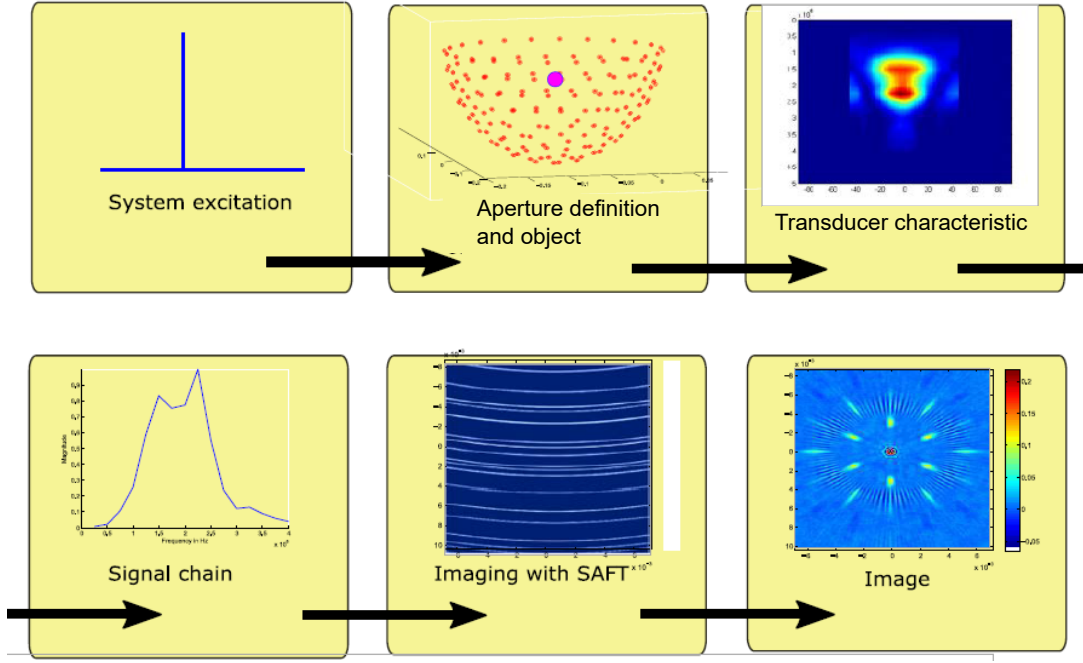


Figure 5.40: 3D USCT II and III imaging chain: from excitation over transducer and system frequency characteristic to a SAFT image.

$$\frac{114.3}{400} \approx 0.556, \quad (5.27)$$

i.e. a 556% increase in the measurement-window duration.

Achieving the same SNR gain by averaging requires repeating the entire measurement window leads to the following increase in measurement time

$$4.3^2 = (18.5 - 1) \Rightarrow 1750\% \text{ longer measurement time.} \quad (5.28)$$

Therefore a CE extension would lead to only $\approx 55.6\%$ overall measurement time increase, compared to an equivalent averaging approach with an $\approx 1,750\%$ measurement time increase. Thus, CE extension is dramatically more time-efficient than averaging and should be preferred whenever possible. Coded-excitation extensions provide a highly time-efficient means for SNR improvement and should be maximised to minimise the need for averaging.

The following design consequences arise as hardware implications. The us-

able measurement area before the arrival of reflections increases from ≈ 26 cm (3D USCT II) to approximately 40 cm (desired), with the implemented compromise of 35 cm driven by clinical constraints (water volume, patient access, door width). The excitation chain must support significantly longer chirps than the 3D USCT II limit of $12.7 \mu\text{s}$ (511 samples at a 40 MHz DAC buffer). A minimum $18\times$ increase is required; 3D USCT III implements a 32k-sample buffer, satisfying this criterion.

5.8 Summary

In this chapter core aspects of the system design are described and how they are decided, typically by an imaging simulation driven approach. The next chapter describes the actual realisation.

Decisions are made with respect to the aperture diameter, transducer aperture geometry and DAQ channel and MUX channel matchings. Questions concerning the optimisation of specialised emitter and receiver transducers and mechanical translation and rotation degrees of freedom are discussed.

The following design decisions for the system are made:

- RoI adjustment addressed the requirements for extending the RoI, while maintaining adequate imaging quality of the system as whole.
- Design of an optimal aperture for the updated RoI requirements.
 - Optimisation analysis of emitter-receiver ratios.
 - Analysis and design of an optimal DAQ system with regard to measurement time, and cost efficiency.
- Systematic analysis of the influence of frequencies bands for the imaging system as a whole. Resulting in upper and lower frequencies required, ≈ 0.9 MHz and ≈ 3.8 MHz for the aimed object structures.
- Transducer's form (aperture):
 - circular and planar transducers.

- Newly designed transducer system leads to two central improvements of the system:
 - Increased number of useable emitter-receiver combinations despite reduction of number of TAS, being more cost efficient and achieving a higher integration density.
 - Largely reduced number of necessary motor positions to reach required number of AScans which reduces the measurement time.

TAS requirements had been accumulated, going significantly beyond the established dice-and-fill piezo technology available at the IPE. Implementing these requirements represented one of the largest engineering challenges of the entire project:

- homogeneous emission and reception fields (ideally using round transducers),
- increase of the opening angle from $\approx 37^\circ$ to $\approx 60^\circ$,
- if possible, increase of bandwidth,
- irregular and arbitrary positioning of transducer locations,
- unified emitter-receiver design to support algorithms relying on identical transmitter and receiver behaviour and position.

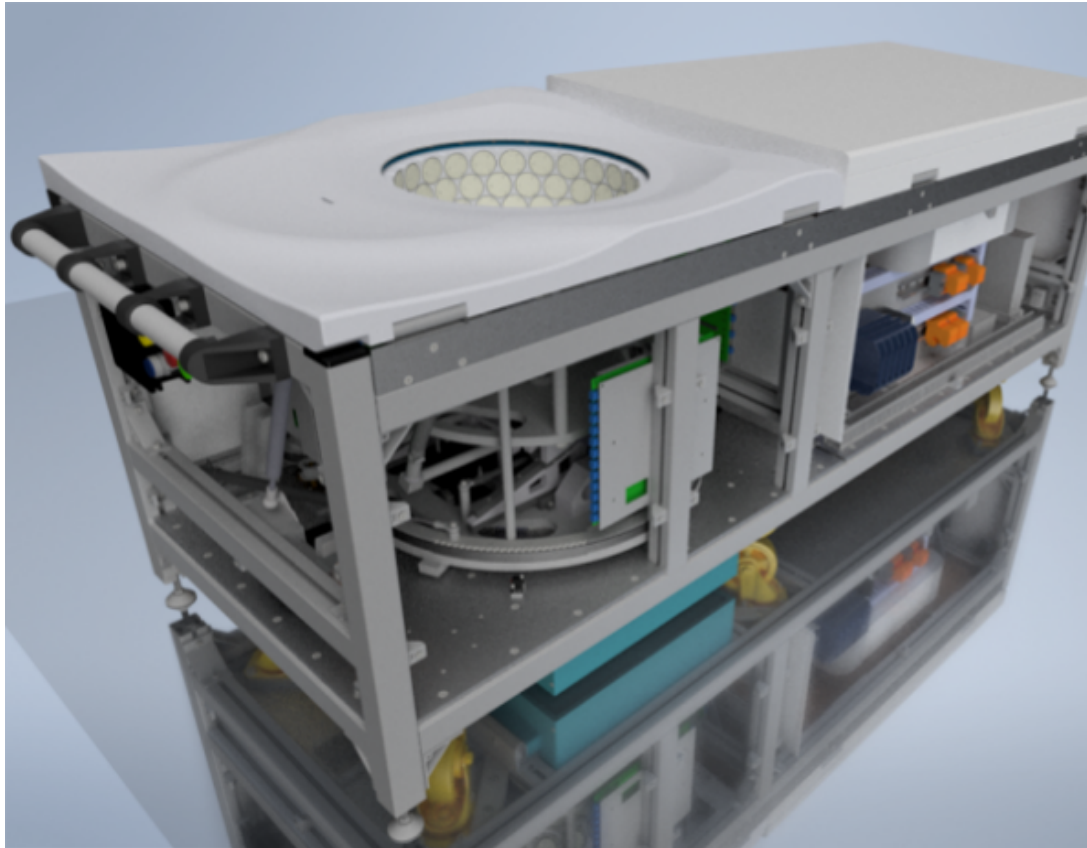


Figure 6.1: *3D USCT III early CAD rendering of the mechanical setup, housing and patient bed. Vision integrating previous design ideas, implementation pending.*

6 KIT 3D USCT III Implementation

This chapter addresses the actual implementation based on the previous established abstract design concepts. For the components which forms the 3D USCT system implementation, technological solutions need to be selected. Besides the functional aspects also regulatory and safety considerations as well as economical aspects need to be considered.

6.1 System Components

The engineering work began with identifying the critical components, followed by the question what was already available on the market and what existed inside KIT or at IPE.

Following the experience with 3D USCT II the following components would need to be custom-designed: a new digital and analogue DAQ digitization card; a new front-end board (FEB) connecting the transducers to the electrical chain; and, for increased density, an IPE in-house ASIC solution, since a discrete solution was not sufficiently integrated. Moreover, an intermediate board for channel decoupling and channel distribution for the many reception and emission channels (the “distribution board”) was again required. A high-density cabling solution between the FEB and the distribution board needed to be developed, as well as a digital serial protocol between the FEB and the DAQ. In addition, μ C firmware for the FEBs, acting as “smart sensors”, as well as FPGA firmware for the DAQ FPGAs and DAQ software integrating all components were required. Ideally, the software should provide output data compatible with the previous USCT generation.

As result, the following two categories were identified:

- Custom components: mechanical system, transducer arrays, FEB, DAQ cards, distribution board, firmware, software.
- Stock hardware components: power supply, DAQ crate and backplane, DAQ PC module, fans, motors, etc.

As DAQ solution a μ TCAv4 platform based on telecommunication industry standards was selected. The transducer design described in Chapter 3 and Sec. 5.8, goes significantly beyond the established dice-and-fill piezo technology available at IPE. Implementing these requirements represented one of the largest engineering challenges of the entire project.

6.2 Medical Device Design and Regulatory Considerations

Besides the technical implementation, significant constraints imposed by medical-device requirements had to be considered early on in order to avoid costly dead ends and time-consuming delays caused by regulatory approval issues following from law and regulations.

Based on specialist feedback and course material on medical-device regulatory requirements for software and hardware, it was concluded that compliance with the international standard IEC 60601 would be crucial for the development process in both the hardware and software domains. The medical safety concept for the 3D USCT III must therefore ensure redundant and double-layered protection according to the IEC 60601 standard across all relevant domains, including electrical safety, US safety, mechanical risks, and, increasingly important, the medical software. The system design should follow the established hierarchy of prioritising inherently safe design over safety mechanisms and, only as a last resort, instructions for safe use.

A central part of the electrical safety concept is the strict electrical isolation between the patient area (direct and indirect patient contact) and the general electrical components. This separation provides a robust barrier between the patient and all high-voltage or DAQ components. In the patient area, full insulation is implemented wherever possible; where insulation cannot be applied, protective earth is used as a secondary safety measure, making the 3D USCT III a hybrid Class I / Class II electrical device.

A core component here is the TAS: the insulation of the TAS itself was significantly reinforced compared to the design of 3D USCT II. In 3D USCT II, the patient-facing TMM4 layer (insulating aluminium oxide) did not cover the complete TAS surface and was therefore prone to water ingress at the boundary regions, see Fig. 4.6. Another disadvantage was that the 3D USCT II transducers exhibited small leakage currents, which could cause unintended malfunctions if not fully isolated. Material selection played an important role: the stainless-steel housings of 3D USCT II were insufficiently insulated, and several short-circuited TAS units



Figure 6.2: (Left) 3D USCT II stainless-steel housing. (Center) 3D USCT II.5 prototyped housing from PEEK material. (Right) Final mechanical TAS housing for 3D USCT III, fully integrated and laser marked.

in the 3D USCT II system illustrated this problem. This weakness was solved in 3D USCT III by applying a TMM4 insulation layer covering the entire TAS face up to the housing rim and by introducing a double-bonding process.

Other changes on the TAS housing were the switch from stainless-steel body material, which failed the electrical isolation safety concept (Class I) of 3D USCT II, to PEEK (Polyether ether ketone), see Fig. 6.2 center. PEEK is a high-performance polymer that can be milled and drilled similarly to steel, while also being biocompatible and electrically insulating, fulfilling the medical requirements.

Most internal system supply voltages remain at or below 48 V DC, ensuring inherent electrical safety. The only unavoidable exception being the transmit (TX) signals for functional reason. These signals exceed 48 V, since this level is required to reach sufficient US emission pressure.

As the central electrical supply component, a medical power supply - a TDK lambda QM8 was chosen - solving challenges left from 3D USCT II related to medical electrical certification and leakage currents. The QM8 provides, in the 3D USCT III configuration, ± 48 V, 24 V, and ± 5 V, with a peak power of 1500 W.

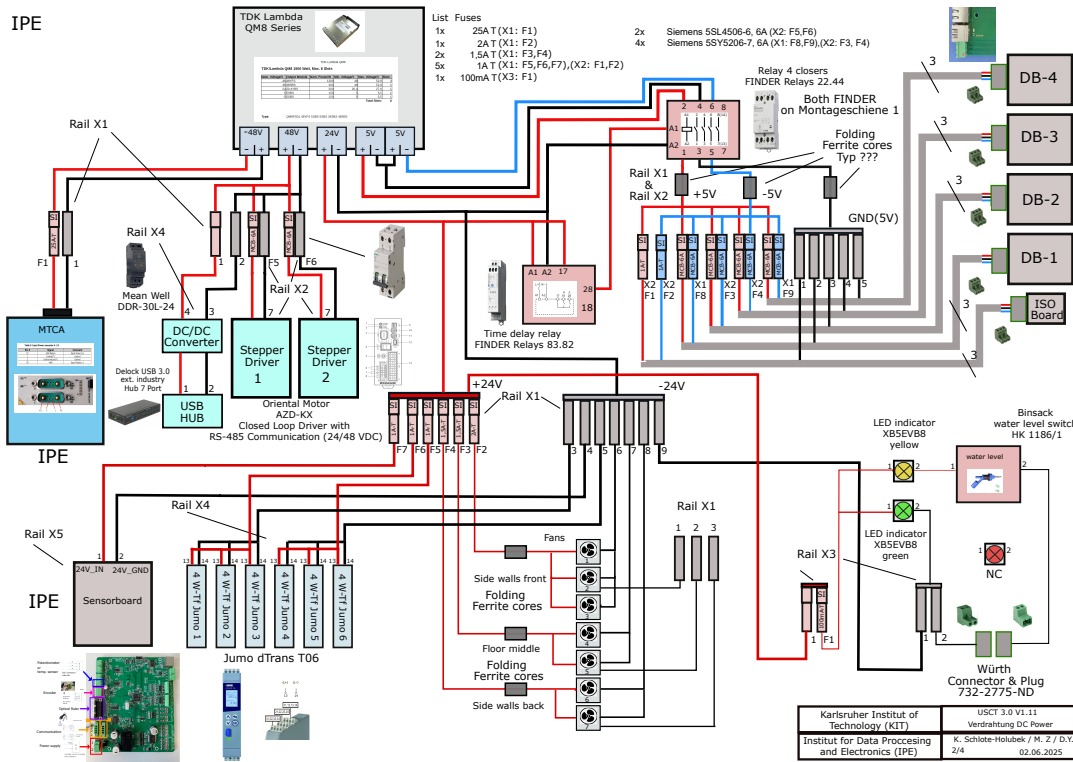


Figure 6.3: Block diagram electronic components of 3D USCT III: top left component is the medical-grade power-supply TDK-Lambda QM8 as core component of the medical electrical safety concept.

As it can be seen in Fig. 6.3 the QM8 provides centrally the secondary DC based supply for the many diverse components and sits directly after the AC power inlet with power line filter.

Furthermore, the DAQ, FPGA, and microcontroller subsystems are designed such that no software behaviour or configuration can lead to a hazardous situation for either the patient or the operator. According to IEC 62304, this classifies all DAQ-related software items (measurement scripts, C code, μ C firmware, VHDL) as *Software Safety Class A*, since a failure of these software items will not result in unacceptable risk for users, patients, or test subjects. The corresponding verification activities included a structured code review. Safety in general is achieved either inherently or via mechanical or electrical safety measures; safety is never dependent on configuration or software state.

Improvements over 3D USCT II include a mechanical clutch limiting motor torque

as an active safety mechanism, while the reduction of mechanical gaps and an IPX-conforming mechanical layout represent inherent safety measures. Another practical change from 3D USCT II for the new generation was to switch the mounting of the transducers from backside mount (from the cabling side inside the cramped machine) to front-side mount, enabling easier inspection, maintainability, and replaceability.

One radical design change from 3D USCT II to the new system, based on hygienic feedback and experience from the Hospital Mannheim, was the complete externalisation of the water-preparation components. Besides the reduction of complexity and saving space, defining these components as medically non-critical “accessories” decouples the core medical device from additional regulatory responsibilities. Hospitals typically have this infrastructure and expertise in-house. Water of drinking-quality is prepared by hospital personnel, degassed, and pre-heated to the required application temperature of around 35 °C. No pump is required; only passive water drainage is utilised.

Hygiene standards are further improved by maintaining a clear separation between fresh and waste water (unlike 3D USCT II), minimising mechanical gaps, water pipes, and functional interfaces so that all patient-contacting components can be easily inspected and cleaned. Reducing areas that are difficult to access or to clean mechanically has been a core design goal for the next-generation USCT system.

After integration and commissioning, the 3D USCT III device was evaluated by an external testing laboratory, to confirm conformity with the technical state-of-the-art according to the relevant IEC 60601 safety requirements. The majority of the technical safety requirements have been achieved. Remaining tasks include electromagnetic compatibility (EMC) certification as well as the clinical study required to demonstrate overall effectiveness and validity of the method.

A multicenter study is planned to demonstrate clinical viability as the final step of the medical regulatory process, serving as application validation before market introduction with the self-applied CE mark of both method and device.

The safety concept was comprehensively described and discussed in the Master’s thesis of Michaela Hofbauer [Hof21].

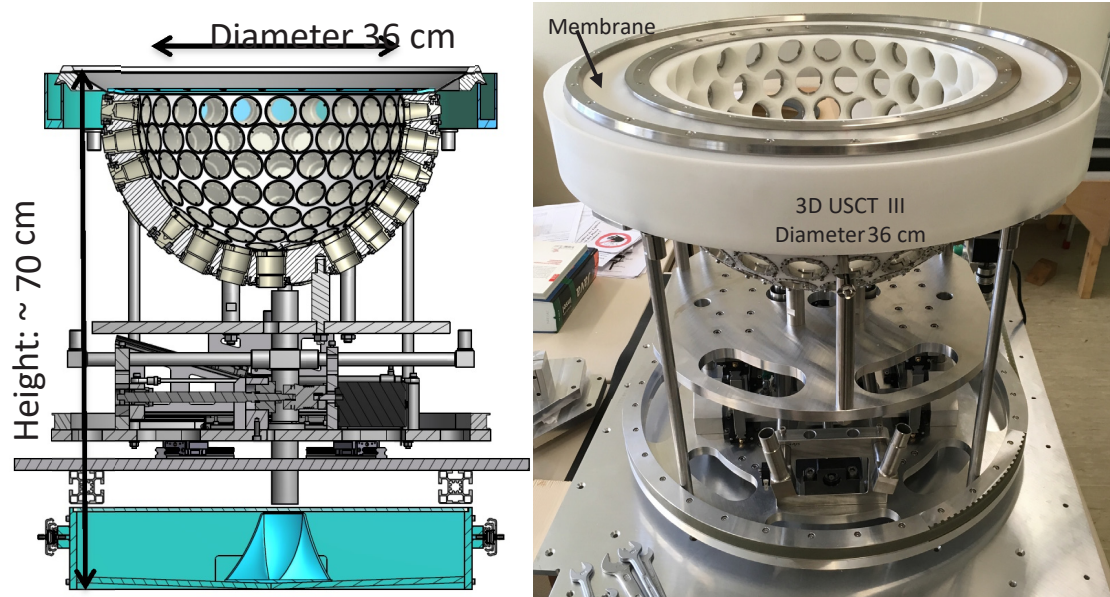


Figure 6.4: (Left) CAD water container and mechanical movement system of 3D USCT III. (Right) realisation. Notable is the thin silicone membrane system removing the 2.5 cm dead-space from 3D USCT II aperture, see Fig. 4.2.

6.3 Mechanical Components and Movement System

In Fig. 6.4 left, the CAD design of the mechanical container and the motor-driven movement system is shown. On the right, the produced and partly integrated mechanical system is visible, still without TAS installed in the mounting holes. Unlike in 3D USCT II, the order of lift and rotational mechanics was reversed: the stainless-steel pillars, also used for installing the distribution board, no longer rotate or lift. The decoupling of movement now occurs entirely via the Hirose cable connecting the FEBs with the distribution board. In the old 3D USCT II system, the mechanical weight of the many LAN cables attached to the rotating distribution board caused slow and unreliable rotational movement.

Another core feature is the silicone membrane (see Fig. 6.4 right), decoupling the rotation and lift movement with very low internal friction while being water-tight. This was a significant improvement over 3D USCT II (see Fig. 4.2), which suffered from high friction, leading to slow and unreliable movement. This membrane resolves also 3D USCT II system's problem of having a large distance in the z-axis between the patient bed and the first transducer. This “dead space” limited

the achievable imaging z-depth of the immersed breast or required patients to bend deeper into the aperture, which was challenging and uncomfortable.

6.3.1 Mechanical Movement System

The mechanical movement system of 3D USCT III was planned as a significant upgrade from 3D USCT II. The repositioning of the aperture over two degrees of freedom – lift and rotation – created additional time overhead. In 3D USCT II, this movement time accounted for around 70 % of the overall measurement duration, see Appendix A.5.1. One identified weak point of 3D USCT II was that the rotation and lift mechanics were prone to sticking due to the expansion and compression of the telescopic, segmented, O-ring-sealed container system. Because this system had high friction and tended to be unreliable, it was replaced by a single-component silicone membrane. This flexible and simpler solution greatly reduced mechanical stiffness and increased the response speed of the movement system.

This means that, unlike in the previous system, the mechanical load on the lift motors is primarily defined by the weight of the medium; the patient is decoupled by the bed, and the breast floats, while the silicone membrane introduces only minimal friction.

This improved the mechanical movement system, achieving a fourfold increase in lift speed and reducing the translation time to only a few seconds, compared to dozens of seconds in 3D USCT II.

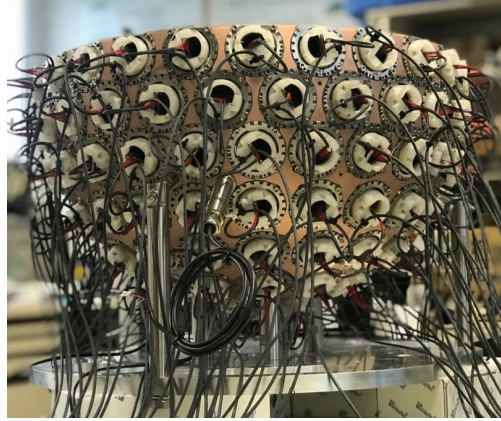
6.3.2 Mechanical Components: Production and Integration

Production and assembly were carried out in parallel wherever possible, mostly with KIT workshop resources. The mechanical frame made from Maytec aluminium and the wheel system were the first components completed.

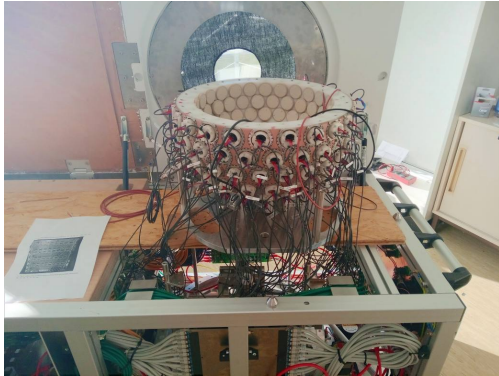
One particularly challenging component was the transducer holder, milled from a 0.5 m diameter POM rod; its fabrication on a 5-axis milling machine took several weeks, see Fig. 6.5, top left. The TAS PEEK housings (see Fig. 6.2) were fabricated



Movement mechanics and measurement container



TAS mounting and cabling



Integration of all components in the housing Cabling device and secondary components

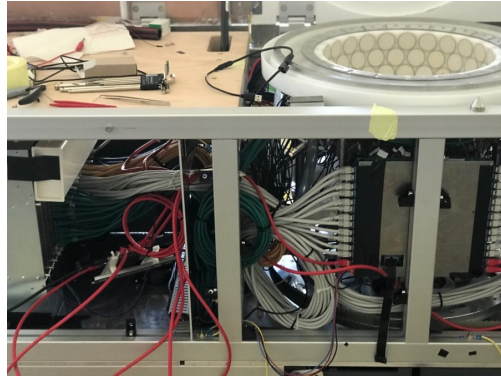


Figure 6.5: KIT 3D USCT III.R hardware integration milestones. (Top left) mechanical movement system and water and TAS container. (Top right) cabling of mounted TAS in the water container, (Bottom left) integration of water container into device. (Bottom right) finalising electrical connectivity and secondary components.

by an external company.

Finally, the ergonomic patient bed was designed to be made from glass-fiber-reinforced polymers (GFRP), see Fig. 6.1, yet it was implemented as simpler flat surface after practicability tests with volunteers, see Fig. 6.41.

After defining and prototyping the custom-made electrical components (FEB, AMC, RTM, Isoboard) and after establishing a cabling solution (Ethernet cable and high-density microcoax cable with Hirose plug), the TAS were mounted into the system and cabled with the Hirose lines, see Fig. 6.5 top right. In Fig. 6.5 bottom left, the measurement container is shown, ready for lift-down into the device. In Fig. 6.5 bottom right, the integration into the patient bed is being installed, and around the water container the waste-water spillage trench is visible. In parallel, the DAQ cards (AMC, RTM, FMC) as well as the firmware and software components were developed, produced and tested.

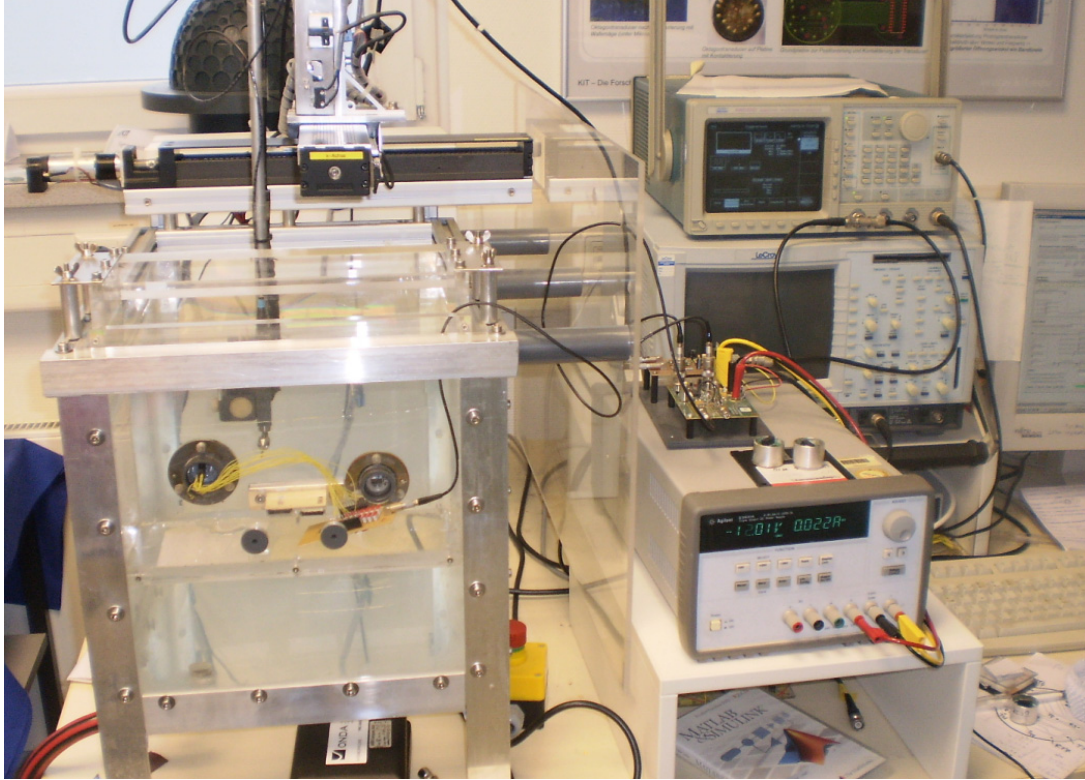


Figure 6.6: 3-axis transducer measurement place at KIT, IPE. (left) water-filled measurement container; (far right) control and DAQ PC, (center) arbitrary waveform generator (Tektronix AWG2021).

6.4 Transducer Prototyping

One of the core components missing for the 3D USCT III are suitable transducers to fulfill the design goals defined earlier. Three approaches were followed and evaluated before a decision was taken. Simulations were applied and prototypes produced and their performance quantified in their electrical and US performance.

The transducer characterisation setup[Pet05] includes a water tank (the measurement container) with a calibrated and movable hydrophone (Onda HNC-400 [Ond]), featuring a 6 dB bandwidth from 1 to 10 MHz and a 60° opening angle at 5 MHz. Ports on the container's front allow the test transducers to be screwed in, as shown in Fig. 6.6.

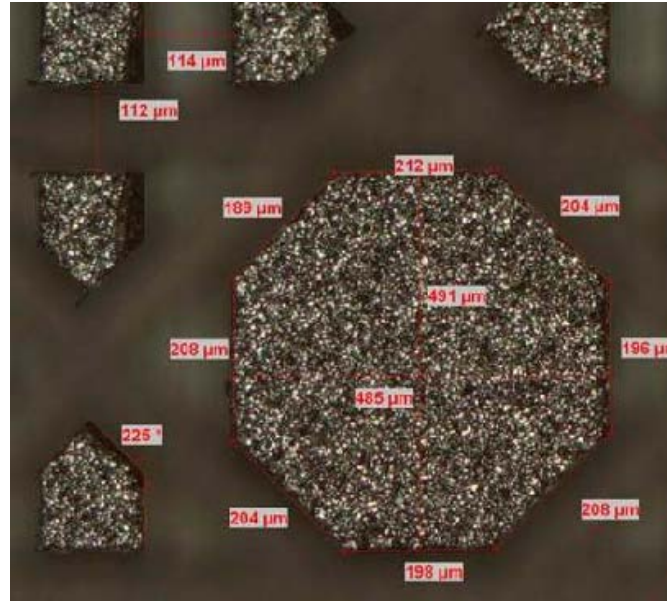


Figure 6.7: *IPE octagon transducer prototype, top-down view with a microscope.*

6.4.1 Prototyping: KIT Octagon Dice-and-Fill Approach

With the 3D USCT I and II the commonly applied dice and fill transducer technology was employed because the designs have a regular layout with rectangular substructures. However, for the 3D USCT III design several innovations are required at the cost of an increase in manufacturing complexity. Here, each single octagon shaped transducer element is placed at a unique position on a TMM4 PCB using a pick-and-placer machine and glued using a conductive silver glue,

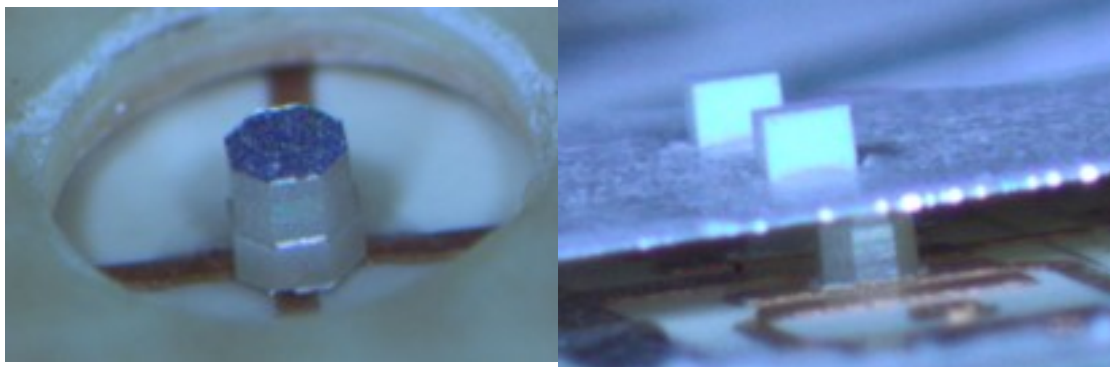


Figure 6.8: *IPE reverse Octagon Transducer - prototype 27. (Left) Octagon transducer element glued on TMM4 PCB. (Right) Octagon transducer prototype glued on TMM4 PCB, top thin aluminum sheet as electrode with TMM4 cubes as matching.*

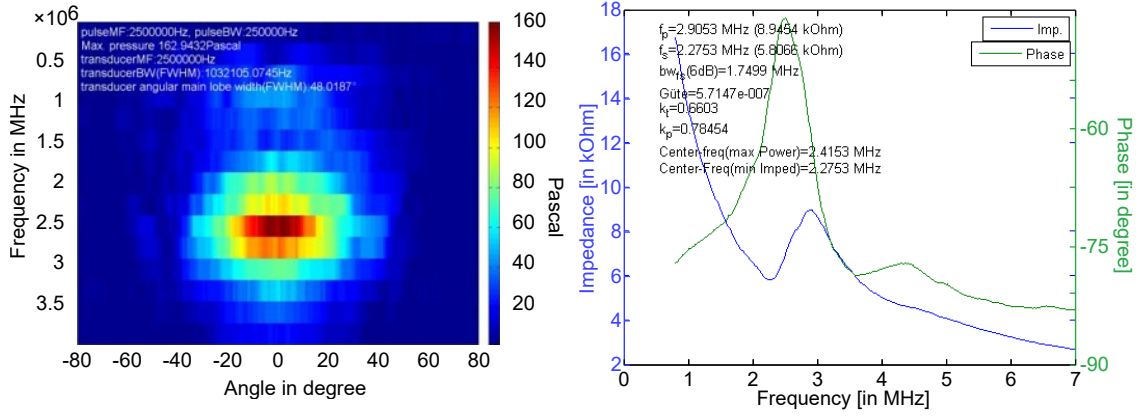


Figure 6.9: IPE reverse octagon transducer prototype #27. Pressure over angle and frequency measured over a semi-circle of 60 mm diameter. (Left) emitter element 2. (Right) impedance-phase curve for element 2. Green curve is phase, blue curve is magnitude. The result show a resonance frequency at 2.5 MHz.

see Fig. 6.8. The octagon shape is needed to achieve a homogeneous radiation pattern over both the elevation and lateral directions and therefore mimicking a circular element. To obtain the octagonal shape, additional cuts under an angle of 45° are made, see Fig. 6.7 and Fig. 6.8.

The fabrication process was evaluated for KIT transducer #27 which features also an reverses stacking order compared to previous generation's design - before with the PCB facing the water, now "forward" with the piezo facing in the water direction. Stack is comprised from a non-conductive matching layer structured with TMM4, followed by the piezo in octagon shape, and finished by a backing with TMM4 and PU. The transducer had an opening angle of $\approx 58^\circ$ on 3 dB for 2.5 MHz and a bandwidth ≈ 1.5 MHz on 3 dB (≈ 2.0 MHz on 6 dB) see Fig. 6.9.

Given its performance and the challenges faced during the production, it was decided not to continue with this approach.

6.4.2 TU Delft Single-Element Transducer Approach

The next transducer design was made in collaboration with the TU Delft. The Department of Imaging Physics (Faculty of Applied Sciences) there had expertise and facilities for ultrasound (US) transducer simulation, development and fabrication.

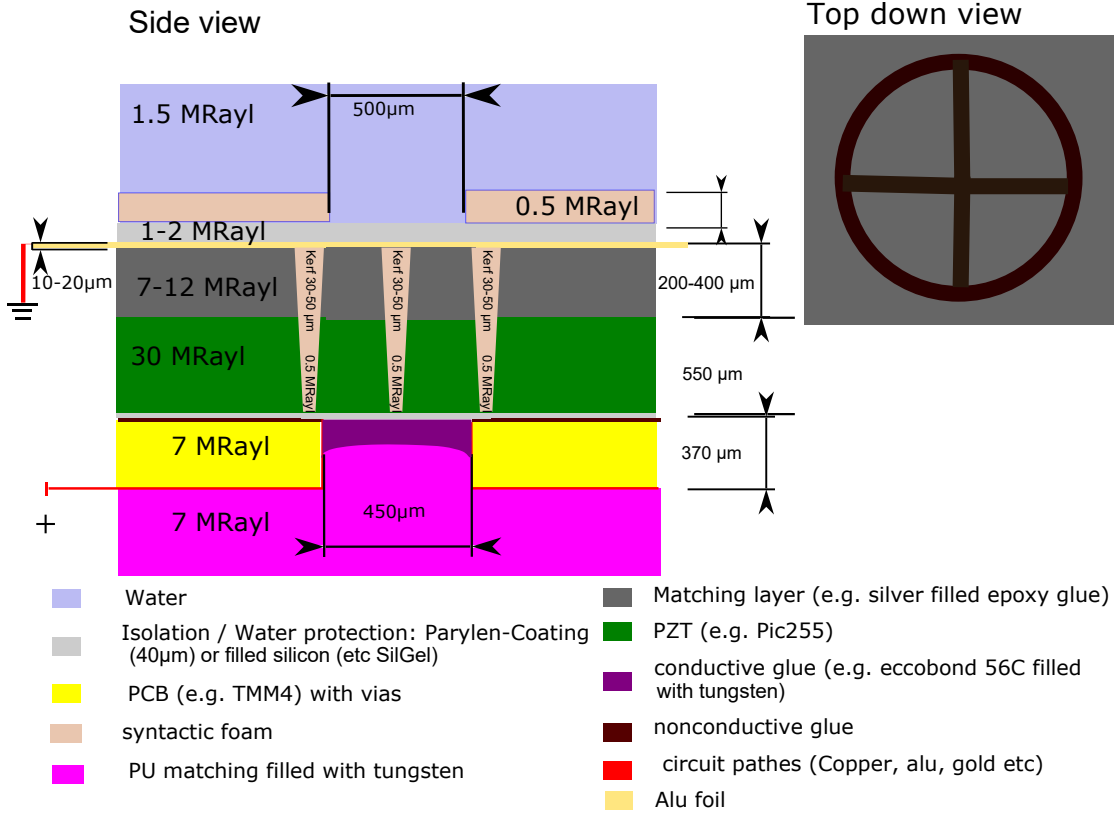


Figure 6.10: *Delft transducer design concept. (Left) schematic side view with materials and dimensions. (Right) top view showing the kerf structure used to segment the PZT element and suppress unwanted oscillation modes.*

Various transducer designs were simulated and optimisation with the PZFlex simulation framework (see Fig. 6.11). This fast Fourier-domain finite-element based simulation framework allows for 3D US waveform modelling suited therefore also for large 3D setups, like the 3D USCT III. This computational efficient setup enabled for the first time the analysis of transducer behaviour in the 3D USCT III context beyond simplified axial piston-mode assumptions, including lateral and higher-order oscillation modes.

Based on on gained knowledge with previous in-house developed transducers and the presence of available infrastructure the following manufacturing process was implemented. A PZT slab of defined dimensions is bonded onto the PCB using a conductive matching layer (e.g. Eccobond 56c). This step has moderate lateral accuracy requirements ($\gtrsim 50 \mu\text{m}$), as precise positioning is achieved in a subsequent laser micromachining step. Fiducial markers on the PCB are used to align the

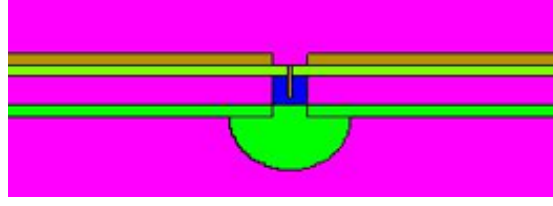


Figure 6.11: *PZFlex simulation model (2D cut). Side view of the transducer stack: backing on the bottom, water-facing side on top.*

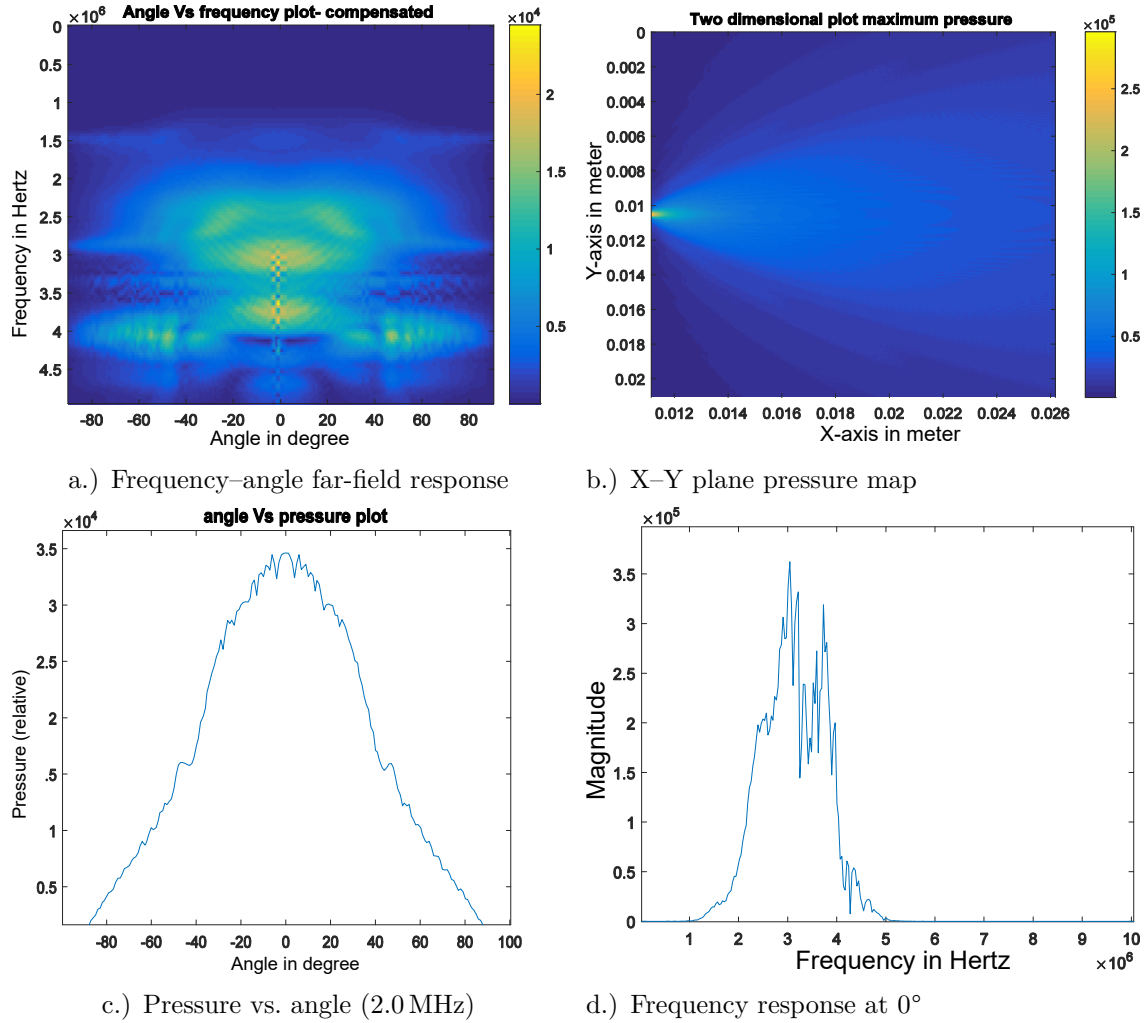


Figure 6.12: *PZFlex simulation results for the Delft transducer design.*

laser cutting process, which defines the individual transducer elements. Electrical interconnection is realised via pre-fabricated vias in the PCB, providing contact to each element.

For the water-facing protection layer, parylene coating was initially evaluated but

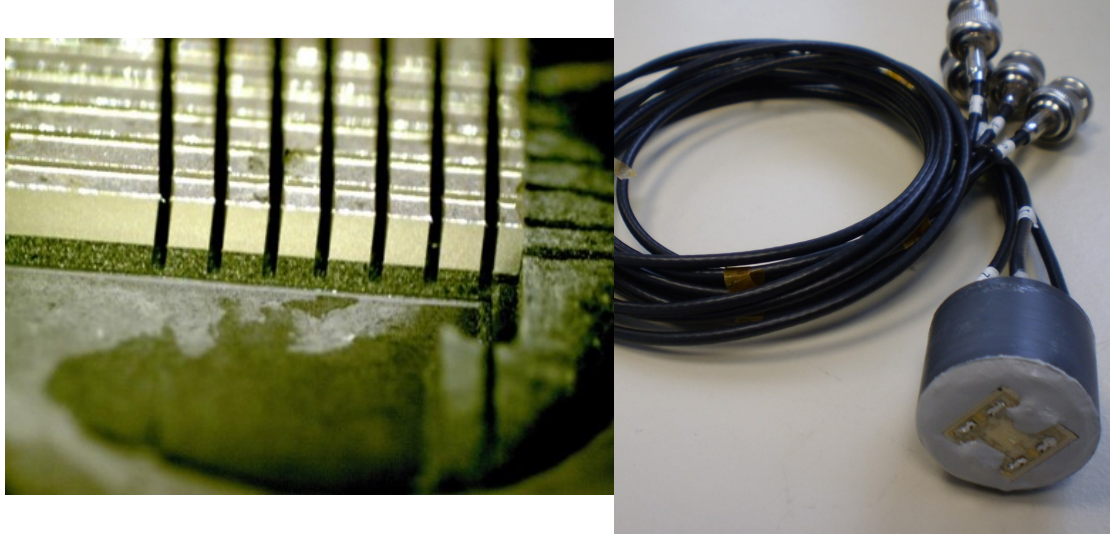


Figure 6.13: *TU Delft transducer prototype. (Left) structured PZT slab after laser processing. (Right) fully assembled device.*

proved unsuitable due to limited mechanical robustness and process reproducibility. It was therefore replaced by a manually applied epoxy layer (Araldite), which provided improved durability.

To reduce lateral acoustic crosstalk, air-filled kerfs were initially considered. While offering favourable acoustic isolation, this approach resulted in insufficient mechanical stability. Consequently, the kerfs were filled with polyurethane (PU), consistent with the established approach used in 3D USCT II.

Several prototypes were fabricated; an example is shown in Fig. 6.13. Water-tank measurements (Fig. 6.14) indicate an opening angle of approximately 60° at the -3 dB level and a bandwidth of about 1 MHz at -3 dB (approximately 1.5 MHz at -6 dB), centred around 2.0 MHz.

