

Motion-Robust Neonatal Coronary Imaging Using Photon-Counting CT

Optimizing Coronary Origin Visualization in
Transposition of the Great Arteries

Master Thesis

S.S.E. Stam

Motion-Robust Neonatal Coronary Imaging Using Photon-Counting CT

Optimizing Coronary Origin Visualization in
Transposition of the Great Arteries

by

S.S.E. Stam

to obtain the degree of Master of Science

at the Delft University of Technology,

to be defended publicly on Tuesday July 14, 2026 at 15:00.

Student number:	5876753
Project duration:	November 17, 2025 – July 14, 2026
Thesis committee:	Dr. ir. M.C. Goorden, TU Delft, supervisor
	Dr. ir. D. Lathouwers, TU Delft
	Dr. ir. D. Dodou, TU Delft
	Dr. ir. M. van Straten, Erasmus University Medical Center, supervisor
	T. van der Laan MSc, Erasmus University Medical Center, advisor

An electronic version of this thesis is available at <http://repository.tudelft.nl/>.

Preface

This thesis marks the completion of my graduation project and represents an important period of learning, persistence, and personal development. During this project, I had the opportunity to work at the intersection of technical imaging physics, clinical relevance, and quantitative image analysis. Investigating photon-counting CT for visualization of the coronary origins in neonates with transposition of the great arteries made me realize how closely technical choices in image acquisition and reconstruction are connected to clinical decision-making.

Throughout this project, I learned much more than how to design and perform a controlled phantom experiment. I learned how important it is to remain critical of quantitative results, especially when working with very small structures close to the limits of image resolution. The process required careful consideration of acquisition strategy, cardiac motion, reconstruction settings, image-quality metrics, and radiation dose. It also taught me that strong scientific conclusions are not built on a single result, but on the consistency between measurements, visual observations, physical reasoning, and clinical relevance.

This project was both challenging and rewarding. There were moments in which the data raised more questions than answers, and moments in which small technical details had a large influence on the interpretation of the results. Working through these challenges helped me to develop a deeper understanding of medical imaging research and taught me to approach scientific problems with patience, structure, and nuance.

I would like to express my sincere gratitude to Marcel, Thom, and Marlies for their guidance, feedback, and support throughout this project. Their expertise, critical questions, and encouragement helped me to improve the quality of this thesis and to better understand both the technical and clinical aspects of the research. I am especially grateful for the time they took to discuss the results and challenge my interpretation, which helped me to strengthen the final conclusions of this work.

I also want to thank Marcel Dijkshoorn for his valuable insights and the helpful discussions about acquisition and reconstruction options for neonatal coronary visualization. These discussions contributed greatly to my understanding of the practical choices involved in neonatal PCCT imaging. In addition, I would like to thank Marcel Segbers for kindly providing access to the `simple_orthanc` DICOM processing tool, which supported the data processing workflow used in this study.

Finally, I would like to thank my family, friends, roommates, and everyone who supported me during this period. Their encouragement, patience, and understanding have meant a great deal to me throughout the process of completing this thesis.

I hope that this work contributes to the further optimization of neonatal cardiac imaging using PCCT and provides a useful foundation for future studies focused on improving coronary-origin visualization in neonates with congenital heart disease.

*S.S.E. Stam
Delft, June 2026*

Summary

Introduction

Reliable visualization of the coronary origins is essential in neonates with transposition of the great arteries (TGA), because coronary anatomy directly influences surgical planning for the arterial switch operation. Photon-counting computed tomography (PCCT) offers high spatial resolution, spectral imaging, and potential dose efficiency, making it promising for neonatal coronary assessment. However, quantitative evidence guiding the choice of electrocardiogram (ECG) synchronized acquisition strategy, virtual monoenergetic imaging (VMI) level, and reconstruction phase remains limited. This study evaluated which PCCT acquisition and reconstruction strategy provides the most favorable balance between coronary origin visibility, motion robustness, and radiation dose under neonatal cardiac motion.

Methods

A neonatal coronary artery phantom was developed to assess small-vessel visibility under controlled cardiac motion. The phantom contained contrast-filled tubes of 1.1 and 2.0 mm diameter, oriented along three spatial directions in a polymethyl methacrylate background. Tube attenuation was selected to approximate iodine enhancement in large vessels in vivo. Cardiac motion was generated using an ECG-synchronized motion simulator at 120, 130, 140, and 150 bpm. Two ECG-synchronized PCCT strategies were evaluated: high-pitch flash acquisition and prospective sequential acquisition. Protocols were compared under matched conditions at 53 keV. Flash acquisitions were additionally reconstructed at 40 and 47 keV to assess the effect of VMI level, whereas prospective acquisitions were reconstructed at 53 keV across cardiac phases from 25% to 80% of the R–R interval to assess phase influence. Image quality was quantified using segmentation-derived metrics describing tube geometry, cross-sectional area, shape, edge definition, attenuation, spatial overlap and contrast-to-noise ratio (CNR). Results were evaluated with respect to acquisition strategy, reconstruction energy or phase, heart rate, tube orientation, z-position, and tube diameter. Radiation exposure was assessed using scanner-reported CT dose index volume, dose-length product and size-specific dose estimate.

Results

Under matched dynamic conditions at 53 keV, flash and prospective acquisition showed broadly comparable preservation of tube size, cross-sectional area, shape, edge definition, attenuation, and spatial overlap. The clearest matched protocol-related difference was observed for CNR, which was more favorable for flash acquisition. When the most favorable prospective reconstruction phase was selected, prospective acquisition improved overlap-based preservation, reflected by higher spatial overlap. Within the flash protocol, 40 keV provided the strongest iodine contrast and highest CNR, whereas 53 keV reduced equivalent diameter and cross-sectional area errors. Within the prospective protocol, reconstruction phase was the dominant optimization parameter, with later phases around 70–80% of the R–R interval generally more favorable than mid-cycle phases. Prospective acquisition resulted in substantially higher scanner-reported dose indices than flash. Heart-rate effects were non-monotonic and depended on acquisition timing, reconstruction phase, tube orientation and z-position. The 1.1 mm tubes represented the most demanding condition, with greater proportional errors, reduced overlap, lower CNR and higher segmentation instability.

Conclusion

High-pitch ECG-triggered flash PCCT is supported as the preferred practical acquisition protocol for neonatal coronary origin visualization in neonatal with TGA. For the primary visual task, 40 keV VMI is preferred because it provided the strongest iodine contrast and highest CNR, while 53 keV should be retained as a complementary reconstruction when geometric preservation or size-based assessment is important. Prospective acquisition may be reserved for selected cases in which phase-resolved information is expected to improve interpretation sufficiently to justify the additional dose and complexity.

Contents

Preface	i
Summary	ii
Nomenclature	vi
1 Introduction	1
1.1 Clinical context: transposition of the great arteries in neonates	1
1.2 Technical background	1
1.2.1 Photon counting CT	1
1.2.2 Virtual monoenergetic imaging	3
1.2.3 ECG-synchronized cardiac acquisition	3
1.2.4 Motion and image quality in neonatal cardiac PCCT	5
1.3 Problem statement	5
1.4 Study objective	6
1.5 Research questions and hypotheses	6
1.6 Scope and thesis outline	7
2 Methods	8
2.1 Phantom development	8
2.1.1 Design requirements	8
2.1.2 Coronary artery phantom design and materials	8
2.1.3 Contrast preparation and dilution	8
2.2 Experimental setup	9
2.2.1 Phantom setup in the motion platform	9
2.2.2 Motion generation and ECG synchronization	9
2.3 PCCT acquisition protocols and image reconstruction	11
2.3.1 General scanner and reconstruction settings	11
2.3.2 High-pitch flash ECG-triggered protocol	12
2.3.3 Prospective sequential ECG-gated protocol	12
2.4 Image analysis	13
2.4.1 Comparison framework	13
2.4.2 Workflow	13
2.4.3 Quantitative image quality	14
2.5 Radiation dose assessment	18
2.6 Statistical analysis	18
2.6.1 Endpoints and general approach	18
2.6.2 Descriptive analysis	18
2.6.3 Inferential analysis	19
2.6.4 Interpretation of endpoints	19
2.6.5 Post-hoc comparison	20
3 Results	21
3.1 Validation of the phantom and experimental setup	21
3.1.1 Visualization of the scans	21
3.1.2 Visual validation of segmentation and registration	22
3.1.3 Repeatability of the experimental setup	25
3.2 Quantitative image analysis	29
3.2.1 Static inter-protocol comparison	29
3.2.2 Cardiac motion inter-protocol comparison	30
3.2.3 Within-protocol motion-robustness analysis: flash	33

3.2.4	Within-protocol motion-robustness analysis: prospective	35
3.2.5	Post-hoc comparisons	37
3.3	Radiation dose results	38
4	Discussion	40
4.1	Summary of main findings	40
4.2	Interpretation of the repeatability analysis	40
4.3	Interpretation of the static inter-protocol comparison	41
4.4	Interpretation of the dynamic inter-protocol comparison	42
4.4.1	Z-matched phase inter-protocol comparison	42
4.4.2	Best-phase inter-protocol comparison	43
4.5	Interpretation of the within-protocol analyses	44
4.5.1	Flash within-protocol motion-robustness analysis	44
4.5.2	Prospective within-protocol motion-robustness analysis	44
4.6	Dose and protocol trade-off	45
4.7	Protocol recommendation	45
4.8	Limitations	47
5	Conclusion and future direction	48
	References	49
A	Literature Review	53
B	Phantom design	68
C	Cardiac motion profiles generation	69
D	Segmentation and registration code	73
D.1	Selected MATLAB functions used for image analysis	73
D.1.1	Recursive DICOM import and metadata extraction	73
D.1.2	Slice position calculation	73
D.1.3	HU-based volume construction	74
D.1.4	Initial tube mask generation	74
D.1.5	Centroid-based alignment	74
D.1.6	DSC-based overlap analysis	74
D.1.7	Cross-sectional metric extraction	75
D.1.8	PMMA-based contrast reference	75
E	Statistical analysis	76
E.1	Selected MATLAB functions used for statistical analysis	76
E.1.1	Reading and cleaning workbook sheets	76
E.1.2	Standardization of variable names	76
E.1.3	Preparation of categorical analysis variables	77
E.1.4	Derivation of tube-specific grouping variables	77
E.1.5	Selection of comparison-specific subsets	77
E.1.6	Definition of the analyzed metrics	77
E.1.7	Descriptive statistical summary	78
E.1.8	Mixed-effects modelling	78
E.1.9	Model structure for the dynamic analyses	78
E.1.10	Model structure for the static baseline analysis	79
E.1.11	Model estimability checks	79
E.1.12	Extraction of coefficient estimates and omnibus tests	79
F	Model diagnostics	81
F.1	Residual diagnostics of representative mixed-effects models	81
F.2	Observed-versus-fitted assessment	81
F.3	Exploratory assessment of quadratic terms	87
G	Results	89
G.1	Visualization of the scans	89