This Delft-KIT experiment series found that the following design was optimal within our boundaries:

- Transducer aperture is structured into four sub-elements, each with a surface area of $0.5 \text{ mm} \times 0.5 \text{ mm}$ (rectangular),
- Use standard USCT PZT slabs,
- Delft matching layer formula (silver-filled epoxy),

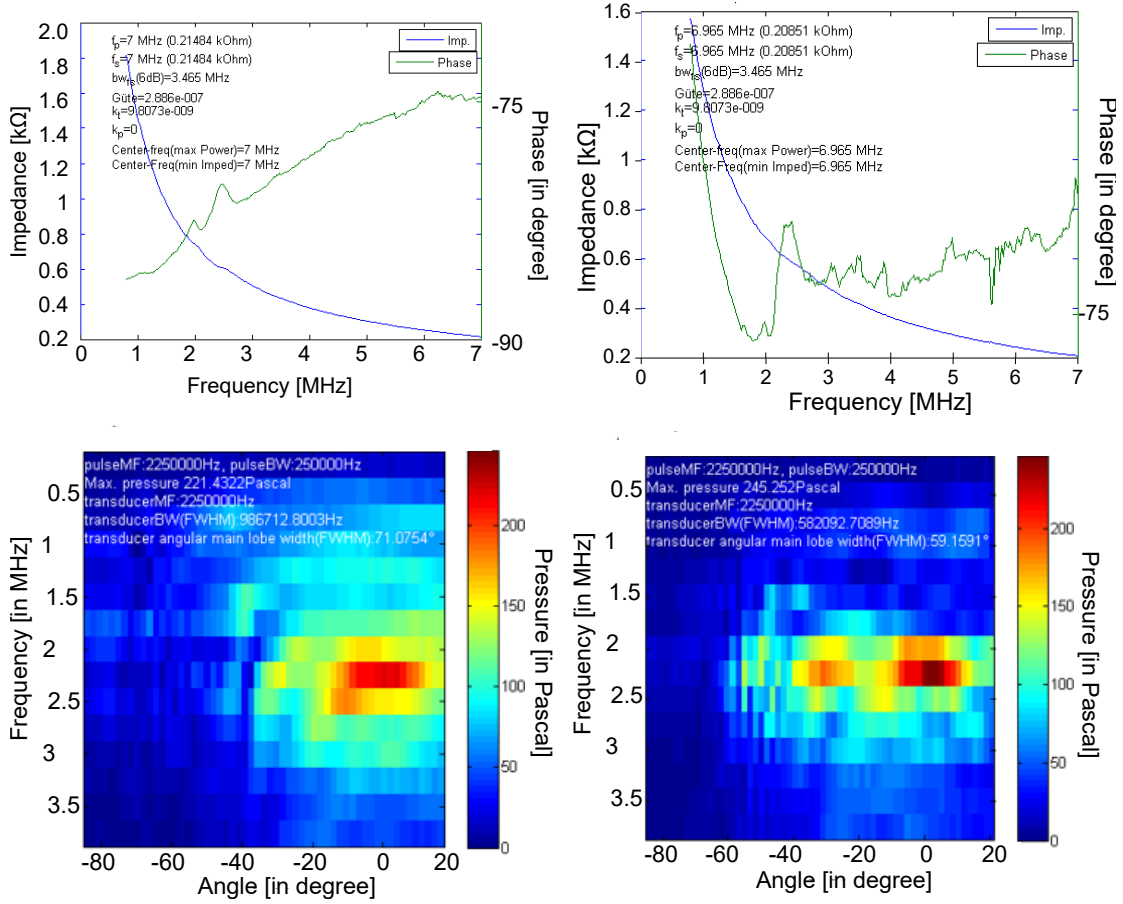


Figure 6.14: Delft transducer measurement results, for prototype #2 element 1 on left and element 2 on the right. (Top row) impedance phase measurements. (Bottom row) pressure in Pascal over frequency and angle.

- Backing made of the USCT default TMM4 and PU with tungsten filling,
- Water-protective coating made of Araldite.

Potential degrees of freedom in spatial dimensions, materials, and structural variations were simulated and analysed. The selected design is shown in Fig. 6.10, which was evaluated from simulated results for the relevant metrics: opening-angle performance and bandwidth, see Fig. 6.12.

With these promising results for opening angle and bandwidth, several prototypes were built. The third and final iteration used 0.5 mm PZT with a single kerf filled with PU. A PU with tungsten mixture in a 50:50 ratio was used as backing and filling material.

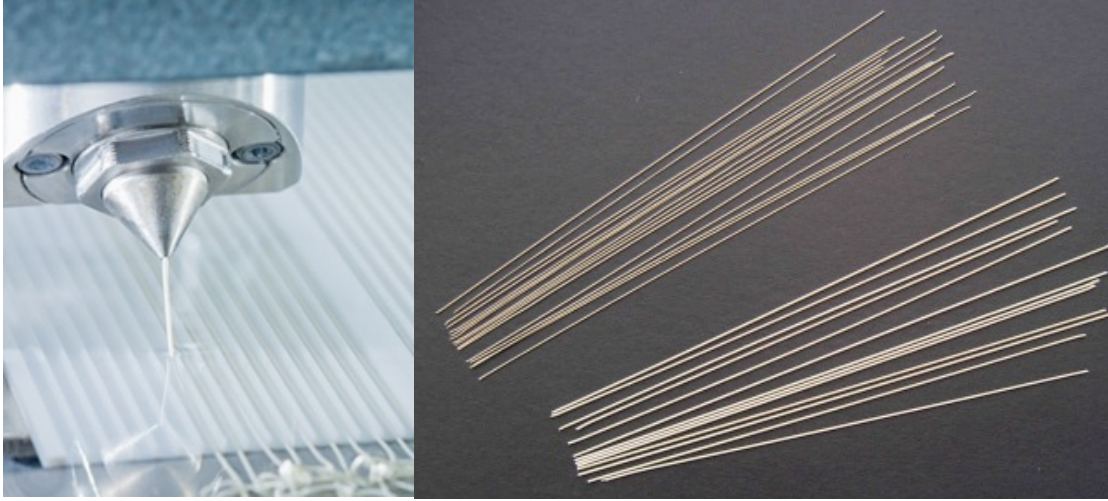


Figure 6.15: *FhG IKTS piezofibre technology. (Left) piezofibre spinning technology process. (Right) cured and sintered piezofibres.*

Since the performance of the prototypes was not ideal and especially the robustness of the design was not sufficient (coating layer), other design approaches were followed up.

6.4.3 Fraunhofer IKTS Piezofibre-Based Transducer Approach

An alternative transducer concept investigated was based on piezofibre technology. This approach inherently enables circular or quasi-circular apertures, which are advantageous for achieving more symmetric emission characteristics compared to conventionally diced PZT elements.

Collaboration was established with the Fraunhofer Institute for Ceramic Technologies and Systems IKTS (FhG IKTS, Dresden), which has extensive expertise in piezofibre fabrication via fibre spinning processes (see Fig. 6.15). Their established manufacturing capabilities, see Fig. 6.19 enabled the exploration of piezofibre-based transducer concepts tailored to the requirements of 3D USCT III, including initial prototype development.

The piezofibre transducer technology offers interesting advantages in the production process over the techniques explored earlier. One core vision is the production of transducer arrays in an efficient, automisable, and reproducible way in larger

batches, see Fig. 6.16.

Furthermore, piezoceramic fibre technology offers design characteristics that meet several key requirements of 3D USCT III: arbitrary positioning of fibres using a mask; inherently round transducer apertures; and improved reproducibility of transducer positions by a “disc-sawing process,” which reproduces the same layout repeatedly, see Fig. 6.17. The centre frequency can also be tuned simply by selecting the thickness of the discs sawn from the fibre rod.

Simulation started with 1-D KLM simulations to define the matching-layer thickness, see Fig. 6.18 top left. To simulate and model additional lateral or shear-wave effects, higher-dimensional simulations were employed. 2-D and 3-D FE simulations were performed with PZFlex (Weidlinger Associates Ltd., Glasgow, UK). The spatial mesh was chosen at least three times finer than the minimal Nyquist criterion, and the temporal sampling was selected automatically by the solver. From these simulations, the far field at several centimetres distance as well as angular and spectral responses were obtained (see Fig. 6.18 bottom left).

The simulations confirmed that PZT fibres with a diameter of $460\text{ }\mu\text{m}$ and a thickness of $750\text{ }\mu\text{m}$ provide the desired behaviour. The frequency response matches the expected angular range (about 25°) over a bandwidth from 1 MHz to 5 MHz. At these frequencies the pressure field remains largely homogeneous without unwanted modes.

To produce circular transducer elements with a diameter $460\text{ }\mu\text{m}$, PZT fibres were fabricated by the fibre-spinning process [HGSM15]. A PZT slurry (Sonox P505, CeramTec GmbH) was extruded into a water bath, solidified by phase inversion, cut, dried, and sintered. Thirteen fibres were placed into a mechanical mask and embedded in epoxy (EPO-TEK 301-2FL). The composite strand (180 mm long, 23 mm diameter) was sliced into 40 discs with thicknesses from $400\text{ }\mu\text{m}$ to $2050\text{ }\mu\text{m}$. Afterwards, the discs were sputtered with a 150 nm Au layer, poled at 2 kV mm^{-1} , and equipped with structured top electrodes for individual fibre access. Silver glue (PC3000, Heraeus) ensured electrical connection.

Geometrical properties were verified via optical inspection and micrometer measurements by the IKTS. Electrical behaviour was evaluated with a phase-impedance analyser (HP 4149A), deriving resonance frequency and k_t from

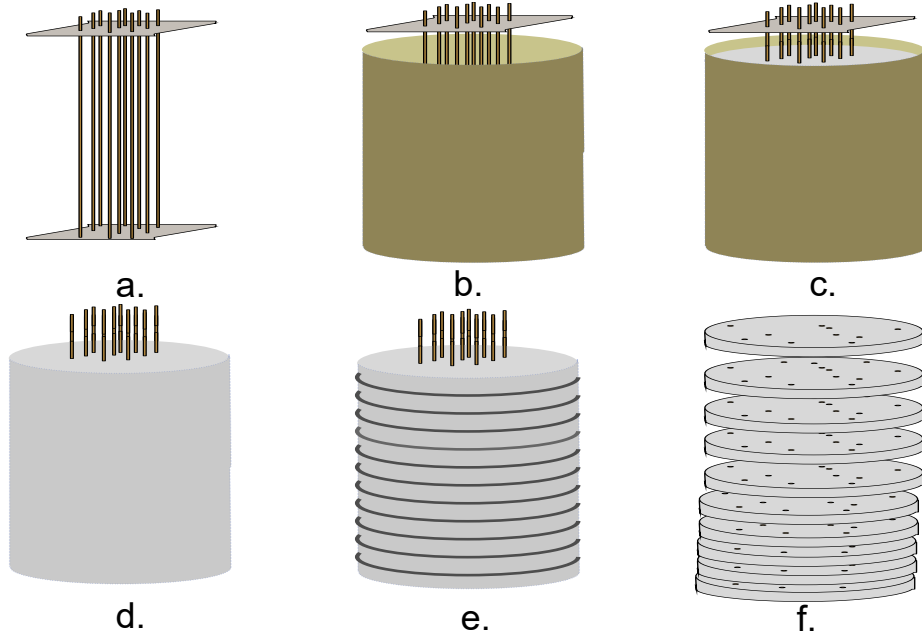


Figure 6.16: *Piezo fibre production vision: from top left to right and bottom: (a.) Metal mask with holes allows positioning of piezofibre in arbitrary positions. (b.) The fibre group is placed in a curing container and (c.) filled with an epoxy. (d.) After curing, the rod can be (e.) sawed and (f.) milled into discs of defined arbitrary thickness.*

impedance spectra using a 0.1 MHz to 15 MHz sweep at 0.5 V.

The fibre diameter of $460\text{ }\mu\text{m}$ was achieved within $\pm 2\%$, and the disc thicknesses also varied by only $\pm 2\%$. Fibre placement accuracy was within a few micrometres, indicating that the technology reliably produces the required geometry.

Integration into TAS and the electrical connectivity is remaining for KIT before evaluation. A flexprint PCB interface was developed and utilised successfully replacing labour-intensive and error prone manual bonding process of previous generation transducer and TAS production. The PCB includes alignment holes ($\approx 0.05\text{ mm}$) and openings for bubble-free casting. The combined transducers with PCBs were mounted in aluminium housings and filled with a tungsten–polyurethane backing. To suppress unwanted modes and improve bandwidth, a polyurethane-based backing (VOSSFlex 2k PU + tungsten powder, particle size $10\text{ }\mu\text{m}$ [Pfi17]) was formulated. After degassing (30 min, $\approx 50\text{ mbar}$), it yielded an acoustic impedance of about 11 MRayl . A single $370\text{ }\mu\text{m}$ matching layer (TMM4) was added, see Fig. 6.22 d.) which parameters were based on KLM results, see

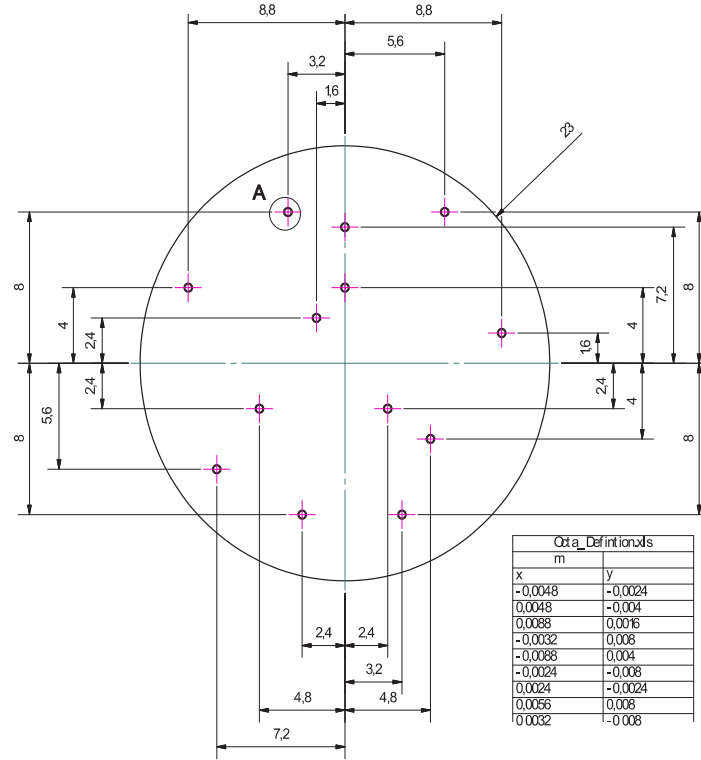


Figure 6.17: Design schematic of the defined transducer positions for the production of the thirteen transducer prototype piezo fibre array. Individual piezo elements are marked with pink crosses at the position. Resulting produced disc is shown under Fig. 6.20.

Fig. 6.18 a.).

After several fabrication runs with varying the production parameters for disc thickness, fibre configuration, connectivity etc. the overall technology was evaluated. For this, the fibre rods were cut into discs of varying thicknesses (450, 500, 550, and 600 μm), metallised, electrical connectivity printed and then polarised.

Electrical evaluation of the transducers was done at the IKTS, the integration and final performance evaluation was then done at the IPE.

US characterisation utilized a three-axis hydrophone setup in a 30 cm $\times$ 30 cm $\times$ 50 cm tank, see Fig. 6.6. Hundreds of positions on a semi-circular plane were measured using chirps from 0.5 MHz to 5.5 MHz in 250 kHz steps at $\pm 100 V_{pp}$. A calibrated Onda HNC-400 hydrophone (1–10 MHz, +20 dB) was used. Signals were sampled at 20 MHz with a record length of 400 μs and 16-fold averaging. MATLAB-based evaluation extracted the acoustic

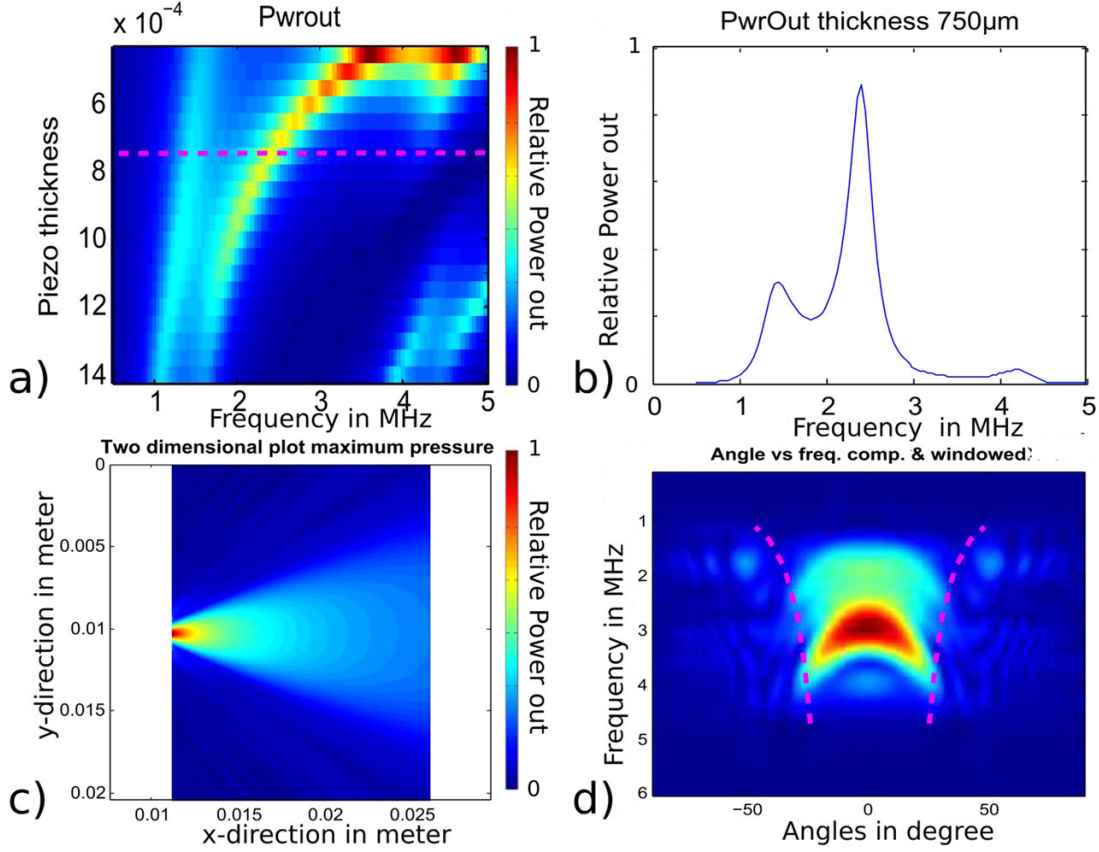


Figure 6.18: *Transducer simulation: (Top left) KLM simulation over frequency and piezo thickness with matchinglayer. Red marked the thickness of 750 μm . (Top right) resulting bandwidth. (Bottom left), 2D x-y plane cut of 3D wave field simulation with PZflex. (Bottom right) frequency (on y-axis) over angular (on x-axis) response. Colour coded the relative power output.*

metrics, see Fig. 6.21.

The TAS impedance (disc + backing + matching) was measured from 500 kHz to 6 MHz. Dependencies of k_t and $f_{R,t}$ were found to be dependent on thickness, see Tab. 6.1. Below 0.75 mm, k_t drops ($<60\%$) due to stronger coupling to planar modes, so smaller thicknesses should be avoided. The integrated k_t includes all assembly effects. A yield metric was introduced (counting elements with sufficient coupling across three TAS). For 400, 550, and 750 μm discs, 37/39 elements (95%) worked.

Only the 400, 550, and 750 μm variants were evaluated further, because thicker samples showed reduced coupling and low pressure. The measured opening angles were about 50° . This is in agreement with the matching layer simulations and

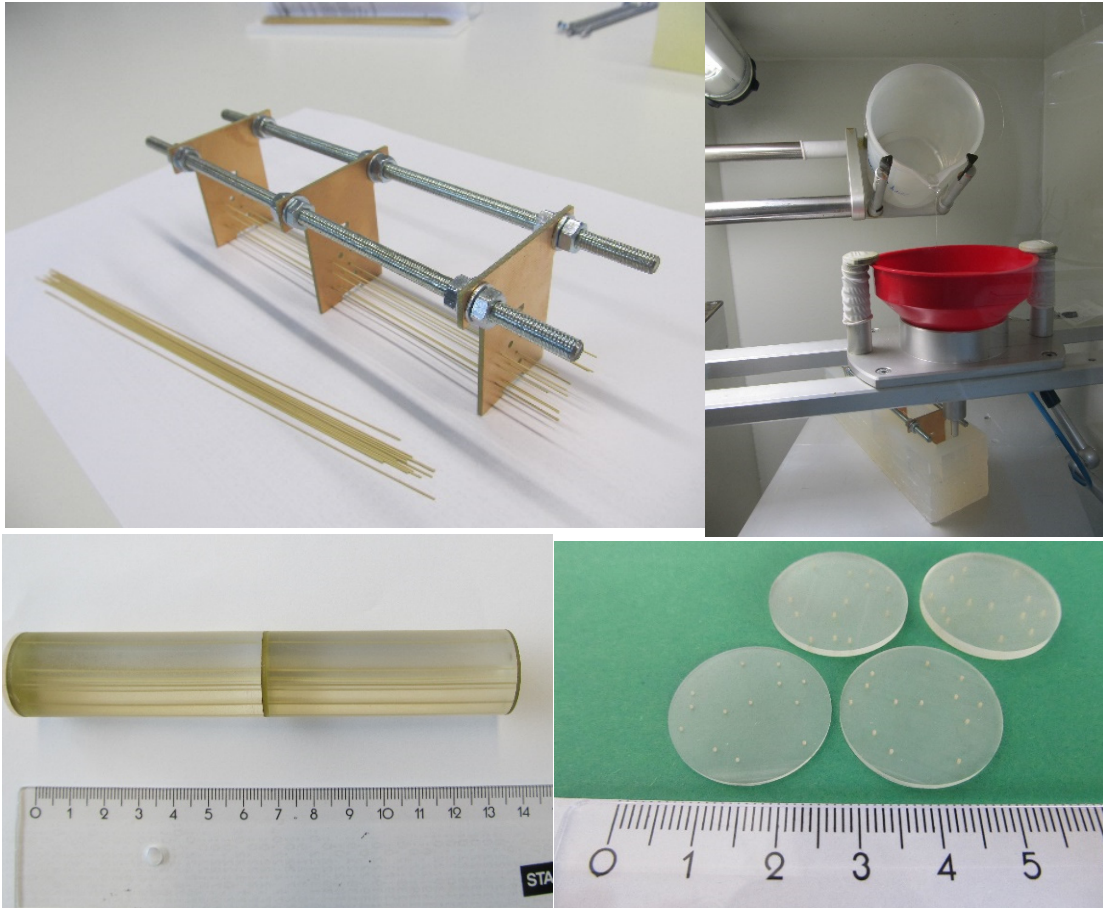


Figure 6.19: Piezo fibre transducer production process: (top left) metal mask with holes allows positioning of piezofibre in arbitrary positions. (Top right) metal mask is placed in a container and filled with an epoxy. (Bottom left) cured block milled into a round rod. (Bottom right) rod is sawed with an in-hole saw into accurate discs of defined thickness.

shows an improvement over older dice-and-fill transducers (36°). Absolute pressure was $\approx 25\%$ of 3D USCT II, mainly due to smaller aperture and stronger backing. Bandwidth stayed near 50%, and the centre frequency was 2 MHz (slightly lower than USCT II at 2.5 MHz). The final 750 μm TAS shows the expected improved opening angle across the entire band.

Overall, the piezocomposite approach was successfully evaluated, design parameter chosen for full production. The approach provides circular, addressable transducers with predictable geometry and good reproducibility. It is cost-efficient, scalable, and compatible with automated electrical assembly. The system demonstrates high yield, good bandwidth, and improved angular performance compared

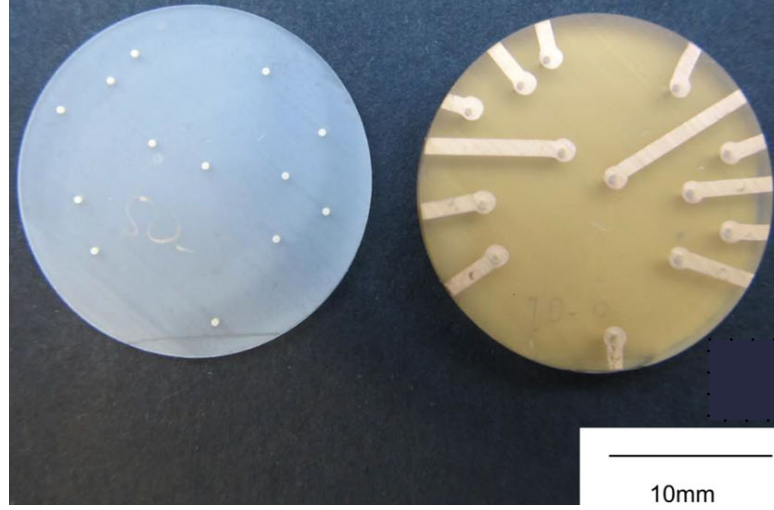


Figure 6.20: Piezo fibre prototype disk for 3D USCT. (Left) sawed disk visible as dots the piezos in the epoxy matrix material. (Right) the individual discs are printed with silver glue and are then polarised with high voltage.

Table 6.1: Piezofibre TAS prototype performance for various thicknesses.

Thick- ness [μm]	Center freq. [MHz] (mean/std.)	Bandwidth (3dB/6dB) [MHz]	Phase (deg) at max. power)	k_t (mean/std.)	Working ($k_t > 0.1$)
400	3.01 (0.24)	1.69 / 2.34	-69.98°	0.56 (0.17)	11/13
550	2.54 (0.07)	1.57 / 1.99	-72.24°	0.84 (0.06)	12/13
750	2.06 (0.11)	1.23 / 1.34	-74.53°	0.90 (0.01)	13/13
1050	3.56 (1.29)	0.17 / 0.17	-85.08°	0.91 (0.03)	4/13
1450	3.34 (0.07)	1.62 / 2.54	-85.22°	0.11 (0.24)	1/13
2050	2.53 (0.04)	1.25 / 1.73	-85.30°	0.00 (0.00)	1/13

to traditional dice-and-fill transducers.

6.4.4 Transducer Integration and Production

After successfully prototyping the piezo fibre technology and evaluating and selecting design parameters with a first prototype production run, there were still several questions and challenges left for integration and for a full industrial quality production run.

For instance for full batch size production run the question of the optimal and reliable electrical connectivity to the PCB was still open as the increased number

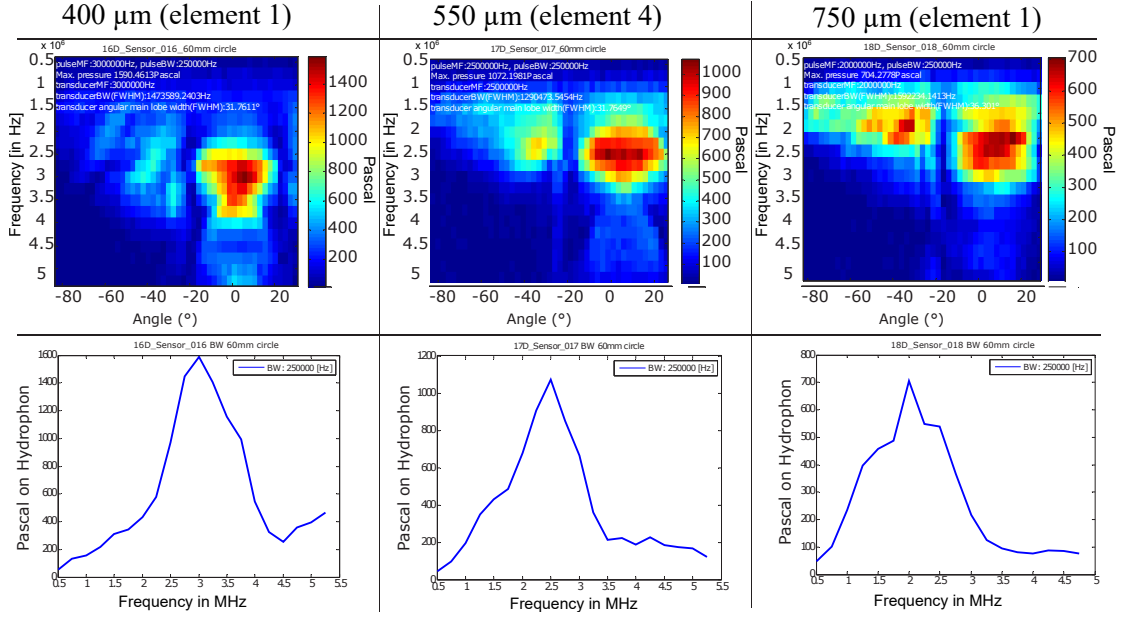


Figure 6.21: Water tank measurement: Bandwidth and opening angle for the three considered thicknesses. (Left) 400 μm thickness piezo, (center) 550 μm thickness piezo, (right) 750 μm thickness piezo. (Top row) angle over frequency pressure measurement. (Bottom row) Frequency for angle 0°.

(13 to 18) of TX/RX channels increased the requirements of the solution. Considered were hard PCBs (TMM4, FR4) and flex PCBs (Kapton) as platform for the transducers. After prototyping with flex PCBs, the increased cost and practicality challenges lead to a more conventional FR4 approach – a very thick PCB of maximum thickness (several cm) available for reasonable price with holes which were filled with the backing material. The backing development is described in further detail in the MSc thesis of Patrick Pfister [Pfi17].

The task of developing the full batch size production was handled by Martin Angerer and is described in his PhD thesis, “Transducer Arrays for 3D USCT” [Ang22]. The full production run was finalised in 2022, one exemplary TAS can be seen in Fig. 6.2.

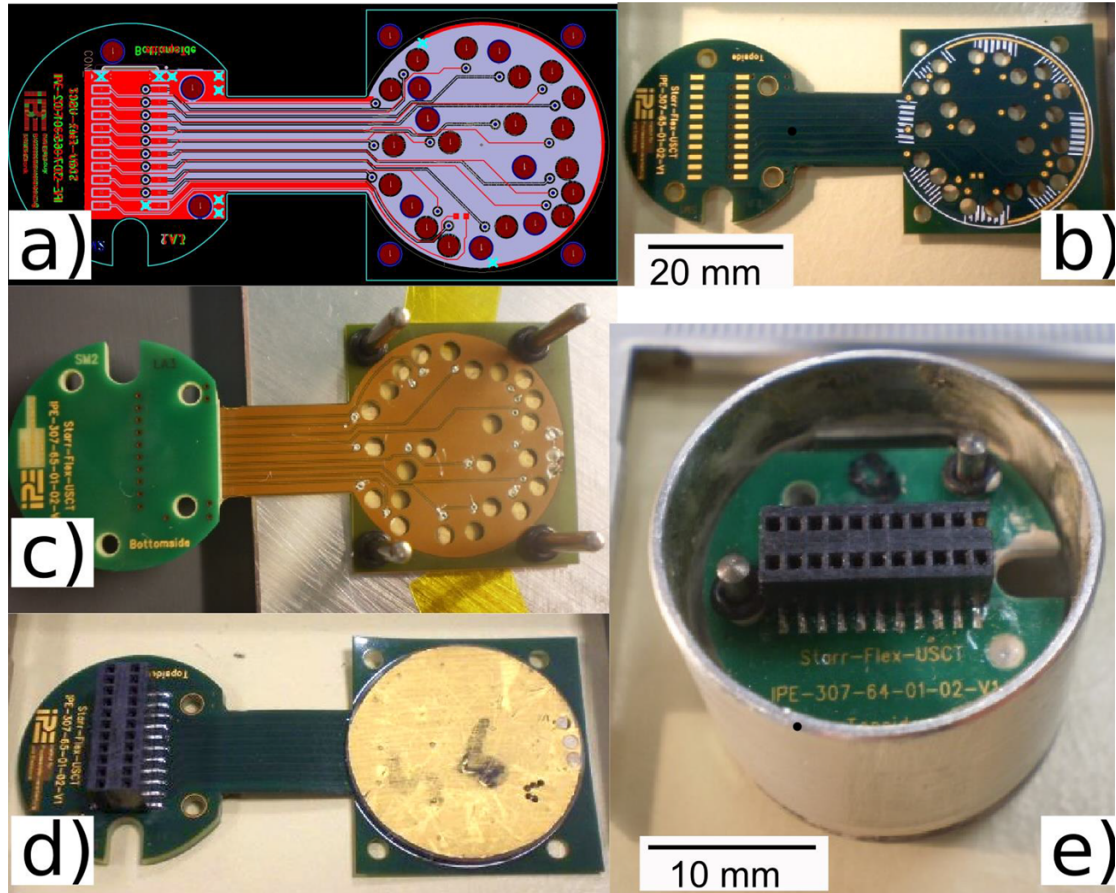


Figure 6.22: Final prototype at KIT IPE produced. a.) PZ-flex schematic, b.) produced PZ flex board. c.) flex board and piezofibre disc combined by a jig and pins. d.) glued together and electrical connected. e.) final prototype with plug, ready for evaluation and characterization.

6.5 Analogue Components: Connecting Transducers and the DAQ

The connectivity between the transducer to the digitisation and DAQ with its analog channels and digital communication posed many challenges. Sepcifically the first part, the parallel 18 low voltage reception channels and the 18 high voltage excitation channels opened challenging questions on interconnective wiring, switching and integration from the transducer to the first stage electrical implementation.

6.5.1 Front-End Improvements: IPE ASIC for 3D USCT

Optimal excitation chain

As discussed in earlier chapters, the analog excitation chain is primarily defined by its bandwidth and achievable peak-to-peak voltage. In the 3D USCT II system, the maximum excitation amplitude was 32 Vpp, with a maximum excitation length of 12.7 μ s, digitally encoded via a DAC.

For the 3D USCT III system, the target specifications were significantly extended: an excitation amplitude of ≈ 100 Vpp and a maximum excitation length of ≥ 400 μ s for chirp-based encoding. These extensions are required to compensate for SNR losses introduced by the enlarged aperture and modified transducer design.

In the 3D USCT II discrete front-end electronics, four TX channels and nine RX channels were implemented using multiplexers, excitation drivers, and amplifiers. Scaling this architecture to a 9/18 TX/RX channel configuration using discrete components would have been challenging due to PCB footprint constraints and cost considerations (approximately 40 k€ for the required number of TAS units). Therefore, an integrated solution was pursued.

The adopted approach replaces the discrete TAS front-end electronics with an application-specific integrated circuit (ASIC), integrating TX/RX multiplexing and amplification. With the availability of the ADL facility and the establishment of the professorship of Ivan Perić at IPE, the development of such a dedicated USCT ASIC became feasible. The ASIC thus forms the core of the analog front-end (AFE) board for the 3D USCT III system.

The ASIC incorporates a three-stage class-B amplifier with an overall gain of up to $\mathcal{O}(10^4)$ and a broadband response from approximately 0.5 MHz to at least 4.5 MHz, see Fig. 6.23. A digital control interface enables extensive parameterisation and calibration, including auxiliary features such as test channels for channel verification, see Fig. 6.25.

The chip is fabricated using AMS 0.35 μ m HV-CMOS technology, enabling high-voltage operation within a compact footprint. The total chip area is approximately 12 mm² [LBSH⁺21, GBH⁺18, Bla19a].

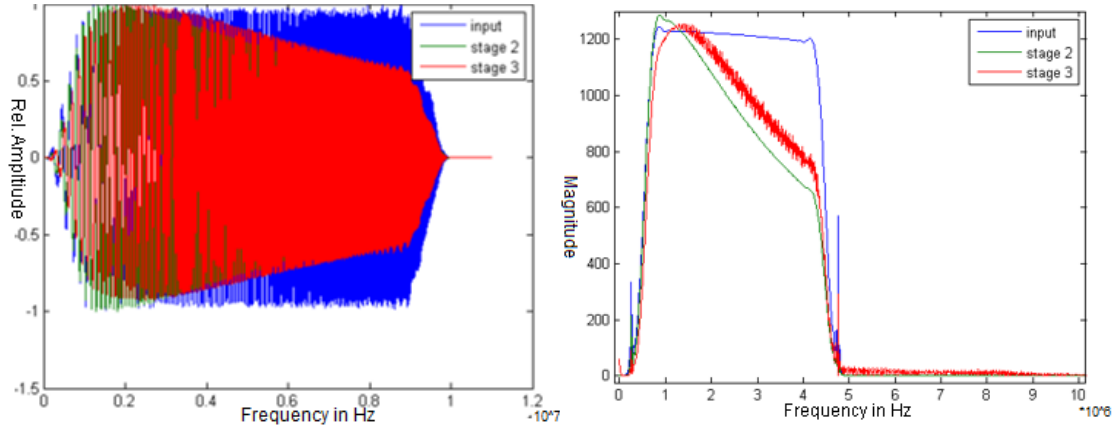


Figure 6.23: ASIC simulation impact on broadband excitation chirp (2016). Blue: input excitation signal. Signals are amplitude-normalised. Green: ASIC stage 2 response. Red: ASIC stage 3 response. (Left) Time domain. (Right) Frequency domain.

The design target was a 9-channel ASIC supporting unified TX/RX operation over a bandwidth of 0.5–5 MHz. The architecture combines a multi-stage amplifier chain with variable gain (up to several thousand) and a high-voltage output stage capable of delivering ± 100 Vpp. From a cost perspective, the target was approximately 100 € per chip, resulting in an estimated system cost of ≈ 20 k€ for 200 ASICs.

For operation of the ASIC, a dedicated front-end board (FEB) was developed to provide power distribution, signal routing, and interfacing. The implementation is shown in Fig. 6.26.

Further details on ASIC design, implementation, commissioning, and optimisation are documented in the PhD theses of Roberto Blanco and Zewei Lu, as well as several Master’s theses within the USCT project [Bla19b, Myl16, Ima16, Bec18].

6.5.2 Bridging the Gap: Distribution Board and Isolation Board

Integration of the transducers and the FEB with the DAQ and digitising hardware needs quite some more analogue bridging.

For instance, finding a physical cabling with high voltage and high density plugs

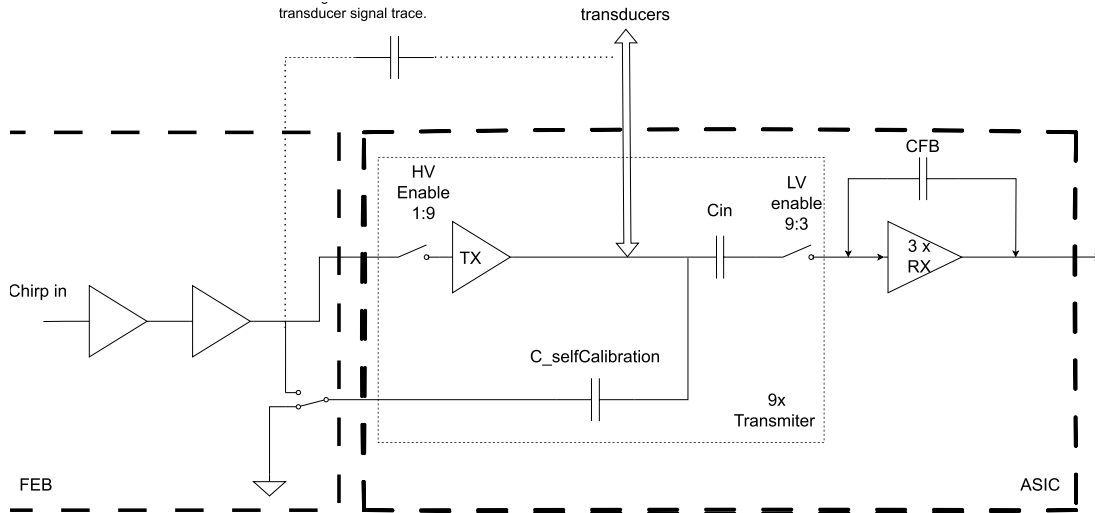


Figure 6.24: *3D USCT III ASIC9c schematic. Left: excitation signal path (TX). Right: receiver signal path (RX).*

while being mechanical moveable was a challenge. Also, the cables have to carry many analogue, supply and digital communication and signal lines in a small form factor.

After several iterations, as solution here micro-coax cable with Hirose plugs and sockets were chosen. They provide sufficient isolation and cable density to bridge the FEB with the next stage. Next stage are the distribution boards, which act as accumulator. This board provides mechanical decoupling, power supply and managing features power supply, serial digital communication, synchronisations lines offered their fair share of complexity, see Fig. 6.27.

This complete topic is described in more detail in the PhD of Zewei Lu “*Development of Methods to control a massively parallel 3D USCT system*” [Lu24].

6.5.3 DAQ Hardware Architecture

DAQ Platform: μ TCAv4 as Standard DAQ Platform

The USCT system employs a dedicated data-acquisition (DAQ) architecture based on the μ TCA.4 standard, an established telecommunication platform widely used within the Helmholtz Association and supported by commercial infrastructure

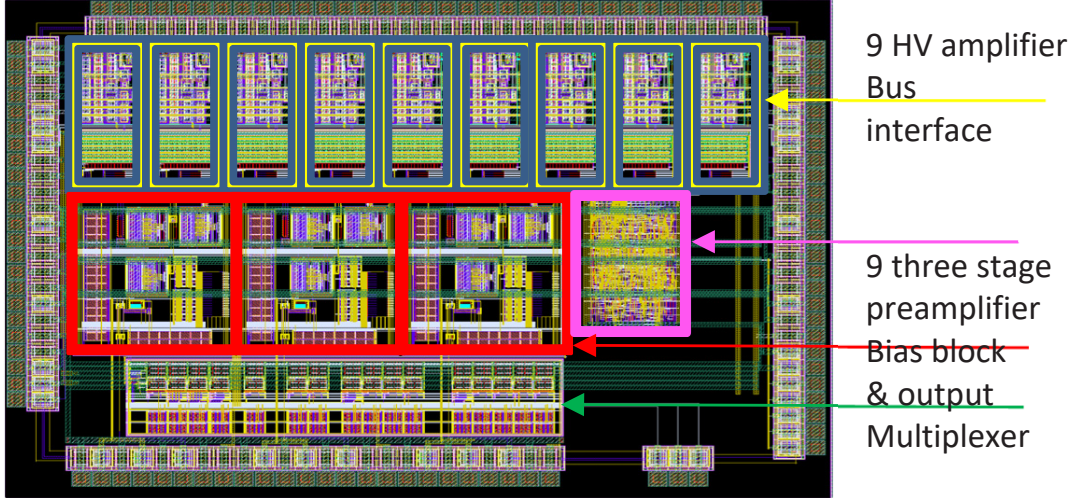


Figure 6.25: 3D USCT III ASIC 9c structure with colour-coded functional blocks.

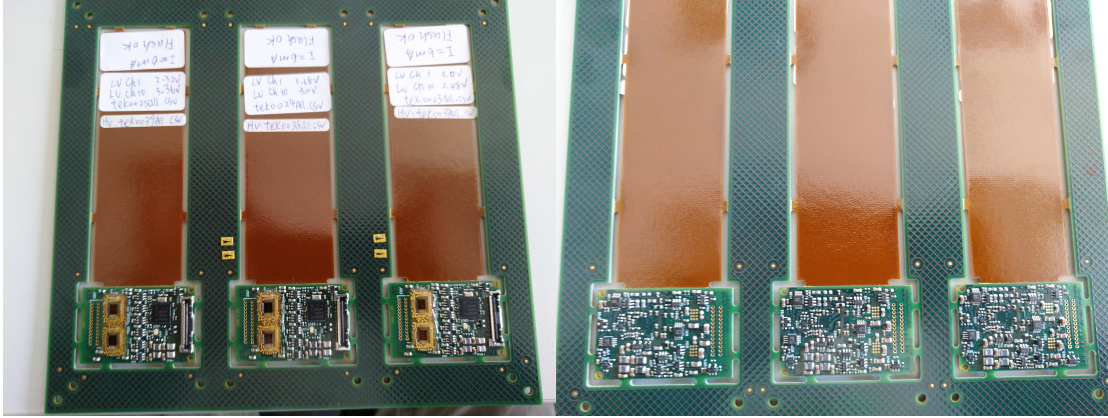


Figure 6.26: Front-end board (FEB) for the ASIC: (left) front side with two bonded and glob-topped ASICs, (right) back side.

providers (e.g., N.A.T. GmbH, SCHROFF GmbH). This standard provides a scalable and modular framework suitable for high-channel-count measurement systems.

The implemented DAQ solution, internally referred to as DAQv5, is centred around a 12-slot μ TCA.4 crate, see Fig. 6.28. Each slot hosts a DAQ card with 32 digitisation channels, enabling a total of 384 channels per crate. This configuration represents a practical upper limit in channel density under the constraints of power consumption and PCB space within the μ TCA.4 standard. The architecture follows a modular design separating analogue and digital functionality, facilitating reuse and adaptation across different projects at IPE and KIT.

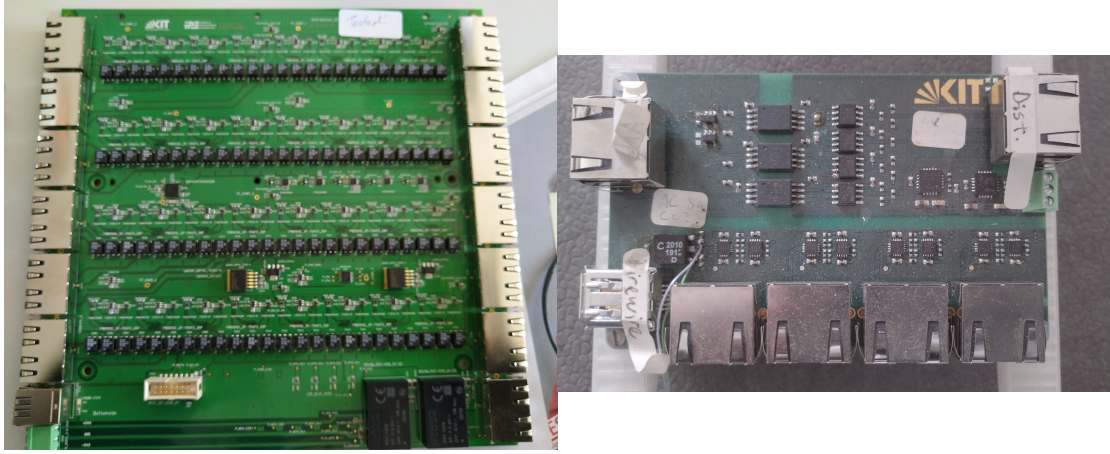
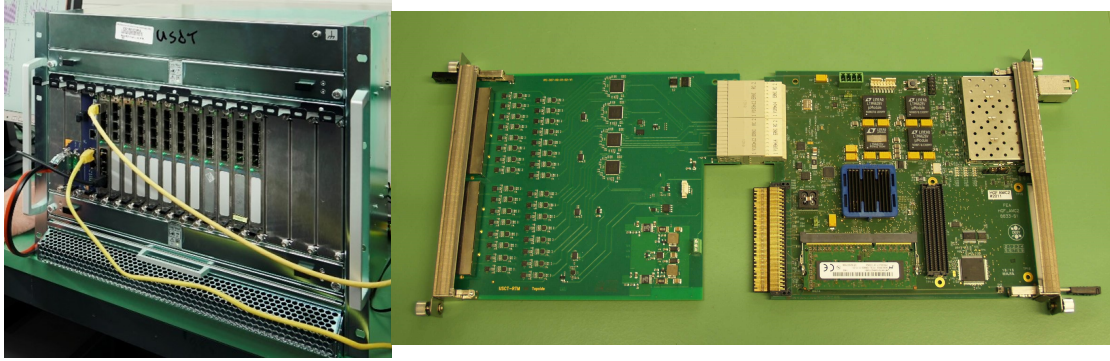


Figure 6.27: Connectivity between the FEB and the DAQ: (left) distribution board, four of them were installed in the 3D USCT III system, (right) Isolation board, distributing the excitation signal to the four distribution boards, the step-ack signal and the serial communication.



μTCAv4 Crate HGF/IPE RTM (left) and AMC (right)

Figure 6.28: DAQ solution for 3D USCT III, DAQv5 μTCA v4 based

The digital processing is performed by Advanced Mezzanine Cards (AMCs), see Fig. 6.28 right, derived from developments in the DESY environment and further extended at IPE. The analogue front-end is realised using Rear Transition Modules (RTMs), which were specifically designed for the 3D USCT III system. These RTMs implement application-specific frequency characteristics, coupling schemes, and analogue anti-alias filtering.

Although the total number of channels per crate (384) is lower than in the previous system generation (480 channels), this reduction is compensated by an improved acquisition scheme and enhanced measurement strategies enabled by the significantly increased processing capabilities of the AMC hardware. Furthermore, the

system supports synchronous operation of multiple crates, allowing the total channel count to be extended to 768 for future scalability.