G.2	Repeatability and quality control	90
G.2.1	Prospective repeated measurements	90
G.2.2	Additional repeatability plots	92
G.2.3	Prospective repeatability	93
G.2.4	Prospective repeatability: coefficient of variation	95
G.2.5	Phase-resolved prospective repeatability coefficients	97
G.3	Additional quantitative data plots	97
G.3.1	Static inter-protocol analysis	98
G.3.2	Z-matched phase inter-protocol analysis	98
G.3.3	Best phase inter-protocol analysis	99
G.3.4	Flash within-protocol analysis	100
G.3.5	Prospective within-protocol analysis	100
G.4	Post-hoc analysis	101

Nomenclature

Abbreviations

Abbreviation	Definition
ALARA	As Low As Reasonably Achievable
ASO	Arterial Switch Operation
CARE	Combined Applications to Reduce Exposure
CHD	Congenital Heart Disease
CT	Computed Tomography
DICOM	Digital Imaging and Communications in Medicine
ECG	Electrocardiogram
EID-CT	Energy-Integrating Detector CT
HU	Hounsfield Unit
IQ	Image Quality
PCCT	Photon-Counting CT
PMMA	Polymethyl Methacrylate
QIR	Quantum Iterative Reconstruction
ROI	Region of Interest
TGA	Transposition of the Great Arteries
VMI	Virtual Monoenergetic Image

Symbols

Symbol	Definition	Unit
A	Cross-sectional area	mm^2
A_{ref}	Reference cross-sectional area	mm^2
B_{HU}	Attenuation bias	HU
C	Circularity	–
CNR	Contrast-to-noise ratio	–
$CTDI_{\text{vol}}$	Volume CT dose index	mGy
CV	Coefficient of variation	%
D_{eq}	Equivalent diameter	mm
$D_{\text{eq,ref}}$	Reference equivalent diameter	mm
D_w	Water-equivalent diameter	cm
DLP	Dose-length product	mGy·cm
DSC	Dice similarity coefficient	–
E_A	Cross-sectional area error	mm^2
$E_{D_{\text{eq}}}$	Equivalent diameter error	mm
ERD	Edge rise distance	mm
L_{eff}	Effective scan length	cm
RC	Repeatability coefficient	metric-dependent
$SSDE$	Size-specific dose estimate	mGy
μ	Mean	metric-dependent
σ	Standard deviation	metric-dependent

Introduction

1.1. Clinical context: transposition of the great arteries in neonates

Congenital heart disease (CHD) is a major contributor to neonatal mortality, and high-quality cardiac imaging is required to support clinical decision-making in early stages [1]. Echocardiography is the primary imaging modality; however, image quality depends on the availability of suitable acoustic windows and may be limited when assessing very small or complex structures, such as the coronary arteries. Therefore, cross-sectional imaging plays an important role in the evaluation of neonatal CHD [2]. Cardiac computed tomography (CT) has become an established technique for the evaluation of congenital cardiovascular anatomy because it provides a high-resolution, three-dimensional depiction of cardiovascular anatomy in very short acquisition times. This rapid volumetric imaging is valuable when magnetic resonance imaging is not feasible due to longer scan times, monitoring requirements, or limited patient stability [3].

Among the critical forms of CHD, transposition of the great arteries (TGA) is a life-threatening defect characterized by the anatomical switch of the aorta and the pulmonary artery. Treatment is performed early in life using the arterial switch operation (ASO), shown in Figure 1.1 [4]. During the ASO, the great arteries are reconnected to the appropriate ventricles and the coronary arteries are transferred. This coronary transfer is a critical step, because coronary artery variants are relatively common in TGA. Examples include both coronary arteries arising from the same aortic sinus or a single coronary artery pattern, which can influence operative strategy, surgical risk, and postoperative outcomes [5]. Accurate preoperative identification of the coronary origins and proximal course is therefore essential, and imaging must provide sufficient diagnostic confidence at the level of the coronary origins [6].

In neonates with TGA, the diagnostic task is therefore not general cardiac visualization, but reliable identification of the coronary origins and proximal coronary course before surgical repair. This task is technically demanding because the coronary arteries are small, move rapidly, and must be visualized while radiation exposure must be kept as low as reasonably achievable (ALARA).

1.2. Technical background

1.2.1. Photon counting CT

Recent advances introduced photon-counting CT (PCCT), which provides intrinsic spectral information along with higher spatial resolution and improved dose efficiency compared to conventional energy-integrating detector CT (EID-CT) [8].

Photon-counting CT differs from conventional EID-CT in the way X-ray photons are detected and processed. A detailed explanation of the detector principle and related technical aspects is described in the literature review, provided in Appendix A. In brief, photon-counting detectors register individual photons and classify these events into energy bins based on pulse height, which is related to photon energy [9]. As a result, spectral information is available in each acquisition, allowing reconstruction of energy-dependent image types such as virtual monoenergetic images (VMIs).