Each DAQ card integrates on-board memory buffers, a high-performance FPGA, and an FMC interface. The FMC module enables combined transmit (TX) and acquisition functionality on a single board, thereby reducing slot usage compared to the previous architecture, where separate master (SLT) and front-end (FLT) cards were required.

The analogue RTM (ADC32-RTM) incorporates a logarithmic variable gain amplifier (VGA) with a dynamic range of 0–80 dB (AD8338, controlled via a 12-bit DAC LTC2636) to extend the usable signal dynamic range. A fifth-order analogue anti-aliasing filter (AAF) is implemented ahead of the digitisation stage. The system employs four eight-channel ADCs, resulting in 32 parallel digitisation channels operating at 20 MHz.

The AMC master board does the orchestration, buffering, digital AAF (anti-aliasing-filtering) before the digitisation step. The complete block diagram is given in appendix A.4. The HGF-AMC board has an Xilinx Kintex-7 420 FPGA (pin compatible variant XC7K355T later utilised), PCI express for external connectivity, SODIMM 8 GByte DDR III as data buffer and serial interfaces to the front-end electronic for parametrisation and configuration. On the FPGA the crucial DAQ procedure and digital data processing is implemented featuring a 209th order digital filter for decimation and sharp AAF.

With the hardware ready, the features implemented in FPGA firmware and software needs to be specified. Base here was the 3D USCT II functionality which was ported and extended for 3D USCT III.

6.6 3D USCT III Data Acquisition

As the 3D USCT III aims to acquire high pressure US signals from measurement paths practically without attenuation (waves missing the object) together with highly attenuated measurement paths, the DAQ system should be capable to cover a great dynamic range, see Sec. 5.2.1.

6.6.1 DAQ Requirements: Signal Dynamic Range and Data Types

To address the need for a wide dynamic range of ≈ 35 -bit, an enormous signal dynamic range beyond normal ADC dynamic ranges, an advanced more complex signal and data chain providing this range is required, see schematic in Fig. 6.31. Also, unlike conventional sonography systems which typically use planar transducer arrays and apply time-gain control (TGC) schemes to compensate for depth-dependent attenuation, the 3D USCT III system employs a surrounding hemispherical array. The resulting differences in propagation distance and timing between the various emitter-receiver pairs are too large for a common TGC scheme. Therefore, a more general approach was adopted.

As described in the dynamic consideration Sec. 5.2.1, the base of the signal dynamic range is the ADC with 12-bit. After that comes the so-called “Auto Gain”-feature implemented in the FPGA in the AMC. In a pre-shot step, the peak-to-peak value of the signal in every channel individually is determined and a variable gain amplifier is set to an amplification value up to 10,000 to compensate low-pressure signals and maximise the utilisation of the 12-bit dynamic range of the ADC [Tch23]. This is set individually per channel and per shot and the gain values are stored in the header per AScan. The mapping of the amplitude peak-to-peak detection to the amplification is shown in Fig. 6.29 left. The goal of the mechanism is to maximise the signals to 90% of the dynamic range of the 12-bit ADC.

After that, an anti-aliasing filter (AAF) chain of 209th order provides a sharp AAF filtering window and decimation to 10 MHz for space-effective storage of the data. The data comes with 20-bit output from the filter and is fed into an averaging mechanism with a maximum of 1024. The averaging adds 10 bits more. The now potentially 30-bit data after averaging and filtering needs to be mapped to a smaller storage format. Two storage formats were implemented: an integer data type with 16-bit (INT16) and a half-float data type with also 16-bit (called FP16, not fully ISO compatible). The half-float has 4-bit exponent, 10-bit mantissa, and 1-bit sign. The signal dynamic range of this data type corresponds to 27-bit. The complete range of the data type is shown in Fig. 6.30 left.

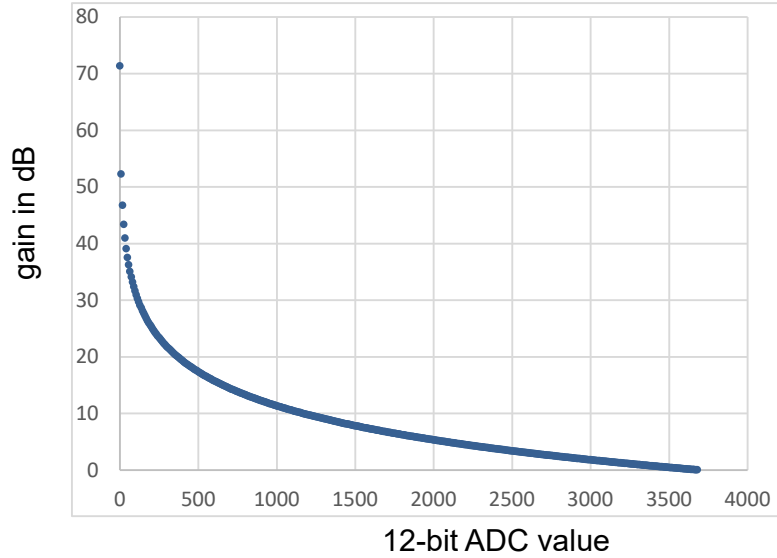


Figure 6.29: AMC Auto gain mechanism for dynamic range extension per channel. *X-axis value is peak-to-peak value from the ADC which defines the VGA's amplification given on the y-axis.*

To save as many bits as possible, especially for the weak reflection signals, the FP16 is the default data mode. The downside of this and any float data format is the varying resolution, which changes over the data type's range, see histogram in Fig. 6.30 right. As here only the already large transmission signal would benefit from INT16, and the smaller and low-SNR reflection signals benefit from FP16, we consider the trade-off in favor of the FP16. Also, the output of 30-bit fits well to the 10-bit plus 27-bit range of the FP16 format, reducing bit loss significantly compared to INT16. Overall, we can now achieve a signal dynamic range of maximum 30-bit – extra bits beyond the 12-bit ADC come from the digital filter (8-bits maximum) and averaging (10-bits maximum). On top, we have the +80 dB gain range from the auto gain procedure with the VGA, which corresponds to 13-bit. Overall, we can achieve in the best case scenario a 43-bit signal dynamic range for the US signal with these data types and processing chain.

After this step, the processing happens on servers and is not limited any more by the 3D USCT III's limited internal computational resources. Therefore, further steps in the processing chain transfer the data from FP16 to a default double-accuracy floating-point data type (floating point 64-bit, called FP64), which inherently offers the signal extended dynamic range for further processing and dynamic range extensions or SNR gains. The signal dynamic range is further increased by

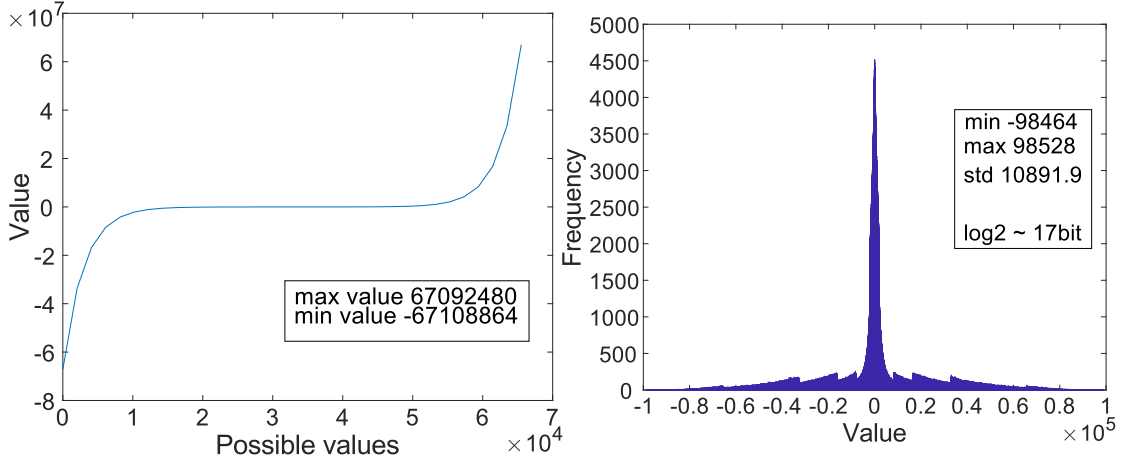


Figure 6.30: (Left) FP16 data type as implemented for 3D USCT III, given all encodable values in order. X-axis are the possible bit-values, y-axis shows the encoded value. (Right) histogram of a 3D USCT BScan with 2304 AScans with all arising sample values.

matched filtering as a later signal processing step – the SNR gain is $n^{0.5}$ proportional to the excitation length with n the length in samples. For instance, when we consider a significant coded excitation length extension from the 3D USCT II default of 12.7 μs to 600 μs , we would achieve an SNR gain of a factor of 6.87, which corresponds to 2.7-bit.

The last signal dynamic range and SNR gain potential is in the SAFT image reconstruction itself, the method we built our system around. When we consider the aimed-for $10 \cdot 10^6$ AScans and perfect data and calibration of the system, the transducers fully omni-directional and all components fully operational, we can assume perfect focusing of all AScans. Then, the SNR gain and dynamic extension can be considered as $(10 \cdot 10^6)^{0.5} = 3162$ factor, corresponding to 11.6-bit.

Overall, the 3D USCT III system therefore has a best-case signal dynamic range of all these components combined: 43-bit + 2.7-bit + 11.6-bit = 57.3-bit. This is beyond the worst case 35-bit dynamic range required in Sec. 5.2.1. The vision is that the multitude of dynamic range extending methods will offer enough headspace, which will hopefully lead to robust behaviour also under suboptimal real world conditions for all kind of tissues and structures. How much of this can be realised under realistic situation (noise etc) will be analysed later.

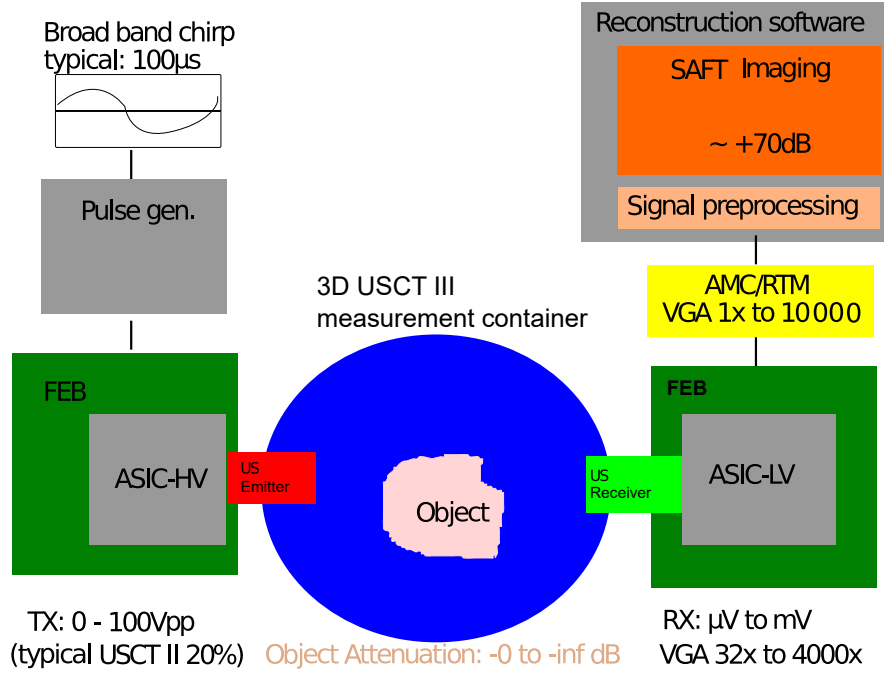


Figure 6.31: 3D USCT III schematic signal chain and dynamic, excitation and reception. All hardware components including imaging as software component.

6.6.2 3D USCT III DAQ Procedure

In the clinical trial in Hospital Jena with the 3D USCT III the measurement procedure took roughly 20 min, where most time was spent on the mechanical movement system, 70% according to profiles, see Appendix A.7.

With the defined DAQv5 solution performance numbers and the defined aperture size and transducer element numbers the following performance is expected: 384 DAQ acquisition channels with a mux of 6 can address the 2304 physical reception channels. The ADC sample rate was kept 10 MHz. Due to increased memory available as buffer memory amount and the strong requirement for more bandwidth the 3D USCT II bandpass undersampling approach was not taken but a classical oversampling approach. Two times ADC oversampling with 20 MHz, bandpass filtering between 0.5 to 4.5 MHz with and Nyquist frequency of 10 MHz.

The measurement container size and its associated RoI of $20 \times 20 \times 20$ cm, derived from the aperture and RoI simulations, result in a measurement phase time of 400 µs. Typical fade out time is a factor two of this. Unified transducer elements are 18, TAS number is 128. Assumed average number is four. As a result, the

total measurement time is

$$T_{\text{meas}} = N_{\text{MUX}} \cdot N_{\text{TAS}} \cdot N_{\text{el}} \cdot (T_{\text{fade}} + T_{\text{scan}}) \cdot N_{\text{avg}} = 6 \cdot 128 \cdot 18 \cdot (400 \mu\text{s} + 400 \mu\text{s}) \cdot 4 = 44.23 \text{ s.} \quad (6.1)$$

With the assumed number of virtual aperture positions of two, this system end up with a pure measurement time of 88 seconds, without overhead and positioning time of the motor driven rotation.

Motor positions and lift transmissions was optimised so the expected mechanical movement time was now expected to be reduced to few seconds per MP.

6.6.3 Measurement Process for 3D USCT III

The overall measurement process involves several tasks on different time-scales and urgencies. While the process is a derivative of the 3D USCT II procedure, some improvements were implemented:

- Hardware parametrisation phase, without patient.
- Hardware initialization and deinitialization phase, without patient.
- Measurement phase, with patient.
- Serialisation of data, without patient.

The most critical phase is the measurement phase, which should be as fast as possible for optimal, unblurred images of the patient. While the initialisation and deinitialisation phase does not impede the imaging performance it might impede the patient throughput in a hospital setting, which is an important economical parameter. The last phase, i.e. the serialisation of the data, is decoupled intentionally from the measurement. Rational here is that from previous experience on utilising hospital IT infrastructure and also by feedback from commercial interested parties an autonomous medical device which can carry the measurement for the patients of one day is preferred, e.g. for the use case operating in rural areas in developing countries.

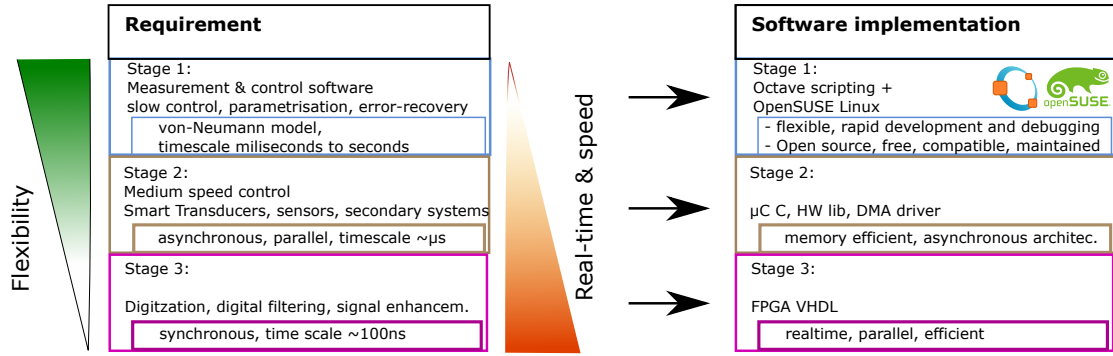


Figure 6.32: Block diagram of the DAQ software architecture concept (left) to implementation hierarchy (right).

There are parametrisation, slow control, high-speed data taking and serialisation and storage tasks. Technical data logging and error recovery and tolerance were defined goals for the overall system. Project requirements were also the compatibility with software and firmware already available at the IPE and project.

6.6.4 Software Architecture

The principle architectural concept for the 3D USCT III software stack is shown in Fig. 6.32.

The new 3D USCT III TAS requires a μC for digital communication and parametrisation with the electronic components in the TAS like the ASIC. A Texas Instrument MSP430 μC was chosen to be compatible with previous generation, allowing the reuse of existing code. The serial bus protocol was changed to RS485 to be capable of having more communication resilience and error detecting capabilities on the data transfer and parametrisation step. Beside being the bridge to the ASIC configuration concurrent asynchronous temperature tracking was another task put onto the μC . The core of this implementation was done well by Zewei Lu and is described in his thesis[Lu24].

For higher-level software components, an open-source approach was deliberately selected. The primary motivation is the ability to perform transparent and systematic code reviews, which are a key element of software verification and validation (V&V) in medical device development. Open access to the source code facilitates independent inspection, traceability of changes, and reproducibility of results, all

of which are essential for regulatory assessment and long-term maintainability.

A second motivation is the commercial and translational perspective: reliance on proprietary or restrictive licenses can constitute a barrier to technology transfer and may introduce significant cost and dependency risks during large-scale production and clinical deployment.

As conclusion from this considerations, and after talking with their support branch, SUSE Linux 15.3 was chosen as PC operating system, also due to the minimal license fee burden and the safety about long term support, planned obsolescence like with proprietary OSes (e.g. the case Windows XP) is impossible. This Linux based operation system is known to be a versatile and accessible platform by having included a wide range of drivers to access all kinds of standard utility hardware like SSDs, monitors, keyboards and communication interfaces like Ethernet, USB, serial ports etc.

To be compatibility with the MathWorks MATLAB software infrastructure and expertise of the IPE USCT project, GNU OCTAVE[GNU26] an open-source clone was chosen as scripting platform for parametrisation, slow control, logging, debugging and visualisation and other high level tasks.

Low level tasks, like bridging the gap to non-standard hardware like our AMC FPGA boards, were implemented in the C programming language.

6.6.5 Data Acquisition Procedure

After the base system architecture is defined and the capabilities for communication to the components is established the time critical components should be considered, implemented mostly in the Xilinx 425 and 350 FPGAs. It was made sure that the capabilities of the FPGA in multipliers, required for our AAF filter, and logic resources and RAM are sufficient to all the design of a measurement scheme which can cover a large SNR. This flexibility allowed us later the flexible adaption to specific measurement and object requirements, yet unknown to us.

The FPGA DAQ system was designed flexible with a large range of SNR gain in mind. Fastest measurement mode without averaging to 1024 averaging should be

possible, offering a maximum factor of 32 of SNR gain on signal level. Additionally the CE length can be chosen between 1 and 32k samples of 40 MHz, offering more potential for SNR gain. Both parameters will come with a measurement time cost, the optimum between both needs to be defined.

The SNR equivalent measurement procedures, starting from naive porting 3D USCT II measurement procedure to improved procedures, utilising all capabilities for 3D USCT III are shown in Fig. 6.33. It can be seen that with the last scheme for the 3D USCT III, the measurement time for a cycle is significant decreased from 51 ms (scheme 0) to 6 ms (scheme 3), which is faster than the 3D USCT II (38 ms), see Tab. 6.2.

6.6.6 Definition of the Technical Measurement Procedure for the QUSTOM study

For the QUSTOM project and the associated planned study, see Sec. 8.1, there should be up to three to four patients per measurement day in a clinical setup be measured with the 3D USCT III.

Planned was a 3D USCT III measurement procedure of approximately 10 million AScan per breast. Both breast should be measured, therefore with a measurement window of 400 μ s and utilisation of all transducers and two motor positions this would yields a data amount of 81 GByte. This data amount fits into the 96 GByte buffer memory on the AMC cards. That this measurement procedure fits completely in the volatile buffer memory allows for a decoupling of the serialisation (non-volatile data storage) to a permanent storage (like SSD) after the critical measurement time with the patient.

The hospital had IT policies, demanding no direct Ethernet connection for non approved devices to the hospital's network. Possible was only internal and local storage solutions. The medical isolated 1 GBit Ethernet port and the SSDs internally were therefore the only options.

Short before the project started, two core changes were required to have optimal performance for FWI: the measurement should produce 20 million ASans per

Table 6.2: 3D USCT III measurement schemes: signal-to-noise ratio versus measurement time. Overhead due to movement, preparation, and data transfer is neglected.

Measurement mode	Averaging & CE length	Rel. SNR gain	Effective time window	Time per shot	Time for 10^7 AScans
USCT II	Averaging: 64 CE: 12.7 μ s Motor positions: 12	1.0	300 μ s	600 μ s	14.5 min
USCT III scheme 0	Averaging: 64 CE: 12.7 μ s Motor positions: 2	1.0	400 μ s	800 μ s	23.6 min
USCT III scheme 1	Averaging: 16 CE: 52.2 μ s Motor positions: 2	1.0	400 μ s	851.2 μ s	6.28 min
USCT III scheme 2	Averaging: 16 CE: 52.2 μ s Motor positions: 2 DAC delay: -100μ s	1.0	400 μ s	751.2 μ s	5.5 min
USCT III scheme 3	Averaging: 16 CE: 52.2 μ s Motor positions: 2 DAC delay: -100μ s Fade-out: 20 + 10 μ s	1.0	400 μ s	381.2 μ s	2.81 min
USCT III scheme (max. SNR)	Averaging: 1024 CE: 519.2 μ s Motor positions: 2 DAC delay: -100μ s Fade-out: 20 + 10 μ s	25.7	400 μ s	849.2 μ s	400.7 min



Figure 6.33: Measurement modes realised in 3D USCT III.

Table 6.3: 3D USCT III measurement schemes for Barcelona: signal-to-noise ratio versus measurement time. Movement, preparation, and data transfer are neglected.

Measurement mode	Averaging & CE length	SNR gain	Eff. window	Time per shot	AScans	Overall time in min
USCT III scheme Barcelona	Averaging: 8 CE: 400 μ s Samples: 5120 Motor positions: 4	1.98	212 μ s	542 μ s	20×10^6	4.00 min
USCT III alternative scheme						
(proposed, not selected)	Averaging: 4 CE: 707.2 μ s Samples: 8192 Motor positions: 4	1.87	212 μ s	849.2 μ s	20×10^6	3.13 min
USCT III original scheme						
(proposed, not selected)	Averaging: 8 CE: 519.2 μ s Samples: 8192 Motor positions: 2	2.26	400 μ s	849.2 μ s	10×10^6	3.13 min

breast with four motor positions. The overall measurement time defined for the patient part was given as 20 min. This resulted in a challenge as the RAM memory can only store 2.37 motor positions at once. A delayed and decoupled readout of the data was not a feasible solution anymore. The sequential storing on the internal SSD with 500 MB/s bandwidth would alone take 6.9 min extra time. Hence, it would be challenging to stay inside the planned time budget per patient. Therefore, a newly implemented solution was a concurrent read-out pattern, which brought the internal CPU under quite some stress but was effective. The resulting measurement process is depicted in Fig. 6.35. The resulting measurement times are given in Tab. 6.3.

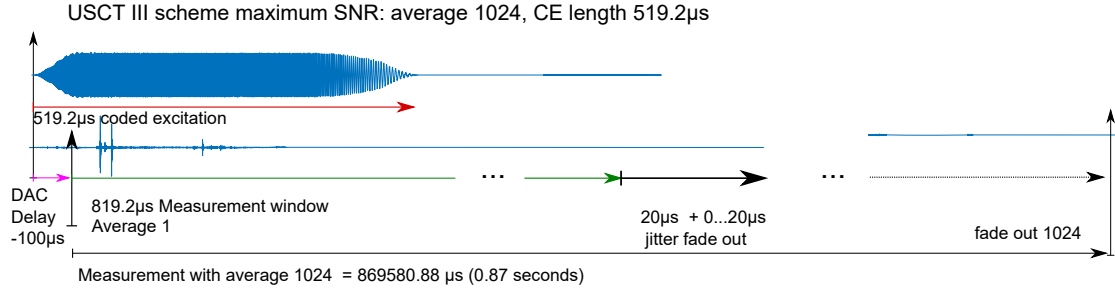


Figure 6.34: *Maximum SNR measurement modes technically realised in 3D USCT III, factor 23.1 over default modes see Fig. 6.33.*

6.6.7 SNR Optimal Measurement Procedure

SNR gain process can be considered as the process of reproducing the intended information extraction process in a coherent reproducible way while the non-coherent probabilistic noise, is averaged out. As a result, the SNR gain for well behaving Gaussian noise is $SNR_{gain} = \sqrt{n}$ with n as the number of measurements and SNR_{gain} the ratio in increase.

While a good trade-off between measurement length, coded excitation length and aperture diameter was already discussed at Sec. 5.1, we now extend the perspective including the average and balancing these two options for optimal SNR gain while minimising measurement time.

The lengthening of the excitation is always more time efficient for time-budget effective than averaging, see Fig. 6.36. Therefore, it was made possible for this generation of 3D USCT that the capability to excite very long coded excitation chirps was provided. It was realised, as a DAC of 32k samples with 40 MHz sampling rate, it became feasible to excite a arbitrary encoded CE of 80 ms length. This is far longer than any needed measurement window length. The longest technical possible measurement windows length in 3D USCT III is 819.2 µs, limited currently by FPGA resources. Unfortunately, this maximum measurement window length limits also the maximum usable CE length, as the complete pulse train for the complete window needed to be measured and fit into the measurement window. Therefore the the FPGA resources where the measurement window was implemented for the digital AAF processing becomes the bottleneck. The maximum is 819.2 µs which is roughly a doubling of the minimum needed measurement

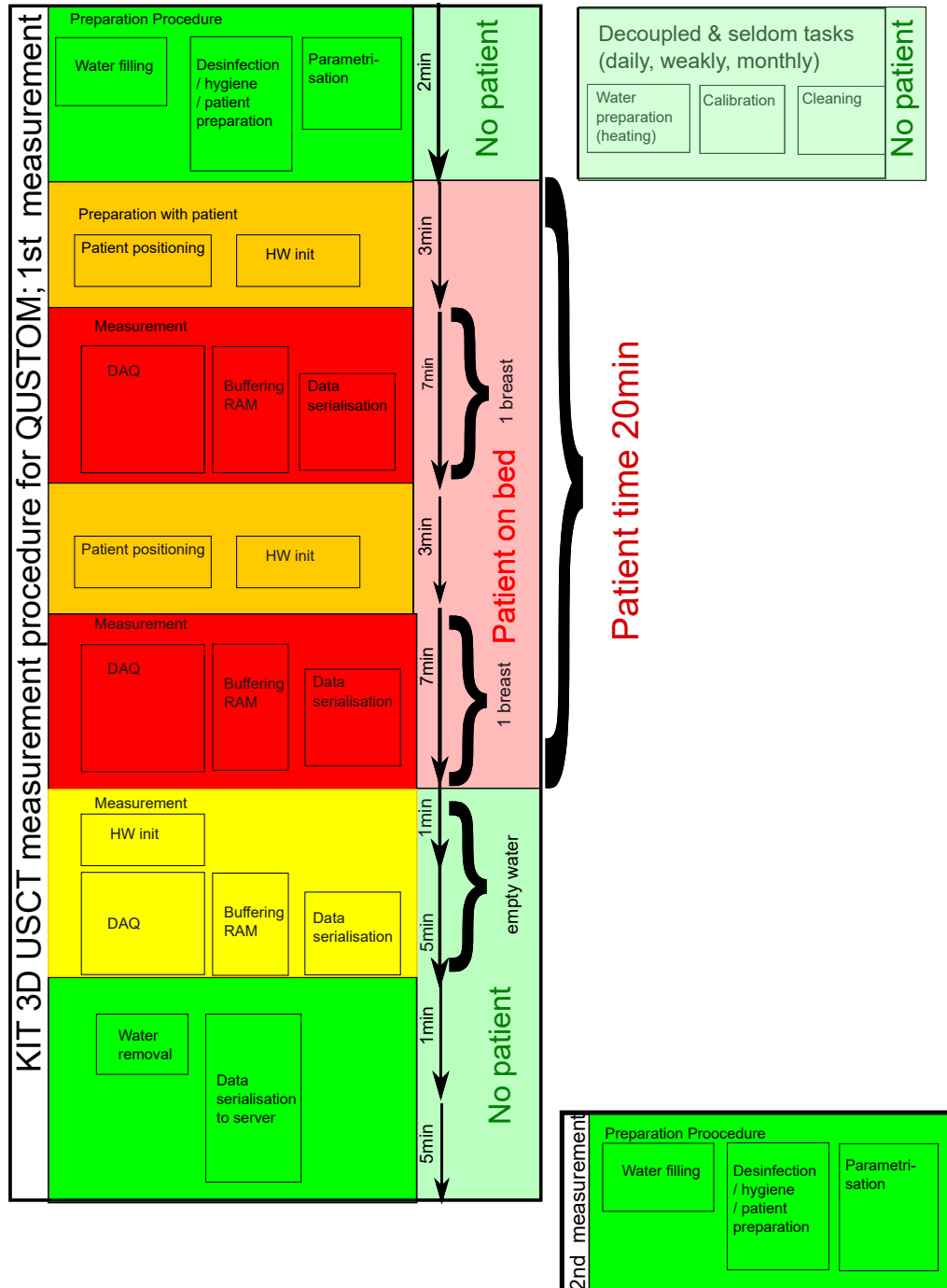


Figure 6.35: Measurement procedure for the KIT 3D USCT III QUSTOM study in Barcelona, extended to four motor positions. The y-axis represents the temporal progression of the procedure. Along the y-axis, the diagram is structured into functional blocks, some of which can be executed in parallel. Green blocks are not time-critical. Red blocks denote the time-critical measurement phases. Orange blocks represent periods during which the patient remains positioned in the device, but no data acquisition is performed. Yellow blocks indicate reference measurements performed with water.

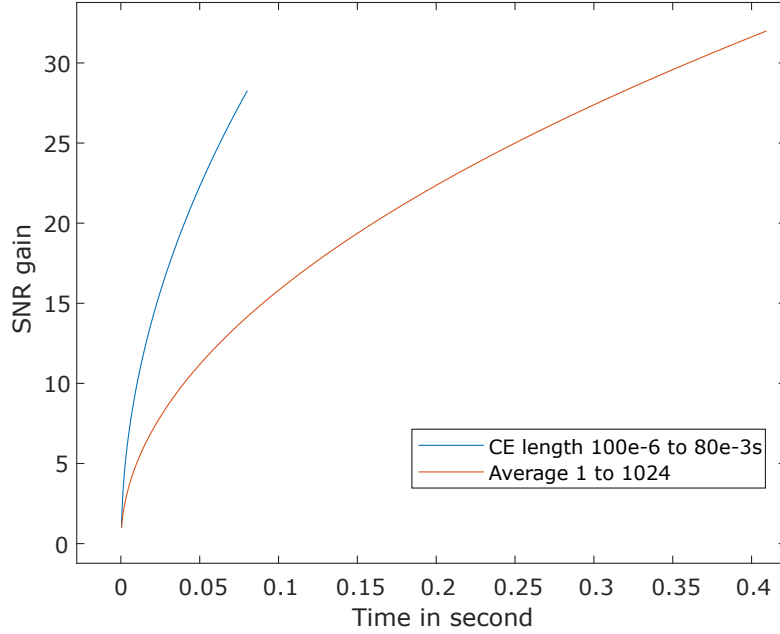


Figure 6.36: Optimal relative SNR gain, compared to base CE length of $100\ \mu\text{s}$ and a measurement window of $400\ \mu\text{s}$. X-axis overall measurement time, y-axis SNR gain. Blue curve, equivalent excitation length extension over time. Red curve, equivalent average increase over time.

window of $400\ \mu\text{s}$. This is much smaller than the potential time of the DAC. The maximum SNR gain achievable by this is therefore a factor two, when we consider a base CE length of $100\ \mu\text{s}$.

To overcome aforementioned limitations, another trick was implemented, the decoupling of the measurement window and the excitation window. This utilises the property of the system that in the first $100\ \mu\text{s}$, see BScan Fig. 6.39, there is little valuable content in these time windows before the transmission pulse. Extreme emitter receiver combinations where the transmission pulse fall into this time window, typically not contribute to the information gathering as these transmission waves have not passed any object in the RoI.

By decoupling the measurement start time and the excitation time point we can gain $\approx 100\ \mu\text{s}$ in possible measurement length. This allows to extend the measurement window to $9182\ \mu\text{s}$, which allows to extend the CE length by another $100\ \mu\text{s}$ and increasing the maximum SNR gain to factor 2.23 (for a base of $100\ \mu\text{s}$ CE length). Implemented is this feature technical as “DAC delay” to decouple the

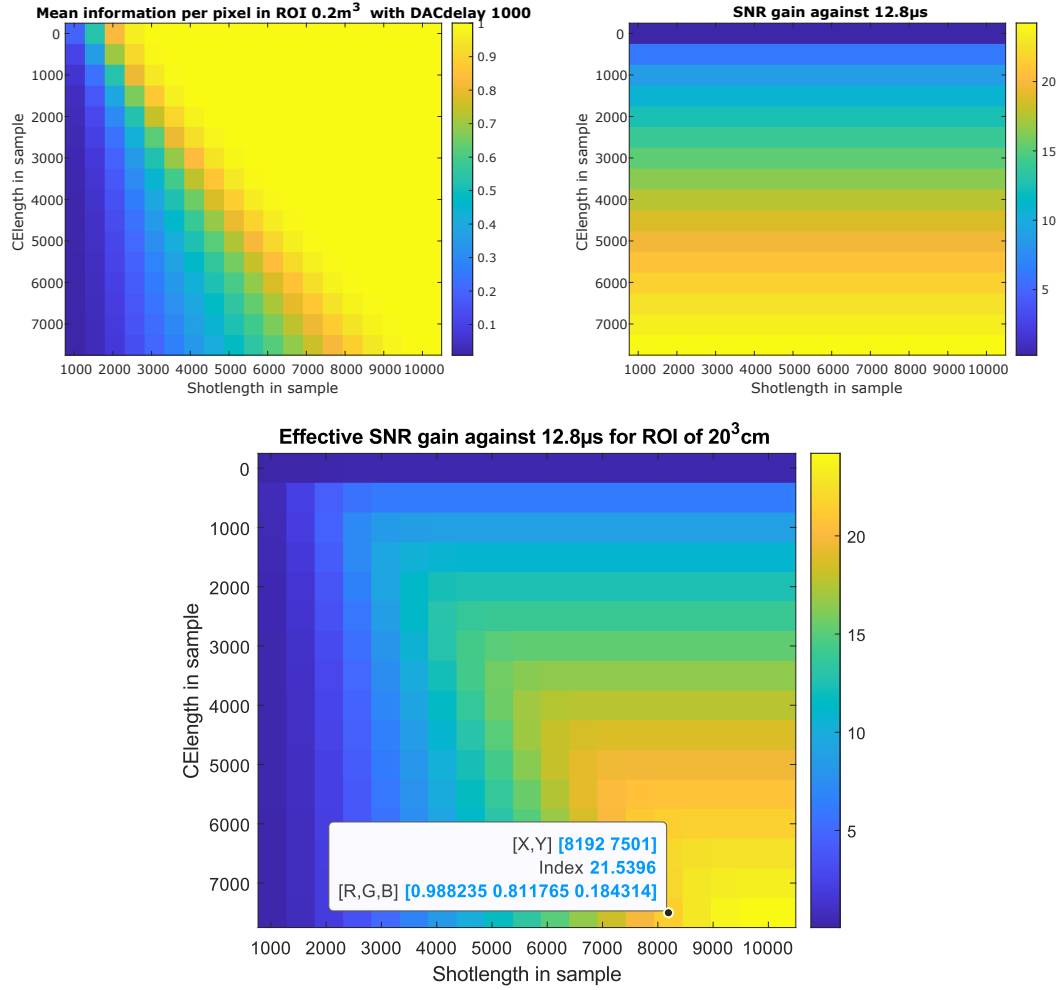


Figure 6.37: *Effective SNR gain considering all frequency components as information carrying equally from the RoI and excitation. The information gathered by the CElength window, potentially cutted by the shot length window, effectiveness colorcoded as yellow for 1 (all info, not cutted) and dark blue 0 (no info, cutted before RoI) due to CE cutting. (Top left) CE length over shotlength. (Top right), SNR gain over CE length. (Bottom) Combination of both, the effective SNR gain achievable.*

excitation from the measurement window, which is represented in Fig. 6.39. One example with no DAC delay (0 µs, left column) and one with DAC delay 500 samples (-50 µs, right column) is depicted. This is significant more efficient than a averaging based SNR gain which would have needed an extra 400 µs to achieve the same SNR gain, see Fig. 6.37 and Fig. 6.38. Here, all frequency contributions are considered equal valuable information. With increasing CE length the achieved SNR gain is rising, beyond a point that the lost information is bigger than the SNR gain, forming an optimum for each shotlength. In Fig. 6.38 we see on the left

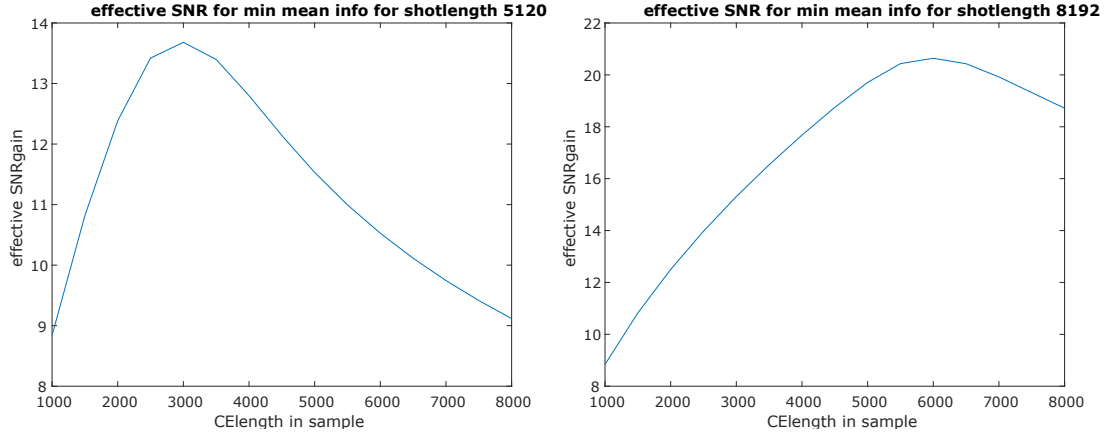


Figure 6.38: Maximising SNR gain under information carrying consideration for a fixed shot length window and 3D USCT III RoI and aperture but varying CELength. (Left) graph for shotlength 5120, SNR maximum is at around CELength 3000 10 MHz samples. (Right) graph for shotlength 8192 (hardware maximum), the optimum is around shotlength 6000 10 MHz samples.

the maximum SNR gain is achieved for a CELength of 3000 samples while having a fixed shotlength of 5120 sample, while on the right graph the maximum SNR is achieved for a CELength of 6000 sample for a fixed shotlength of 8192, which is the hardware maximum.

6.6.8 Optimal Bandwidth Excitation for Flat-Spectrum Response

The inherent frequency response of the 3D USCT III system is not flat across the intended measurement range. In particular, the low frequencies below 1 MHz rise above the noise floor only under extensive averaging, resulting in a limited usability for low-frequency imaging tasks.

Low-frequency performance was a major requirement for the Barcelona Full-Waveform Inversion (FWI) measurement scheme. FWI is computationally intensive but relies primarily on low-frequency components, as these are less sensitive to modelling errors and penetrate tissue more robustly than higher frequencies.

To recover more low-frequency energy from the system, a non-flat-spectrum coded excitation was implemented. The goal was to compensate for the system's non-uniform frequency response by pre-emphasising frequencies that are inherently

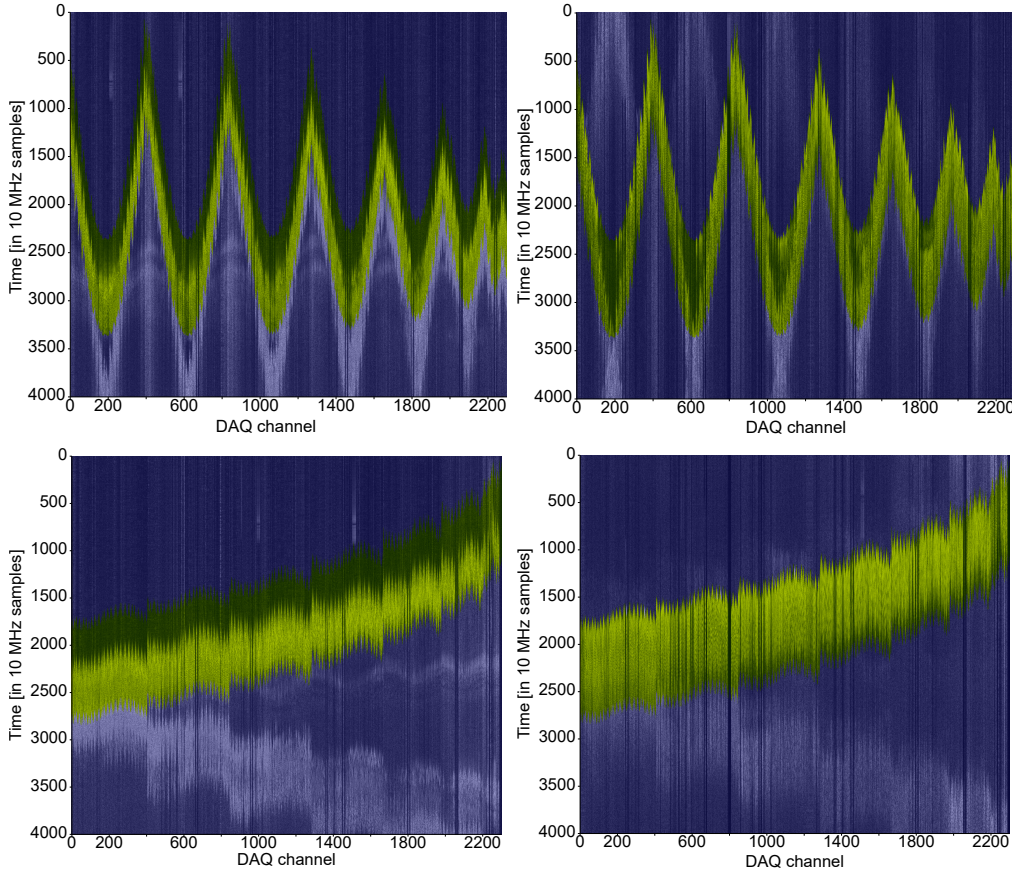


Figure 6.39: *BScans from Experiment exp597 with DAC delay 0 (left column) and exp835 and DAC delay 500 (right column). X-axis are the DAQ channels, y-axis is time, $t=0$ on top. Top row, transducer 1 element 5 as emitter, bottom row TAS 128, element 15. X-axis is channels, y-axis is time in 10 MHz samples. Colour coded green is the RoI, brightness indicates the AScans amplitude. Visible is the shift of the transmission pulse, optimising the utilised and saved data window.*

attenuated by the electronics and transducers.

The procedure is as follows: an empty-water measurement was acquired using a flat-spectrum chirp, and the mean system response over all transducers was computed (Fig. 6.40, left). Its reciprocal was then calculated and clipped at frequencies below 0.5 MHz to avoid excessive amplification of noise-only regions (Fig. 6.40, right). Multiplying this compensated curve with the original chirp yielded a modified coded excitation with a substantially flatter spectrum. Due to the 10-bit DAC used for excitation, highly attenuated frequencies in the centre of the band were encoded with only about two effective bits, limiting the achievable dynamic range.

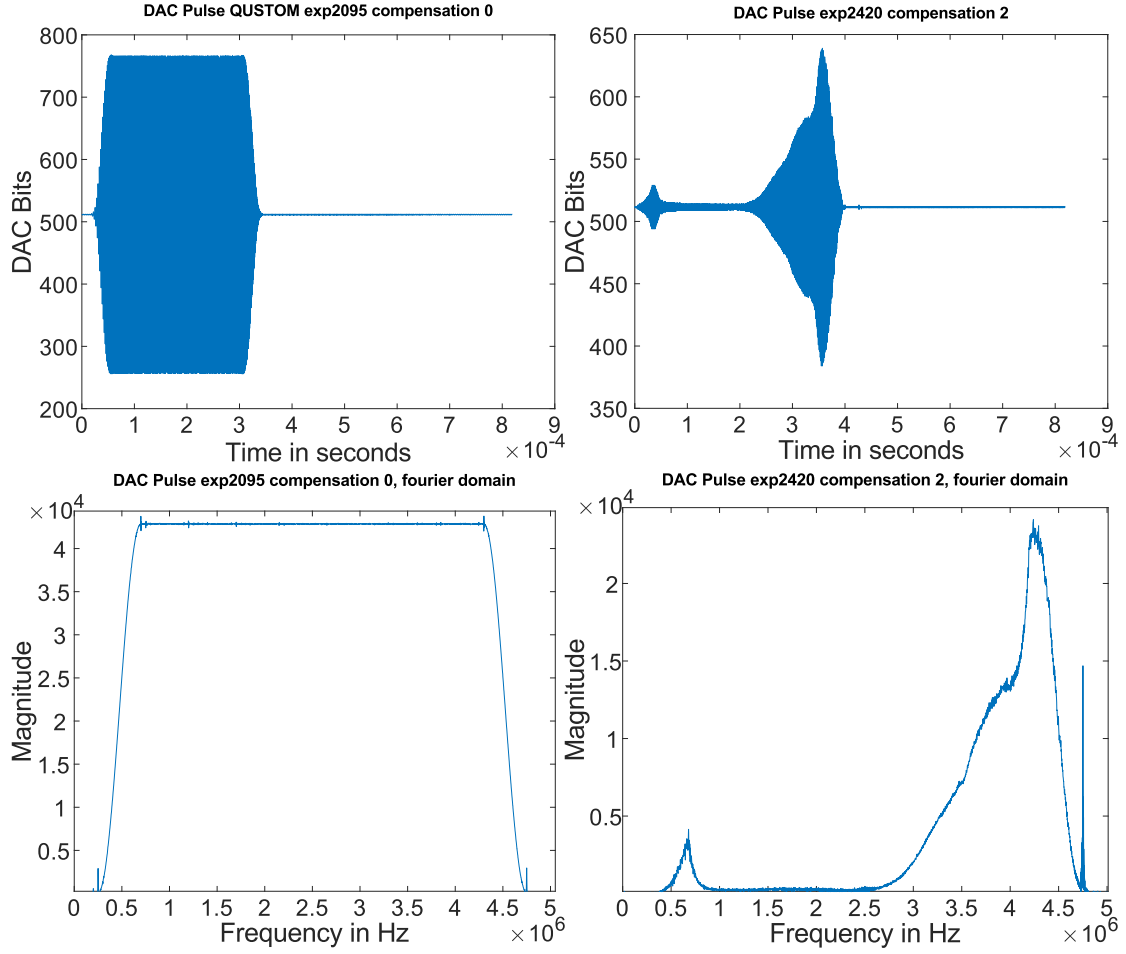


Figure 6.40: *Compensated chirp approach for Barcelona’s FWI-focused measurement scheme. (Left column) uncompensated chirp. (Right column) compensated chirp. (Top row) time domain. (Bottom row) magnitude in the Fourier domain.*

A representative compensated measurement is shown in Fig. 6.40 (bottom right). The high frequencies are significantly enhanced, and, aside from an overcompensation peak around 3 MHz, the resulting spectrum (blue) more closely follows the ideal flat coded-excitation shape (yellow). However, because the centre-frequency components had to be attenuated to stay within hardware voltage limits, the overall SNR decreased across the frequency band. The elevated noise floor (red) demonstrates that the SNR reserve is now being actively utilised to push beyond the system’s native -3 dB or -6 dB bandwidth.

While these long-averaging, compensated-excitation modes are unsuitable for clinical measurements due to their extended acquisition time, they remain highly valuable for research and phantom studies where bandwidth is important.