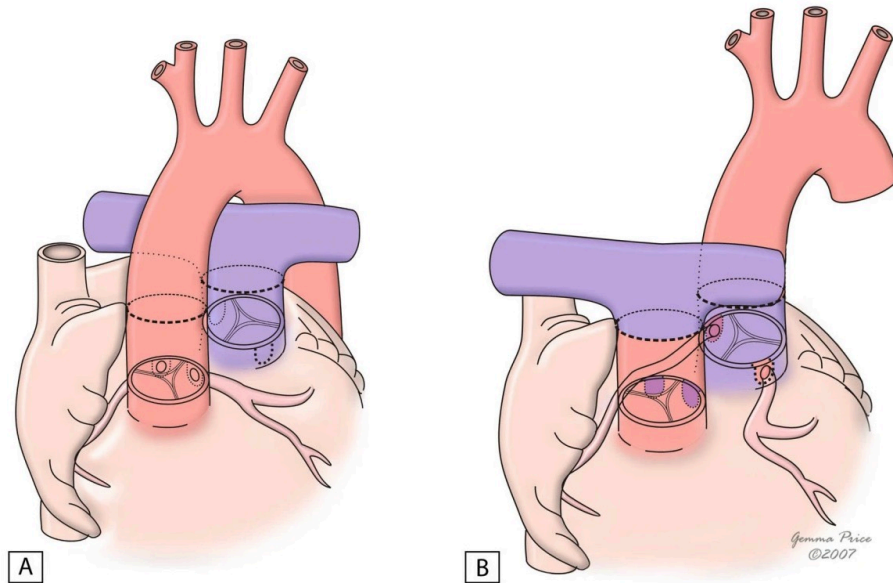


Figure 1.1: Schematic illustration of TGA and the surgical concept of the arterial switch operation. (A) With the arterial switch operation, the great arteries are reconnected to their correct ventricles. The preoperatively indicated coronary arteries are implanted into the replaced aorta. (B) In neonates with TGA, the aorta (pink) originates from the right ventricle and the pulmonary artery (purple) from the left ventricle, resulting in a parallel instead of serial circulation. The coronary arteries originate from different sinuses of the aortic root [7].

This detector-level energy discrimination further distinguishes PCCT from dual-source CT, in which spectral information is obtained by acquiring or combining data from two different X-ray sources. Although dual-source CT also enables material differentiation and VMI reconstruction, it remains based on energy-integrating detector measurements [10]. In PCCT, energy information is obtained directly at the detector level by sorting individual photons into predefined energy bins.

From an image-quality perspective, these characteristics are relevant for neonatal coronary imaging. In contrast-enhanced angiographic imaging, iodine is used as the intravascular contrast agent. Because iodine attenuation is energy dependent, spectral separation can be used to increase the contribution of lower-energy photons, thereby improving iodine contrast and vascular visibility [9]. This is especially important in neonates, where clinically relevant structures such as the coronary arteries are very small and may approach the spatial resolution limits of CT.

The depiction of such small, high-contrast structures depends not only on contrast and noise, but also on spatial resolution, edge sharpness, and partial-volume effects. Partial-volume effects occur when a voxel contains both contrast-filled blood and surrounding tissue, causing the measured attenuation to represent an average of these tissues. This can reduce vessel visibility, blur vessel boundaries, and affect measurements of vessel size and shape. PCCT may offer advantages in this respect because its small detector pixel size and direct-conversion detector design reduce detector-related blurring compared with conventional EID-CT. However, spatial resolution is not determined by detector design alone, but is also influenced by focal spot size, reconstruction kernel, slice thickness, matrix size, and image noise.

Neonatal cardiac PCCT protocol optimization, therefore, requires a balance between iodine contrast, spatial resolution, image noise, motion sensitivity, and radiation dose. Because neonatal imaging is performed under strict dose constraints, improvements in coronary visualization must be achieved without unnecessarily increasing radiation exposure [11]. In this context, PCCT combined with appropriate electrocardiogram (ECG) synchronization and spectral reconstruction settings provides a promising framework for optimizing coronary artery visualization in neonates.

1.2.2. Virtual monoenergetic imaging

Virtual monoenergetic imaging (VMI) reconstructs images that correspond to a specified photon energy (keV), simulating the attenuation of a monoenergetic X-ray beam. In practice, VMIs are derived from spectral information by, post-acquisition, combining energy-resolved measurements to produce images with effective attenuation properties corresponding to a chosen energy, without requiring multiple scans [9]. VMI energy selection allows retrospective optimization of image contrast and noise characteristics and therefore directly impacts task performance.

In PCCT, keV selection directly affects vessel contrast because iodine attenuation is energy-dependent. Lower-keV VMIs generally increase iodine attenuation and can therefore increase contrast relative to the surrounding tissue, which is advantageous for delineating small structures [12]. This effect is illustrated in Figure 1.2, which shows cardiac PCCT images reconstructed at increasing VMI energy levels. The iodine-enhanced coronary lumen and cardiac chambers show high signal intensity at low keV, with a progressive reduction in iodine signal and vascular conspicuity at higher-keV levels.

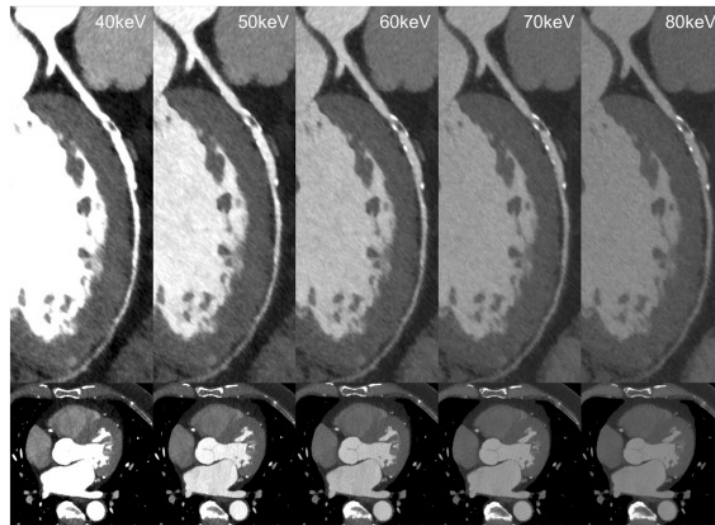


Figure 1.2: Effect of VMI energy selection on iodine-enhanced cardiac PCCT images acquired at 120 kVp. The lower row shows axial cross-sectional images, and the upper row shows corresponding coronary artery reconstructions from 40 to 80 keV. At lower-VMI energies, iodine attenuation is higher, producing greater contrast within the coronary lumen and cardiac chambers. At higher-VMI energies, the iodine signal decreases, reducing vascular contrast [13].

The higher attenuation at a lower keV is related to the increased photoelectric absorption near the iodine K-edge, where photon energies approach the binding energy of the inner electron shell [14, 15]. However, lower-keV reconstructions can also show increased noise because fewer photons contribute effectively to the reconstructed image and because of spectral processing. On the other hand, higher-keV VMIs reduce iodine contrast due to the reduced photoelectric effect when the VMI level is significantly above the K-edge of iodine, while decreasing noise and mitigating spectral artifacts such as beam hardening [12, 14]. Therefore, the optimal VMI energy for cardiac visualization is a balance between iodine-related contrast and noise.

1.2.3. ECG-synchronized cardiac acquisition

Cardiac PCCT strategies reduce motion artifacts by synchronizing data acquisition to the cardiac cycle, using the ECG as a timing reference. With ECG-synchronized imaging, acquisition is timed relative to the R–R interval to sample periods with relatively reduced cardiac motion [16]. The cardiac phase is commonly expressed as a percentage or time instance of the R–R interval, enabling phase targeting across different heart rates. In this context, ECG triggering refers to using the ECG signal to determine the moment at which a scan is started, whereas ECG gating refers to data acquisition at a predefined phase window of the cardiac cycle. Figure 1.4 provides an overview of the cardiac cycle in relation to the ECG signal.

Two ECG-synchronized acquisition approaches are central to this study: high-pitch flash and prospective sequential acquisition (Figure 1.3). In a high-pitch flash acquisition, the acquisition time is very short, allowing the cardiac volume to be covered within a limited time window of approximately 66 ms. Because the table moves continuously through the scanner during this acquisition, different positions along the scan direction, referred to as the z-direction, are acquired at slightly different times within the cardiac cycle. As a result, the effective cardiac phase depends on the z-position within the scanned volume. This reduces susceptibility to motion because the data for a given anatomical region are acquired over a short time interval [16, 17].

In prospective sequential acquisition, data are acquired in discrete slabs that are gated at predefined cardiac phases. The cardiac volume is therefore built up over multiple ECG-gated acquisitions, with each slab targeted to a selected phase of the R–R interval. This approach can potentially improve small-vessel depiction by targeting phases of the cardiac cycle with relatively reduced motion.