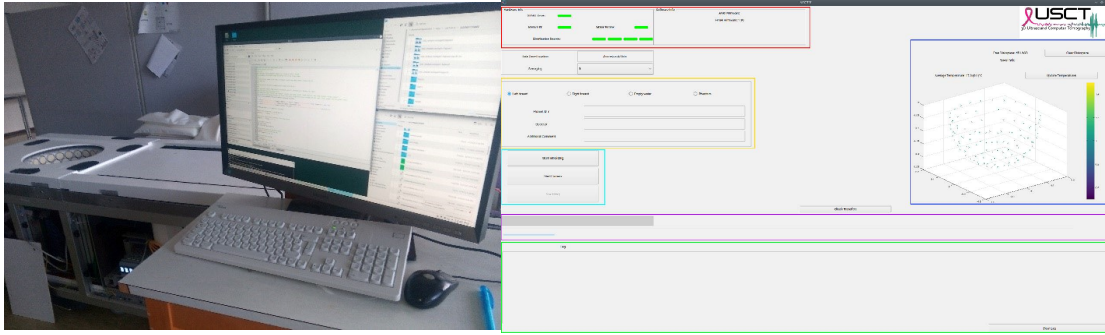


Figure 6.41: (Left) 3D USCT III.R DAQ running and measuring. Right on the monitor the open Octave UI with the “Autoshot script”. (Right) Software GUI, work flow is top to down: top row system health status (red box), operational marked green. The right side indicates TAS health status and temperature (dark blue box). The left middle area (yellow box) are the parametrisation fields. In the middle are, in bright blue, command buttons. On the bottom is the message area (green box).

A proposed but unused refined iteration of the compensated excitation, designed to mitigate the over excitation around 3 MHz, is shown in Fig. 8.9 together with further bandwidth-optimisation strategies in Sec. 8.3.3.

6.6.9 Software and GUI

One core goal from the beginning was that the data formats and meta data ideally would be compatible with the 3D USCT II reconstruction frame-work.

To achieve this the software has the following interdependent core components:

- Autoshot script (hidden from operator, maintainance)
- GUI (for medical operator usage)
- Parser (compatiblity layer)

and additional various system visualisation and analysis components (partly later introduced under system evaluation).

The “Autoshot script” is the core module for parametrisation, controlling and using the 3D USCT III hardware and system. Written in Octave, an open source variant of MATLAB, it offers easy scriptability, familiarity with the syntax and compatibility with the mostly MATLAB based software environment. Addition-

ally, being free open source software it offers benefits in being an easily reviewable medical software component (unlike proprietary and closed-source MATLAB). This is also important for later commercialisation, with its well-understood license model, which avoids additional license costs per device or per patient or other licensing (compatibility) complexities. The Autoshot script can be utilised for automatised technical test series and parameter explorations using the 3D USCT III as an experimental research device via a flexible software interface.

Also, the standardised measurement procedure for medical measurements is supported via this interface. In general the script is not meant for medical operators, but for researchers or maintenance use cases. For medical operators another software component was developed.

On the next abstraction level is the “GUI” which offers a suitable interface for non-technical medical device operators and offers just the subset of technical functionality required in clinical application in hospitals, see Fig. 6.41 right [Sch23]. Patient measurement relevant parameters like the “measurement type” (empty or left or right breast side), “patient ID”, “operator name” and data serialisation place (internal, network) can be set. Then the system status is represented in the GUI, checking and indicating suitable system health of components after boot up and initialisation. After initialisation and starting the measurement it gives feedback about current measurement status and successful or non-successful measurements process for fast repetition if needed.

After storage and transfer of the measurement data to the processing and imaging server, the last utilized software component is utilised, the “parser”. The parser performs data and meta-data checks on the serialised data from the experiment and converts and transfers the data into a format compatible with the USCT project’s reconstruction software.

Overall, the integration and commissioning of the complex hardware system could be achieved quite fast in the first iteration after availability of the integrated device. Also, the software framework proved to be robust enough to handle several upgrade and extension cycles in features and hardware, while keeping the measurement process and data export formats compatible.

7 Evaluation of 3D USCT III

7.1 System Self-Calibration and Error-Detection

One property intended for the 3D USCT III, motivated from the experience of 3D USCT II and its production challenges, is the system self-calibration capability. The 3D USCT device as complex research device composed from many components and having many degrees of freedoms, many influences on the assumed operation are possible.

Classical application of external tools for system assessment like oscilloscope or other external measurement tools is time consuming, expensive and very impractical in clinical environment.

Therefore, system self-checking and characterisation capabilities are considered essential to be able to identify problems fast in the commissioning stage and also later in the clinical operation phase. Methods and systems for self-characterisation on hardware and software level were planned and integrated in the system from the beginning.

Correct operation of the complex 3D USCT III system and technical setup should be assessed for correct operation with a test run or a calibration procedure.

Require capabilities in hardware and software were planned and developed to achieve this feature.

7.2 System Self-Characterisation and Evaluation via ASIC

One core mechanism for testing the complete DAQ and control chain is the internal coded excitation measurement capability, called CEMeasure, implemented in the ASIC. This CEMeasure characterisation involves the full chain from the coded excitation digital definition as 10-bit DAC values over 32k sample, the excitation triggering time, the analogue chain with amplification and switches chain until the front-end and the ASIC. In the ASIC the connection to the receiving channels is implemented as bypass via a small capacitive coupling component. This capacitive coupling connects the excitation input channel and the three output channels of low voltage receiver channels with a high attenuation, mimicking the transducer load and bypassing the HV amplification, see Fig. 6.24. The low voltage signal analogue signal is then transferred via the analogue cables until DAQ systems amplifier, filtering and digitisation stage.

Originally it was planned to utilise the ASIC even further and also implemented that the ASIC has the capability to create test signals on defined output channels to help identifying the reception signal chain by setting digitally an ASIC channel (almost the begin of the reception channel) to a specific value, which can than easily on the other end (DAQ software, digitisation at AMC, reconstruction software) identified and this reception channel correctly associated. Unfortunately, this feature was not functional and other means needed to be found to check these cases (see Sec. 7.10 with an algorithmic approach).

In the complete 3D USCT III measurement process this part of measurement is taken on the end after the time-critical patient measurement was finished. It is done parallel for the excitation physical lines, all TAS and all receiving channels with its muxes. Overall there are 2304 CEMeasure AScans, characterising the TAS's 128 excitation chains and the 384 reception lines and the FEBs six muxing setting combined response. The individual CEMeasure AScans are taken with the same digitisation parameters like the water AScans.

This self-characterization signal serves several purposes, the fast testing of the excitation and DAQ chain (independend of transducer performance or even hav-

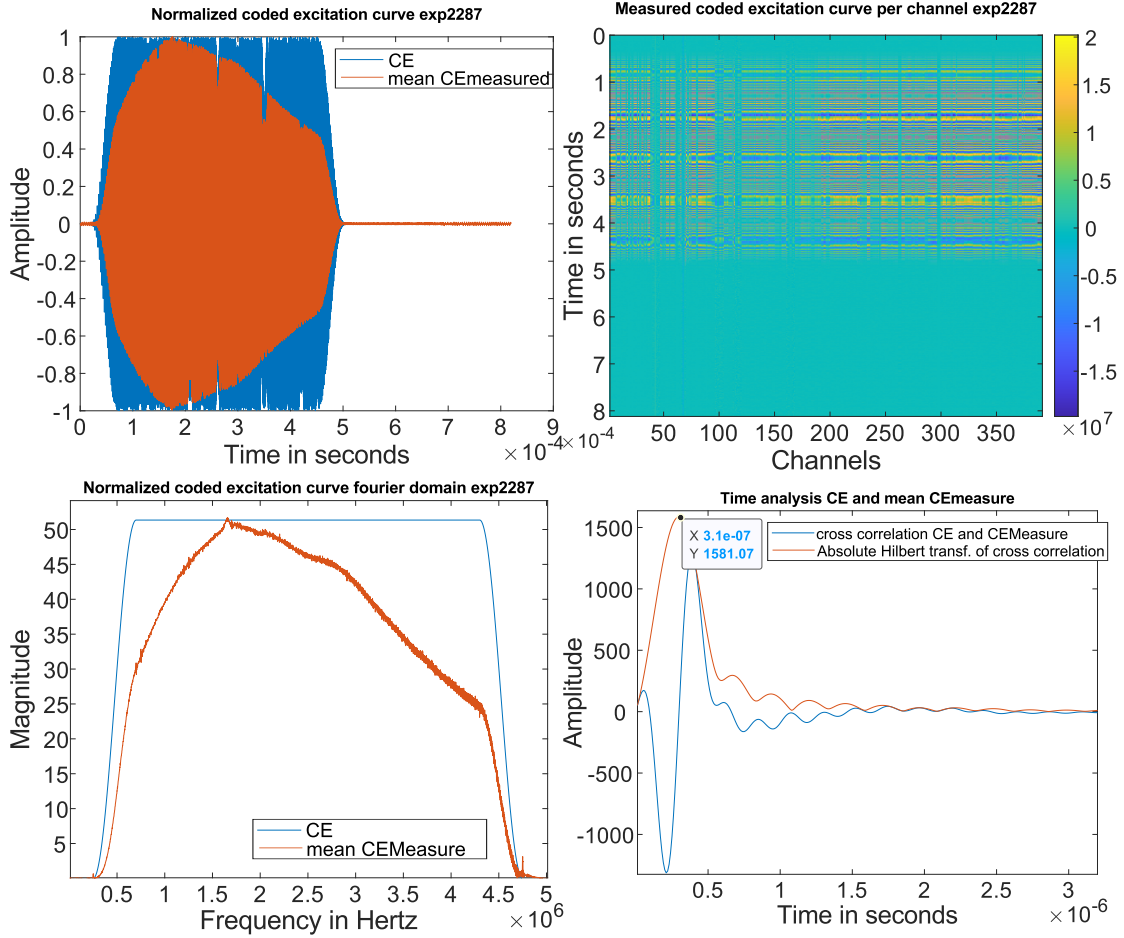


Figure 7.1: *CE measure and analysis. (Top left) blue theoretical excitation pulse, red measured CEMeasure pulse. (Top right) all CEMeasure channels, x-axis channels, y-axis time. (Bottom left) Frequency response, blue theoretical excitation pulse, red CEMeasure. (Bottom right) time response after filtering, blue optimal cross correlation response, red hilbert crosscorrelation response.*

ing them mounted or unmounted), the measurement of the analogue and digital chains delay and frequency behaviour and also for timing and jitter analysis and evaluation, see Fig. 7.1.

The biggest benefit in the reconstruction chain is the direct usability as system representing linear filter in the matched filtering step.

Definition of the effective SNR gains for the reference and measured matched-filter responses and the resulting relative SNR gain in percent. The SNR gains $G_{\text{SNR}}^{(\text{CE})}$ and $G_{\text{SNR}}^{(\text{CEM})}$ are computed as the ratio of the signal peak to the standard deviation of the first 500 samples, representing the noise baseline. The relative gain $G_{\text{rel}}^{[\%]}$

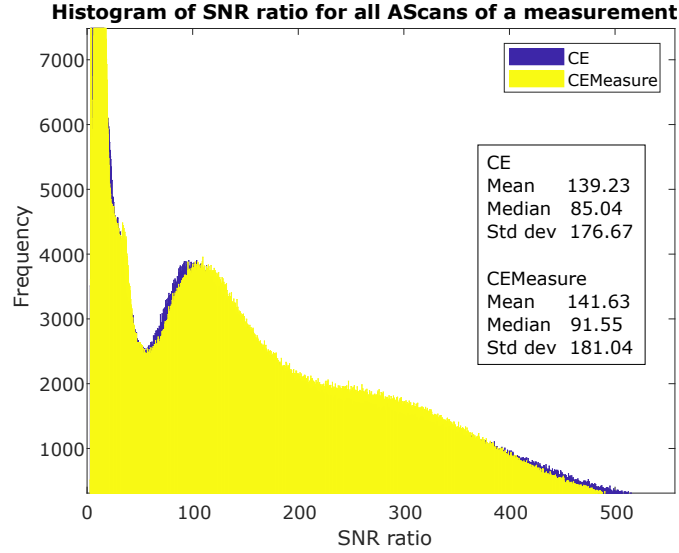


Figure 7.2: Histogram of SNR ratios for all AScans of experiment 2742. Yellow curve is matched-filtered response with CEMeasure, blue curve is matched filtered with the reference CE.

expresses the measured-to-reference ratio as a percentage, providing an intuitive measure of relative improvement, see Eq. 7.1,7.2,7.3.

The effectiveness of the CEMeasure signal as better system representing filter in the matched filtering process can be seen by an increased SNR ratio, see the extraction for a complete experiment Fig. 7.2.

$$G_{\text{SNR}}^{(\text{CE})} = \frac{\max(C_E^{\text{mf}}(n))}{\text{std}(C_E^{\text{mf}}(n), n \in [1, 500])}, \quad (7.1)$$

$$G_{\text{SNR}}^{(\text{CEM})} = \frac{\max(C_{EM}^{\text{mf}}(n))}{\text{std}(C_{EM}^{\text{mf}}(n), n \in [1, 500])}, \quad (7.2)$$

$$G_{\text{rel}}^{[\%]} = 100 \frac{G_{\text{SNR}}^{(\text{CEM})}}{G_{\text{SNR}}^{(\text{CE})}}. \quad (7.3)$$

For the timing checks, shown in Fig. 7.1 a matched-filter response was computed by correlating the measured excitation $C_{E\text{Measured}}(n)$ with the reference signal $C_E(n)$ in the frequency domain through complex conjugation of the reference spectrum. The resulting cross-correlation $r(n)$ (Eq. 7.4) represents the matched-filter output.

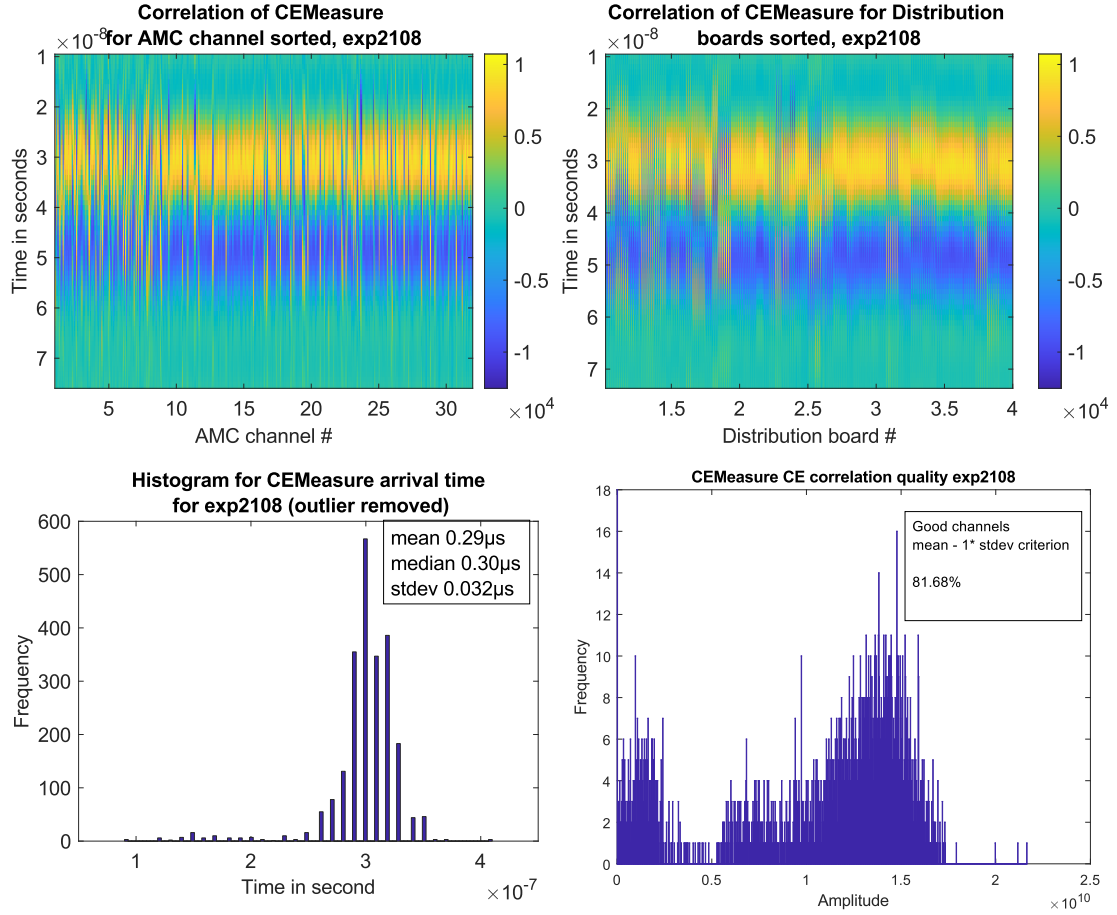


Figure 7.3: 3D USCT III system response for Barcelona study parameter: CEMeasured analysis signal synchrony exp2108 from QUSTOM.

To obtain an amplitude-independent representation of the signal response, the analytic envelope $a(n)$ was derived as the magnitude of the Hilbert transform of the matched-filter output (Eq. 7.5). Together, these operations form the basis for the detection and evaluation of the effective SNR gain. Computation of the matched-filter cross-correlation $r(n)$ and its analytic envelope $a(n)$.

$$r(n) = \mathcal{F}^{-1} \left\{ \mathcal{F} \left[\frac{-C_{E\text{Measured}}(n)}{\max |C_{E\text{Measured}}|} \right] \cdot \mathcal{F} \left[\frac{C_E(n)}{\max |C_E|} \right]^* \right\}, \quad (7.4)$$

$$a(n) = |\mathcal{A}\{r(n)\}| = \sqrt{r(n)^2 + (\mathcal{H}\{r(n)\})^2}. \quad (7.5)$$

For the 3D USCT's performance it was important to achieve an uninterrupted,

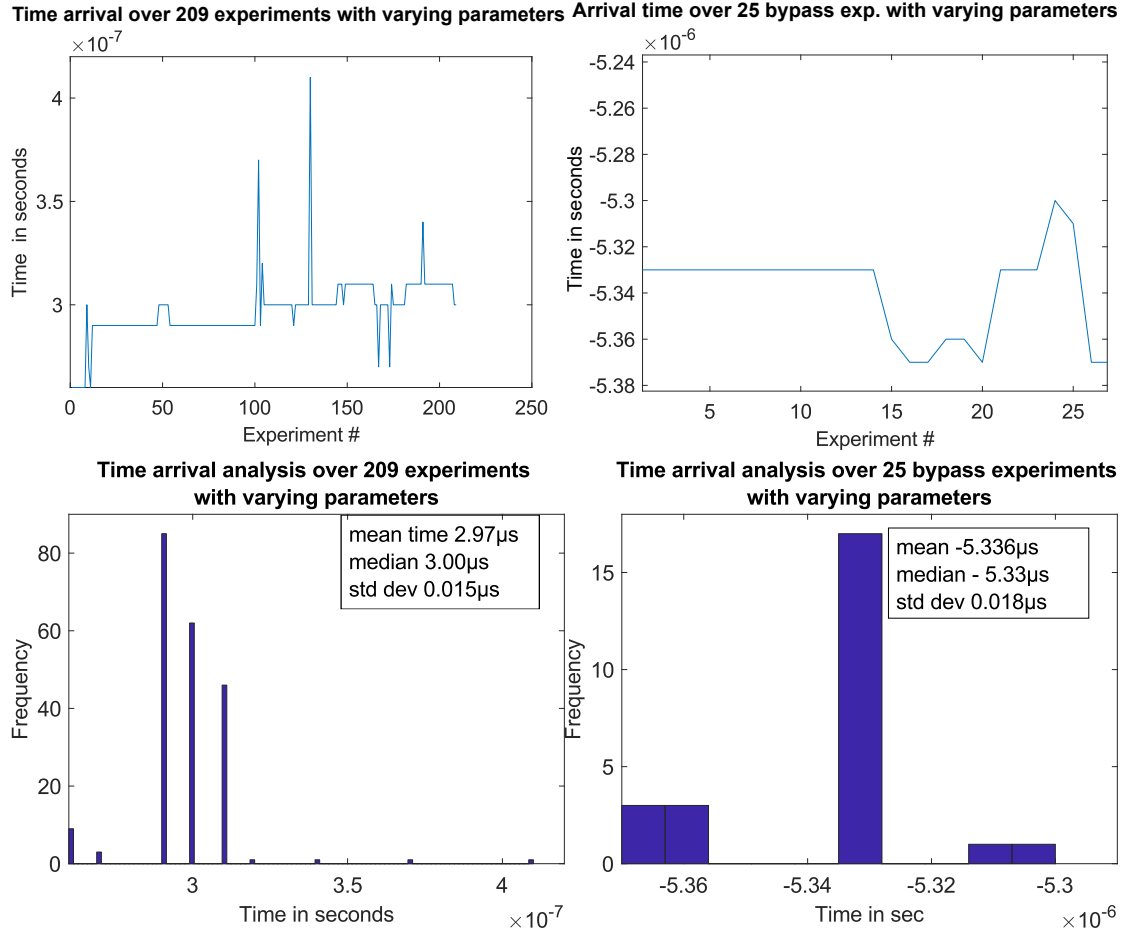


Figure 7.4: 3D USCT III synchrony analysis over multiple experiments. (Top row) CEMeasure arrival time for anti-aliasing-filter measurements on the left, on the right anti-aliasing-filter disabled measurements. (Bottom row) histogram analysis of both groups.

synchrony measurement procedure - unlike 3D USCT II which proved here to be error prone and unreliable. Unnoticed errors here would lead to impaired imaging quality or completely failing imaging procedure. As evaluation, additionally to the internal logging and checking mechanisms, the synchrony of the parallel measured signals was evaluated by comparing their temporal alignment and jitter characteristics, utilising the self-calibration capability of CEMeasure, see Fig. 7.3. This provides a quantitative measure of the channel-to-channel delay stability and test that the DAQ system maintains phase-aligned sampling across all 384 channels and MUXes within sub-microsecond tolerances, ensuring the phase fidelity required for coherent reconstruction.

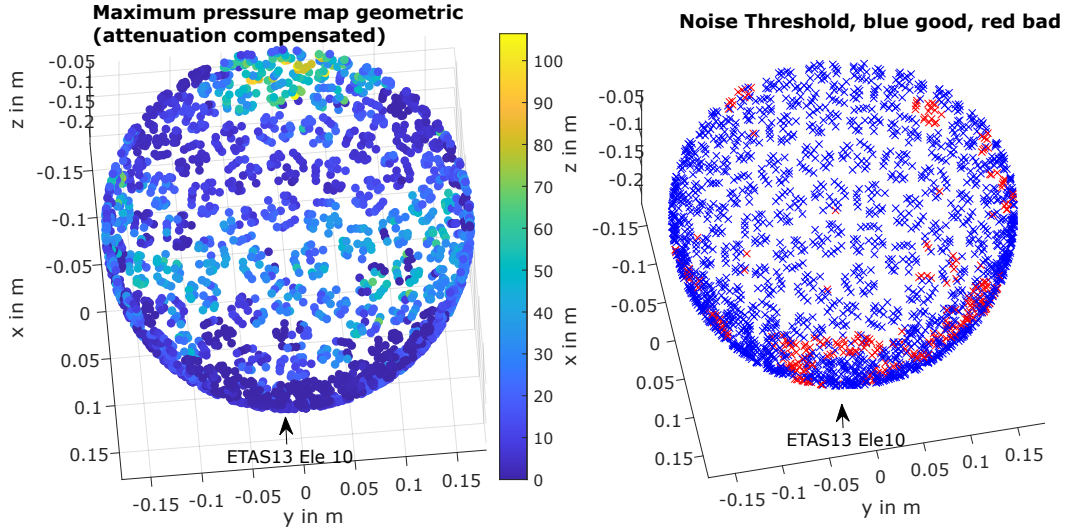


Figure 7.5: 3D USCT III system response, for emitter TAS13 element 10 from experiment 2095 from the QUSTOM study. Exemplary for emitter TAS13 element 10. Pressure is geometric attenuation corrected and normalised by the standard deviation. (Left) Sound pressure, colour coding is going from blue to yellow with increasing pressure. (Right) receiver transducer data (see Fig. 7.6) below 6σ (Eq. 7.7) SNR are red marked as bad, above blue and good.

The extracted mean value is $0.3 \mu\text{s}$ with a standard deviation of only $0.032 \mu\text{s}$, see Fig. 7.3. This variation is significantly smaller than the temporal half-wavelength criterion corresponding to the highest employed frequency of 4.5 MHz, which amounts to

$$\frac{0.5}{4.5 \cdot 10^6} = 0.111 \mu\text{s}. \quad (7.6)$$

Hence, the observed timing fluctuation utilises only approximately 29% of the permissible half-wavelength tolerance, confirming excellent temporal stability of the system.

Another result extractable is that 81.7% of the 2304 emitter receiver chain combinations were functional, see Fig. 7.4. The metric was, a channel a is considered good if its amplitude $A_{\text{ch},a}$ is smaller than one standard deviation below the mean of all channel amplitudes, i.e. $A_{\text{ch},a} < \sigma_A - \mu_A$.

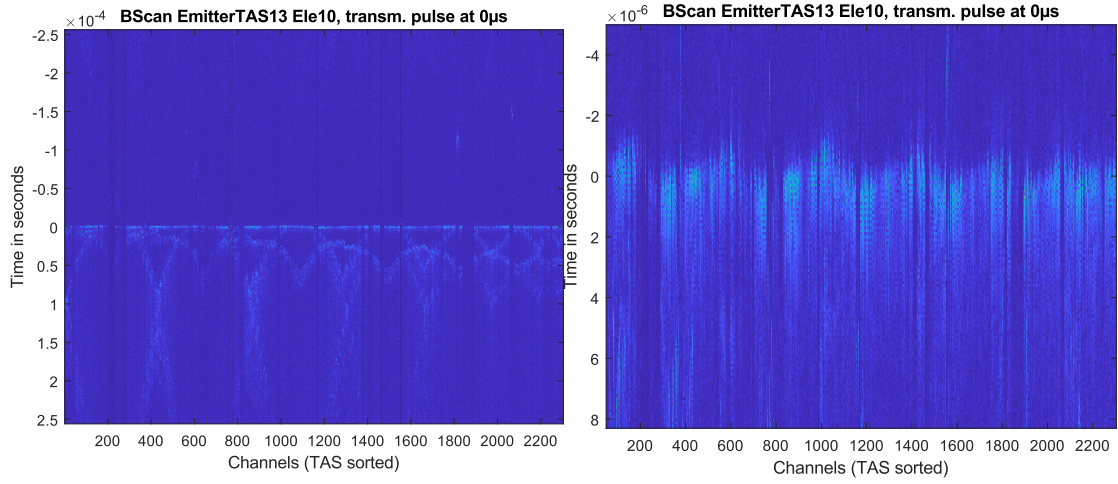


Figure 7.6: 3D USCT III system response, for emitter TAS13 element 10 from experiment 2095 from the QUSTOM study. (Top left) BScan with sorted channels for TAS. All AScans geometrically shifted, straight line represents then the transmission signal at $t = 0$ s. (Top right) zoom into BScan transmission pulse.

7.3 Transducer and Opening Angle Performance

For the evaluation of the transducer performance an empty experiment from the QUSTOM study campaign in Barcelona was selected, exp2095. Some example data and 2D BScan graphs for emitter TAS 13 and element 10 and all receivers are given in Fig. 7.6 and Fig. 7.7.

To define a binary transducer quality the following definition is formulated, see Eq. 7.7. This transducer-level quality definition, is applied for receiver and emitter each. For each transducer pair (k, ℓ) , a binary mask $T_{k\ell}$ indicates whether the signal-to-noise ratio $\text{SNR}_{k\ell}$ exceeds 6. The mean performance P_k of each transducer is compared to the global mean minus one standard deviation. Transducers satisfying this condition are classified as good, and the percentage of good units is given by P_{good} , see for instance Fig. 7.5 right. Indices k, ℓ denote transducers, and n the discrete time sample index.

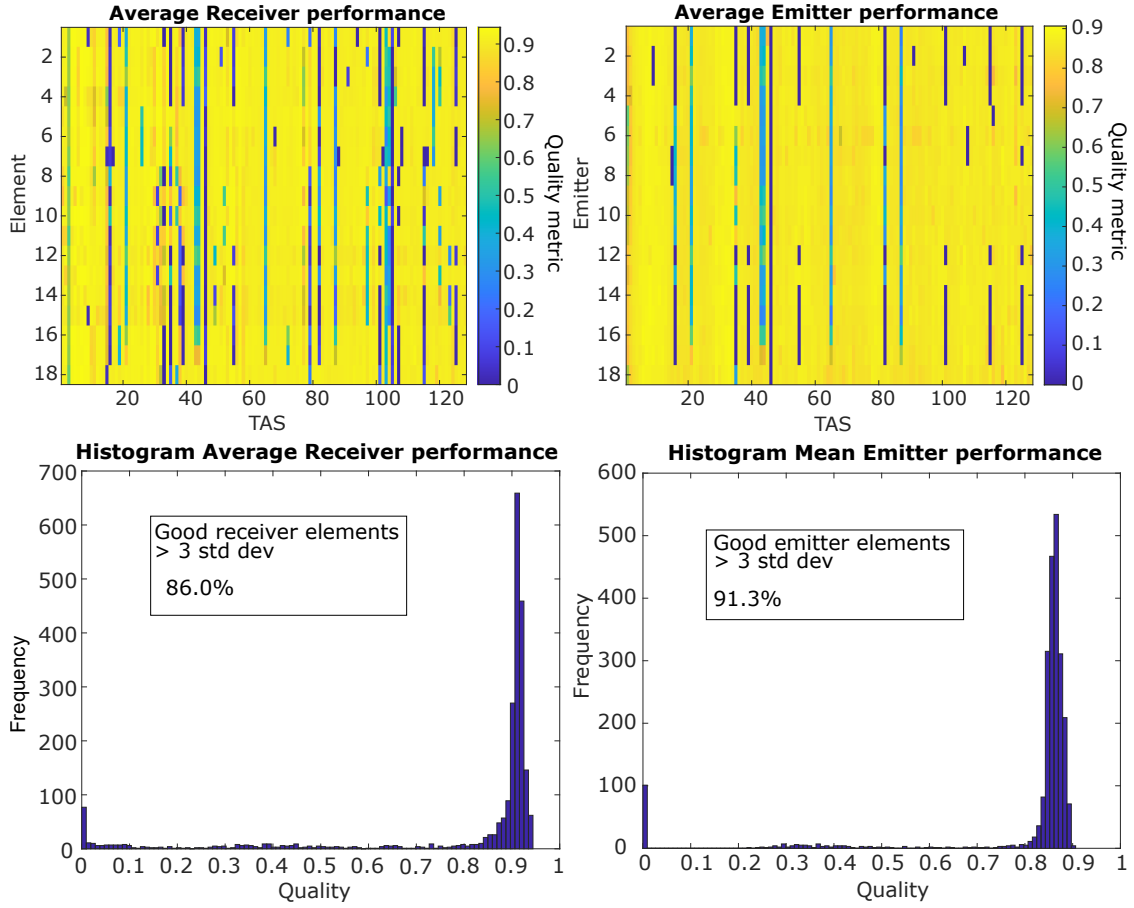


Figure 7.7: 3D USCT III system response, for experiment 2095 from the QUSTOM study. (Left column) receiver channel performance. (Right column) emitter channel performance. (Top row) association with TAS (y-axis) and transducer element (x-axis). (Bottom row) histogram.

$$\text{SNR}_{k\ell} = \frac{\max_{n \in \Omega_{\text{pulse}}} A_{k\ell}(n)}{\sigma(A_{k\ell}(n))_{n \in \Omega_{\text{noise}}}},$$

$$\Omega_{\text{noise}} = [2300, 2500], \quad \Omega_{\text{pulse}} = [2500, 2600],$$

$$T_{k\ell} = \begin{cases} 1, & \text{SNR}_{k\ell} > 6, \\ 0, & \text{otherwise,} \end{cases} \quad (7.7)$$

$$P_k = \frac{1}{|\mathcal{N}(k)|} \sum_{\ell \in \mathcal{N}(k)} T_{k\ell},$$

$$\text{Transducer } k \text{ good} \iff P_k > \bar{P} - \sigma(P),$$

$$P_{\text{good}} = 100 \frac{\sum_{k=1}^{N_T} \mathbf{1}\{P_k > \bar{P} - \sigma(P)\}}{N_T}.$$

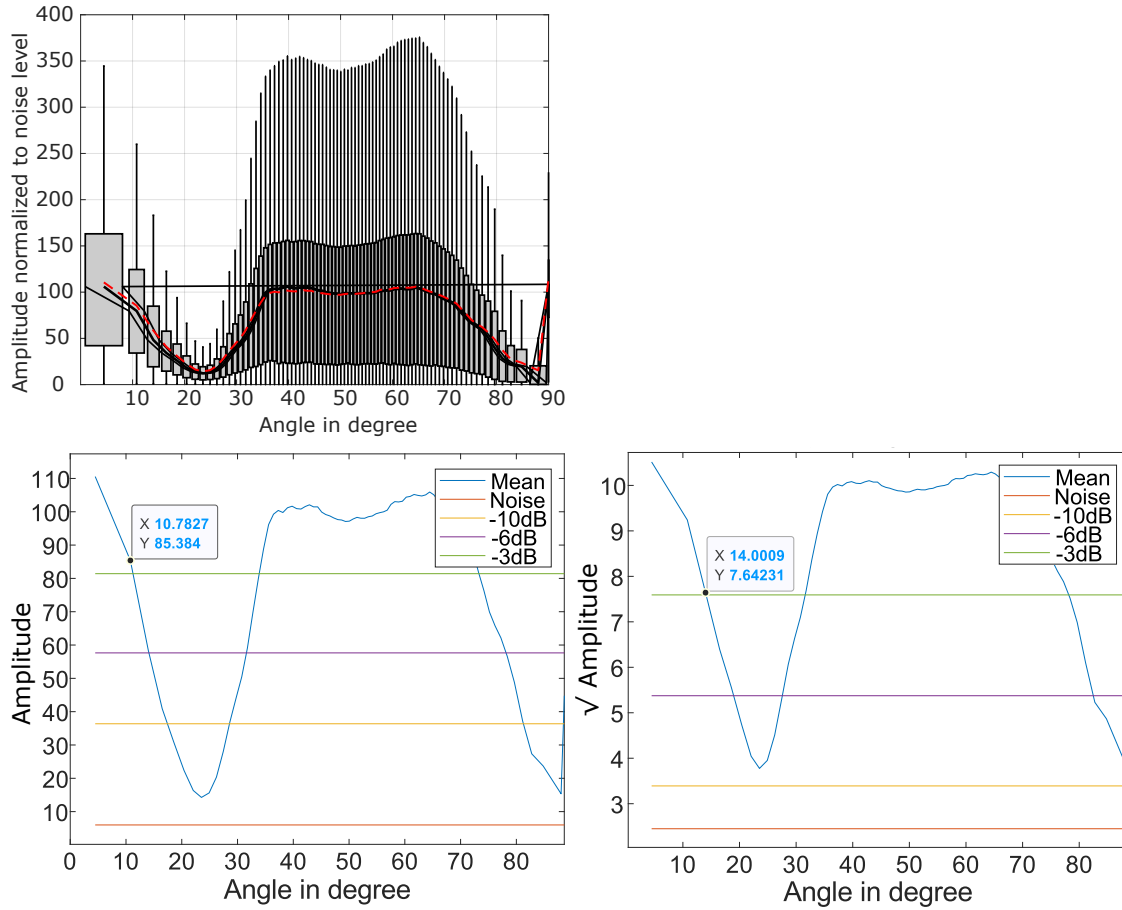


Figure 7.8: Averaged 3D USCT III pressure response versus angle relative to the transducer surface normal (experiment 2095, QUSTOM study). (Top left) Box plot of individual angle measurements with emitter and receiver responses combined. (Bottom left) Angularly averaged combined transducer response. (Bottom right) Estimated emitter and receiver responses obtained by square-root decomposition of the combined response assuming symmetric directivity.

The averaged 3D USCT III system response, for experiment 2095 from QUSTOM study is given under Fig. 7.8 The averaged angular sensitivity combined for 3D USCT III emitter and receiver, averaged over experiment 2095 is presented in Tab. 8.2 and the decoupled, approximative representing the transducer performance is given under Tab. 8.3.

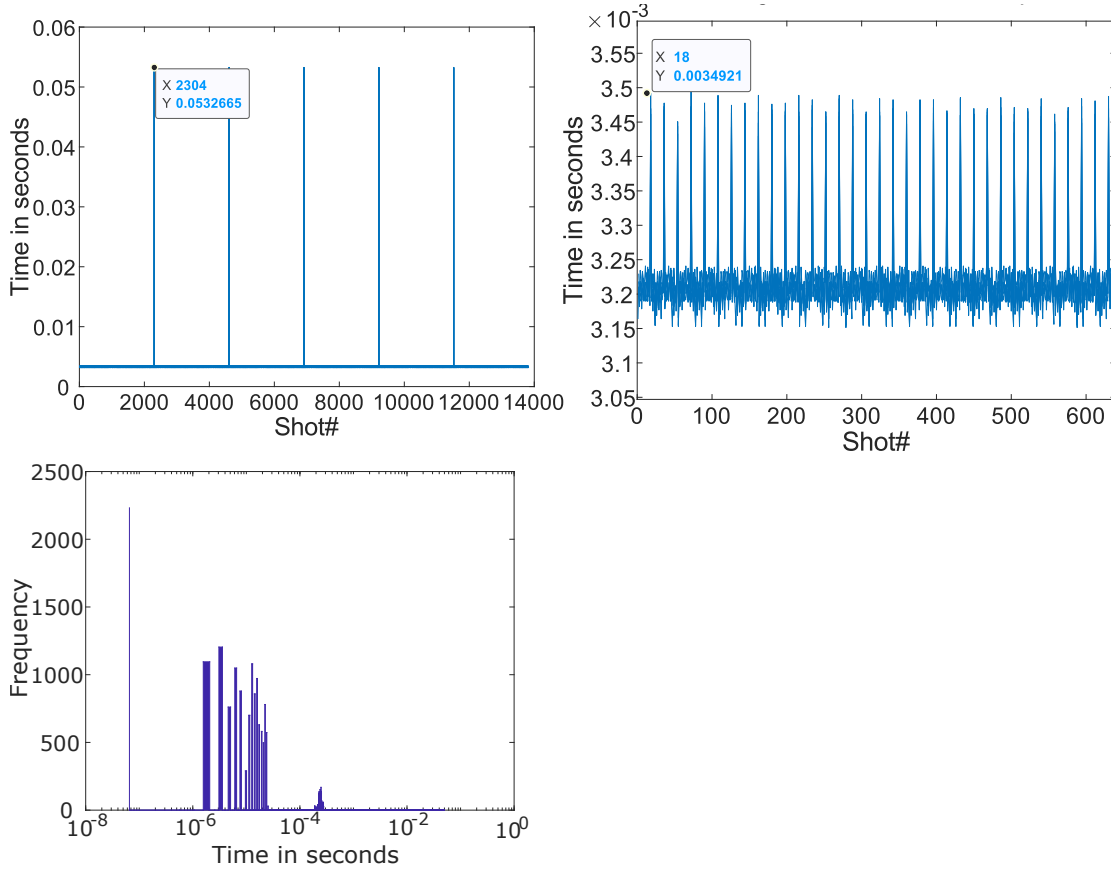


Figure 7.9: 3D USCT III measurement process logging for exp2094, measurement shots and measurement time. (Top left) measurement procedure in 20 MHz sample tracking of the master AMC. (Top right) zoom into the first 620 shots/measurement steps. (Bottom) histogram of the measurement steps, three distinctive timing groups are visible.

7.4 Measurement Procedure Evaluation

The measurement procedure is a complex process involving the parametrisation of dozens of devices. The time critical measurement procedure relies on the step-ack measurement procedure where the AMC master card sends a signal per broadcast to the distribution boards and then the TAS reconfiguring them for the next shot. The ACK per TAS is integrated on the distribution board level and sent back to the AMC master card who waits for completion. There are delay mechanisms integrated to accommodate the required settle times of the various switches and digital μC driven procedures. Also, there is variability due to the digital communication and retry and recover mechanisms. As last line of defence there are time out times defined on every interface to be able to continue also in case of a

complete component failure.

To test the synchronisation with a non-disruptive and concurrent profiling mechanism was established. It is based on an FPGA timer and a concurrent to the DAQ process running process on the internal DAQ PC. This can track without overhead the step-ack procedure to the 20 MHz resolution of the FPGA, and allows debugging and tracking of the procedure in the commissioning and also later in clinical environment. An example is shown in Fig. 7.9 where the procedure went flawless and on the right the three distinctive set timings for switching the transmission channel, switching the reception channel and switching between transmission and reception operation can be seen.

7.5 Dynamic Range Evaluation in the Data

To evaluate the expected signal dynamic range supporting the imaging algorithms, and in particular the SAFT algorithm, which relies on weak reflection signals in contrast to transmission tomography that utilises the strongest signals, the values considered in Sec. 6.6.1 were verified using a QUSTOM patient measurement.

See Fig. 7.10, in the top left graph, which depicts the VGA amplification of each AScan of experiment 2216, 9-bits out of the possible 13-bits due to the autogain mechanism were utilised. The top right graph depicts the signal dynamic range of all AScans without autogain, defined as *max to min* and then converted to an integer-16 format. The FP16 dynamic range utilisation would require as integer equivalent of 18-bits in median, meaning that with FP16 we saved approximately 2-bits extra compared to the implemented, alternative 16-bit integer data format.

The bottom left graph shows the maximum amplitude of the AScans after autogain correction, representing the signal and its analog dynamic. Between the lowest maximum AScan and the highest amplitude AScan there is a difference of 21-bits, illustrating the enormous signal dynamic range in the 3D USCT III setup.

Regarding the effectiveness of the dynamic-range-increasing schemes: from the 13-bits possible through the VGA, only 9-bits are utilised. This indicates that the peak-to-peak detection process on the unfiltered AScans is quite limited by the

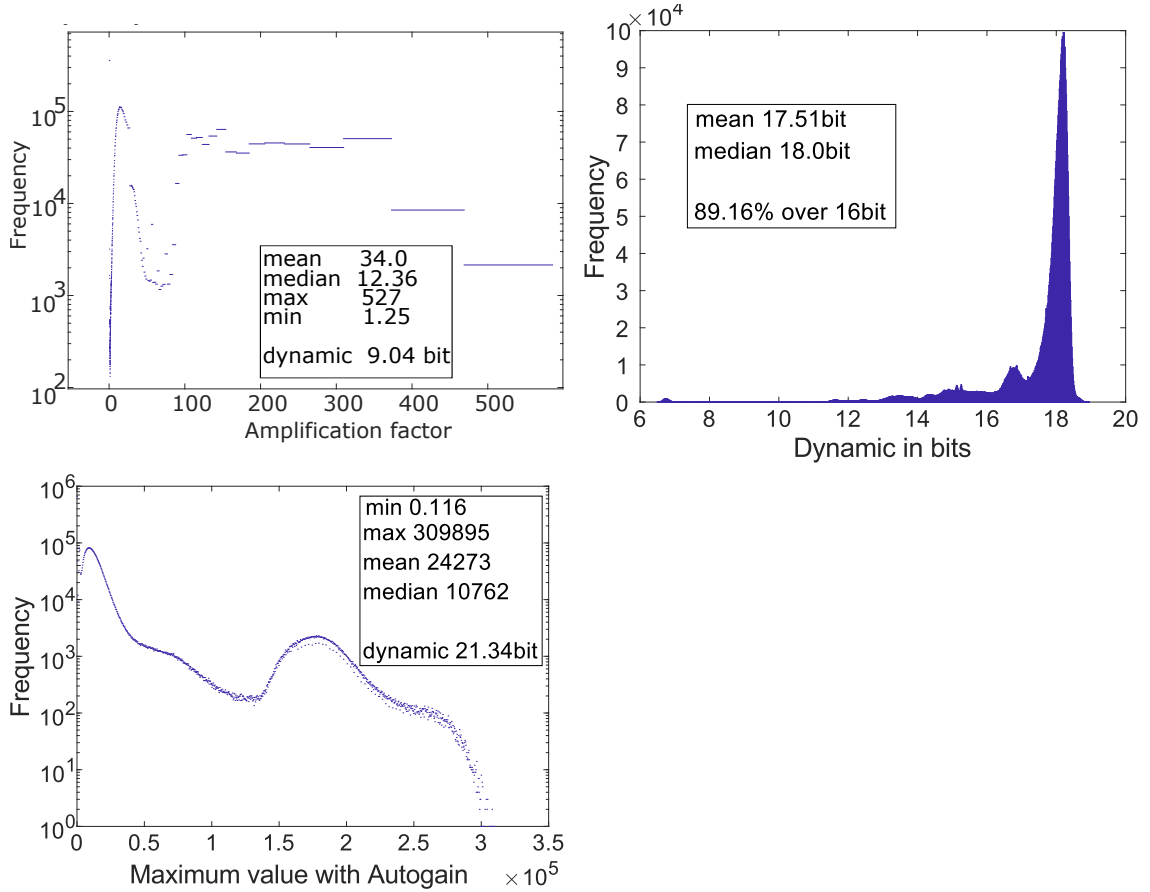


Figure 7.10: Data and signal dynamic range analysis for experiment 2116 from the QUSTOM study. (Top left) Histogram of the variable-gain-amplifier’s amplification factors. (Top right) Signal dynamic range per AScan, shown as a histogram of the required equivalent integer data type. (Bottom left) Signal dynamic range between AScans, represented by the histogram of the maximum values across all AScans.

noise level. In the FP16 data format, the exponent bits are only partially used; this results from the limited averaging factor employed for the QUSTOM study. With higher averaging, the exponent bits would be fully utilised. Alternatively, another FP16 variant could be defined with more mantissa bits and fewer exponent bits.

Overall, the goal of extending the signal dynamic range beyond the 12-bits of the ADC and achieving efficient bit-budget utilisation was successfully realised with the developed DAQ scheme. Compared to the need of 140 dB (≈ 35 -bit), formulated in Sec. 5.2.1 the implemented dynamic range in Sec. 6.6.1 seems to be sufficient for dynamic range arising and observed in the real data.

7.6 Signal-based SNR Gain Capacity of the 3D USCT III

As the 3D USCT III system design follows a SAFT based imaging approach where SNR gain in the hardware is compromised for larger opening angle, the SNR gain capabilities are of greatest importance.

As it was already introduced, ideally, these mechanisms should increase the SNR by $\sqrt{n}$ with n being the averaging number or the relative effective length of an excitation. In reality, the prerequisite of perfectly uncorrelated and well behaving noise might be compromised by correlated system or physical components.

For evaluating the expected SNR improvement capabilities a specific measurement series was conducted, covering the whole range of the SNR improving capabilities of the 3D USCT III in the digital DAQ process. The individual experiment's parameter list is given in Tab. A.2 in the appendix.

The analogue components and parameters were optimised already. Three parameters are of core importance here: Averaging, the coded excitation length and the pause windows jitter feature. The parameter domains analysed are: Averaging 1, 4, 16, 64, 256, 1024. Coded excitations of 10, 20, 40, 80, 160, 320 μs length. Experiments conducted were all permutations and with jitter bit enabled and disabled. The expected SNR gain as factor should range from 1 for the reference case averaging 1 and CE length 10 μs to 181.02 μs for the case average 1024 and a CE length of 320 μs . For this evaluation 72 experiments were conducted, the first 36 having the jitter feature disabled, the second group of 36 experiments having it disabled. Inside these two groups coded excitation length is increase from 10 μs to 320 μs in six steps. The averaging is increased from 1 to 1024 in six steps. The experiments utilised for this were exp1022 to exp1092 before Barcelona. Per experiment 27 AScans from varying transducer combinations were utilised (Emitter TAS 1, emitter element 5, 9, 17, receiver TAS 16, 64, 128, receiver element 5, 8, 9). A visual representation of the last AScan (ETAS 128, element 9) for all experiments is given in Fig. 7.11, top left.