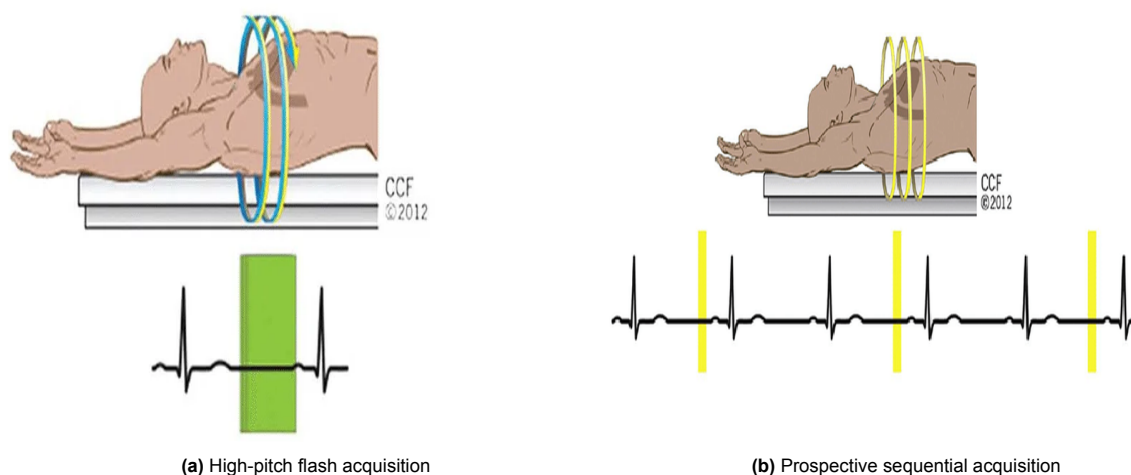


Figure 1.3: Schematic illustration of the two ECG-synchronized cardiac PCCT acquisition approaches used in this study [18]. (a) In high-pitch flash acquisition, the ECG-trigger initiates one short continuous helical scan during a predefined acquisition window. As the table moves continuously through the scanner, different z-positions of the heart are acquired at slightly different moments within the same cardiac cycle. (b) In prospective sequential acquisition, the cardiac volume is acquired in separate axial slabs over multiple heartbeats. Each slab is acquired during a predefined ECG-gated acquisition window, while table movement occurs between successive slabs.

In neonatal cardiac imaging, selection of the cardiac phase is important because coronary image quality depends strongly on the amount of cardiac motion during data acquisition. The cardiac cycle is commonly described using the R–R interval, defined as the time between two consecutive R-peaks on the ECG. The R-peak corresponds to the tall peak of the ECG signal and is commonly used as the temporal reference point for ECG-gated cardiac imaging. One R–R interval therefore represents one complete heartbeat, with cardiac phases expressed as percentages of this interval, from 0% at the R-peak to 100% just before the next R-peak.

Physiologically, the cardiac cycle alternates between systole and diastole. Systole is the contraction phase, during which the ventricles eject blood from the heart, whereas diastole is the relaxation and filling phase of the ventricles. These ECG-related phases are schematically shown in Figure 1.4. Because coronary arteries move throughout the cardiac cycle, image quality depends on whether data acquisition coincides with a relatively low-motion period. This is particularly relevant in neonates, where high heart rates shorten the R–R interval and reduce the time available for motion-robust imaging.

This timing directly affects the two acquisition approaches used in this study. In prospective sequential acquisitions, the target phase or reconstruction phase can be selected within the available ECG-gated acquisition window. In the Flash protocol, phase control is more constrained and is mainly determined by the ECG-trigger timing and the z-position of the acquisition window.

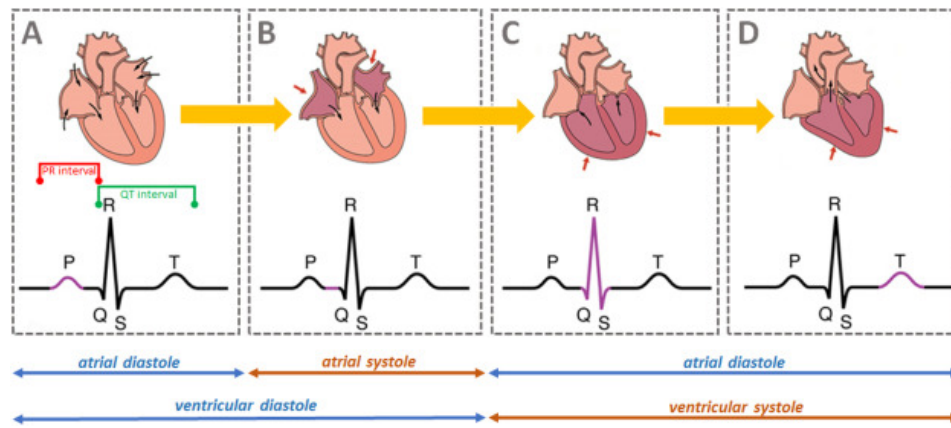


Figure 1.4: Schematic overview of the cardiac cycle in relation to the ECG signal [19]. The R–R interval, defined as the time between two consecutive R-peaks, represents one complete heartbeat and is used to express cardiac phases as percentages of the cardiac cycle. Systole refers to ventricular contraction and blood ejection, whereas diastole refers to ventricular relaxation and filling.

1.2.4. Motion and image quality in neonatal cardiac PCCT

Motion during CT acquisition degrades image quality through mechanisms that differ from quantum noise. Quantum noise refers to random image noise arising from statistical fluctuations in the number of detected X-ray photons. In contrast, for periodic cardiac motion, temporal inconsistencies between projections can produce edge blurring, duplication or deformation of small structures, and streak-like artifacts. These motion-related effects reduce effective spatial resolution and can introduce a structured, non-stationary signal that may be perceived as noise by human observers.

From a frequency-domain perspective, motion can be interpreted as a reduction of higher spatial frequencies, thereby reducing edge sharpness and small-structure visibility. In addition, motion-induced artifacts are often anisotropic and non-stationary, meaning that simple white-noise assumptions and scalar noise measures may not fully capture their impact on interpretability. Because ECG-synchronized acquisition strategies sample the cardiac cycle differently, the severity and appearance of motion-induced degradation are protocol dependent.

Motion may also influence the visual effect of energy selection in spectral reconstructions. Lower-keV VMI settings increase iodine-related signal, but they may also make motion-related blurring or artifacts more apparent because the contrast of iodine-enhanced structures is higher. Therefore, the benefit of low-keV reconstruction under dynamic conditions depends not only on iodine contrast, but also on the extent to which motion affects vessel sharpness and visibility [12, 20].