In Fig. 7.11 (bottom left), the effect of the jitter mechanism, described in the

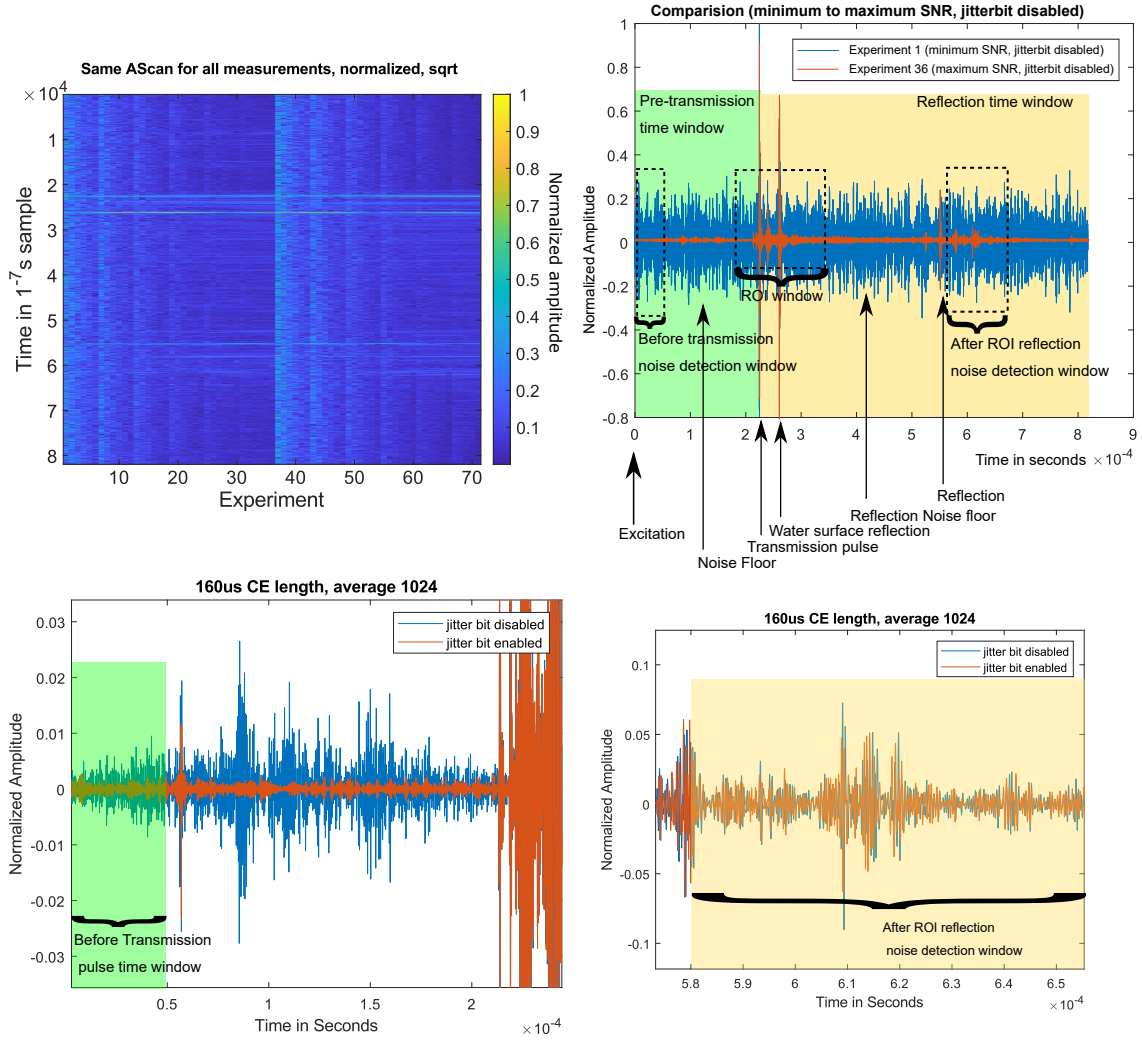


Figure 7.11: SNR gain dynamic of 3D USCT III evaluated. Uses exp1022 to exp1092. Receiver TAS 1, emitter element 5, 9, 17, receiver TAS 16, 64, 128, receiver element 5, 8, 9. Overall 5,832 AScans were utilised in this evaluation. (Top left) For all 72 experiments (x -axis) same normalised AScan, y -axis time. (Top right) comparison minimum to maximum SNR AScan for Jitterbit off (exp 1 and exp35). The averaging raises the SNR effective and the transmissions signals get reconstructed from the noise background. (Bottom left) comparison same AScan with maximum SNR image with normalised amplitude maximum SNR with and without jitterbit in the first noise window zoomed in. Jitter mechanism is effective due to the time probabilistic / shot relative noise origin. (Bottom right) comparison same AScan with maximum SNR image with normalised amplitude maximum SNR with and without jitterbit in the second noise detection window zoomed in. Jitter mechanism is ineffective due to a coherent noise effect which is previous shot uncorrelated (e.g. geometric reflection, crosstalk).

Sec. 7.6 on measurement procedures, is visualised. The noise floor before the transmission pulse (green window until transmission signal) is noticeably lower when the feature is enabled.

The general impact of the SNR improving mechanism is exemplary depicted in Fig. 7.11 top right graph the minimum SNR gain scheme was used for measurement, averaging one and coded excitation length of 10 μ s and depicted as blue curve. The transmission pulse is barely visible against the background noise. The red curve depicts the maximum SNR gaining procedure with average 1024 and a coded excitation length of 320 μ s. The transmission pulse, and later reflections become visible. Also, there is a distinctive difference in the noise floor behaviour with the jitter bit enabled for the first time window compared to the disabled jitter measurement scheme (bottom left graph). For the second window there is no improvement visible, the noise source must have a non-probabilistic nature but time deterministic which can't be broken with the jitter mechanism randomization. Effects like system wall reflections and crosstalk are potential causes.

A SNR gain value $G_{\text{SNR}}(i)$ is created from the AScan data $A_S(n, i)$. The maximum signal amplitude $A_{\text{max}}(i)$ is determined across all samples n as

$$A_{\text{max}}(i) = \max_n(A_S(n, i)), \quad (7.8)$$

while the standard deviation $\sigma_A(i)$ is evaluated within a predefined noise window as

$$\sigma_A(i) = \text{std}(A_S(n, i), n \in \text{window}), \quad (7.9)$$

Their ratio yields the signal-to-noise ratio gain for each AScan i :

$$G_{\text{SNR}}(i) = \frac{A_{\text{max}}(i)}{\sigma_A(i)}, \quad (7.10)$$

In Fig. 7.11 in the top right figure the green and yellow marked time windows, selected for noise floor extraction are utilised for SNR graph creation. Fig. 7.12 depicts the SNR performance for the green time window, before the transmission signal. The both top row graphs for all 27 ASans over all experiments, right column with jitter feature enabled, left disabled. Middle row depicts the median

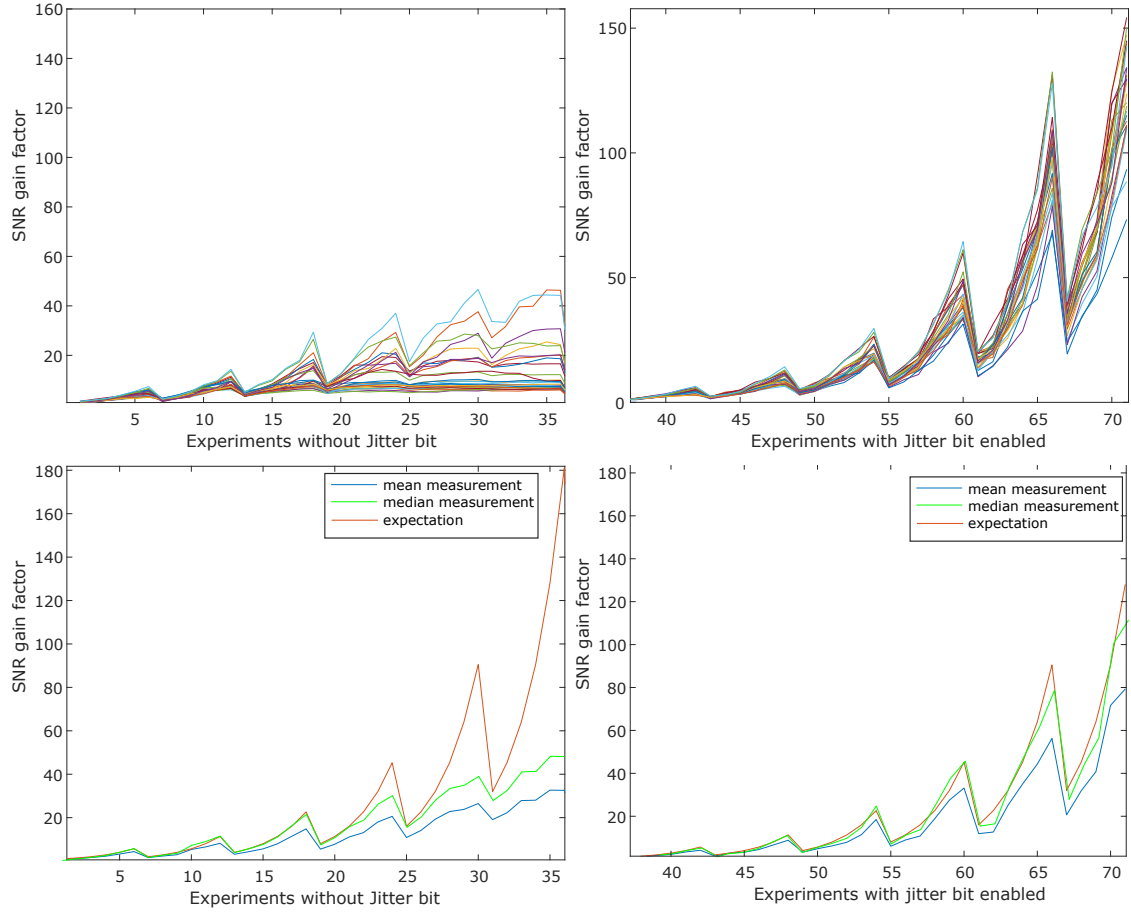


Figure 7.12: SNR gain of the 3D USCT III, system noise window before transmission signal (0 to 50 μ s, see Fig. 7.11). This analysis uses exp1022 to exp1092 with composed from 5,832 AScans. (Right column) improvement by jitter mechanism. (Left column) without jitter bit. (Top row) SNR gain factor for Emitter TAS 1, Emitter Element 5, 9, 17, Receiver TAS 16, 64, 128, Receiver Element 5, 8, 9. (Bottom row) mean and median SNR gain factor.

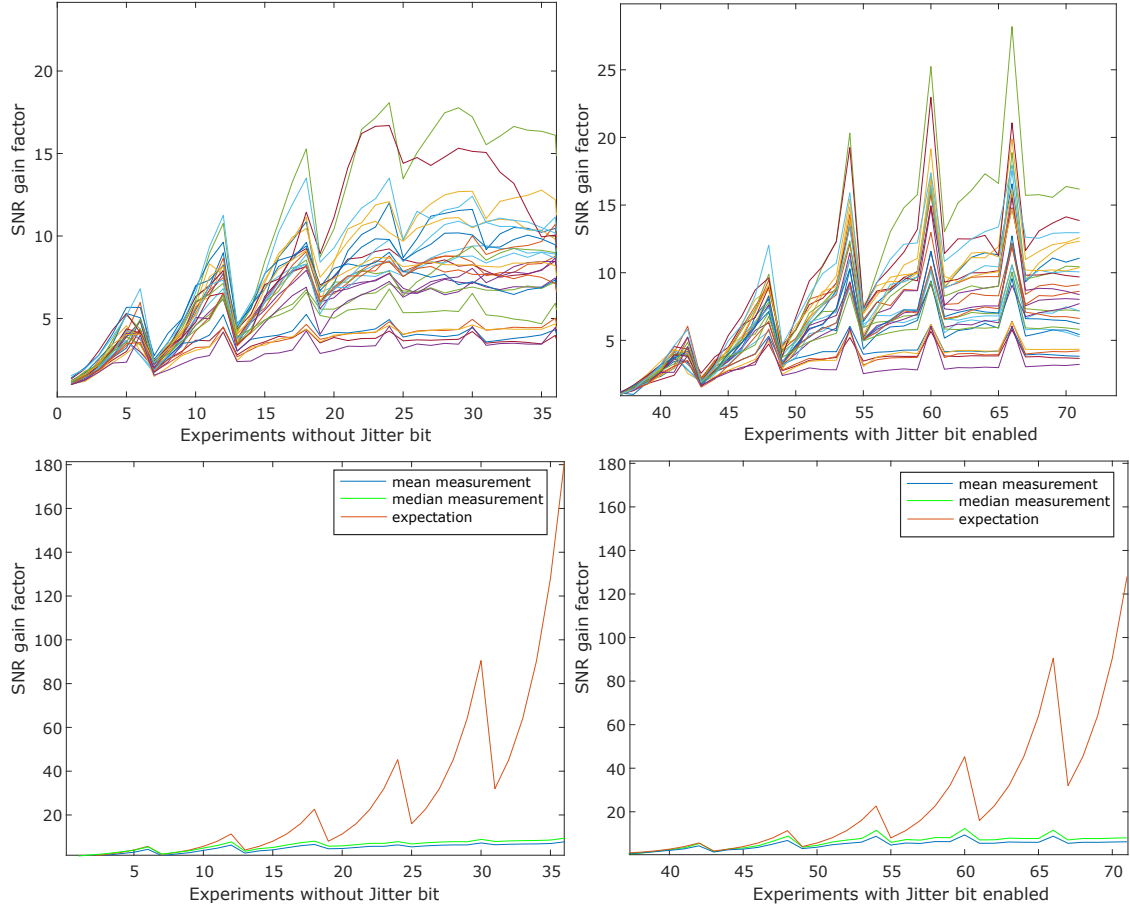


Figure 7.13: SNR gain of the 3D USCT III system, noise window after transmission signal (time window 580 to 680 μ s, see Fig. 7.11). Uses exp1022 to exp1092 with overall 5,832 AScans. (Right column) improvement by jitter mechanism. (Left column) without jitter bit. (Top row) SNR gain factor for Emitter TAS 1, Emitter Element 5, 9, 17, Receiver TAS 16, 64, 128, Receiver Element 5, 8, 9. (Bottom row) mean and median SNR gain factor.

from all AScans, bottom row the mean. It is notable that without the jitter feature the maximum SNR gain stalls around factor 40, only one third of the theoretical maximum. With jitter feature enabled the potential could be almost fully realised, hinting that as expected a correlation formed from the reflections between individual shots in the averaging process exist and could be effectively broken and randomised.

Fig. 7.13 depicts the SNR performance for the blue time window, after the transmission signal. The both row top graphs for all 27 AScans over all experiments, right column with jitter feature enabled, left disabled. Middle row depicts the me-

dian from all AScans, bottom row the mean. It is notable that with and without the jitter feature the maximum SNR gain stalls around factor 20, only one sixth of the theoretical maximum. The overall interpretation from both graphs is, that while the digital DAQ system mechanism are effective and the noise floor introduced by the analogue electronics and transducers is well, there is some correlated noise component.

It is evident that a systematic, correlated noise component persists after the transmission pulse, in the crucial RoI window, which cannot be averaged out by either direct averaging (phase-correlated) or jittering (between-shot correlated).

Two effects are under investigation as the primary sources of this correlation:

1. Electromagnetic interference (EMI) coupling into multiple channels simultaneously.
2. System internal crosstalk.
3. Acoustic effects, diffuse reflections from container walls or transducer surfaces coherently overlapping within the measurement window.

Methods to identify and reduce these correlated noise effects are discussed in the following chapter and future work section Sec. 7.8.1, see Fig. 7.26.

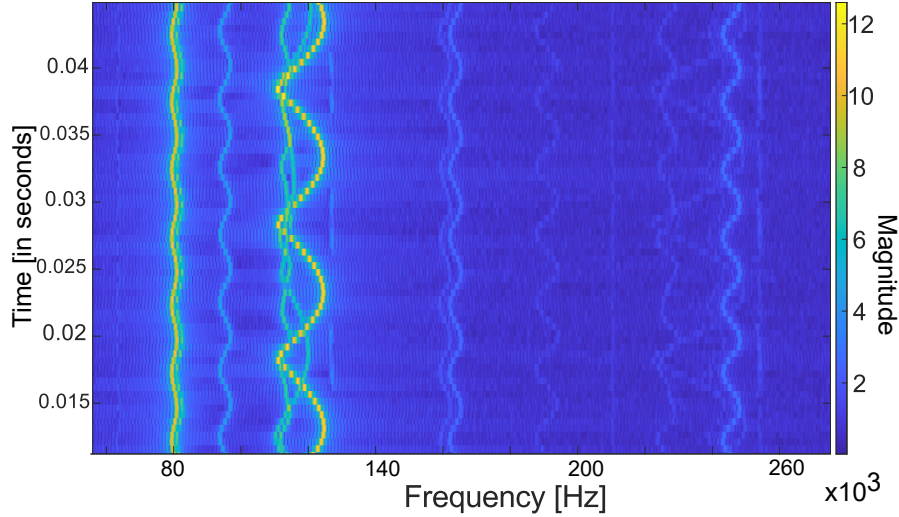


Figure 7.14: Oscilloscope measurement on 5 V supply: short time Fourier spectrum of TDK Lambda QM8 power supply without load, time interval of $0.5 \mu\text{s}$ [Kra24].

7.7 EMI Noise Spectrum Analysis

Frequency analysis of the 3D USCT III data revealed strong low frequency (beginning around 300 kHz) noise like structures preventing sufficient positive SNR in for these frequencies. Some analysis revealed them to be originating in the TDK Lambda switching power supply and being distributed via the 5 V supply lines which have high current flows[Kra24].

Analysis of this EMI is done as noise visualisation in the fourier-domain taken from empty air noise measurements from the 3D USCT III to clearly differentiate between US wave effects from EMI, electrical chain and DAQ effects. This can be differentiate as phase coherent and non-coherent noise content. The formulation of the non-coherent and coherent noise spectra across M channels

$$S_{\text{noncoh}}(f) = \frac{1}{M} \sum_{i=1}^M |X_i(f)|, \quad (\text{non-coherent noise spectrum}), \quad (7.11)$$

$$S_{\text{coh}}(f) = \left| \frac{1}{M} \sum_{i=1}^M X_i(f) \right|, \quad (\text{coherent noise spectrum}), \quad (7.12)$$

with $X_i(f) = \mathcal{F}\{x_i(t)\}$ denotes the Fourier transform of the i -th channel. The non-coherent spectrum averages the magnitude of each FFT, whereas the coherent spectrum takes the magnitude of the averaged complex FFT, emphasising

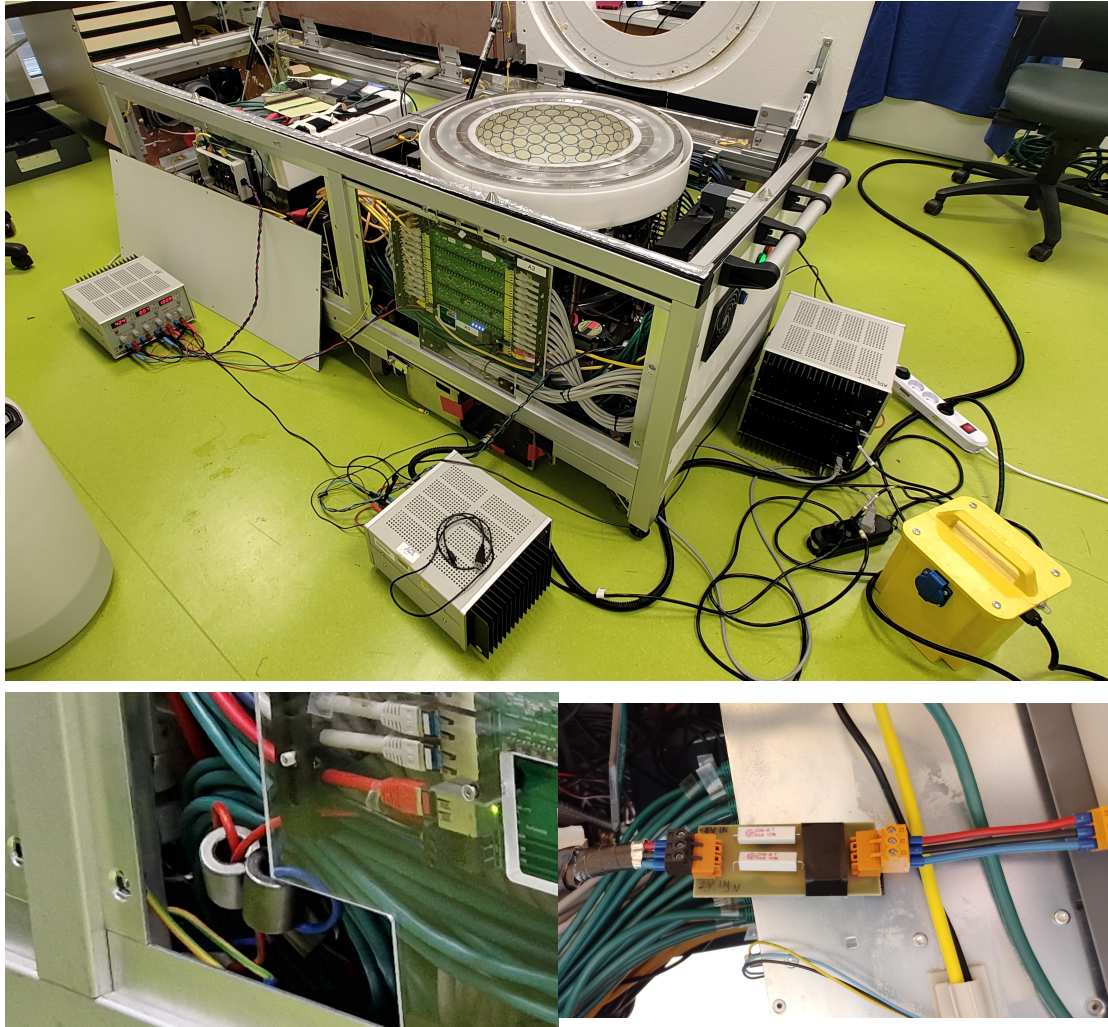


Figure 7.15: *Experiments with alternative supplies. Top picture: supply setup of experiment 2320. Bottom left picture: Chokes for supply line EMI suppression in the critical pm5 V supply line for a distribution board, test before Barcelona study. Bottom right: Filter for the pm5 V power supply for the ISO board, test before Barcelona study.*

phase-aligned components. Results are given in Fig. 7.16. In both figures a lowest distortion frequency of around 310 kHz is noticable and many higher harmonics until 1.5 MHz for the TDK Lambda measurement (blue curve) are visible. A measurement with alternative powering setup with linear laboratory power supplies (red curve) shows encouraging results with significantly reduced low frequency distortions, additionally the general noise floor is much lower in the phase-coherent metric, see Fig. 7.16 right. Sporadic non-coherent distortion frequencies from other sources in the system (motor drivers, USB etc) are in the non-coherent metrics graph, like the spikes around 1.1 MHz and 3.1 MHz in Fig. 7.16 right. These are non-noticable in the non-coherent metrics graph due to being averaged out, see Fig. 7.16 left.

The power supply is a AC-DC switching based and utilising spread-spectrum blurring techniques spreading the discrete switching frequency distortion over a broader bandwidth. Unfortunately, this mechanism can not be turned off and the switching frequency can not be easily filtered with a sharp band-pass filter approach. Analog approaches were tried to, like EMI suppression chokes on the supply lines, see Fig. 7.15, proved to be not effective enough.

The relevancy of the phase-coherent noise compared to the non-coherent is that for the 3D USCT III's almost perfectly spherical aperture this coherent spikes sum up also coherently in the SAFT imaging as ring artifacts, see Fig. 7.26 in Sec. 7.9.3, with reduced suppression by adding more AScans. While the 3D USCT III was designed and built following compressive sampling approaches with randomisation to suppress non-correlated noise sources, one symmetry was kept and not randomised, the spherical aperture with its constant distance of all TAS to the RoI. Adding here at the transducers or aperture also a randomisation element would help suppress also such phase-coherent noise in an improved 3D USCT system.

While for the current system, the EMI actual, pathway of catching the supply noise in the AScans were not fully understand, it is known that the ASICs PSSR (Power supply suppression ratio) is low[Bla19b, Lu24]. Therefore the electronics components were designed from the beginning measures to accomodate for that: filters, shielding and separating of the ground lines. Also later extra components like EMI filtering chokes were tested, see Fig. 7.15. The low frequency distor-

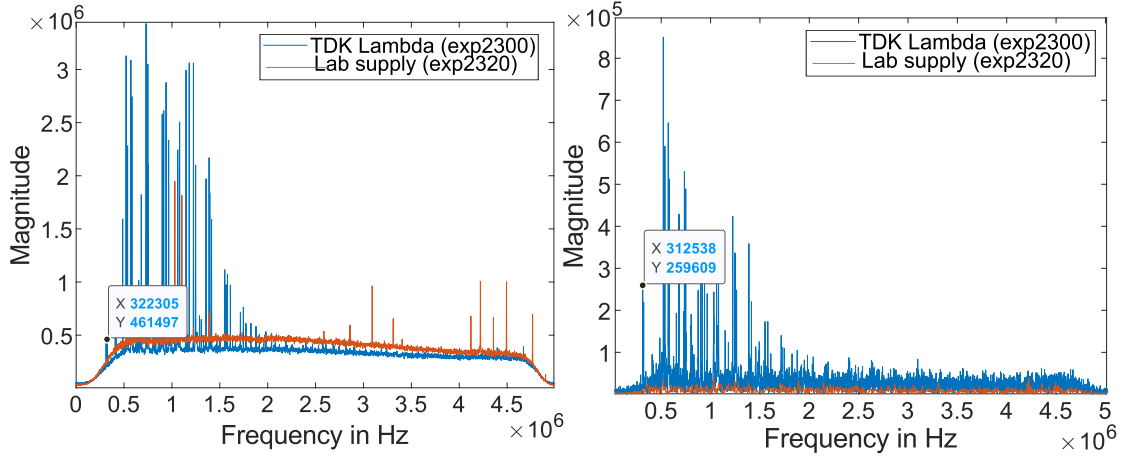


Figure 7.16: *Experiments with alternative power supply for cleaner supply voltage. (Left) Mean Fourier domain response of 3D USCT III over all channels for MUX1. Phase in-sensitive (non-coherent) noise from Eq. 7.12. Blue graph, powered with the default medical TDK Lambda QM8 switching power supply, experiment 2300. Red graph from exp2320 3D USCT III powered with lab power supplies replacements. (Right) Mean Fourier domain response of 3D USCT III over all channels for MUX1. Phase coherent noise from Eq. 7.11. Blue graph, powered with the default medical TDK Lambda QM8 switching power supply, experiment 2300. Red graph from exp2320 3D USCT III powered with lab power supplies as replacements.*

tions could not fully removed from the signal for the Barcelona study but were considered toleratable by the project partners applying the FWI method. Also, this behaviour can be positively interpreted as another example of the system's excellent capability of taking up the even smallest of signals through its SNR enhancing mechanisms.

To verify inter-channel isolation and detect possible crosstalk, the normalised cross-correlation coefficient ρ_{ij} between all digitised AScans $s_i(t)$ and $s_j(t)$ was computed:

$$\rho_{ij} = \frac{\sum_t (s_i(t) - \mu_i)(s_j(t) - \mu_j)}{\sqrt{\sum_t (s_i(t) - \mu_i)^2 \sum_t (s_j(t) - \mu_j)^2}}, \quad (7.13)$$

where $\mu_i = \frac{1}{T} \sum_t s_i(t)$ and $\mu_j = \frac{1}{T} \sum_t s_j(t)$ denote the mean signal amplitudes of each channel i and j , respectively. The correlation coefficient satisfies $\rho_{ij} \in [-1, 1]$.

Values close to zero indicate statistically independent channels (no coupling),

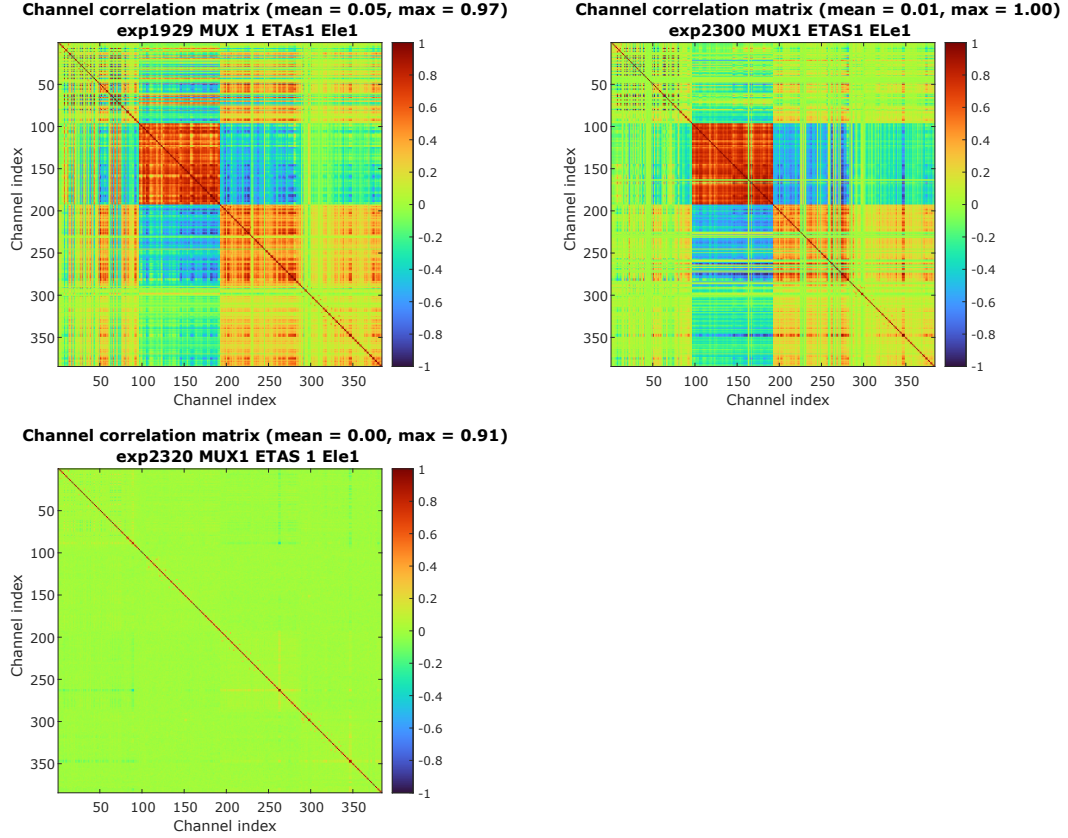


Figure 7.17: *Between channel correlation analysis. (Top left) Air empty measurement before Barcelona with TDK Lambda, (Top right) Air empty measurement after Barcelona with TDK Lambda, (Bottom left) water empty measurement after Barcelona with lab power supply, the strong correlation associated with distribution board two are not visible any more.*

whereas higher absolute correlations ($|\rho_{ij}| > 0.2$) suggest electrical or acoustic cross-interference between channels. The computation was performed for all channel pairs (i, j) .

For an overall quantitative measure of the inter-channel independence, the mean absolute off-diagonal correlation was computed as:

$$\bar{\rho}_{\text{off}} = \frac{2}{N(N-1)} \sum_{i < j} |\rho_{ij}|, \quad (7.14)$$

where N is the number of channels. A small value (e.g., $\bar{\rho}_{\text{off}} < 0.05$) indicates a well-isolated, low-crosstalk acquisition chain. Future outlook is discussed in Sec. 8.3.1.

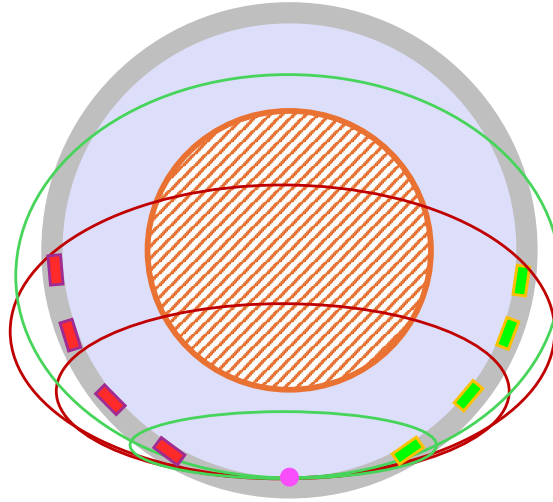


Figure 7.18: *Measurement container side wall reflection schematic, top down view on the 3D USCT device. Blue disc is measurement container with water, orange striped disc the RoI. Red rectangles represent US emitters, green ones US receivers. Ellipses represent the SAFT projections. From the pink side wall reflection point the ellipses are projected into the image domain in the measurement container. Red ellipses, formed by specific emitter and receiver combinations, impede the crucial RoI, green ones doesn't impact the RoI.*

7.8 Side-wall reflection evaluation

While the opening angle was increased in 3D USCT III an unintended side effect was an increase reflection amount on the side walls reflecting back and overlapping for specific angles combinations of receiver and emitter transducers in the reconstructed RoI, see Fig. 7.18. Current analysis of this effect hints to the measurement containers walls and transducer components, which create many diffuse reflections overlaying coherent in the signal measurement window.

Another experiment was done to explore the impact of the side-walls reflections impact on the imaging with experiment 2710 an water empty measurement with experimental absorption mats and TAS caps, see Fig. 7.20 left.

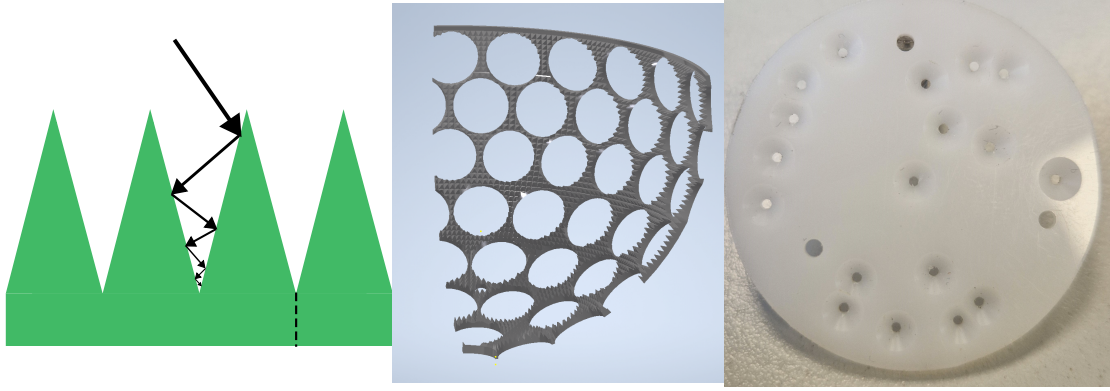


Figure 7.19: (Left) wave diffusion and absorption concept. (Center) 3D printed absorber pyramidal structure for lining the measurement container insed. (Right) US transducer wave shielding mask.

7.8.1 Side Wall Reflection Suppression: Attenuating and Beam Limitation

While the opening angle was increased for 3D USCT III as intended, an unintended and unforeseen side effect was an increased reflection amount on the side walls, reflecting back and overlapping for specific angle combinations of receiver and emitter transducers in the reconstructed RoI, see Fig. 7.18, limiting signal SNR and impacting reflectivity imaging contrast, see Sec. 7.6.

This noise floor can also be seen in the limitation of the SNR gain after the transmission signal, see Fig. 7.11 or in the SNR analysis on the spherical projection method for the upper part see top Fig. A.4.

An approach under investigation is the implementation of 3D filament TPU-printed pyramidal absorption structures, combined with a transducer shielding mask restricting the waves from angular response for emission and transmission, see Fig. 7.19 [Kra25]. Overall, in first practical experiments, see Fig. 7.27 right, with a breast-mimicking phantom, attenuation values similar to the simulation predictions (-20 dB) could be observed, see Fig. 7.21.



Figure 7.20: (Left), Experiment 2710, water measurement with TAS caps and absorption mats. (Right), another absorption experimental setup with a CIRS phantom.

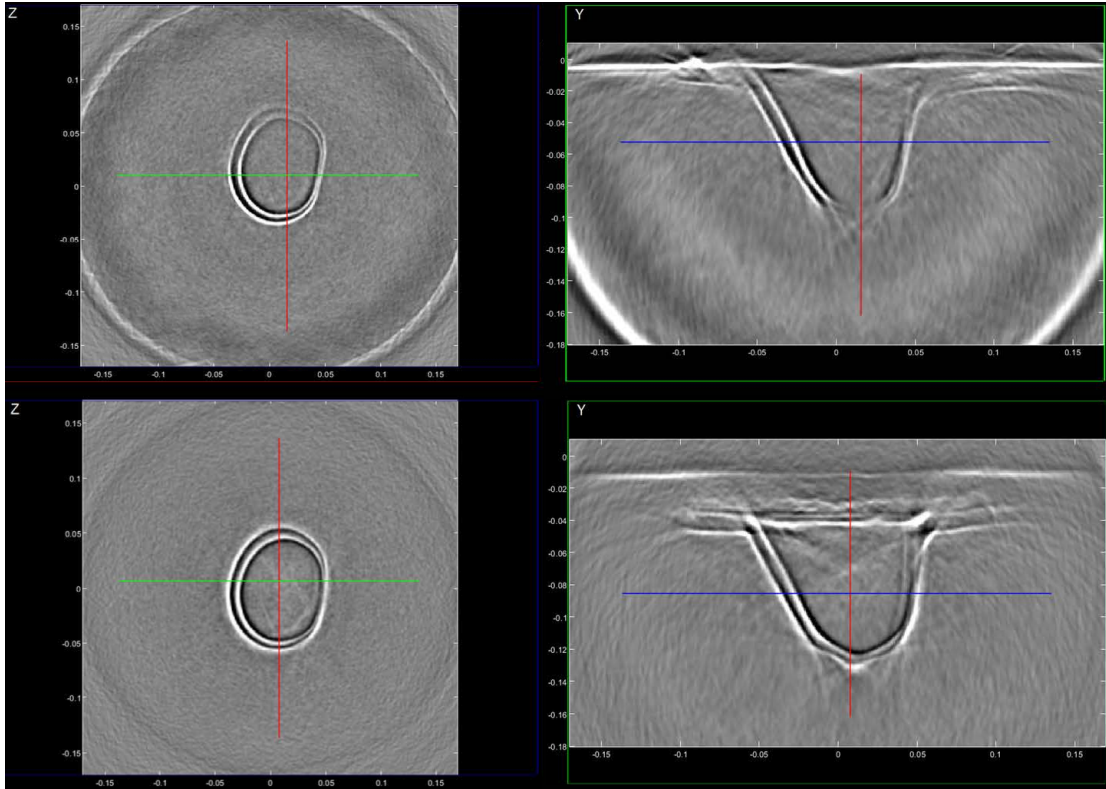


Figure 7.21: (Top) CIRS reflectivity reconstruction images in the z and y plane with strong side wall reflections and side wall visibility. (Bottom) CIRS reflectivity reconstruction with absorber mats and transducer shields applied.

7.9 Image-based SNR Gain Potential of the 3D USCT III

As shown in Sec. 7.6, the 3D USCT III system demonstrates effective and robust SNR gain at the signal level, even in the presence of noise and certain classes of distortions. In particular, the implemented jitter mechanism enables an effective temporal randomization of excitation-correlated reflections (“backwall reflections”), which can be reduced through averaging (see Fig. 7.12).

However, additional distortion mechanisms and correlated noise sources remain, such as inter-channel correlations and other system-related effects. These contributions limit the achievable SNR gain at the signal level, as illustrated in Fig. 7.13.

While these effects appear strongly correlated in the one-dimensional signal domain—evidenced by their weak attenuation and saturation behavior with increasing averaging—they may exhibit reduced or negligible correlation in other domains. In particular, the three-dimensional spatial reconstruction domain provides additional degrees of freedom. The system aperture was specifically designed to exploit spatial diversity and thereby reduce such correlations.

A central claim of 3D USCT systems is their capability to perform synthetic aperture focusing (SAFT) in software rather than relying on hardware beamforming, as is common in conventional sonography systems. The evaluation of the imaging performance of the 3D USCT III system—comprising the aperture, analog and digital signal chain, and DAQ system—therefore focuses on the SNR gain achievable through the SAFT process (see Sec. 5.2.5).

The pseudo-randomized aperture is expected to act as a robust mechanism for decorrelating device-induced noise contributions.

While a comprehensive evaluation using phantoms, patient data, and proband measurements is ongoing—particularly due to calibration-related effects—an alternative approach is employed here to assess the system’s SNR gain potential without relying on ground truth or complex test objects.

The ability to suppress system noise with an increasing number of AScan contributions can be demonstrated under the assumption that the system behaves

approximately linearly and that residual noise contributions are sufficiently uncorrelated. In this regime, distortion effects are assumed to be minimal.

Under these assumptions—namely uncorrelated, well-behaved Gaussian noise and negligible distortion introduced by the analog and digital DAQ chain—the SNR gain of coherent imaging is expected to follow a $\sqrt{n}$ relationship.

7.9.1 Coherent SAFT Test Setup

In addition to the standard USCT spatial-domain SAFT imaging algorithm, conventional linear signal processing steps are applied, including per-channel gain normalization, matched filtering, and DAC delay compensation.

An additional normalization step is introduced: without it, AScans with elevated noise floors would dominate the coherent summation, biasing the SAFT result and corrupting SNR measurements. Variations in noise power across A-scans can arise from differences in acoustic coupling, hardware variations, or defective channels (see previous chapter, where more than 10% of transducers exhibit malfunctioning behavior).

Normalization to the individual noise floor ensures that each AScan contributes equally to subsequent processing steps, such as coherent summation in SAFT, cross-correlation, and SNR estimation.

$$\text{AScans}_j^{\text{norm}} = \frac{\text{median}_k(\text{std}(\text{AScans}_k))}{\text{std}(\text{AScans}_j)} \text{AScans}_j, \quad j = 1, \dots, N. \quad (7.15)$$

To see evaluate the performance of the real 3D USCT III system, its aperture and DAQ system minimal extra pre-processing is applied, only linear operations: no cutting out of signal components (transmissions pulses etc), no AScan rejection by SNR filters or bistatic angle is applied. The as image an 2D z-slice in depth 5 cm, and covering the complete 3D USCT III diameter of 35 cm. Resolution is 0.35 mm, the AScans are low-pass filtered accordingly. For the SNR analysis a σ is taken for the whole z-slice and one for a $0.03 \times 0.03\text{m}^2$ RoI piece in the center.

First, a ground truth with the imaging setup as established with ideal gaussian

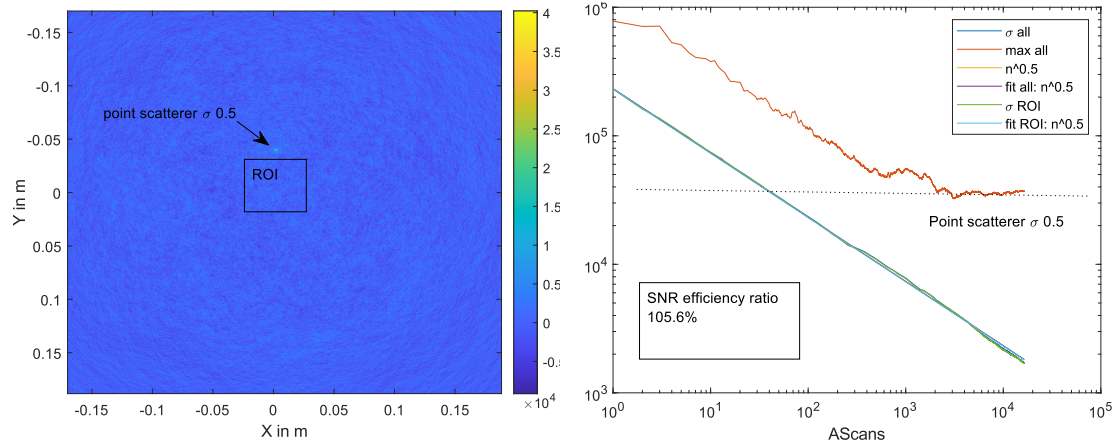


Figure 7.22: *Evaluation of imaging SNR gain capability using coherent SAFT reflectivity imaging. Simulated ASans with an ideal uncorrelated noise pattern. (Left) reflectivity image. (Right) corresponding metrics – red: maximum reflectivity for each ASan contribution; blue: standard deviation per ASan; yellow: expected standard deviation under the $\sqrt{n}$ assumption; violet: fitted function. Green: standard deviation in RoI; cyan: fitted standard deviation for RoI.*

uncorrelated noise. Artificial ASans were created and a point scatterer reflector positioned at $[x = -0.04\text{m}, y = 0.002\text{m}, z = -0.05\text{m}]$ with an amplitude of 0.5σ of the noise ASan, as marker. A subset of 16,384 ASans was utilised covering all transducer and emitter TAS. The metrics are created for every addition of an individual ASan, the resulting imaging is normalised for the metrics by the ASan number. In this ideal graph in Fig. 7.22, it can be seen that the coherent SAFT reflectivity imaging with the 3D USCT III aperture manages to follow the ideal SNR gain curve, even slightly outperform it. A fit found instead of $n^{0.5}$ an exponent of $n^{0.5035}$. In a SNR gain effectiveness metric, defined as $\frac{\sigma_{achieved}}{\sigma_{expected}}$ for the 16,384 ASans contributing, of 105.6 %. Beside this quirk, the performance is as expected, the point scatterer becomes visible and defines the maximum starting with ASan number thousand, the SNR gain curve for both RoI and complete image fit the expectation perfect.

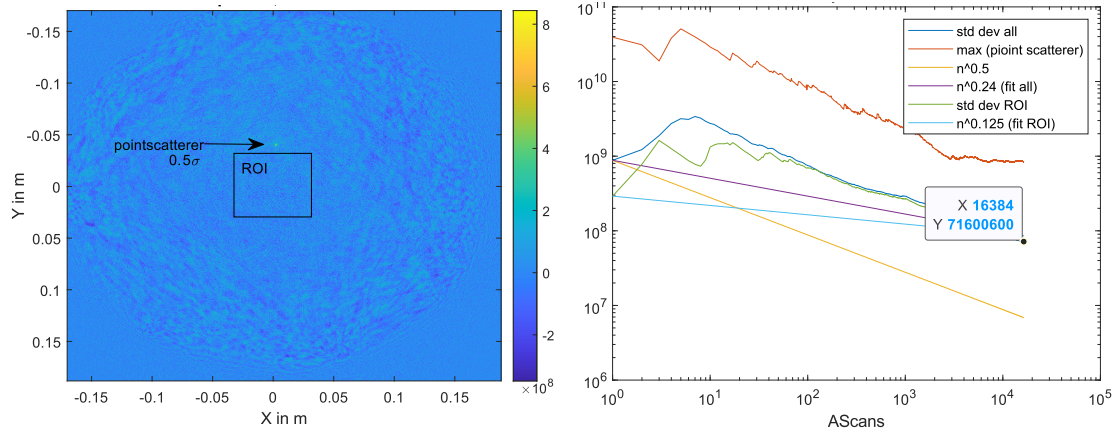


Figure 7.23: 3D USCT III imaging SNR gain potential, ideal performance compared to the measured SNR gain capacity. Water empty experiment 2580 with average 1024. Sixteen thousand ASans added. One perfect point scatterer with 0.5σ was added at $X = -0.04$ m, $Y = 0.002$ m, $Z = -0.05$ m.

7.9.2 Complete 3D USCT III system SAFT SNR gain capacity evaluation

Next step is utilising of real data including US data. As we have seen in previous ASan based SNR analysis, while the system is capable of SNR gain increases on signal level, there are correlated components after the transmission pulse limiting the potential SNR gain. Yet, in the image domain, with 3D USCT III highly irregular aperture, this components might and should become uncorrelated.

To test this capacity of the 3D USCT III without additionally hard to gather ground-truth and perfect calibration of the system it was chosen to take a pure water measurement (exp2580) with maximum signal SNR already, where the signal based SNR gain capacity is already exhausted, e.g. saturated.

While there is a significant SNR gain demonstrable against the artificially place point scatterer. Calculating the SNR gain ratio as $\frac{\sigma_n}{\sigma_{ideal}}$ we achieve an SNR gain of 184.17 compared to the expected factor 2303.2, a SNR gain efficiency of 8.54% only. Similarly, when utilising the fitted values $\frac{5.3e6^{0.325}}{5.3e6^{0.5}}$ a relative SNR gain of 153.246 and efficiency ratio 6.6% can be deducted.

The reason for this underperforming of the 3D USCT III must be correlated components in the noise signal as the performance of the complete chain with artificial

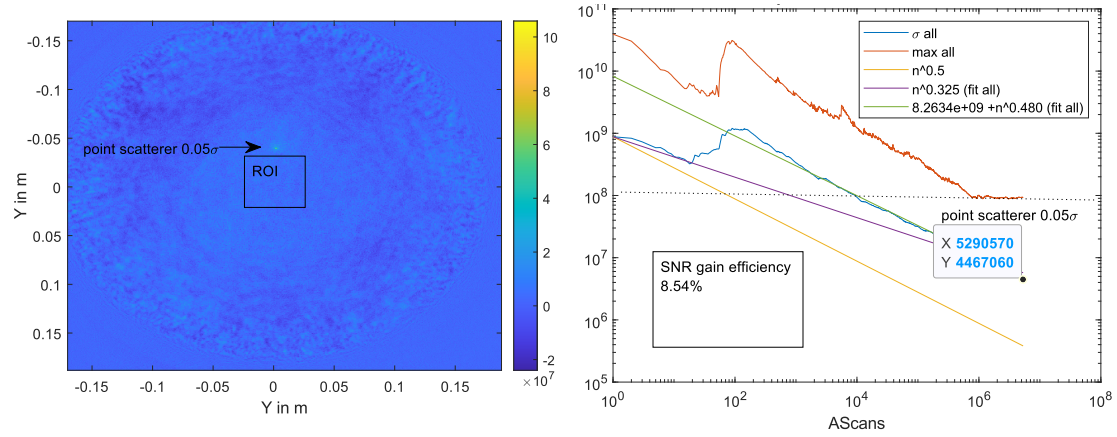


Figure 7.24: 3D USCT III data SNR gain potential, with experiment 2580 water empty measurement with average 1024. With 5 millionen ASans added. One perfect point scatterer with 0.05σ was added at $X = -0.04\text{ m}$, $Y = 0.002\text{ m}$, $Z = -0.05\text{ m}$.

ideal noise was ideal and as expected. Sources of such correlated effects currently considered identified can be crosstalk between the channels or correlated EMI noise sources, or other system inherent effects can be ruled out with calibration.

7.9.3 Electrical correlated noise impact on SAFT

SAFT analysis of the aperture for different power supply noise setups to evaluate the impact of the correlated noise as continuation of the EMI noise impact analysis from Sec. 7.7.

To separate and explore the impact of the different correlated noise sources experiments were conducted. An air empty measurement was set up as experiment, replacing the water medium with air, allowing to separate physical wave effects (reflections etc) from electrical and other system effects. As air medium can not transfer MHz US waves this therefore isolate purely system-level noise and cross channel correlations as source. In Fig. 7.25 a channel cross correlation metric was created from the measured data. Indeed some strong correlated noise was found, see graph right (air measurement), compared to the left (water measurement). Additionally there are strong correlation 2D blocks visible in the resulting graph. These blocks can be associate to hardware features, they correlate here exactly to the distribution boards. Analysis is ongoing here and improvement proposals are in the future work section.

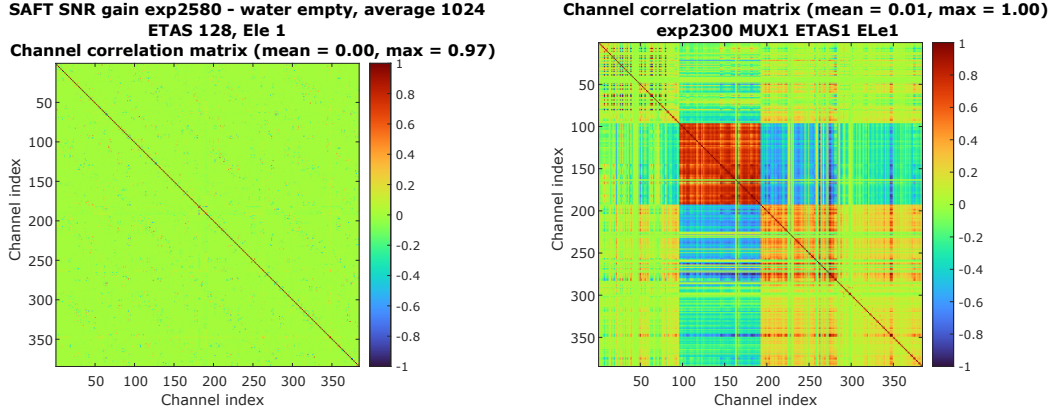


Figure 7.25: *Between channel correlation analysis, near zero denotes uncorrelated channels, 1 and -1 correlated. (Left) exp2580 water measurement for reference, much less correlated. (Right) exp2300 pure air measurement exposing correlated noise components.*

To explore if the 3D USCT III pseudo-random aperture and coherent SAFT imaging is impeded by this real system effect, coherent SAFT imaging was conducted. 16,384 AScans were utilised from all TAS, evaluated with z-slice SAFT image on 5 cm depth and covering in x and y the complete aperture and RoI with 0.35 mm resolution, see Fig. 7.27 on the response with system measurement air data.

For the air empty measurement an SNR gain curve of 3D the curve fit was $n^{0.461}$ and the SNR gain effectiveness was 71.19% for 16,384 AScans contributing. This is significant better than fit for the 16k AScans water empty measurement experiment which had a fit of $n^{0.125}$ see Fig. 7.23.

While this noise is clearly channel correlated it seems to be uncorrelated quit well in the imaging domain and seems to be not the core source of correlated noise impeding the complete systems SNR gain performance.

7.9.4 Wall-reflection impact analysis on SAFT imaging

Coherent SAFT imaging analysis is applied like with previous experiments and the resulting image can be seen at Fig. 7.28. Compared with the resulting Fig. 7.24 from the water experiment Sec. 7.9.2 without extra attenuation mats and TAS caps, we see an significant changed background noise pattern.

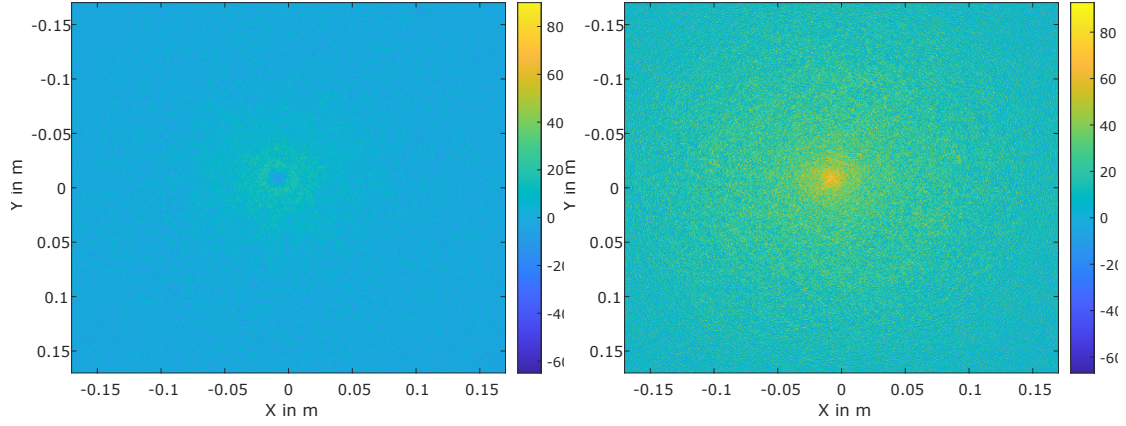


Figure 7.26: SAFT reflectivity imaging with channel correlated extracted coherent noise ASans from two empty air experiments. Single MUX, a parallel measurement phase only used. (Left) from experiment 2320 lab power supplies and reduced coherent noise. (Right) from experiment 2300 with standard TDK Lambda.

The noise floor in circular shape following the aperture surface is gone and is replaced by a central noise floor. For 5.3 million ASans the best SNR fit (without constant component) was $n^{0.325}$ and improved to $n^{0.38}$ for the later experiment with side-wall suppression mechanisms included. This indicates that the correlated noise component could have been successfully reduced with the measure, while there is still optimisation potential.

7.10 System Evaluation: Coherent Linear Projection-based System Detection

As discussed earlier in the CEmeasure Sec. 7.2, one feature intended to be implemented in the ASIC was unfortunately not available to reliably identify specific transmission and reception chains throughout the system. Therefore, a software-based replacement algorithm was implemented, with capabilities beyond the original channel-association evaluation use case.

While there are similarities and overlap with earlier geometry detection approaches [Zap03, Hop03, ZRF⁺07, Tan17], the core goal and motivation of this algorithm are different. In contrast to previous methods, a method is required which allows the exploration not only of global maxima but the exploration of

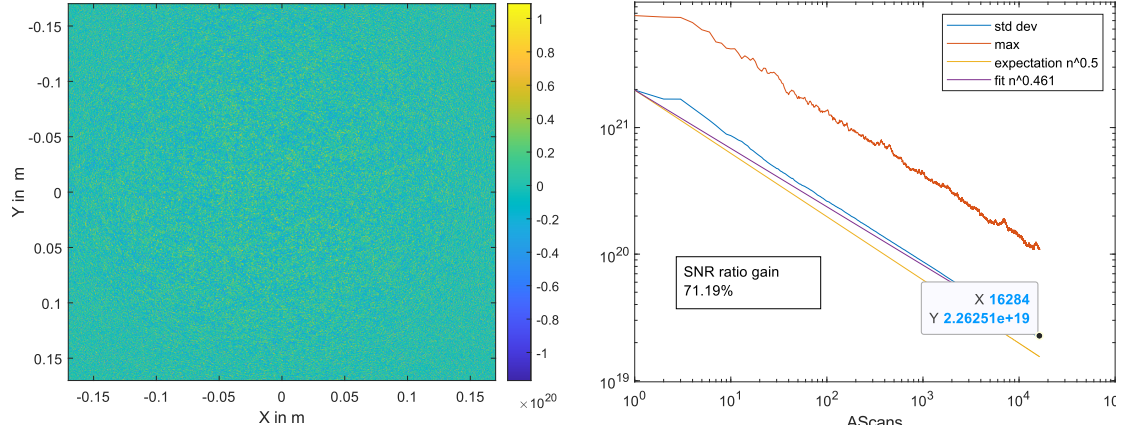


Figure 7.27: Evaluation of imaging SNR gain capability using coherent SAFT reflectivity imaging. Air measurement (exp2300) using a medical TDK-Lambda QM8 switching power supply. (Left) reflectivity image. (Right) corresponding metrics, red curve: maximum reflectivity for each AScan contribution; blue: standard deviation per AScan; yellow: expected standard deviation under the $\sqrt{n}$ assumption; violet: fitted function.

side and local maxima, see Fig. 7.29. Such local maxima might be originating in electrical or mechanical crosstalk or bonding errors on the ASIC assembly and are therefore crucial to identify. Also, as there are many challenges on signal based level detections utilised also by previous approaches, an approach was chosen bringing the detection to another domain - the spatial imaging domain.

Motivation: 3D USCT III is a complex, parallel, multi-channel hardware

The general motivation is the complexity of the 3D USCT III hardware, see Fig. 7.30. The excitation chain alone contains at least three hardware levels with multiple boards involved (FEB, distribution board, isolation board, FMC). The excitation chain includes a 1-to-2304 multiplexing structure implemented across three hardware layers with additional power-control capabilities. The reception chain consists, on the board level, of four layers of components (FEB, distribution board, RTM, AMC). Digitally controlled multiplexing of 2304 physical transducers into 384 digitisation channels is performed across two hardware levels.

Hundreds of discrete LAN cables, microcoax lines, and their corresponding connectors transport the analogue signals, synchronisation signals, and digital com-

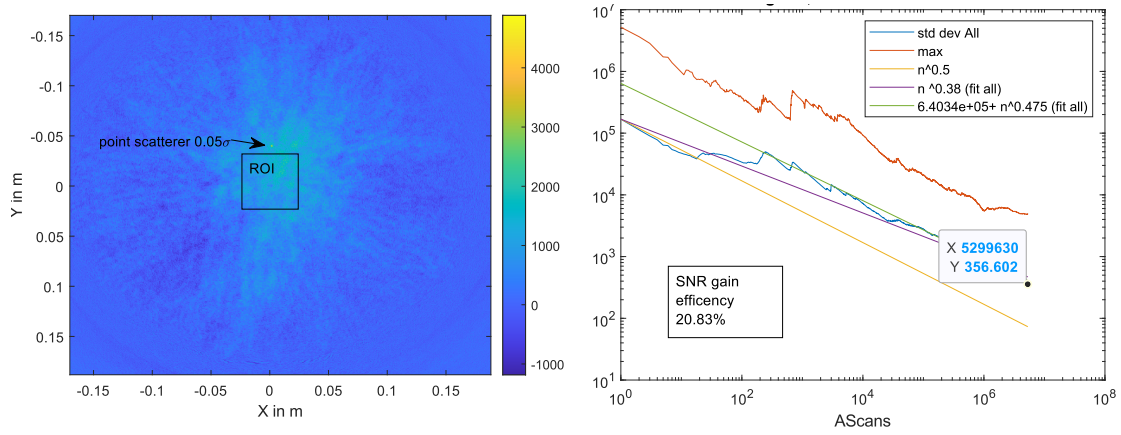


Figure 7.28: Evaluation of imaging SNR gain capability using coherent SAFT reflectivity imaging. Water empty measurement (exp2710) using a medical TDK-Lambda QM8 switching power supply. Average 16. (Left) reflectivity image. (Right) corresponding metrics. red curve: maximum reflectivity for each AScan contribution; blue: standard deviation per AScan; yellow: expected standard deviation under the $\sqrt{n}$ assumption; violet: fitted function.

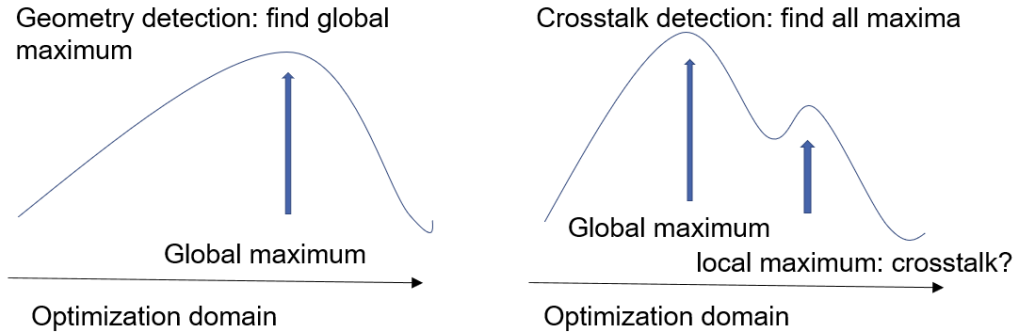


Figure 7.29: Spherical SAFT projection motivation: global detection vs local maxima detection. For illustration purposes only 1D, in the real algorithm 3D in the spatial domain.

munication required for parametrisation.

Potential error cases are manifold: mis-cabling, misconfiguration of multiplexers, and crosstalk between signal lines [KRKZ25b].

Algorithmic approach

While there are already on software level many hardware specific checks were implemented, a fully comprehensive general test was not directly available from the existing DAQ software and protocols. One way forward would have been the de-

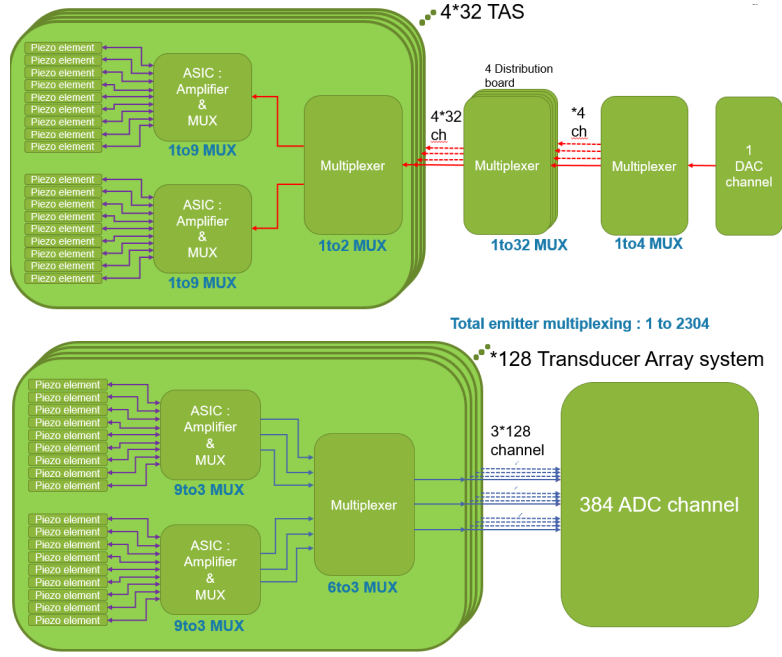


Figure 7.30: Schematic of the signal-chain architecture of the 3D USCT III, many potential sources for crosstalk. (Top) Excitation chain and its components. (Bottom) Reception chain and its components.

development of hardware-specific test routines and maybe hardware extensions. It was decided for later improved clinical practicability for aiming for a full algorithmic software solution. This solution should work without the need for doing extra measurement and procedures in hardware and DAQ software, reducing complexity, required time and cost.

A more general, robust approach was implemented utilising the already existing empty water measurement with a standard measurement protocol. To evaluate the complex system, and its independent receiving and emitter chain, a system wide approach was implemented as general detection.

As there are many challenges on signal based level detections, an approach was chosen bringing the detection to an another domain - the spatial domain.

Inspired by algorithms of the SAFT family, the sum-and-delay reflectivity imaging, a spherical imaging algorithm was implemented[ZRKD25][KRKZ25a].

In the following, $R(\vec{p})_{\vec{e}_j}$ denotes a reconstruction in the image domain for a fixed emitting index j , obtained by spherically backprojecting the corresponding AS-

cans A_{jk} over all receiver positions $\vec{r}_k$ - we call this emitter detection, see Eq. 7.16. Conversely, $R(\vec{p})_{\vec{r}_k}$ denotes a reconstruction in the image domain for a fixed receiver index k , using all emitting positions $\vec{e}_j$ - this we call receiver detection, see Eq. 7.17.

The signal $A_{jk}(t)$ is the AScan (over time t) measured for the emitter receiver combination $(\vec{e}_j, \vec{r}_k)$. $T()$ summarises the applied pre-processing steps. For a given candidate point $\vec{p}$ in the image domain, the one-way travel-time variables $t_{e_j}(\vec{p})$ and $t_{r_k}(\vec{p})$ describe spherical propagation from $\vec{p}$ to all receiver positions or from all emitter positions to $\vec{p}$, respectively, with c denoting the assumed sound speed.

$$R(\vec{p})_{\vec{e}_j} = \sum_{k=1}^{N_{\text{Receiver}}} T(A_{jk}(t_{e_j}(\vec{p}))), \quad (7.16)$$

$$t_{e_j}(\vec{p}) = \frac{\|\vec{p} - \vec{r}_k\|}{c}.$$

$$R(\vec{p})_{\vec{r}_k} = \sum_{j=1}^{N_{\text{Emitter}}} T(A_{jk}(t_{r_k}(\vec{p}))), \quad (7.17)$$

$$t_{r_k}(\vec{p}) = \frac{\|\vec{p} - \vec{e}_j\|}{c}.$$

As the detection method separates emitter and receiver contributions, conclusions can be drawn independently for the receiver chain and the emitter chain. This makes it possible to identify cabling issues, mounting errors, electrical multiplexing faults, as well as software and configuration problems.

Configuration and software errors should be mostly associated with the measurement procedure, which orchestrates the emission order and configuration – mistakes here should be visible in emitter detection. Obviously, the electrical emitter chain, emission cabling, and US transducer hardware errors are represented here too.

The reception chain is less susceptible to software configuration, as it is massively parallel and therefore fixed depending on the static cabling. Dynamical aspects, such as the later data reorganisation in the server-side data parser, are checked by the detection; if there were mistakes, these would be visible in the detection.

Moreover, the method can be used as an alternative for geometry detection, with the major advantage of not being restricted to signal-level evaluations, unlike all previous approaches [Zap03, Hop03, ZRF⁺07, Tan17]. Therefore, the method can exploit the $\sqrt{n}$ SNR gain provided by averaging in the imaging domain, allowing it to overcome SNR limitations present at the raw signal level. For the 3D USCT III system, this corresponds to a maximum of 2304 contributions for emitter and receiver detection, resulting in an ideal SNR improvement factor of 48. In addition, this approach allows the use of negative-SNR signal data and works not only with high-SNR empty measurements but also with measurements containing objects (attenuating signal) or generally low-SNR conditions. The robustness against detection distortions caused by objects also comes from the wide angular coverage of all transducers, which means there are always emitter–receiver combinations that remain unimpeded and bypass the object. The second capability is the utilisation of low and very low frequencies, which provides greater tolerance against uncertainties and errors. Lower frequencies exhibit slower phase variation per distance and therefore remain coherent over larger spatial deviations, making the detection less sensitive to geometric distortions or local sound-speed fluctuations. The spherical projection was successfully utilised below 1 MHz, which corresponds to one millimetre of tolerance

$$\lambda_{\text{coherent}} = \frac{0.5 \cdot 1520 \text{ m/s}}{0.75 \cdot 10^6 \text{ Hz}} \approx 1 \text{ mm}, \quad (7.18)$$

which helps compensate sound-speed variations while still fulfilling the coherent half-wavelength criterion. For even greater tolerance, but on cost of resolution, the analytic signal can be used and the absolute value of the Hilbert transform is taken. The resulting envelope is defined by its bandwidth which can be reduced significantly by low-pass filtering e.g. to 100 kHz. When the bandwidth is on the order of 100 kHz, the corresponding coherence scale becomes

$$\lambda_{\text{coherent}} = \frac{c}{B} \approx \frac{1520 \text{ m/s}}{100 \cdot 10^3 \text{ Hz}} \approx 15.2 \text{ mm}, \quad (7.19)$$

which yields a half-coherence interval of

$$\lambda_{\text{coherent}}/2 \approx 7.6 \text{ mm}. \quad (7.20)$$

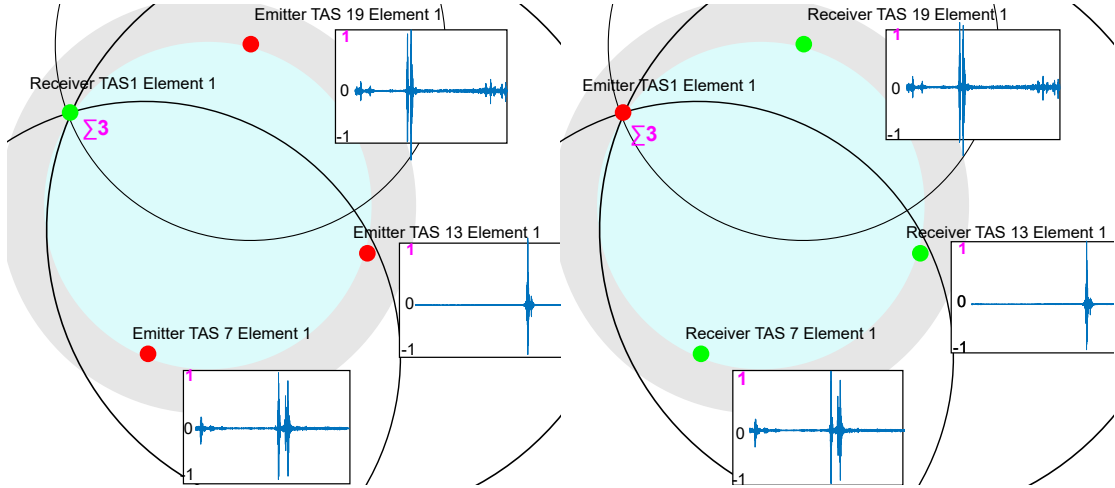


Figure 7.31: *Spherical back projection approach: top down view schematic of the approach in 2D top-down view, red dots denote origin of emitter transducer, green dots origin of receiver transducer. Left side, receiver detection case for assumed receiver TAS 1 and element 1. Utilising the ASans from emitter TAS 7 element 1, TAS 13 element 1 and TAS 19 element 1 associated with receiver TAS 1 element 1. Spherical distance projection given is as black circle for the transmission signal, created from the associated ASan. The circles overlap ideally in the expected receiver position. On the right, the analogue situation for emitter detection is described, searched is Emitter TAS 1 element 1, utilised is receiver TAS 7 element 1, receiver TAS 13 element 1 and receiver TAS 19 element 1.*

Compared to the ASans coherent half-wavelength tolerance of only ≈ 1.0 mm at 0.75 MHz, the envelope-based representation is therefore roughly an order of magnitude more tolerant with respect to geometric errors, transducer-position uncertainties, and local sound-speed variations. This increased robustness directly supports the detection performance under low-SNR conditions and in the presence of objects, where coherent signal constraints are too strict.

Additionally, this method allows the exploration not only of global maxima but the exploration of side and local maxima. This is possible due to the exploration in the full 3D image domain. Interpreted can be such maxima for instance as electrical or mechanical crosstalk, or as shortcuts in the ASIC bonding.

In Fig. 7.31 on the exemplary iterative projection process is shown in a 2D case. Left shown is a receiver detection for TAS 1 element 1, denoted as green dot. In the detection the data from Emitter TAS 7 element 1, TAS 13 element 1 and TAS 19 element 1 (red dots are origin) are projected spherical in the imaging

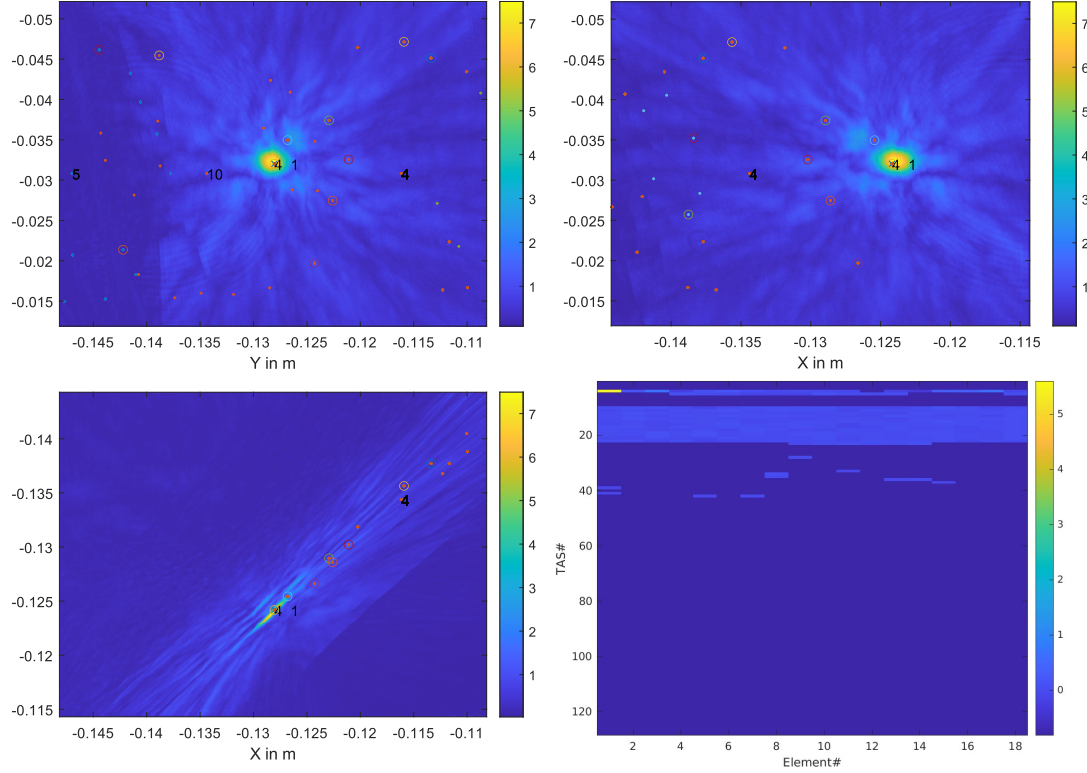


Figure 7.32: *Spherical detection for EQ9 device experiment 0020ceff713b_20241126T110935_mat Receiver detection for TAS4 element 1, three maximum intensity projections along the axis. Overlaid are the expected positions of transducers. (Top left) x-axis. (Top right) y-axis (Bottom left) z-axis. All axis are in meter. Bottom right, association of spatial position back to transducers.*

domain - denoted as black circle for the transmission signal, carrying the relevant information. When the system is well enough calibrated and the sound speed is known good enough, the circles overlap ideally in the expected receiver position summing up coherently. In the shown case, the three ASans with transmission amplitude one would sum up to an amplitude of three the image domain. On the right is the analogue case described for the emitter detection. For assumed TAS 1 element 1, ASans associated with receiver TAS 7 element 1, receiver TAS 13 element 1 and TAS 19 element 1 were projected spherical and their signal amplitude overlaps in this case too coherently to an image domain amplitude of three.

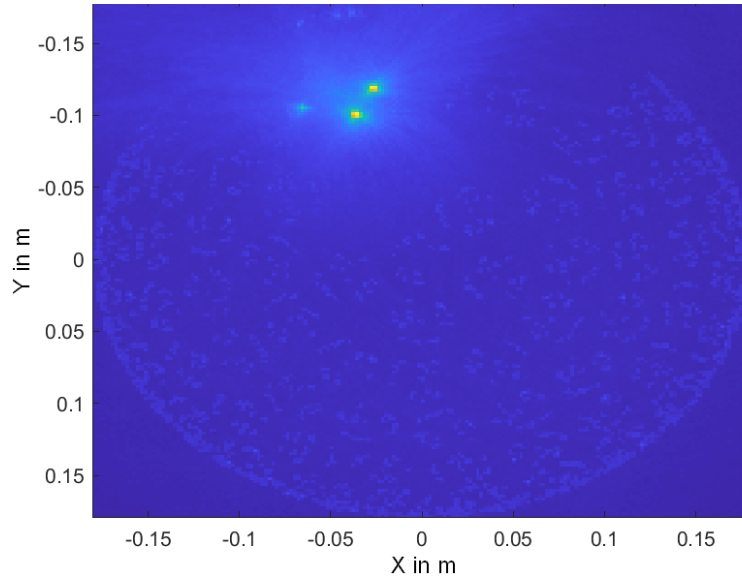


Figure 7.33: Receiver crosstalk detection for TAS 128 element 1. EQ9 experiment 0020ceff713b_20241126T110935_mat: case low SNR, three TAS channels are strongly visible and all other positions too. Maximum intensity projection along the z-axis. X- and y-axis in meter.

Application for system evaluation

In Fig. 7.32 one example is given for a 3D USCT III device, in this case the collaboration partner's device, which was analysed for errors and potential cross talk which might impede the quality. Shown are is a receiver detection for TAS4 element 1 in the spatial domain, as maximum intensity projection cut outs along X, Y, Z. Visible is a strong peak near the expected positions in all three dimensions. Bottom left, is a back association and detection to potential transducer elements, the highest peak fits the expectation TAS4 element 1 ($\approx 5 \cdot 10^8$), the next peak for another transducer would have a factor five lower peak, a suitable SNR.

In Fig. 7.33 from also an EQ9 dataset is shown for a receiver detection with an identified crosstalk case. In this case the transducer was very weak working and the two neighbouring channels were dominating the signal of this channel. Reason for three channels are the physical three channels implemented on the FEB and are the parallel output of the ASIC.

In general, crosstalk in the 3D USCT III system can be estimated to be below approximately -30 dB, based on the observed difference between measurements

obtained with properly functioning hardware and those affected by faulty channels. In the faulty cases, crosstalk becomes a dominant contribution within the signal band and effectively defines the measured signal level, albeit at an amplitude approximately 30 dB lower than that of nominal signals. From this observation, an upper bound on the inter-channel coupling strength of about -30 dB can be inferred. This is consistent with expected coupling mechanisms, such as capacitive coupling between adjacent parallel channels [KRKZ25a, ZRKD25].

Quantification of calibration status

As side effect the linear projection based detection can be used for quantifying the coherent calibration status of the complete system. There is an overlap with earlier works on geometry detection, but this is not identical.

This is a crucial feature to be capable to focus the images to best resolution and optimal SNR and dynamic range, see Sec. 5.2.5.

Core problem is missing ground truth and overlapping error sources in our complex device. One approach is evaluating the capability to achieve coherent overlapping according to the half-wavelength criteria - evaluated for all frequencies independently from our bandwidth.

Coherence is then defined as:

$$\text{Coherence}_{\%}(f) = 100 \cdot \frac{\sum_{i=1}^{n_{\text{ascan}}} \max(\text{image})}{\max(\text{ascan_filtered}_{i,f})} \quad (7.21)$$

The algorithm allow iterative processing, refining the results be reusing already existing detection geometrical results e.g. starting with a receiver detection and using it for the next emitter detection, this is currently under investigation.

7.10.1 3D USCT III System Health Status Evaluation

To assess the 3D USCT III device's operational state, an empty-water measurement is typically using default parameters. Such acquisition, registered as ex-

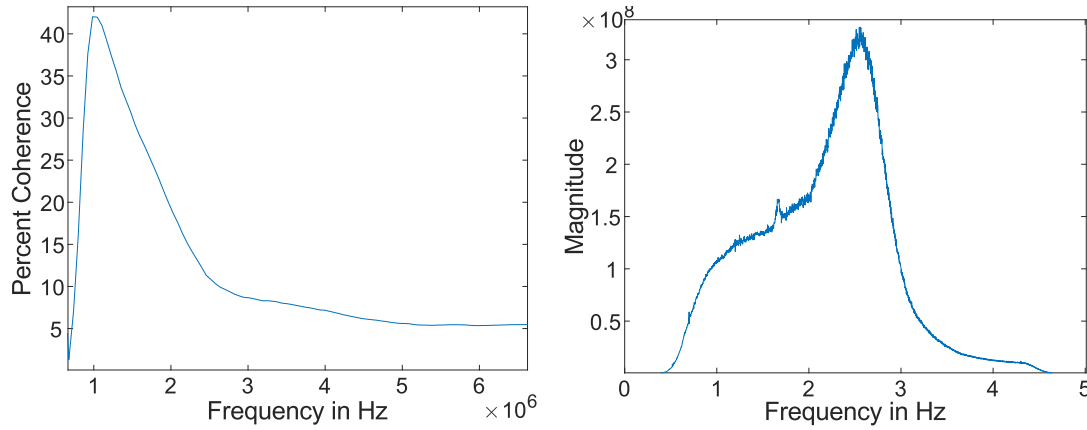


Figure 7.34: (Left graph) coherent contribution for experiment 2635, a thread phantom, measurement utilised for geometry detection. Analysed is the coherent contribution for TAS 1 element 1 in its capability as receiver. The maximum coherent response does not correlate with the maximal contributing frequency (2.5 MHz) but with a lower frequency due to more relaxed coherent criteria for lower frequencies. (Right) bandwidth for the same experiment 2635 as average Fourier spectrum of from Emitter TAS 1 Element 1, all 2304 AScans contributing.

periment 1938 (exp1938), was conducted without incident. For data quality and system health assessment, the measurement data was subsequently analysed offline.

The linear spherical back-projection method (see Sec. 7.10) is employed for this evaluation. The rationale is that if the system can successfully self-calibrate, both the hardware integrity and the measurement procedure can be considered reliable. This method also enables verification of several hardware-related features. The algorithmic setup used a full-aperture spatial search region of $0.35 \times 0.35 \times 0.20 \text{ m}^3$ to detect potential cabling or geometric errors such as TAS misalignment. Emitter detection was performed to identify or exclude DAQ misconfiguration or procedural issues, such as missing synchronisation signals. The imaging resolution was set to 3 mm, which is sufficient for robust transducer localisation. All AScans were low-pass filtered AAF accordingly.

The performance of the emitter detection run can be seen in Fig. 7.35. The histogram of the normalised maximum amplitudes in the image-domain for all transducers shows a centred Gaussian distribution; on the very left near zero is a peak with emitter transducer elements with very low detection response. There are many potential reasons from electronic components to transducers themselves,

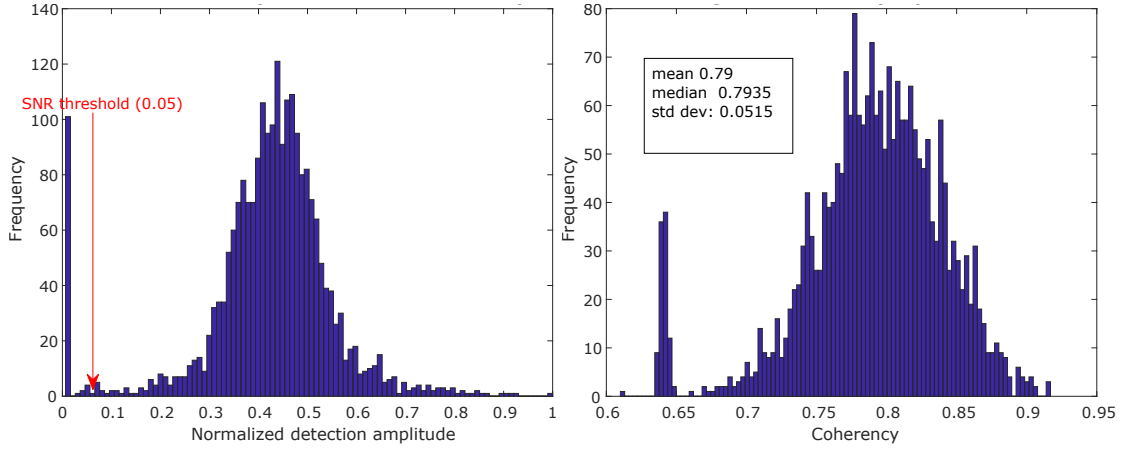


Figure 7.35: Geometrical emitter detection via spherical projection for experiment 1938. (Left) histogram of the normalised maximum amplitudes in the image-domain for all 2304 emitter transducers shows a centred Gaussian distribution and on the very left near zero a distinct peak with emitter transducer elements without or very low SNR. (Right) Histogram of coherency for all 2304 emitters detected. It has a large mean value of 0.79 (optimal value is 1). Also here, broken and weak elements differentiate themselves on the left clearly from the normal distribution of the functional elements.

therefore their origin is analysed later. The coherency, the metric which denotes the effectiveness of the synchrony and image focusing capability of the system, has a high mean value of 0.79 (optimal value is 1.0), see the graph on the right. This indicates that the measurement is suitable for further imaging usage .

The resulting detected emitter spatial positions are shown in Fig. 7.36 as vectors, pointing from the expected position to the detected one. Therefore, short or invisible vectors denote good fitting with expectation. Marked as black circles are the transducer elements with very weak detection response, defined as < 0.05 of the maximum detection response of all transducer elements. In cyan are the elements which were associated with an other transducer than the expected one, indicating a mis-detection. First, mis-detections inside a TAS are tried to be fixed by a brute-force fit by a TAS rotational transformation around the TAS's rotational axis. Assumption is that these deviations are of spatial origin, specifically the TAS housing and its mounting. This model is correct when most transducer elements from a TAS benefit from such fit and the overall deviation goes therefore significantly down. Four transducers were identified this way; as example TAS 44 is shown in Fig. 7.37 with an detected rotation of 116.8° . For the TAS's found (21, 43, 44, 87) rotational deviation are in the order of $\approx 120^\circ$, which fits well to the

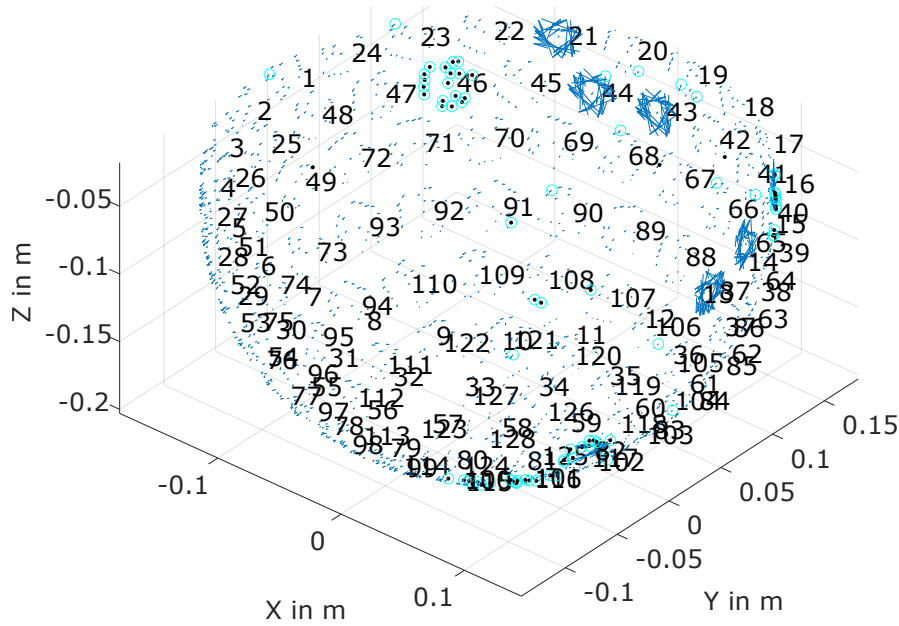


Figure 7.36: *Spatial position detection vectors of the detected emitter positions for exp1938. Blue vectors point from the expected position to the detected positions. Black dots mark low detection SNR (<0.05 max.) results, cyan marked rings denote unexpected detection results.*

hardware feature of the mounting bayonet (see Fig. 5.38), which allows rotating the TAS in steps of 120° , 240° , and 360° .

These identified TAS can now be corrected in hardware or handled by an adapted geometry file coming from a geometry calibration process with higher resolution.

To quantify the detection result, a classification was defined: high-SNR expected detection if having amplitude >0.05 of the maximum amplitude in the detection chosen from the histogram, see Fig. 7.35. Mis-detected if the detected transducer element is more near to another transducer than the expected one.

This gives four classes: high SNR detected, low SNR detected, high SNR mis-detected, low SNR mis-detected. The full classification to TAS and transducer elements is given in the classification Tab. A.2 in the appendix.

In the upper part of Tab. 7.1 the result for these four classes is given; Fig. 7.38 shows a 2D hardware sorted representation as mapping to the TAS and transducer elements associated with the two ASICs. Fortunately 94 % of the transducer elements are of high SNR and are detected at the expected position. 0.13 %

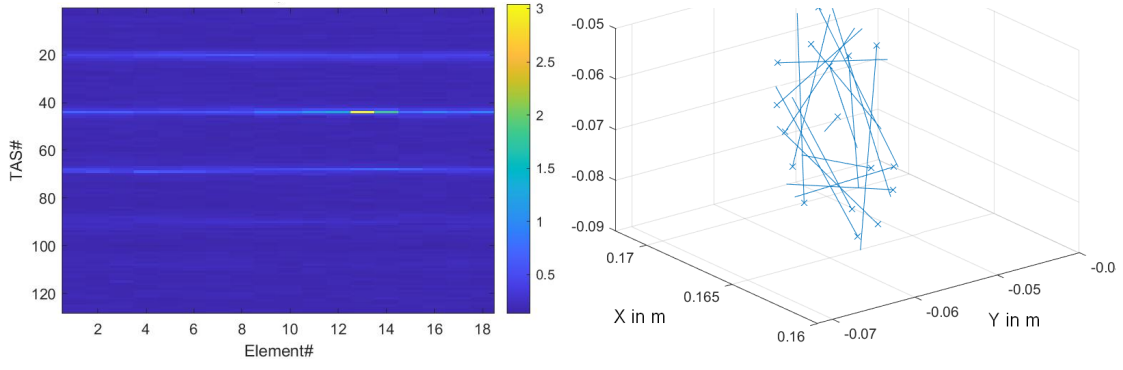


Figure 7.37: Emitter detection for exp1938, soundspeed 1938, delay 0. (Left) TAS 44 and element 1 expectation to detection correlation, highest correlation detection as TAS 44 element 13. (Right), spatial vector direction representation of all the TAS 44 expected positions to the detected position, visible is a clockwise association for all elements. A fit (right image) identified it as TAS rotation error of $\approx 116^\circ$, see Tab. A.3.

Table 7.1: Emitter detection classification for exp1938: Detection outcomes grouped by SNR level (threshold at 5 % of global maximum detection amplitude, see Fig. 7.35).

	Detected expected transducer	Detected unexpected transducer
High SNR	94.14% (2169)	1.30% (30)
Low SNR	0.130% (3)	4.43% (102)
Low SNR, Complete ASIC1 subset	—	3.51% (81)
Low SNR, Complete ASIC2 subset	—	0.390% (9)

(3 transducers) were properly detected but of low SNR. 4.4 % were of low SNR and were mis-detected, they have in fact no SNR and are therefore positioned randomly. Visible are strong horizontal red lines of CClass 3 especially in the left graph for ASIC 1 – all transducer elements for such TAS’s ASIC failed. Nine such ASICs can be identified. For ASIC 2 there seems to be only one TAS ASIC which behaves like this, indicating a specific HW-associated reason in the PCB design or bonding for ASIC 1, not a general ASIC failure. This is consistent with previous observations which showed the ASICs themselves as robust; FEB, bonding, and connectivity were the weak points noticed before. In fact, in the production of the FEBs many of the FEBs were damaged in a reproducible way, impacting always the same area on the PCB, which might be a likely source.

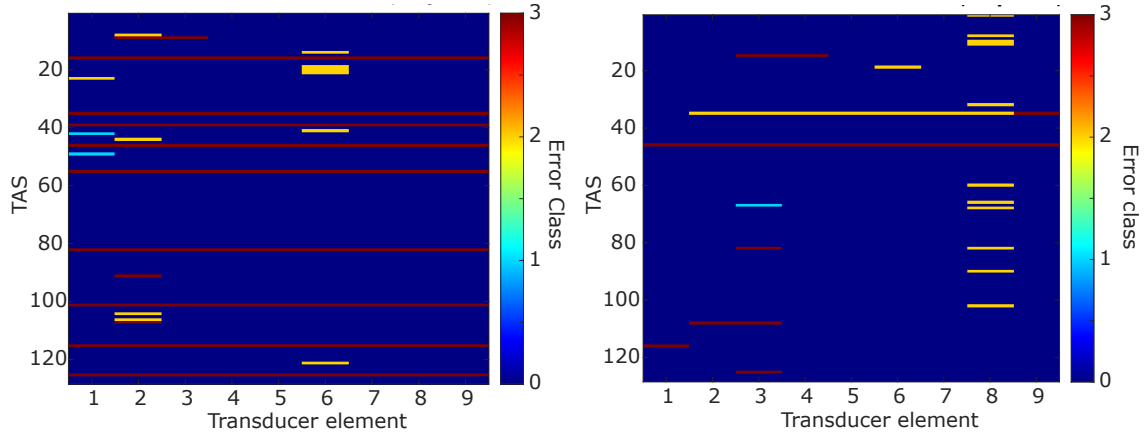


Figure 7.38: *Experiment 1938 emitter detection, separated for hardware feature ASIC 1 (left) and ASIC 2 (right). Detection classification into four classes: Class 0 is high SNR and detected as expected, Class 1 is low SNR and detected as expected, Class 2 is high SNR detected unexpectedly, Class 3 is low SNR detected unexpectedly.*

With the error source found in the ASIC and FEB, already 88% of all low-SNR transducers are explained from the detection. The remaining Class 3 ones seem to be randomly distributed, indicating origin is maybe bonding or packaging and glob-topping harming individual bonds. Interesting cases are Class 2 detections; there is some clustering in transducer element 13 and somewhat in transducer 2, which might be some cross-talk effect; analysis is here pending.