In this study, coronary origin visibility was treated as a task-based image-quality problem. Accurate visualization requires not only high attenuation but also preservation of vessel continuity, size, cross-sectional shape, edge definition, and spatial agreement with the expected vessel position. These different aspects of visibility may respond differently to acquisition mode, VMI level, heart rate, and reconstruction phase.

1.3. Problem statement

After consultation with a cardiovascular radiologist, the main diagnostic challenge was identified as visualization of the coronary origins. Reliable visualization of the coronary origin is a diagnostic requirement in neonates with TGA, since coronary abnormalities can directly affect surgical planning and outcome.

Despite the theoretical advantages of PCCT for small contrast-enhanced vessels, quantitative evidence guiding protocol and reconstruction choices for neonatal coronary assessment remains limited. In particular, it is unclear how ECG-synchronized high-pitch flash and prospective sequential acquisitions compare under neonatal heart-rate conditions, how the reconstruction phase influences prospective imaging performance, and how VMI energy affects motion robustness.

The current evidence base provides limited quantitative recommendations for neonatal cardiac imaging, as further described in the literature review in Appendix A. Neonatal outcomes are often not reported separately, image quality is commonly assessed using subjective scoring, and motion robustness is frequently inferred from acquisition choices rather than quantified directly. Although geometric image-quality metrics such as equivalent diameter, cross-sectional area, and circularity have been used in previous studies, these are not consistently applied as task-based metrics [21, 22].

Consequently, there is insufficient quantitative evidence to support the selection of PCCT protocols and reconstruction settings for coronary assessment in neonates with TGA. Specifically, uncertainty remains regarding (i) whether high-pitch flash or prospective sequential acquisition provides more robust coronary origin visibility across neonatal heart rates, (ii) which VMI level maximizes coronary contrast for each protocol in the presence of cardiac motion, and (iii) for prospective imaging, which cardiac phase yields optimal coronary visualization and how this depends on heart rate.

1.4. Study objective

This study aimed to identify the combination of photon-counting CT acquisition and reconstruction parameters that provided the best coronary origin visibility in neonates while being most robust to cardiac motion.

1.5. Research questions and hypotheses

This thesis addresses the following research questions:

- **Main research question:** Which ECG-synchronized PCCT acquisition strategy provides the best balance between coronary origin visualization and radiation dose in neonates with TGA across clinically relevant heart rates and VMI levels? Within the prospective protocol, which cardiac phase provides the most clinically relevant coronary origin visualization?
- **Subquestion 1:** How does the visibility of the coronary origin change with increasing heart rate for flash versus prospective acquisition, using matched acquisition and reconstruction settings where possible?
- **Subquestion 2:** Which VMI level is most favorable for motion robustness of the coronary origin for the flash protocol, and does the optimal VMI level change with increasing heart rate?
- **Subquestion 3:** Which cardiac phase(s) within the prospective protocol provide the best visibility of the coronary origin at different heart rates?

Based on the literature review and physical considerations, it was hypothesized that prospective ECG-gated PCCT would improve coronary visualization through phase selection, but that this benefit would need to be balanced against dose and acquisition complexity. Within this protocol, image quality was expected to vary across cardiac phases, with reduced motion phases yielding the best visibility of the coronary origins.

In addition, coronary origin visibility was expected to decrease with increasing heart rate for both protocols, due to increased motion. This decline was hypothesized to be more pronounced for the prospective protocol, whereas flash acquisition was expected to be relatively more motion robust at higher heart rates due to the short acquisition time.

It was further hypothesized that VMI level would influence the balance between iodine contrast and apparent size accuracy. Lower-keV reconstructions were expected to increase iodine attenuation and improve contrast-related visibility of the vessel lumen. At the same time, the increased iodine signal may enhance apparent blooming at the vessel boundary. Higher-keV reconstructions were therefore expected to reduce iodine-related blooming, while the spatial resolution of the reconstruction remains unchanged.

1.6. Scope and thesis outline

This study focuses on PCCT protocol optimization for visualization of the coronary arteries, including the coronary origins, in a neonatal TGA imaging context. This phantom-based study was designed to isolate the effects of protocol type, heart rate, reconstruction energy, and cardiac phase selection under controlled conditions. The scope includes objective image-quality measurements alongside commonly reported dose indices.

The remainder of this thesis is organized as follows. Chapter 2 describes the phantom design and setup, acquisition and reconstruction parameters, the objective data extraction and statistical analysis methods. Chapter 3 presents the results, followed by interpretation and implications in Chapter 4. Chapter 5 concludes with answers to the research questions and a protocol recommendation for neonatal coronary PCCT imaging.

2

Methods

2.1. Phantom development

2.1.1. Design requirements

For this study, coronary origin visibility was operationally defined as accurate visibility of the contrast-filled tube combined with preservation of tube geometry. Based on this definition, a dedicated neonatal coronary artery phantom was developed to allow reproducible assessment of small-vessel visibility under simulated neonatal cardiac motion. Requirements for the complete phantom included: (i) neonatal coronary-sized tubes to ensure clinically partial-volume effects and spatial resolution demands; (ii) background attenuation approximating clinical soft-tissue environment; (iii) tubes with stable and repeatable iodine concentration; (iv) periodic and controllable representative cardiac motion; (v) ECG synchronization; and (vi) mechanical repeatability to minimize scan-to-scan variability.