Overall, the number of functioning transducers exceeds the project goal of achieving more than 90% operational transducer elements. Therefore, it was decided not to replace the partially damaged FEBs. On the one hand, the stock of spare FEBs was already low due to limited production yield; on the other hand, it was deemed more important to retain the remaining spares for potential issues that may arise during the clinical trial.

The damaged or suspicious transducers have been identified and can be black-listed for experiments or subsequently filtered out during image-reconstruction processing. More detailed graphs and the full result protocol are provided in Appendix A.3.

From the above results the D USCT III's technical readiness can be concluded for the QUSTOM study at Vall d'Hebron Hospital in Barcelona.

8 Results and Performance

The build-up and commissioning of the first 3D USCT III system (designated *KIT 3D USCT III.1*) were completed at KIT with the goal of preparing a clinically applicable measurement protocol for use in a real hospital environment. In parallel, a second system, the *EQ9 3D USCT III*, was manufactured, assembled, and commissioned in Hangzhou (China) in collaboration with EQ9 as part of a KIT technology transfer project. The finalising multi-centre study in China got delayed due to regulatory complexities. Therefore, the final evaluation of the 3D USCT III device was carried out within the QUSTOM study at the Hospital Vall d’Hebron in Barcelona, Spain.

8.1 QUSTOM: First Clinical Commissioning and Study Results

The EU Innovation Council (EIC) project QUSTom (Quantitative Ultrasound Stochastic Tomography) aims at achieving full 3D FWI imaging. The project combines the FWI and Uncertainty Imaging concepts developed by Imperial College London (UK) and FrontWave (Spain), the optimisation strategies of ARCTUR (Slovenia), and the computational resources of the Barcelona Supercomputing Center (BSC, Spain), all applied to the 3D USCT III device developed at the Karlsruhe Institute of Technology (KIT, Germany). Medical expertise and clinical guidance are provided by the Vall d’Hebron Research Institute in Barcelona.

A clinical feasibility study with 59 patients from the breast screening program of the Vall d’Hebron Women’s Hospital was carried out between May and June 2024. The aim of this study is to validate the reconstructed FWI images and the USCT



Figure 8.1: *EU Horizon 2020 QUSTOM study conducted in Vall d’Hebron between 8 May 2024 and 14 June 2024. (Left) Examination room in the mammography with the setup up KIT 3D USCT III.R ready for the study. (Right) Vall d’Hebron hospital project team and KIT USCT project members.*

device by performing a kappa analysis of typical breast structures, using standard mammography as the reference modality

8.1.1 QUSTOM: Patient Measurement Time Evaluation

For the feasibility study at Vall d’Hebron Hospital in Barcelona, a measurement time budget of 20 minutes per examination was defined by the collaboration partners. The goal was to acquire a sufficient number of ASans for an FWI imaging workflow, specifically around 20 million ASans per patient. Achieving this data volume while keeping the required time lying on the patient bed acceptable was a central challenge.

The overall objective was to examine approximately 60 patients over a period of three months at Hospital Clinic Vall d’Hebron in Barcelona, Spain.

To prepare for the study, the error robustness and reliability of the complete system were successfully optimised. Several measures contributed to this outcome: the error-tolerant and flexible parametrisation workflow operated reliably during the Barcelona study, and the targeted patient cohort could be measured without hardware-related interruptions. This marks a clear improvement in stability

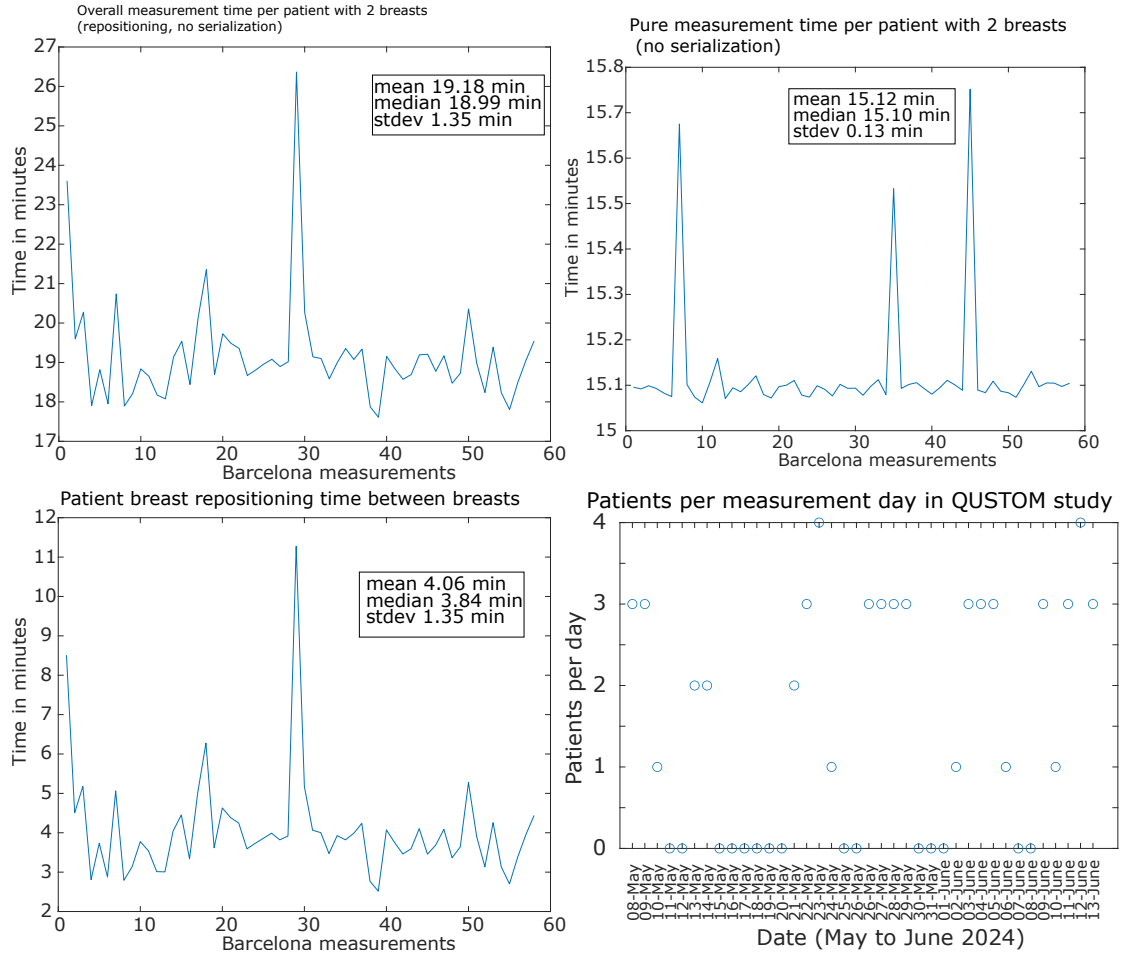


Figure 8.2: Time evaluation of the study at Vall d'Hebron Hospital in Barcelona. Experiment exp2281 was excluded due to an interrupted measurement. In total, 58 experiments were evaluated. The time window spans from 08-May-2024 15:18:07 to 14-Jun-2024 18:57:20.

compared to the previous 3D USCT II system used at Hospital Jena.

In total, 59 patients were successfully examined, corresponding to 117 individual breast scans acquired with the 3D USCT III device. These results demonstrate the practical clinical feasibility of the system, and 3D USCT III reflectivity imaging could be successfully applied [RHZ⁺25], see Fig. 8.3.

8.1.2 QUSTOM: Study Imaging Performance

For each QUSTOM data set, straight-ray based reflectivity, sound-speed, and attenuation volumes were reconstructed for quality assessment and as starting models. All looked for anatomical structures were visible in 64 % of the 117 cases. The skin and nipple were identifiable in 100 % of cases, connective-tissue structures in 86 %, and the chest muscle in 72 % of cases.

Missing internal structures occurred in several examinations. Most affected were A-cup breasts (6 cases), breasts with off-center positioning or low water immersion depth (23 cases), and breasts showing increased patient motion during the measurement (18 cases).

8.1.3 QUSTOM: Clinical Applicability

Patient positioning was generally good, yet still challenging, with numerous outliers (32 % of cases). Overall, 81 % of all examinations remained within the defined imaging volume of the USCT device. The average breast immersion depth was 7.6 cm, but positioning was noticeably more difficult for A-cup breasts.

Air bubbles were present in a large majority of cases ($> 80\%$). Although air bubbles at the patient interface typically do not degrade image quality, bubbles located near the breast tip or along the breast surface can become critical. The unexpectedly high occurrence of air bubbles during this feasibility study contrasts with earlier experience. Consequently, water treatment procedures and purification steps should be standardised and tested on site before future studies and clinical use.

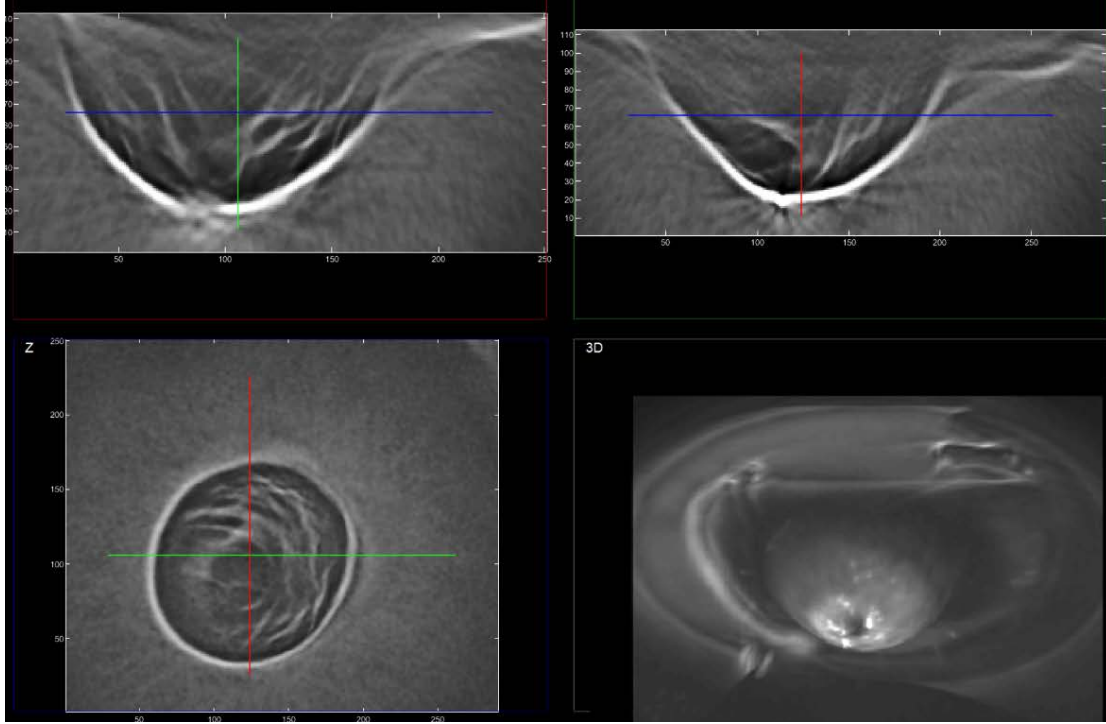


Figure 8.3: *Imaging example from a subject's breast from the QUSTOM study: reflectivity reconstruction (57-year-old volunteer with B-cup and breast density B). Left top to right bottom: Sagittal, transversal, frontal slice and maximum intensity projection.*

A subset of cases exhibited insufficient immersion depth and limited available USCT aperture coverage, especially for small breasts. In these situations, the reconstructed sound-speed volumes may not serve as reliable starting models for FWI. This limitation is largely attributable to the small breast size and suboptimal positioning in the aperture, water immersion depth averaged only 5.7 cm in these cases, compared to 7.6 cm overall.

8.1.4 QUSTOM: Outlook and next steps towards application

On the other hand the ambitious goal to transfer the 3D USCT method to FWI algorithms could not be achieved in the run time of the QUSTOM project. No direct follow project is in preparation by the QUSTOM collaboration, but alternative approaches and collaborations are in progress and in preparation. At KIT IPE a paraxial approximation based imaging scheme, is in development aiming for utilising the QUSTOM data[MGZ⁺22]. Also other collaborations are in prepa-

ration e.g. with the TU Delft's Imaging Physics Faculty of Applied Sciences, a long standing collaboration, to establish successfully the connection of the 3D USCT III measurement method and 3D FWI algorithms.

8.2 Improved Performance of 3D USCT III

While the general clinical applicability was demonstrated with the QUSTOM project and its successful clinical trial in Barcelona, some core technical goals and their impact on the systems imaging capacity needs further evaluation.

In general, the analysis of the QUSTOM measurements imaging performance and calibration efforts are ongoing, direct imaging evaluation is challenging, also due to missing ground truth provided for most cases with the Barcelona study. Therefore, for this work's analysis on imaging performance metrics and conclusions were derived directly from system properties and measurements. As notable advantage, no often hard to acquire ground truth and complex objects are required for this image performance analysis, by that simplifying the evaluation. Several aspects are discussed here based on extracted system metrics and their impact on SAFT imaging, derived from the original motivation with the Hospital Jena study and its analysis: bandwidth, SNR, opening angle and general reliability and technical functionality.

8.2.1 Bandwidth and Frequency Response

The 3D USCT III system's possible excitation (TX) bandwidth was increased from 3D USCT II 1.66 MHz to 3.81 MHz (at -6dB drop), see Fig. 8.4 left. The achieved overall receiver bandwidth in the transmission signal for the Barcelona parameters is given in Tab. 8.1.

3D USCT III has the capacity by utilising its SNR reserves to achieve broader bandwidth, nearing the ideal of a flat spectrum response, see Fig. 8.4 right and the results. In the calibration it could be shown that frequency band components below 2 Hz, until around 1 MHz, can be utilized too despite being under noise floor. Other approaches of improving the bandwidth even further are described in the

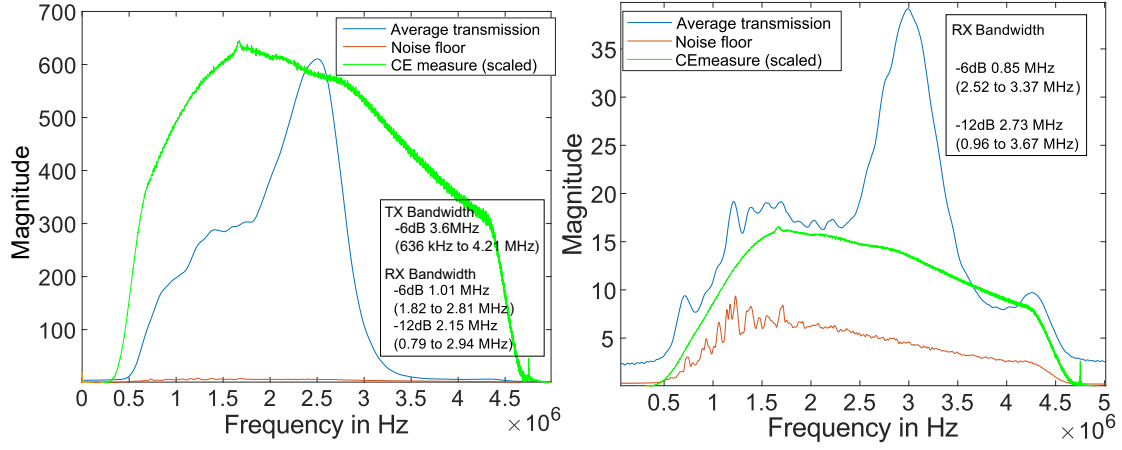


Figure 8.4: 3D USCT III frequency response for the QUSTOM study, excitation parameters linear chirp of 320 μ s length. (Left) TX frequency response according CEmeasure with uncompensated excitation. RX transmission signal response from exp2095. (Right) RX transmission signal response from exp2420 with compensated excitation.

Table 8.1: 3D USCT Transmit and receive bandwidths, see Fig. 8.4.

Signal	Level	Bandwidth (MHz)	Frequency Range (MHz)
3D USCT II TX	−6 dB	1.66	1.66 – 3.33
3D USCT III TX	−6 dB	3.81	0.636 – 4.21
3D USCT III RX (uncomp.)	−6 dB	1.01	1.82 – 2.81
3D USCT III RX (uncomp.)	−12 dB	2.15	0.79 – 2.94
3D USCT III RX (comp.)	−6 dB	0.85	2.52 – 3.37
3D USCT III RX (comp.)	−12 dB	2.73	0.96 – 3.67

future chapter.

8.2.2 Timing and Synchronisation Accuracy of the 3D USCT III DAQ

For the QUSTOM study performance it was important to achieve an uninterrupted, synchrony measurement procedure - unlike 3D USCT II which proved to be significantly more error prone and unreliable. Unnoticed errors here would lead to impaired imaging quality or completely failing imaging procedure. Additionally to the internal logging and checking mechanisms, the synchrony of the parallel

Table 8.2: Averaged angular sensitivity combined for 3D USCT III emitter and receiver, averaged over experiment 2095. Thresholds over noise limit according to Eq. 7.7.

Threshold	Angles until threshold	Combined angles over threshold	Angular space supported
−3 dB	21.6°	104.21°	57.9%
−6 dB	28.0°	123.79°	68.8%
−10 dB	32.99°	142.96°	79.4%

measured signals was evaluated by comparing their temporal alignment and jitter characteristics, utilising the self-calibration capability of CEMeasure, see Fig. 7.3. This provides a quantitative measure of the channel-to-channel delay stability and check that the DAQ system maintains phase-aligned sampling across all 384 channels and MUXes within sub-microsecond tolerances, ensuring the phase fidelity required for coherent reconstruction.

In an analysis of over 209 measurement the mean value is $0.3 \mu\text{s}$ with a standard deviation of only $0.015 \mu\text{s}$, see Fig. 7.4. This variation is significantly smaller than the temporal half-wavelength criterion corresponding to the highest employed frequency of 4.5 MHz, which amounts to

$$\frac{0.5}{4.5 \cdot 10^6} = 0.111 \mu\text{s}. \quad (8.1)$$

Hence, the observed timing fluctuation utilises only approximately 29 % of the permissible half-wavelength tolerance, confirming excellent temporal stability of the system.

Another result extractable is that 81.7% of the 2304 emitter receiver chain combinations were functional, see Fig. 7.3. The metric was, a channel a is considered good if its amplitude $A_{\text{ch},a}$ is smaller than one standard deviation below the mean of all channel amplitudes, i.e. $A_{\text{ch},a} < \sigma_A - \mu_A$.

8.2.3 Transducer and Opening Angle Performance

For the evaluation of the transducer performance an empty experiment from the QUSTOM study campaign in Barcelona was selected, exp2095. Some example data and graphs for emitter TAS 13 and element 10 are given in Fig. 7.7.

Table 8.3: Averaged angular sensitivity for a decoupled 3D USCT III transducer (either receiver or emission) by square root, averaged over experiment 2095. Thresholds over noise limit Eq. 7.7.

Threshold	Angles until threshold	Combined angles over threshold	Angular space supported
−3 dB	28.0°	123.79°	68.8%
−6 dB	37.12°	146.79°	81.6%
−10 dB	180.0°	180.0°	100%

With the binary transducer quality definition formulated in Eq. 7.7 86.0% of receivers and 91.3% of the emitters are classified functional. This is inside the expectations of having a failure rate of $\approx 10\%$ known from the 3D USCT II and, for a first device production run, is acceptable and inside our tolerances for data. Additionally, the flexible design with additional degrees of movement freedom allows to compensate for this missing data.

The averaged 3D USCT III system response, for experiment 2095 from QUSTOM study is given under Fig. 7.8. The averaged angular sensitivity combined for 3D USCT III emitter and receiver, averaged over experiment 2095 is presented in Tab. 8.2 and the estimated decomposited result representing a single transducer's performance is given under Tab. 8.3. While the direct goal of a continuous opening angle of $\approx 60^\circ$ was not achieved ($\approx 37.1^\circ$), there was positive SNR sensitivity achieved over the full range of angular space, see Fig. 7.5 and Fig. 7.6. On the other, while this enormous angular sensitivity helps with the RoI homogeneity it created also a challenge with the spherical aperture and arising side-wall reflections overlapping into the RoI's time signal windows, see Sec. 7.8.1.

8.2.4 Imaging domain SNR gain results for the 3D USCT III

In Fig. 8.5 the SNR gain of the different 3D USCT III experiments is shown together with the theoretical optimum and, as reference, the 3D USCT II system. For 3D USCT II an effective SNR-gain exponent of 0.371 was found in the earlier in-vivo study [RZD⁺12, RZHG13].

Overall, all water-based measurements cluster in the same region and remain far from the theoretical optimum, for 3D USCT II and 3D USCT III alike. The 3D

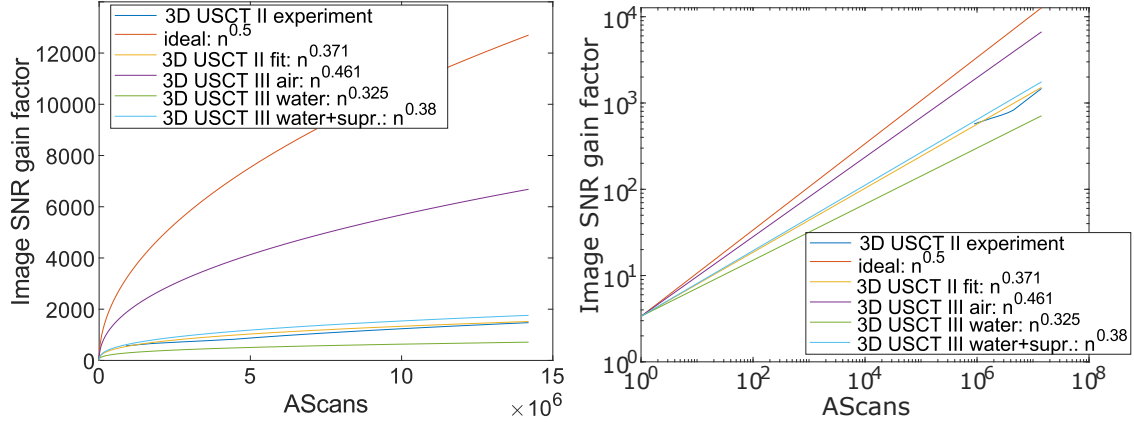


Figure 8.5: 3D USCT III imaging SNR gain potential, ideal performance compared to the measured SNR gain capacity. (Left) Linear representation. (Right) logarithmic representation.

USCT III “air” experiment (exp2300) essentially represents the electrical system performance, including EMI but without the water-related coherent noise contributions. Its significantly better SNR gain indicates that EMI and electrical cross-talk are not the primary reason for the deviation from the theoretical limit in the water experiments.

Experiment 2580 (water) performs worse in terms of SNR gain than 3D USCT II, suggesting that an additional correlated noise component appears in 3D USCT III that was not present to the same extent in the previous system. In contrast, the 3D USCT III experiment with side-wall suppression (exp2710) shows substantially improved performance – better than experiment 2580 and slightly better than 3D USCT II. This strong difference suggests that side-wall reflections are likely one of the major coherent-noise sources limiting the SNR-gain capacity of 3D USCT III. A full separation and quantification of the different noise components is ongoing. The performance metrics are summarised in Tab. 8.4.

Despite these limitations, a substantial SNR gain in the imaging domain was achieved. Importantly, this demonstrates that the conceptual shift of 3D USCT III, moving SNR gain from the time domain, where coherent noise saturates averaging, into the imaging domain via coherent SAFT, was successful. In the imaging domain several coherent noise components lose their coherence and become suppressible, enabling additional SNR gains that are not accessible in the signal domain.

Table 8.4: *SNR scaling exponents, SNR gain for 10 million AScans, and relative performance to the ideal $n^{0.5}$ case.*

System / Condition	Experiment	Fit n^x	SNR gain	Efficiency
Ideal (theoretical)	simulation (7.22)	0.5	3162.3	100%
3D USCT II	[RZD ⁺ 12, RZHG13]	0.371	502.38	12.58%
3D USCT III (air)	2300 (7.27)	0.461	1702.4	53.83%
3D USCT III (water)	2580 (7.24)	0.325	189.28	5.99%
3D USCT III (water + suppression)	2710 (7.28)	0.38	460.18	14.55%

8.2.5 3D USCT III SNR Gain Performance

The evaluated SNR gains achieved from all components mechanisms can be compared against the requirements for SAFT imaging. The central question is how much of the best-case dynamic-range potential of the 3D USCT III signal chain,

$$43\text{-bit} + 2.7\text{-bit} + 11.6\text{-bit} = 57.3\text{-bit}, \quad (8.2)$$

was actually realised. The summarised results are given in Tab. 8.5.

From the QUSTOM experiment we determined that the practically achieved signal dynamic range was 21-bit (Fig. 7.10). The partially overlapping analyses show that the transmission case reaches the full expected SNR gain of about 11.5-bit (Eq. 7.10), while the reflection case saturates at approximately 4.3-bit (Fig. 7.13). In the imaging domain, the best-case side-wall-suppression test yielded a SNR gain of 8.8-bit (Tab. 8.4).

The operational measurement scenario combines transmission and reflection data. Transmission signals define the AScan maximum because they are not clipped, and reflection signals are only visible within the remaining dynamic range. For evaluating the impact we consider both the worst-case two-way travel scenario (resolving highly attenuating structures from back to front) and the more realistic one-way travel scenario, where attenuation up to the breast centre is the limiting factor, making use of the surround-aperture geometry of 3D USCT III.

From this table we derive a combined effective dynamic range of 29.9-bit in the fused case. This is sufficient for the realistic one-way travel scenario of the defined

Table 8.5: Overview of theoretical and realised SNR contributions along the signal chain.

Case	Signal chain (ADC, FP16 datatype, filter, Average, autogain)	Average + Matched filter	Image- domain	Summary
Worst-case Requirement (2-way travel)	—	—	—	35-bit (5.2.1)
Realistic Requirement (1-way travel)	—	—	—	25-bit (5.2.1)
Best case theoretical design	43-bit	8.8-bit + 2.7-bit	11.6-bit	53.3-bit
Data evaluation	21-bit (7.10)	—	—	—
Transmission signal	—	8.8-bit + 2.7-bit (7.10)	—	—
Reflection signal	—	4.3-bit (7.13)	—	—
Noise suppression	—	—	8.8-bit (8.4)	—
Hypothetical fused case	21-bit (7.10)	—	8.8-bit (8.4)	29.9-bit

worst-case tissue parameters, but insufficient for the extreme two-way scenario. The goal of extending the dynamic range beyond the 12-bit ADC was, however, clearly achieved; the designed DAQ architecture (see Sec. 6.6.1) provides more than enough capacity for the dynamics observed in real data.

For 3D USCT III an imaging-domain SNR-gain capacity of 460.2 was demonstrated. Still, only about 14.6 % of the theoretical limit is reached, and the performance level is comparable to 3D USCT II.

From a design perspective, the achieved SNR gains compensate for the compromises introduced in the transducer and aperture design to obtain a larger opening angle and a larger RoI. Further improvement potential and ongoing work are summarised in the discussion and future-work Sec. 8.3.

8.3 Discussion and Future-work

8.3.1 Realising 3D USCT III's Full Potential

Low frequency components

Despite being not an original design goal, the importance of low frequency components became more clear over the life-time of the 3D USCT III project realisation. It became a concrete goal with the QUSTOM project and its imaging approach. QUSTOM aimed at full-wave form inversion (FWI) imaging schemes, which benefit two-fold from low-frequencies. FWI benefits due to reduced computational load, in iterative optimisations schemes in improved convergence and the non-requirement of higher frequencies for sufficient resolution. The last one emerges for FWI by being unbound by traditional linear imaging frequency-to-resolution dependency, for which the 3D USCT III was optimised for with 2.5 MHz center frequency. For the QUSTOM study bandwidth optimised chirps and bandwidth compensated chirp approach were implemented and utilised allowing as lowest frequency 1 MHz to be used and were considered good enough. Yet, as the future of general US imaging and 3D USCT III imaging goes in direction of more complex, non-linear imaging schemes (FWI, paraxial imaging etc), extending the bandwidth to frequencies below 1 MHz is of great importance. One core limitation in the 3D USCT III for achieving this, are found low-frequency EMI distortions, see Sec. 7.7.

Reducing correlated, low frequency EMI distortions from the power supply

For making the 3D USCT III ready for future very low frequency applications, alternative power supply solutions with very low ripple options and linear power supplies as replacements are an considered option. While first evaluations indicates a limited availability of medical grade options of low ripple or non-switching power supplies, some options are under evaluation currently.

Extended model calibration needed for achieving coherent focussing

The emitter detection analysis in Sec. 7.10.1 with an empty water measurement before the Barcelona clinical trial was successful and showed good system health and capability to self-calibrate and detect system status even with low frequencies. Interestingly, when the utilised frequency and bandwidth were increased, a fast drop of the coherency metric (see Fig. 8.6) was observed. This starts already for frequencies far below the aimed operating center frequency of 2.5 MHz, around 0.5 MHz, which is concerning. This indicates that there might be overall error and uncertainty sources left which sum up.

Expressed as a spatial error, the uncertainty might be in the order of $\approx 3 \text{ mm}$ ($\frac{1520 \text{ m/s}}{0.5 \cdot 10^6 \text{ 1/s}}$) or, expressed as a time error, in the order of $\approx 2 \mu\text{s}$ ($20 \times 10 \text{ MHz samples}$), which is significant and will prevent the utilisation of higher frequencies with phase-accurate, requiring imaging algorithms.

The discussion about potential error sources is crucial; several aspects might be originating from the system design as a surrounding aperture SAFT system, whereas sonography systems minimise these effects (see Sec. 5.3). Further, improved calibration might be the solution. There are indications that the strong frequency dependencies in several hardware components might be an error source, as these frequency-independent sources were covered in previous calibration approaches (spatial position, delay, sound speed)[Zap03, Hop03, ZRF⁺07, Tan17].

More advanced calibration might help realise the potential of USCT by bringing the uncertainty down to half wavelength especially for the higher frequencies around the center frequency 2.5 MHz.

The data inherent uncertainty can be modelled as phase shift originating from electronical effects, transducer effects and geometry / spatial effects.

$$\phi_{12} = \phi_{12}^E + \phi_{12}^M$$

The observed / measured phase ϕ_{12} can be formulated as being originating from Emitter and Receiver, which result in an measured AScan. Spatial errors can be

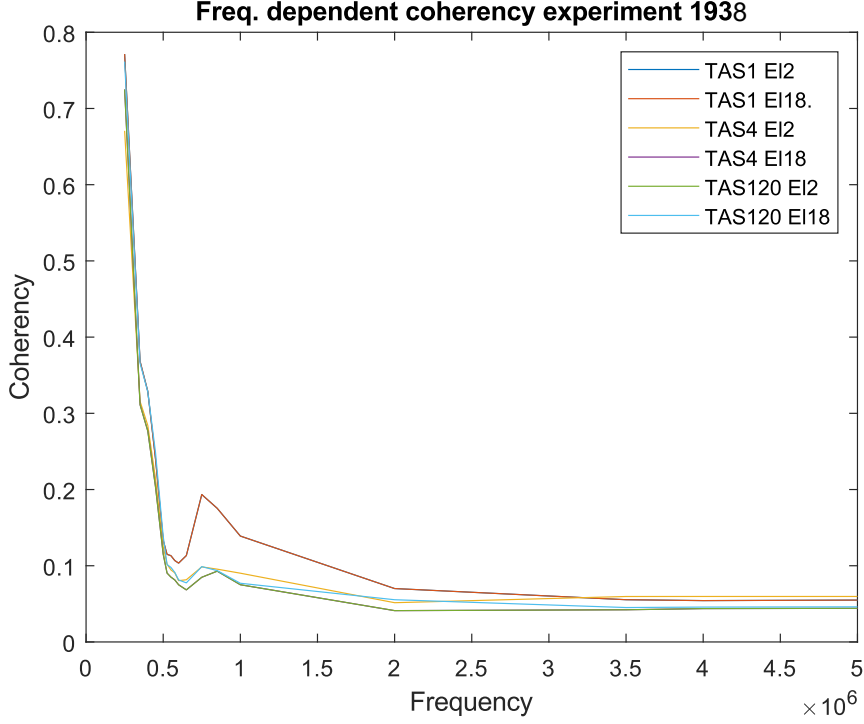


Figure 8.6: Empty water measurement experiment 1938 before the QUSTOM study: Frequency dependent coherency of the emitter linear projection detection. The coherency drops fast in direction of higher frequencies after 0.5 MHz.

due to misalignments and are in case of USCT III composed from X,Y,Z. In this formulation it is reciprocity assumed: ϕ_{12} is equal to ϕ_{21} for both electrical and spatial phase error sources. System phase shift can be modelled as:

$$\begin{pmatrix} \phi_{12}^E \\ \phi_{13}^E \\ \phi_{23}^E \end{pmatrix} = \begin{pmatrix} 1 & 1 & 0 \\ 1 & 0 & 1 \\ 0 & 1 & 1 \end{pmatrix} \begin{pmatrix} \varphi_1^E \\ \varphi_2^E \\ \varphi_3^E \end{pmatrix} \quad (8.3)$$

With such a modelling, ideally separated for individual frequency bands and system components, a sufficient half-wavelength calibration of the system to achieve an coherent focussing with the system should be possible.

8.3.2 Reducing System Uncertainties

Temperature Calibration

A better temperature calibration with a more physical correct model is not yet on the level to reduce the temperature uncertainty under the half-wavelength criterion. Extending promising previous works might be a viable path forward [Fri22].

Improved medium approach

As already discussed in Sec. 5.3.2, the 3D USCT III medium and its soundspeed uncertainty leads to calibration challenges. As alternative to the approach of calibrating or modelling the temperature variation over time and space to compensate the medium's behaviour, instead a medium with more favourable properties could be used. Luckily, there is prior work in the US domain about utilising ethanol–water mixtures, which have in the right proportion the good property of compensating each other's positive and negative temperature coefficients, see Fig. 8.7 left with an quite flat curve. First encouraging experiments were conducted with such mixtures already at KIT[KMS24]. Yet, there are some questions open about the practicability and error margins due to the sensitive response to proportion mixing errors, see Fig. 8.7 right. Already small single-digit percentage errors lead to significant deviations (several degrees) in the desired soundspeed.

System phase compensation approaches

The various contributing errors, in the Fourier domain represented as phase-shifts, coming from various sources can be minimised, covered and tracked by the methods described above or tried directly to compensated following the linear system kernel approach.

As it was shown in Sec. 2.9.3 that the imaging process of an US system can be described in the spatial Fourier domain (k-space) as a linear process if certain assumptions are met.

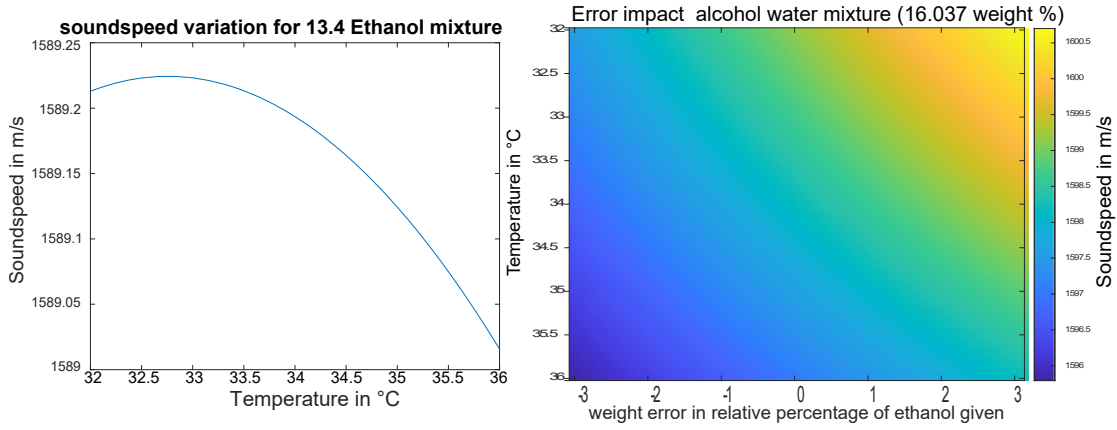


Figure 8.7: (Left) *Ethanol water mixture: x-axis temperature in °C in the 3D USCT III relevant range around 35°C. Y-axis sound speed variation.* (Right) *Deviation in sound-speed due to mixing errors. X-axis relative weight percentage error in ethanol. Y-axis sound speed deviation.*

For SAFT-based imaging, apertures with significant spatial extent relative to the object size, and particularly enclosing apertures, tend to emphasise low spatial frequencies. These over-amplified components, combined with phase shifts, result in image blurring and characteristic sidelobe artifacts. To compensate these effects, a heuristic compensating filter was designed empirically based on general aperture characteristics, and produced encouraging improvements in contrast and artifact suppression simulated and medical data[ZHR16] see Fig. 8.8. An generalised form is directly utilising the the general 3D spatial positions of the 3D USCT transducers in space instead of assuming idealised geometrical structures like spheres.

System Self-Characterisation Capabilities

For 3D USCT III there were several system level self-test and self-characterisation mechanism implemented, allowing faster commissioning and system quality checks.

Especially self-characterisation scheme via CEMeasure, see Sec. 7.2 was successfull and a significant improvement over the situation with 3D USCT II, which allows to track parameter and electronic system delay and phase shift per measurement.

While most parts of the DAQ system excitation and measurement chain are cov-

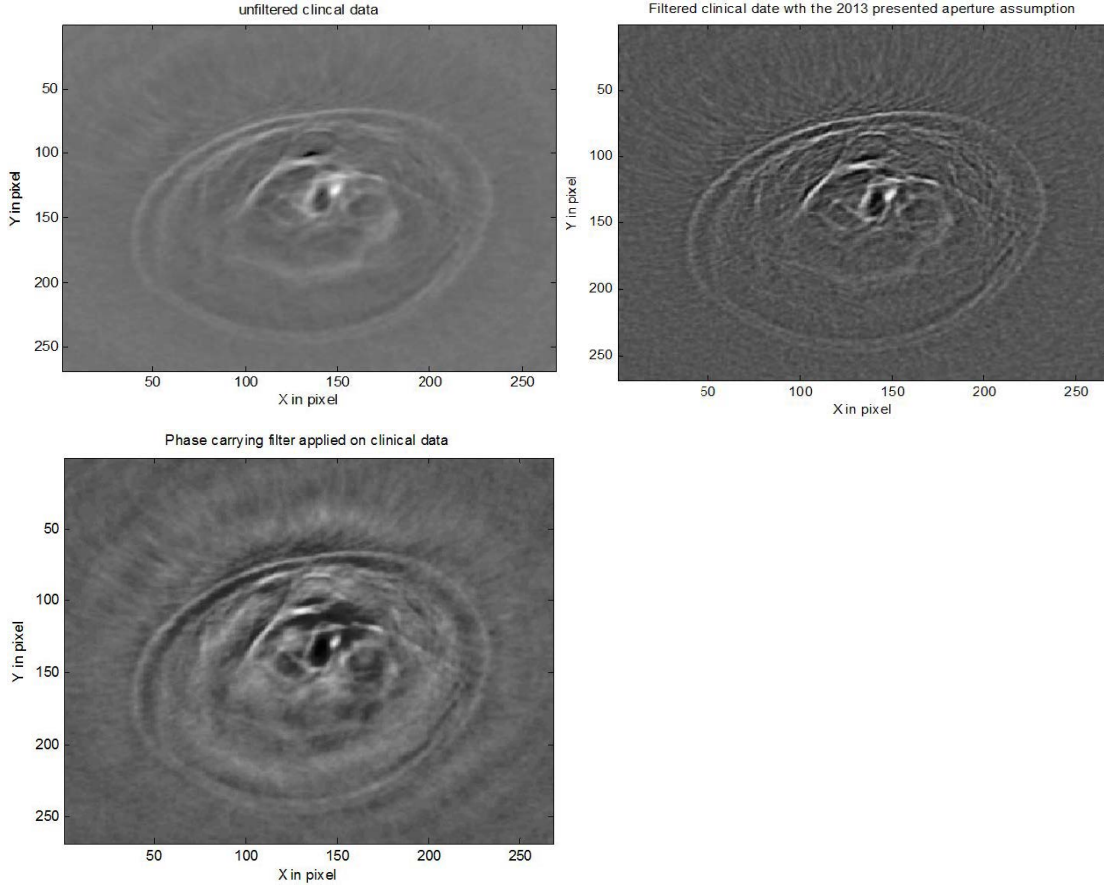


Figure 8.8: 2D frontal slice images from a clinical study (field of view $\approx 12\text{ cm} \times 11\text{ cm}$, contrast enhanced). (Top left) uncompensated reflectivity image with SAFT. (Top right) SAFT image with earlier amplitude-only aperture compensation. (Bottom) result with the new phase-compensating filter applied.

ered by the CEMeasure, it is not covering all components. Most important improvements would be the integration of the ASIC's HV parts and especially the transducers a complex load on it - an attenuator after the HV amplification part to the reception would be a superior approach. Also, an ASIC channel feature for channel test pattern encoding would be very helpful for fast system evaluation.

Another way would be to model the delay and attenuation via the Kramer-Kronig criterion based on water measurements to compensate this system imminent components currently not covered.

With the timelog mechanism acted as concurrent “watch dog” an parallel mechanism to the internal logging identifying hiccups in the sensitive step-ack reconfiguration process between Master FPGA and the μC of the TAS. This part was in

3D USCT II a error prone and hard to debug major weakness, which was fixed with this generation allowing error tracking and error recovery where possible.

8.3.3 Further Bandwidth Improvements

Hardware Improvements for the existing Architecture

A more fundamaental approach for extending bandwidth would be improving on the source, the US transducer. Here happened already testst and improvements im the project which are encouraging. An first set is produced and planned for integration with the 3D USCT III.2[PZR24]. More general, the complete US imaging domain experiments with exciting alternative transducer approaches to traditional PZT designs.

But also in the analog electrical chain there is more potential for improvement, between the emitter and receiver transducers and the FEB's first amplifier stage an added passive component or active matching circuitry promises optimisation potential in bandwidth and SNR without complete redesign need of the system[Yau25].

Bandwidth: Improved coded excitation schemes

The in Sec. 6.6.8 described approach of bandwidth optimisation to generate a flat spectrum response has the downside that the full voltage range is under-utilised in the excitation. Also the 10-bit DAC limits the achievable dynamic range and signal fidelity achievable. While further improvements in this voltage adaption scheme are possible, see proposed chirp in Fig. 8.9 which fixes the 3 MHz over-excitation, the reduced SNR efficiency downside remains.

An better, alternative approach would be an arbitrary non-linear chirp encoding with phase changing speed compensating for the system frequency sensitivity response. Advantage would be full utilisation of the maximum voltage range, therefore creating a flat spectrum response without compromising energy so much.

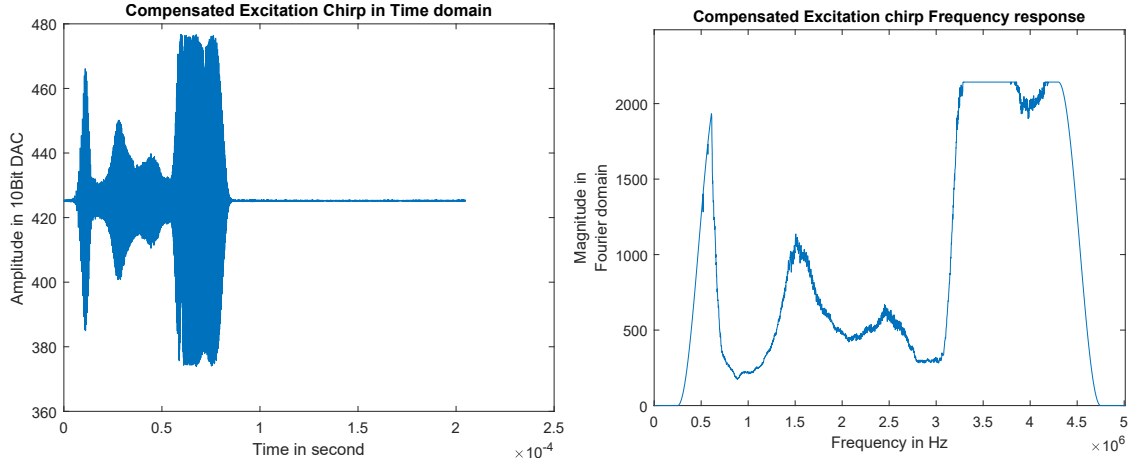


Figure 8.9: *Second-generation compensated chirp for the Barcelona FWI measurement scheme. (Left) time domain. (Right) Fourier domain.*

ASIC parameter optimisation for increased bandwidth and low frequency performance

Another promising approach to increase the bandwidth and improve the lower frequency response is optimising the highly parametrisable ASICs further for broader bandwidth. In a bachelor thesis analysis the potential was evaluated, from the 20 ASIC parameters seven were identified as relevant[Klo24]. The found parameter sets, promise some substantial improvement potential which should be evaluated in future.

3D USCT aperture subset selection

A further idea for handling the side-wall reflections is to select specifically an optimised, less impacted subset of all the emitter–receiver transducer combinations. In the spherical back-projection SNR images, it appears that lower emitter combinations create fewer side-wall reflections. The SNR rises from top to bottom by roughly a factor of two, see Fig. A.4. The standard deviation graph, see Fig. A.5, as noise floor representation illustrates that the noise floor is rising from bottom to top and not the amplitude of the signals is falling, supporting a side-wall reflection interpretation.

Advanced beam-pattern transducer aperture

At least for mono-frequent systems, a perfect wave-front with homogenous sensitivity in a desired open angle range and without sensitivity outside is representable in transducer design, avoiding unwanted side-wall reflection. Some example is given in Fig. 3.19 and the method is described in Sec. 3.7.2.

More advanced imaging methods

Methods not requiring linear behaviour of the system, e.g. by either modelling it in or by being robust against it (iterative solver methods) like from the FWI family, need further investigation.

While the broad bandwidth approach of USCT III should allow a broad range of algorithms and methods to be applied, the inherent 3D-ness of the data is a challenge, complexity and calculation load wise. AI based approaches might be here a way out by offering performance lower than O^3 coming from the spatial 3D dimensionality requirement.

Post-beam forming and k-space imaging, as indicated in earlier chapters are further potential approaches.

9 Summary

This PhD thesis presents a comprehensive investigation into the development of an advanced medical three-dimensional Ultrasound computed tomography (3D USCT) system. At the Institute for Data Processing and Electronics (IPE) of the Karlsruhe Institute of Technology (KIT), a novel 3D USCT method is being developed – constituting the first device of its kind worldwide. Over the course of the project, several demonstrator generations have been built, evolving from a two-dimensional proof of concept and a technical 3D demonstrator to a clinically applicable device (3D USCT II). The present work focuses on the conception and realisation of the third-generation system (3D USCT III), which was specifically designed for early breast cancer detection through innovative imaging methodologies and system design principles. The aim of this generation was to bridge the gap between the first clinical feasibility studies and routine clinical use for independent diagnostic purposes.

Theoretical Framework

A rigorous theoretical foundation for US imaging systems was established based on Fourier optics and k -space analysis. Concepts from electromagnetic wave and Fourier imaging theory were translated into the acoustic domain, resulting in mathematical models that describe US propagation, diffraction, and image formation as linear and shift-invariant systems. This theoretical framework enabled the systematic design and optimisation of complex 3D aperture geometries and the quantitative evaluation of key imaging parameters such as spatial bandwidth, lateral and axial resolution, aperture configuration, and sensor positioning. The derived methods bridge the theoretical imaging performance described in k -space

with the practical constraints of real-world system design and implementation.

Motivation and Design Evolution

The design of the 3D USCT III system directly addresses the limitations identified during a clinical study performed at the University Hospital Jena, where ten patients were examined using the 3D USCT II prototype. Key findings from that study included:

- The region of interest (RoI) was not sufficiently large to accommodate anatomical variability.
- Breast positioning within the hemispherical aperture was challenging, particularly in achieving sufficient submersion and central alignment.
- Patient motion occurred during the several-minute acquisition period, that was required to record approximately 5 million AScans per motor position.
- Operational inefficiencies such as lengthy water preparation and cleaning cycles require improvement.

To address these challenges, the imaging volume was expanded from $10 \times 10 \times 10 \text{ cm}^3$ to $20 \times 20 \times 20 \text{ cm}^3$, for accommodating realistic breast anatomy as well as positioning tolerances. Moreover, the mechanical, procedural, and electronic sub-systems were redesigned to improve robustness, reliability, and manufacturability, essential prerequisites for clinical translation.

Imaging Concept and Transducer Development

The new imaging architecture employs a surrounding aperture configuration that provides homogeneous illumination from multiple directions to mitigate shadowing artifacts common in conventional sonography. Unlike 2.5D systems, the 3D USCT III system achieves true volumetric imaging, enabling the detection of out-of-plane structures.

The transducer design was substantially revised to increase bandwidth and opening angle while maintaining sufficient signal strength. The resulting amplitude loss was compensated by increasing the signal-to-noise ratio (SNR), achieved via faster measurement sequences that allow signal averaging across repeated acquisitions and longer excitations. In addition, the major mechanism of SNR gain by synthetic aperture focussing technique (SAFT) in the imaging domain, which provided several orders of magnitude of SNR, was demonstrated. While the number of transducers was increased, fully dense spatial sampling was not feasible. Compressive sensing principles were therefore employed, demonstrating that semi-randomly distributed sparse apertures can retain nearly equivalent information content compared to Nyquist-dense configurations.

Based on simulation studies, this dissertation defined idealised apertures and transducer geometries to be round and planar point sources of approximately 0.45 mm^2 (see Sec. 5.4.4). These models correlated well with experimental observations from prior generations. A new transducer concept employing PZT piezo-fiber technology (Fraunhofer IKTS, Dresden) was subsequently implemented, allowing arbitrary piezo-transducer positions, a more homogeneous 3D wave field response, high production reproducibility, and efficient production scalability for effective commercialisation into the medical market.

Signal Processing and Data Acquisition

The increase in opening angle (approximately 68° over -3 dB, see Fig. 8.3) and the increased aperture size reduced the effective acoustic amplitude for both transmission and reception by a factor of roughly four, corresponding to an SNR decrease of about 16 times. This amplitude-related loss was compensated by optimised digital signal processing, including coded excitation schemes and adaptive averaging enabled by faster acquisition sequences.

Each motor position acquires approximately 5 million AScans, and the standard measurement protocol involves two positions, resulting in a total of about 10 million AScans per breast. To support advanced extended measurement procedures, a new data acquisition (DAQ) architecture was implemented, based on industry

standard μ TCA v4. Originating from high-energy physics applications at DESY and adapted for USCT at IPE, this architecture provided a substantial upgrade in digital performance compared to previous generations. In addition, a custom USCT-specific ASIC was developed, optimising the analog front-end for the specific requirements of the new system.

Integration and Clinical Evaluation

System integration was conducted in collaboration with the industrial partner Realcan EQ9 in Hangzhou, China, resulting in two systems built in parallel: 3D USCT III.R KIT and 3D USCT III.R EQ9. Despite restrictions during the COVID-19 pandemic, development and validation proceeded successfully through distributed engineering and online coordination. New diagnostic and calibration tools, including a coherence-based imaging quality metric, were developed for system evaluation.

Clinical validation was achieved within the EU Horizon 2020 project QUSTOM at the Val d'Hebron Hospital in Barcelona, Spain, involving 59 patients. The study confirmed the clinical viability of the 3D USCT III, achieving:

- an expanded RoI with improved illumination homogeneity,
- a measurement time of approximately only 5 minutes per 10 million ASans including overheads (Sec. 8.1.1),
- a patient throughput of up to four examinations per day, and
- a system health evaluation, with a resulting coherence of 41 % at 1 MHz (Fig. 7.34), as first quantification of the calibration status and the system's focus capability for sharp images.

The system demonstrated full utilisation of the 10 MHz DAQ bandwidth, successfully implementing extended measurement procedures and achieving an SNR gain potential of up to 111 times in the digital processing chain. Most of the key goals of improving 3D USCT II were therefore achieved with the 3D USCT III system.

Scientific Contributions

This dissertation makes several key contributions to the field of ultrasound tomography and system design:

- 1. Transfer of Fourier Optics Principles to Ultrasound Tomography.** The work adapts foundational concepts from Fourier optics imaging theory to ultrasound tomography, providing a coherent framework linking aperture formation, bandwidth, and spatial frequency coverage to achievable resolution and object representation. This enables predictive system analysis beyond heuristic design rules.
- 2. Multi-Spectral and SNR-Guided System Design.** A design methodology is introduced that defines required frequency bands and signal-to-noise ratios for imaging performance. The influence of spectral content and SNR on resolution, contrast, and reconstruction fidelity is quantified, forming a basis for design trade-offs across transducer technology, analog front-end design, data acquisition, and signal processing.
- 3. System Architecture with Emphasis on Linearity and Information Preservation.** A complete system architecture is proposed to maintain linear behavior throughout the signal chain while maximising recoverable information. Design criteria for excitation, measurement strategies, and hardware ensure compatibility with linear reconstruction, even for complex and varying object geometries.
- 4. Investigation of Irregular Sampling in Transducer Arrays and Apertures.** Irregularly sampled transducer geometries are analysed regarding aperture formation, beam patterns, and image quality. Guidelines are provided to exploit non-uniform sampling for improved spatial coverage and reduced aliasing in complex 3D arrays. A novel piezo-fibre-based design was developed and prototyped.
- 5. Identification and Evaluation of Correlated Noise Sources.** Correlated noise mechanisms in ultrasound tomography acquisition chains are identified and quantified for the first time, and their impact on image for-

mation is systematically evaluated. Strategies for mitigating these effects are proposed to improve robustness and diagnostic image quality.

6. **System Health Monitoring and Calibration.** A calibration and system health framework is introduced, combining a phase- and frequency-sensitive spherical projection approach with frequency-dependent focus analysis. This enables quantitative tracking of system stability and degradation, supporting reproducible high-performance operation.

Status and next steps

While with the 3D USCT III the clinical viability was significantly increased from the previous generation device 3D USCT II (TRL ≈ 6) to at least TRL 7 with the QUSTOM trial.

Consequently, following the successful clinical validation in Barcelona, the 3D USCT III system has entered the regulatory preparation phase for a planned multicenter study in China with the partner EQ9. At the same time, there are ongoing discussions to bring the method to the Europe / Near East region as well, in cooperation with a new commercial and academic collaboration partner. Aiming at full market entrance as a medical device TRL 9 is in reach with the required regulatory and technical improvements in progress with international partners.

Ultimately, this research advances 3D US tomography as a powerful diagnostic tool that enables earlier and more reliable detection of breast cancer as well as improved patient outcomes.

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A Appendix

A.1 Contributions

Engineering and project contributions

- Technical lead and hardware team organiser for KIT 3D USCT III built up and commissioning.
- Definition of the DAQ requirements and architecture.
- Implementing the DAQ software component of the DAQ measurement procedure.
- Commissioning and Evaluation of the device and data quality
- Technical lead of 3D USCT III preparation and adaption for the specific requirements for the QUSTOM study at Val d'Hebron hospital in Barcelona, Spain.
- Successful technology transfer to commercial company Realcan/EQ9 in Hangzhou, China.

Scientific Contributions

- Adapting Fourier optics imaging theory to US.
- System layout - focus on linear system, complexity and bandwidth.
- Irregular sampling transducers and aperture.
- Bandwidth and SNR focus driven system design.

- Identifying correlated noise sources and evaluating impact, proposals for handling.
- System health evaluation and calibration: Linear spherical projecting method and frequency dependent quantification of system focus status.

Publications

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Patents

Patents related to 3D Ultrasound Computer Tomography at KIT involving Michael Zapf.

- *Device for ultrasound-supported reflection and transmission tomography,*
 Patent No.: PCT/EP2015/000369 (KIT File No. A14006),
 Status: Granted in US (US 10,758,208 B2), EP (EP 3107459 B1); applied in DE,
 Inventors: Hartmut Gemmeke, Michael Zapf, Torsten Hopp, Robin Dapp, Nicole Ruiter.
- *Device for actuating and reading a group of ultrasound transducers for ultrasound computer tomography, and ultrasound computer tomography machine,*
 Patent No.: EP 3 688 486 B1,
 Status: DE, JP, CN,
 Inventors: Ivan Peric, Michael Zapf, Hartmut Gemmeke, Richard Leys.
- *Apparatus and method for inspecting an object using ultrasonic waves in the field of material testing,*
 Patent No.: EP 3 239 706 A1,
 Status: EP,
 Inventors: Ahmad Osman, Bernd Valeske, Randolph Hanke, Werner Bähr, Hartmut Gemmeke, Nicole Ruiter, Michael Zapf, Torsten Hopp.
- *Device and method for 3D ultrasound-based reflection and transmission tomography,*
 Patent No.: EP 4 146 081 A1,
 Status: Applied in EP, CN, US; US (US 11,877,892 B2) granted Jan 23, 2024,
 Inventors: Nicole Ruiter, Hartmut Gemmeke, Michael Zapf, Torsten Hopp, Nima Rashvand.

A.2 Appendix of SNR Experiments parameters of 3D USCT III

Table A.1: Configuration of all 72 SNR evaluation experiments (*pause* = 200 μ s, *dacdelay* = 500 ns, *shotlength* = 8192 samples, *exc scale* = 0.5). Loop order: excitation length first, averaging second.

No.	Filename	Averaging	Exc. length [s]	Jitter	Group
1	__exp1022_snr_analysis	1	1e-05	0	A
2	__exp1023_snr_analysis	1	2e-05	0	A
3	__exp1024_snr_analysis	1	4e-05	0	A
4	__exp1025_snr_analysis	1	8e-05	0	A
5	__exp1026_snr_analysis	1	1.6e-04	0	A
6	__exp1027_snr_analysis	1	3.2e-04	0	A
7	__exp1028_snr_analysis	4	1e-05	0	A
8	__exp1029_snr_analysis	4	2e-05	0	A
9	__exp1030_snr_analysis	4	4e-05	0	A
10	__exp1031_snr_analysis	4	8e-05	0	A
11	__exp1032_snr_analysis	4	1.6e-04	0	A
12	__exp1033_snr_analysis	4	3.2e-04	0	A
13	__exp1034_snr_analysis	16	1e-05	0	A
14	__exp1035_snr_analysis	16	2e-05	0	A
15	__exp1036_snr_analysis	16	4e-05	0	A
16	__exp1037_snr_analysis	16	8e-05	0	A
17	__exp1038_snr_analysis	16	1.6e-04	0	A
18	__exp1039_snr_analysis	16	3.2e-04	0	A
19	__exp1040_snr_analysis	64	1e-05	0	A
20	__exp1041_snr_analysis	64	2e-05	0	A
21	__exp1042_snr_analysis	64	4e-05	0	A
22	__exp1043_snr_analysis	64	8e-05	0	A
23	__exp1044_snr_analysis	64	1.6e-04	0	A
24	__exp1045_snr_analysis	64	3.2e-04	0	A
25	__exp1046_snr_analysis	256	1e-05	0	A
26	__exp1047_snr_analysis	256	2e-05	0	A

Table A.1: Configuration of all 72 SNR evaluation experiments (continued)

No.	Filename	Averaging	Exc. length [s]	Jitter	Group
27	_exp1048_snr_analysis	256	4e-05	0	A
28	_exp1049_snr_analysis	256	8e-05	0	A
29	_exp1050_snr_analysis	256	1.6e-04	0	A
30	_exp1051_snr_analysis	256	3.2e-04	0	A
31	_exp1052_snr_analysis	1024	1e-05	0	A
32	_exp1053_snr_analysis	1024	2e-05	0	A
33	_exp1054_snr_analysis	1024	4e-05	0	A
34	_exp1055_snr_analysis	1024	8e-05	0	A
35	_exp1056_snr_analysis	1024	1.6e-04	0	A
36	_exp1057_snr_analysis	1024	3.2e-04	0	A
37	_exp1058_snr_analysis	1	1e-05	1	B
38	_exp1059_snr_analysis	1	2e-05	1	B
39	_exp1060_snr_analysis	1	4e-05	1	B
40	_exp1061_snr_analysis	1	8e-05	1	B
41	_exp1062_snr_analysis	1	1.6e-04	1	B
42	_exp1063_snr_analysis	1	3.2e-04	1	B
43	_exp1064_snr_analysis	4	1e-05	1	B
44	_exp1065_snr_analysis	4	2e-05	1	B
45	_exp1066_snr_analysis	4	4e-05	1	B
46	_exp1067_snr_analysis	4	8e-05	1	B
47	_exp1068_snr_analysis	4	1.6e-04	1	B
48	_exp1069_snr_analysis	4	3.2e-04	1	B
49	_exp1070_snr_analysis	16	1e-05	1	B
50	_exp1071_snr_analysis	16	2e-05	1	B
51	_exp1072_snr_analysis	16	4e-05	1	B
52	_exp1073_snr_analysis	16	8e-05	1	B
53	_exp1074_snr_analysis	16	1.6e-04	1	B
54	_exp1075_snr_analysis	16	3.2e-04	1	B
55	_exp1076_snr_analysis	64	1e-05	1	B
56	_exp1077_snr_analysis	64	2e-05	1	B
57	_exp1078_snr_analysis	64	4e-05	1	B

Table A.1: *Configuration of all 72 SNR evaluation experiments (continued)*

No.	Filename	Averaging	Exc. length [s]	Jitter	Group
58	_exp1079_snr_analysis	64	8e-05	1	B
59	_exp1080_snr_analysis	64	1.6e-04	1	B
60	_exp1081_snr_analysis	64	3.2e-04	1	B
61	_exp1082_snr_analysis	256	1e-05	1	B
62	_exp1083_snr_analysis	256	2e-05	1	B
63	_exp1084_snr_analysis	256	4e-05	1	B
64	_exp1085_snr_analysis	256	8e-05	1	B
65	_exp1086_snr_analysis	256	1.6e-04	1	B
66	_exp1087_snr_analysis	256	3.2e-04	1	B
67	_exp1088_snr_analysis	1024	1e-05	1	B
68	_exp1089_snr_analysis	1024	2e-05	1	B
69	_exp1090_snr_analysis	1024	4e-05	1	B
70	_exp1091_snr_analysis	1024	8e-05	1	B
71	_exp1092_snr_analysis	1024	1.6e-04	1	B
72	_exp1093_snr_analysis	1024	3.2e-04	1	B

A.3 Appendix of Linear Projection Emitter Detection Results for 3D USCT III exp1938

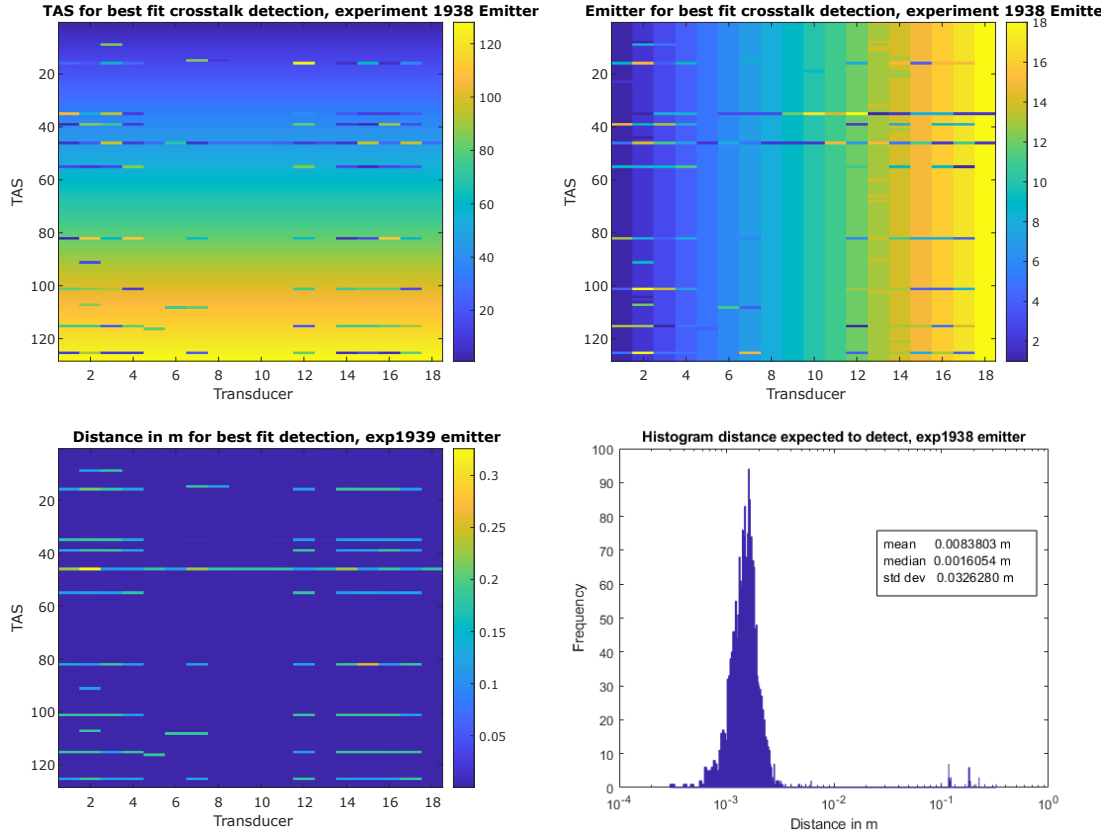


Figure A.1: Experiment 1938 Emitter detection: Per transducer element metrics. (Top left) TAS detection, top right transducer element detection. (Bottom left) distance in m between expectation and detection. Bottom right coherency per element (image resolution 0.003 m)

Low SNR Misdetected (1)		Low SNR Misdetected (2)		Low SNR Misdetected (3)		High SNR misdetected		Low SNR detected	
TAS	Transducer	TAS	Transducer	TAS	Transducer	TAS	Transducer	TAS	Transducer
9	2	46	1	91	2	1	13	42	1
9	3	46	2	101	1	8	2	49	1
15	7	46	3	101	2	8	13	67	7
15	8	46	4	101	3	10	13		
16	1	46	5	101	4	11	13		
16	2	46	6	101	12	14	14		
16	3	46	7	101	14	19	10		
16	4	46	8	101	15	19	14		
16	12	46	9	101	16	20	14		
16	14	46	10	101	17	21	14		
16	15	46	11	107	2	23	1		
16	16	46	12	108	6	32	13		
16	17	46	13	108	7	35	6		
35	1	46	14	115	1	35	7		
35	2	46	15	115	2	35	8		
35	3	46	16	115	3	35	9		
35	4	46	17	115	4	35	10		
35	12	46	18	115	12	35	11		
35	14	55	1	115	14	35	13		
35	15	55	2	115	15	41	14		
35	16	55	3	115	16	44	2		
35	17	55	4	115	17	60	13		
35	18	55	12	116	5	66	13		
39	1	55	14	125	1	68	13		
39	2	55	15	125	2	82	13		
39	3	55	16	125	3	90	13		
39	4	55	17	125	4	102	13		
39	12	82	1	125	7	104	2		
39	14	82	2	125	12	106	2		
39	15	82	3	125	14	121	14		
39	16	82	4	125	15				
39	17	82	7	125	16				
		82	12	125	17				
		82	14						
		82	15						
		82	16						
		82	17						

Table A.2: Detection classification for emitter detection Exp1939.

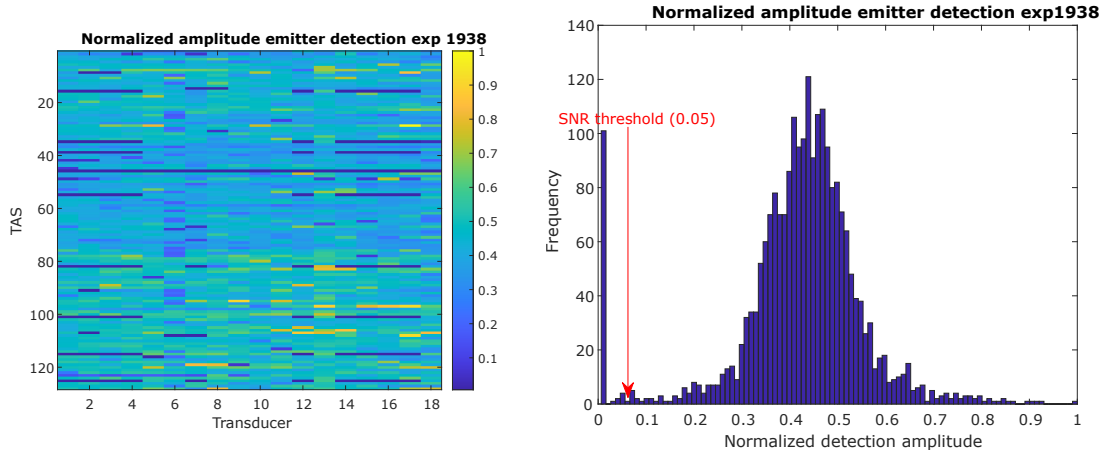


Figure A.2: *exp1938 emitter detection amplitude. Amplitude normalized to maximum value of all transducers. Threshold set to 0.05 according distribution.*

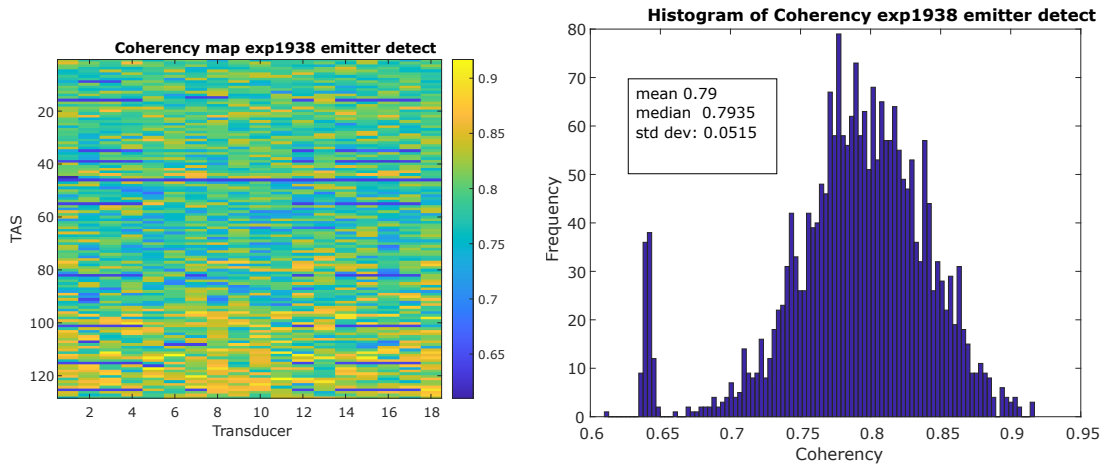


Figure A.3: *exp1938 emitter detection: Coherency for every transducer.*

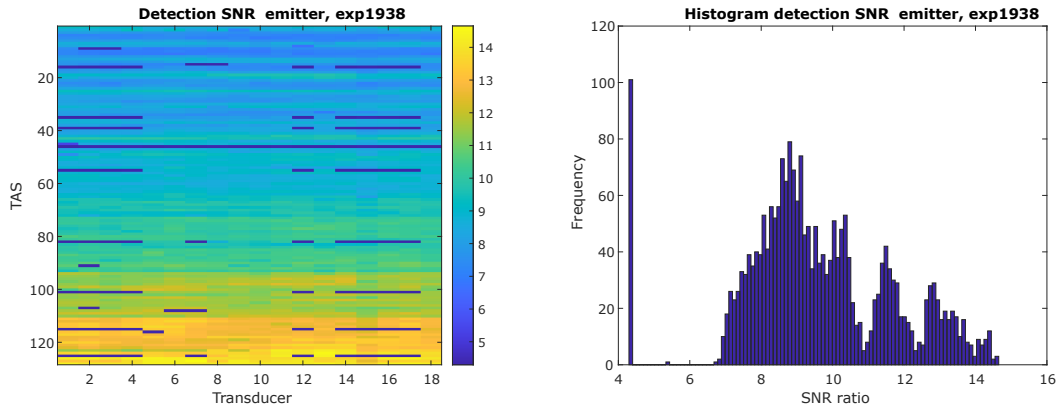


Figure A.4: *exp1938 emitter detection: SNR defined as maximum amplitude over the whole image domain divided by the standard deviation over all the image domain pixels. Projection-domain SNR rises from top (TAS 1) to bottom (TAS 128) by a factor of ≈ 2 .*

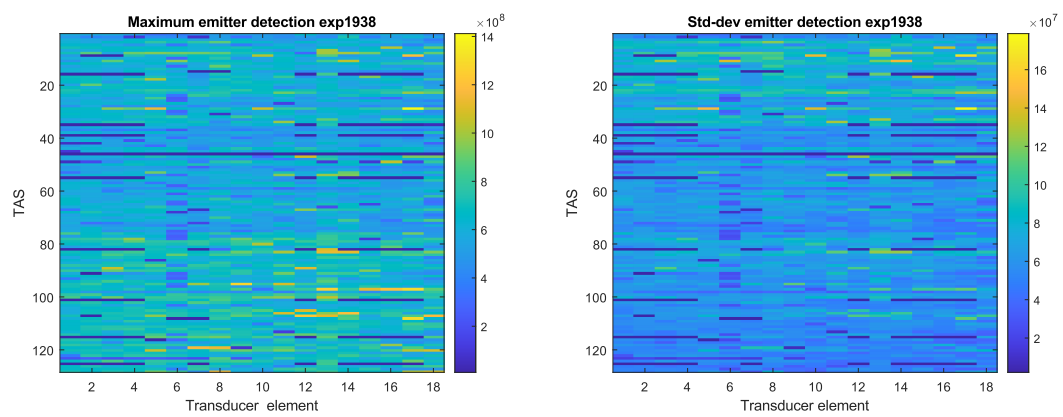


Figure A.5: *exp1938* emitter detection, stdev and maximum which from the SNR metric. (Bottom left) maximum value from the imaging domain. (Bottom right) standard deviation of the whole imaging domain as noise floor approximation.

Table A.3: *Experiment 1938 Emitter detection: TAS metrics, positioning error and rotation statistics.*

TAS	Δr [m]	$\Delta r_{TASaxis}$ [m]	share $r_{TASaxis}$ [%]	$\Delta\phi$ [deg]
1	0.002060	0.000955	46.4	-1.237
2	0.002215	0.001420	65.2	-1.237
3	0.002221	0.001462	67.7	-2.383
4	0.002077	0.001547	75.8	-2.383
5	0.002358	0.001679	72.2	-3.529
6	0.002056	0.001292	65.6	-4.675
7	0.001960	0.001101	56.5	-2.383
8	0.002339	0.000935	41.2	-3.529
9	0.035340	-0.009514	25.2	-5.821
10	0.001944	0.000136	4.8	-3.529
11	0.001843	-0.000186	-13.5	-2.383
12	0.001662	-0.000563	-35.4	-3.529
13	0.001803	-0.000561	-34.7	-4.675
14	0.001691	-0.000671	-41.1	-3.529
15	0.032972	-0.000503	-50.7	-2.383
16	0.212870	-0.143140	-31.4	119.084
17	0.001635	-0.000736	-48.7	-2.383
18	0.001680	-0.000869	-54.4	-3.529
19	0.001906	-0.000583	-32.8	-3.529
20	0.001783	-0.000464	-27.7	-3.529
21	0.022608	-0.000073	0.1	116.792
22	0.001692	0.000135	6.1	-1.237
23	0.001811	0.000381	18.7	-2.383
24	0.001732	0.000719	39.9	-1.237
25	0.001919	0.001325	70.6	-2.383
26	0.002051	0.001518	75.3	-2.383
27	0.002020	0.001551	77.4	-2.383
28	0.002082	0.001592	78.1	-2.383
29	0.002030	0.001530	76.3	-3.529
30	0.002101	0.001486	71.0	-2.383
31	0.001868	0.001215	67.7	-3.529
32	0.002164	0.000895	42.2	-2.383
33	0.001573	0.000420	27.0	-2.383
34	0.001923	0.000247	12.0	-2.383
35	0.173250	-0.081397	-12.8	19.389
36	0.001654	-0.000441	-33.0	-3.529
37	0.001693	-0.000536	-35.2	-3.529
38	0.001629	-0.000579	-37.7	-3.529
39	0.181070	-0.094680	-41.1	-4.675
40	0.001802	-0.000838	-51.5	-4.675
41	0.001612	-0.000566	-35.1	-2.383
42	0.001720	-0.000521	-34.1	-3.529
43	0.022509	-0.000327	-2.6	116.792
44	0.022734	0.000109	2.0	116.792
45	0.001761	0.000309	13.9	-2.383
46	0.387830	-0.216110	-45.3	-133.018
47	0.001874	0.000922	49.0	-3.529
48	0.001737	0.001030	59.7	-2.383
49	0.001750	0.001090	64.4	-3.529
50	0.001700	0.001375	82.2	-3.529
51	0.001831	0.001472	81.5	-3.529
52	0.001920	0.001649	86.2	-3.529
53	0.001922	0.001488	78.0	-3.529
54	0.001856	0.001494	80.5	-2.383
55	0.201390	-0.143740	8.2	1.055
56	0.001773	0.000866	51.5	-3.529
57	0.001632	0.000763	50.3	-2.383
58	0.001569	0.000400	26.4	-3.529
59	0.001666	0.000056	5.0	-2.383
60	0.001735	-0.000090	-8.4	-2.383
61	0.001730	-0.000286	-15.9	-3.529
62	0.001512	-0.000223	-15.8	1.055
63	0.001645	-0.000397	-27.2	1.055
64	0.001572	-0.000673	-46.9	-1.237
65	0.022441	-0.000539	-4.8	-1.237
66	0.001726	-0.000420	-27.7	-1.237
67	0.001595	-0.000478	-33.3	0.000
68	0.001654	-0.000197	-16.6	-1.237
69	0.001855	0.000010	-2.6	-4.675
70	0.001519	0.000443	27.1	-1.237
71	0.001892	0.000734	39.3	-3.529
72	0.001713	0.000884	54.7	-3.529
73	0.001602	0.001079	68.1	-3.529
74	0.001712	0.001357	79.9	-3.529
75	0.001740	0.001498	86.4	-3.529
76	0.001711	0.001463	86.4	-2.383
77	0.001889	0.001529	82.2	-3.529
78	0.001782	0.001336	76.4	-2.383
79	0.001500	0.000888	61.5	-1.237
80	0.001494	0.000621	44.3	-3.529
81	0.001564	0.000560	37.2	-1.237
82	0.181810	-0.056167	-9.1	115.646
83	0.001492	-0.000182	-13.1	-3.529
84	0.001509	-0.000174	-6.8	-2.383
85	0.001569	-0.000149	-10.3	-3.529
86	0.001647	-0.000378	-24.0	-3.529
87	0.023403	-0.000287	-1.6	-122.704
88	0.001386	-0.000361	-32.2	-3.529
89	0.001598	0.000062	0.8	-3.529
90	0.001515	0.000171	10.6	-1.237
91	0.020494	-0.009522	23.8	-3.529
92	0.001449	0.000663	47.9	-2.383
93	0.001653	0.000967	60.2	-3.529
94	0.001665	0.001186	73.2	-3.529
95	0.001663	0.001285	78.0	-2.383
96	0.001601	0.001303	82.7	-3.529
97	0.001636	0.001322	82.1	-2.383
98	0.001866	0.001395	76.1	-3.529
99	0.001811	0.001193	68.6	-2.383
100	0.001684	0.000853	52.9	-3.529
101	0.146890	-0.000135	29.7	-2.383
102	0.001673	0.000219	16.5	-3.529
103	0.001289	0.000251	20.8	-2.383
104	0.001699	-0.000133	-6.3	-3.529
105	0.001449	-0.000071	-11.2	-2.383
106	0.001557	-0.000020	-0.8	-3.529
107	0.015217	0.006256	-2.2	-1.237
108	0.043223	-0.016369	19.5	-2.383
109	0.001627	0.000558	38.3	-3.529
110	0.001621	0.001002	63.3	-2.383
111	0.001561	0.000905	62.7	-2.383
112	0.001762	0.001117	66.3	-3.529
113	0.001782	0.001137	64.5	-3.529
114	0.001548	0.000968	64.5	-2.383
115	0.160250	-0.021090	24.7	-2.383
116	0.013867	0.008120	37.6	-3.529
117	0.001414	0.000107	6.9	-3.529
118	0.001302	0.000183	14.4	-2.383
119	0.001497	0.000147	9.8	-2.383
120	0.001540	0.000152	11.5	-3.529
121	0.001575	0.000455	33.2	-3.529
122	0.001581	0.000630	41.5	-3.529
123	0.001608	0.000946	62.5	-2.383
124	0.001476	0.000712	51.9	-1.237
125	0.191410	-0.078524	-7.7	-6.967
126	0.001365	0.000303	21.5	-2.383
127	0.001363	0.000643	49.9	-2.383
128	0.001522	0.000800	55.3	-1.237

Table A.4: *Experiment 1938 emitter detection: Metrics statistic over all TAS positions.*

Metric	Mean	Median	Std. dev.	Min	Max
Δr in m	0.01572	0.00175	0.05191	0.00129	0.38783
$\Delta r_{\text{TASaxis}}$ in m	-0.00223	0.00073	0.01984	-0.21611	0.00812
$\text{share}_{\text{TASaxis}}$ in %	23.6	27.0	36.4	-54.4	86.4
$\Delta\phi$ in $^\circ$	-3.71	-3.529	27.84	-133.018	119.084
$ \Delta\phi $ in $^\circ$	11.86	3.529	25.19	0.000	133.018

A.5 Appendix of Protocols

A.5.1 3D USCT II Performance Profile (2012)

USCT3Dv2Measurement (1 call, 239.228 sec)

Generated 21-Feb-2012 23:04:20 using cpu time.

M-function in file C:\Dokumente und Einstellungen\zapf\Desktop\USCT-DAQII-rep\USCT3Dv2\USCT3Dv2Measurement.m
[Copy to new window for comparing multiple runs]

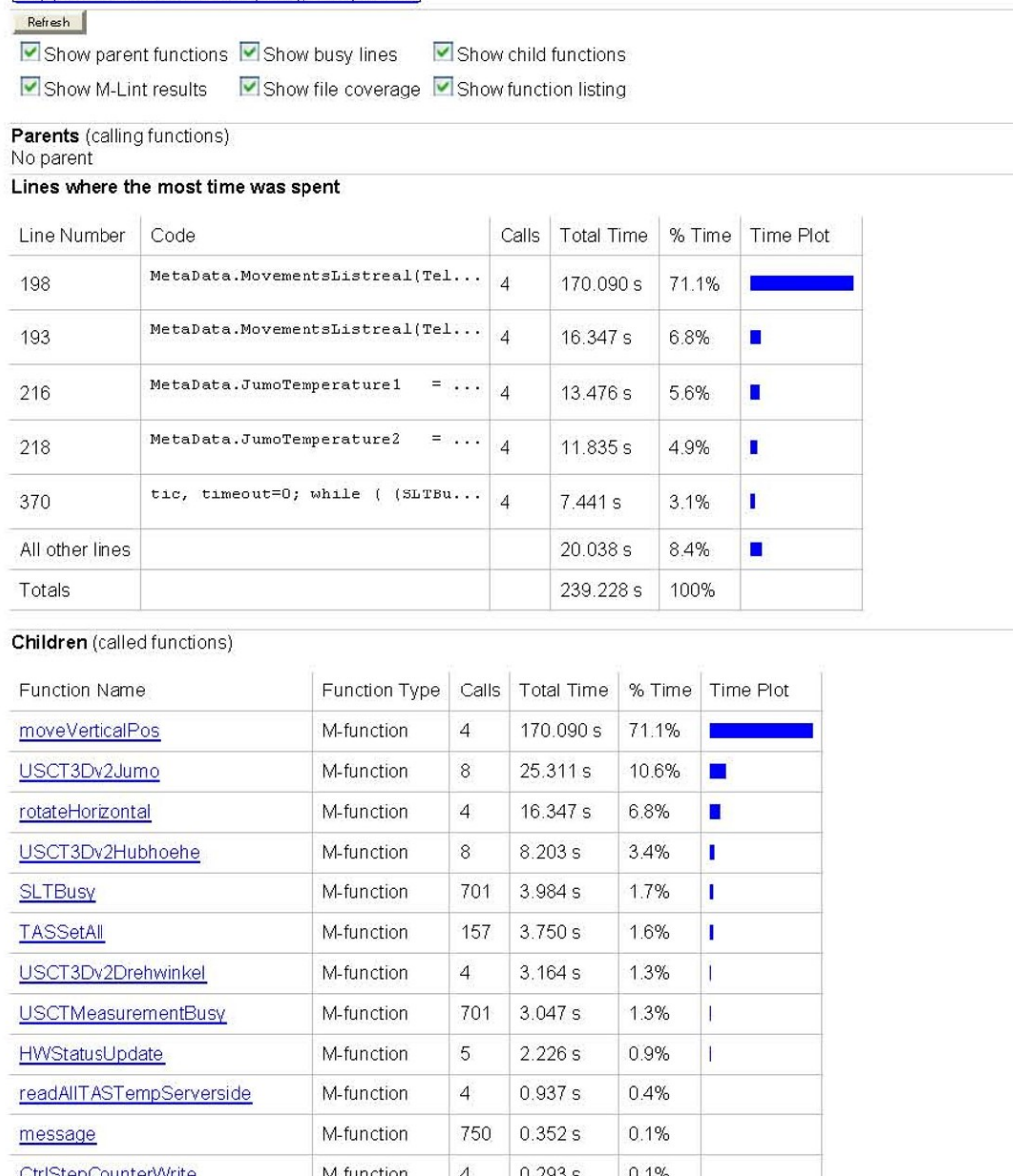


Figure A.7: Performance 3D USCT II: 240 seconds for standard measurement protocol with 4 mps, $\approx 80\%$ was defined by movement

A.6 Safety concept for 3D USCT III

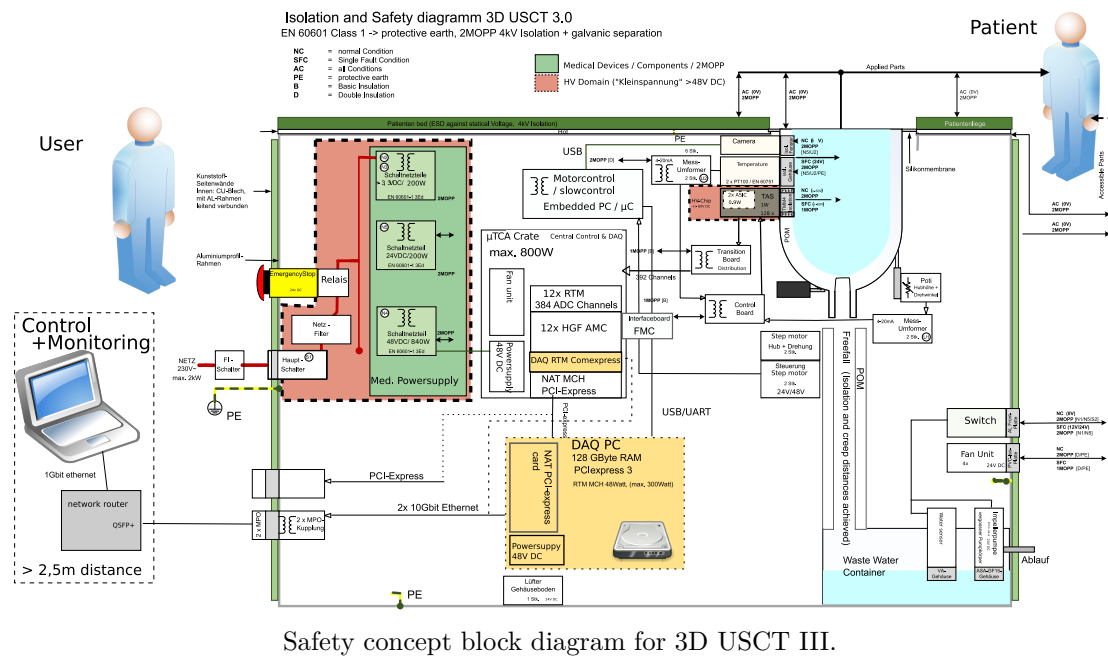
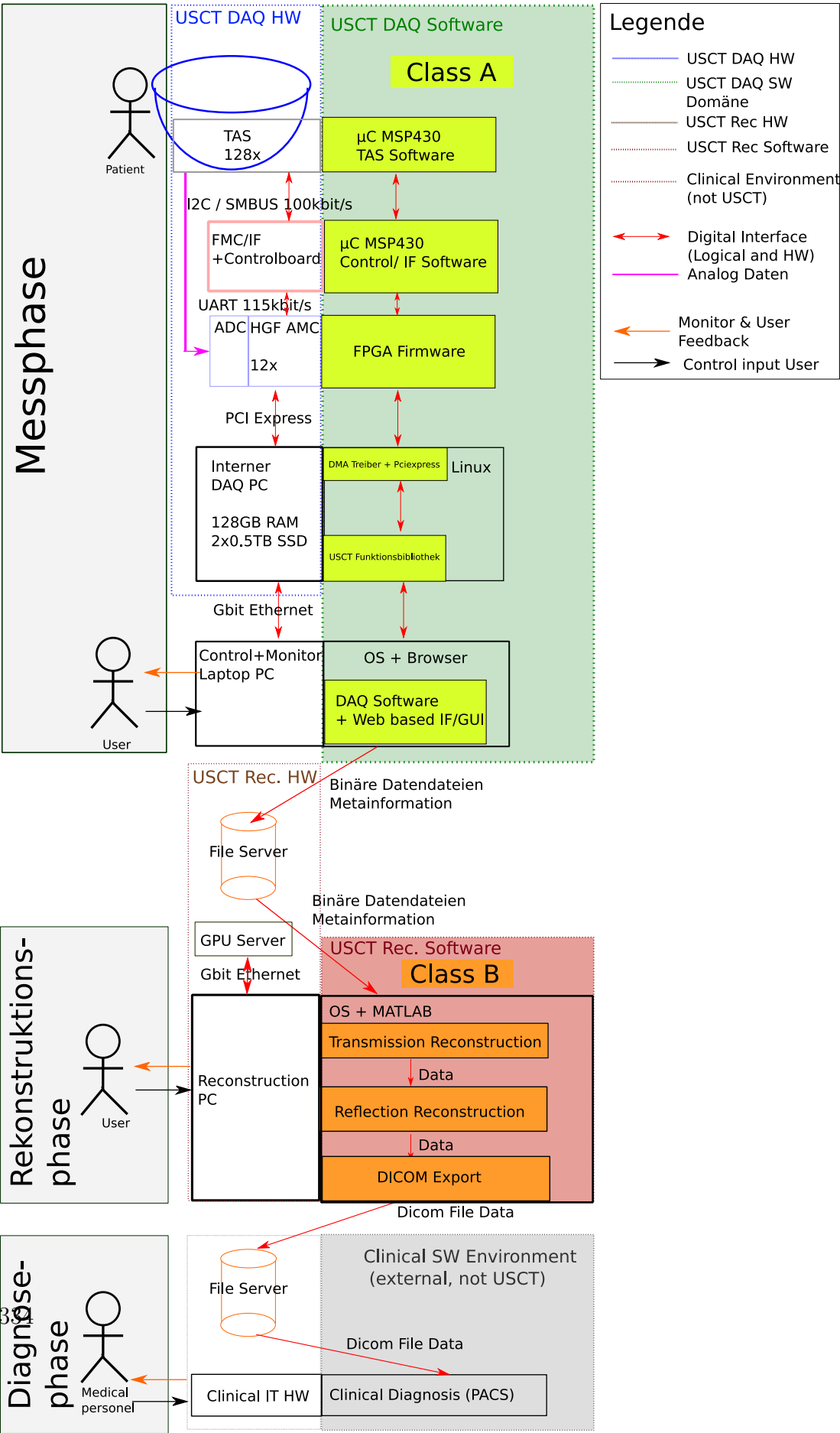


Figure A.8: *Safety concept block diagram for 3D USCT III.*

Software Architecture USCT 3.0



Software block diagram for 3D USCT III.

A.7 Timeline of 3D USCT III

Table A.5: *USCT III timeline, 2008–2012*

2008–2013	Simulation phase for USCT III concepts.
2010 Apr	USCT II Handbook finished.
2011 Mar	Georg G. leaves.
2011 Jul	TÜV Süd USCT II (part 1).
2011 Aug	TÜV Süd USCT II (part 2).
2011 Nov	TÜV Süd USCT II (part 3).
2012 Feb	Review of USCT II.
2012 Jul	Temperature analysis (USCT II).
2012 Aug	EMC measurement USCT II.
2012 Oct	Start of Jena pre-clinical trial.
2012 Nov	First USCT III proposal.

Table A.6: *USCT III timeline, 2013–2015*

2013 Jul	TAS 1.5 housings ordered.
2013 Sep	First TAS 1.5 prototypes.
2013 Nov	μ TCA DAQ proposed; IKTS collaboration started.
2013 Dec	Hygienic review at UMM and piezofibre concept.
2014 Jan	Aperture patent filed.
2014 Feb	ASIC9c brainstorming; TAS 1.5 electronics.
2014 Apr	USCT II Jena image review.
2014 Jul	ASIC9c definition.
2015 Apr	Coded excitation thesis.
2015 Jun	End of USCT II Mannheim trial.
2015 Sep	Octagon transducer prototyping.
2015 Oct	NEO Prize for USCT.

Table A.7: *USCT III timeline, 2016–2017*

2016 Jan	Crowdfunding completed.
2016 Apr	RTM assembled; mechanical design finalized.
2016 Jun	Bandwidth optimization.
2016 Sep	USCT III project review.
2016 Dec	DAQ/software completed; ASIC9c results.
2017 Jan	First PTPiesa piezofibre discs.
2017 Apr	Transducer 2.5 prototypes.
2017 Jul	ASIC9c characterization.
2017 Aug	Mechanical setup completed (USCT III.R).
2017 Oct	ASIC 9c testboard; transducer 2.5 prototypes.
2017 Dec	Multifibre prototypes; calibration thesis.

Table A.8: *USCT III timeline, 2018–2019*

2018 Jan	KIT–Realcan meeting Hangzhou.
2018 Jun	Neuland Prize.
2018 Jul	160 piezofibre discs delivered.
2018 Nov	Visit to Mannheim USCT II.
2019 Jan	Hangzhou meeting; Roberto’s PhD.
2019 Apr	EQ9 meeting at KIT; USCT II returns.
2019 Jun	Heinz retires.
2019 Oct	EQ9 builds first USCT III.R.
2019 Oct	First USCT III DAQ data.

Table A.9: *USCT III timeline, 2020–2021*

2020 May	ASIC18 ordered.
2020 Aug	USCT III hygienic concept.
2020 Oct	Transducer evaluation and build-up.
2021 Jan	FEB assembly and transfer to EQ9.
2021 Mar	TAS integrated into KIT USCT III.
2021 Jun	First full KIT USCT III measurement.
2021 Jul	First EQ9 USCT III measurement.
2021 Sep	ASIC18 evaluation.
2021 Dec	ASIC18 branch terminated.

Table A.10: *USCT III timeline, 2022–2024*

2022 Feb	QUSTOM/Pathfinder kickoff.
2022 Mar	EMC pre-compliance tests.
2022 Oct	USCT III BScan + metric generation.
2022 Nov	Transducer thesis (Angerer).
2023 Feb	Parallelization work.
2023 Apr	Synchronization visit EQ9 Hangzhou.
2023 Jun	ASICs delivered to EQ9.
2023 Sep	Movement system update; Camera/GUI integration.
2023 Nov	Commissioning EQ9 USCT III.
2024 Mar	Transport to Barcelona for QUSTOM study; hardware repair.
2024 Apr	Software hotfixes.
2024 May	QUSTOM study start; Zewei defense.
2024 Jul	Study finished; system returned.
2024 Oct	Medium evaluation; QUSTOM completed.
2024 Nov	EQ9 collaboration meeting in China.