At Erasmus University Medical Center (Erasmus MC), a cardiac motion simulator (Sim4D-VL, QRM GmbH, Germany) was available and was used to fulfill requirements (iv), (v), and (vi). The remaining requirements were addressed through the development of the neonatal coronary artery phantom.

2.1.2. Coronary artery phantom design and materials

The phantom contained six cylindrical tubes representing coronary arteries, arranged in three orthogonal orientations. Neonatal coronary artery diameters are typically in the range of 1–2 mm [23, 24]; therefore, the phantom was designed to represent this clinically relevant size range. Three tubes with an inner diameter of 1.1 mm and three tubes with an inner diameter of 2.0 mm were oriented along the x-, y-, and z-axes. The final design of the coronary artery phantom, including the mounting material for the cardiac motion simulator, is shown in Appendix B.

To mimic the attenuation of the surrounding soft tissues, the phantom was constructed from polymethyl methacrylate (PMMA), a material with an expected CT number at 53 keV in the range of cardiac soft tissue, approximately between 70 and 100 HU [25, 26]. The phantom was manufactured by the Department of Experimental Medical Instrumentation at Erasmus MC.

2.1.3. Contrast preparation and dilution

The tube filling was prepared to approximate the vessel-to-background attenuation difference observed in neonatal coronary PCCT scans. A clinical reference was derived from six neonatal patient scans acquired at Erasmus MC, all reconstructed at the a virtual monoenergetic level of 53 keV. These scans were performed for clinical indications requiring coronary artery visibility, including assessment of a suspected fistula, evaluation of coronary anatomy, and insufficient perfusion of the left coronary artery.

Because direct attenuation measurements in the coronary arteries are strongly affected by partial-volume averaging and limited spatial resolution, these do not provide a reliable estimate of iodine attenuation alone. The aorta was therefore used as a larger vascular reference structure to estimate the amount of iodine needed in the vessels.

The median [interquartile range] attenuation difference between the aorta and surrounding cardiac soft tissue was used to define the clinically representative vessel-to-background contrast. Across the reference scans, the median aortic attenuation was 447 [160.24] HU and the median surrounding cardiac soft-tissue attenuation was 48 [14] HU, resulting in a median vessel-to-background attenuation difference of 408 [157.25] HU.

In the phantom, the surrounding PMMA background had a mean attenuation of 92.9 ± 22 HU (mean $\pm$ standard deviation) at 53 keV. Based on the clinical vessel-to-background difference, the target attenuation of the diluted contrast agent was therefore approximately 500 HU. The initial iodine dilution was estimated based on the linear relationship between iodine concentration and CT attenuation previously reported in the literature [27]. Using the reported attenuation of approximately 800 HU for 15 mg I/mL iodine at 53 keV, this resulted in the following estimate:

$$C_{\text{target}} = 15 \cdot \frac{HU_{\text{target}}}{800}, \quad (2.1)$$

where C_{target} is the estimated iodine concentration in mg I/mL, and HU_{target} is the required tube attenuation at 53 keV.

The calculated concentration only provided an initial estimate, therefore preliminary phantom scans were performed to verify the achieved attenuation experimentally. Two test dilutions resulted in a mean attenuation of approximately 135 HU and 443 HU at 53 keV. Linear interpolation between these measurements was used to prepare an additional dilution, which resulted in a mean attenuation of approximately 523 HU. Because this exceeded the target attenuation of 500 HU and the 443 HU dilution remained within the range observed in the neonatal reference scans, the dilution resulting in approximately 443 HU was selected for the subsequent experiments.

2.2. Experimental setup

2.2.1. Phantom setup in the motion platform

The coronary artery phantom was positioned in a water tank and mounted onto the QRM motion simulator for all dynamic experiments. The total phantom setup, including the PMMA body and surrounding water, had dimensions representative of a neonatal thoracic imaging condition. The water tank was used to approximate the attenuation contribution of surrounding neonatal chest tissues, so that the iodine-filled tubes were imaged within a realistic attenuation environment. The simulator was controlled using a dedicated workstation, on which the predefined motion profiles (Section 2.2.2) were uploaded to generate the desired cardiac motion patterns during scanning. The phantom included mounting material for attachment to a mechanical arm that generated lateral displacement. A representation of the experimental setup is shown in Figure 2.1.

2.2.2. Motion generation and ECG synchronization

Cardiac motion was generated using the QRM cardiac motion simulator, which produced the displacement of the phantom in three dimensions. Neonatal heart rates typically range from 120 to 150 beats per minute [28], therefore four heart rate settings were simulated: 120, 130, 140, and 150 bpm. The 120 bpm motion profile, obtained from previous research, was used as the reference trajectory for generating the higher heart rate motion profiles [29].

The motion profiles for various heart rates were generated in MATLAB by temporally rescaling the 120 bpm reference profile. The reference profile had a cardiac cycle duration of 500 ms and contained discrete displacement coordinates in the x-, y-, and z-directions.

Only translational displacement in the x-, y-, and z-directions was included in the motion model, consistent with the previous study from which the reference trajectory was derived. Although the heart and aortic root may exhibit complex twisting motion during systole, rotational motion was not included. This simplification was considered acceptable because the clinical task was visual identification of the coronary origins, for which translational displacement was assumed to be the dominant motion component affecting visibility.

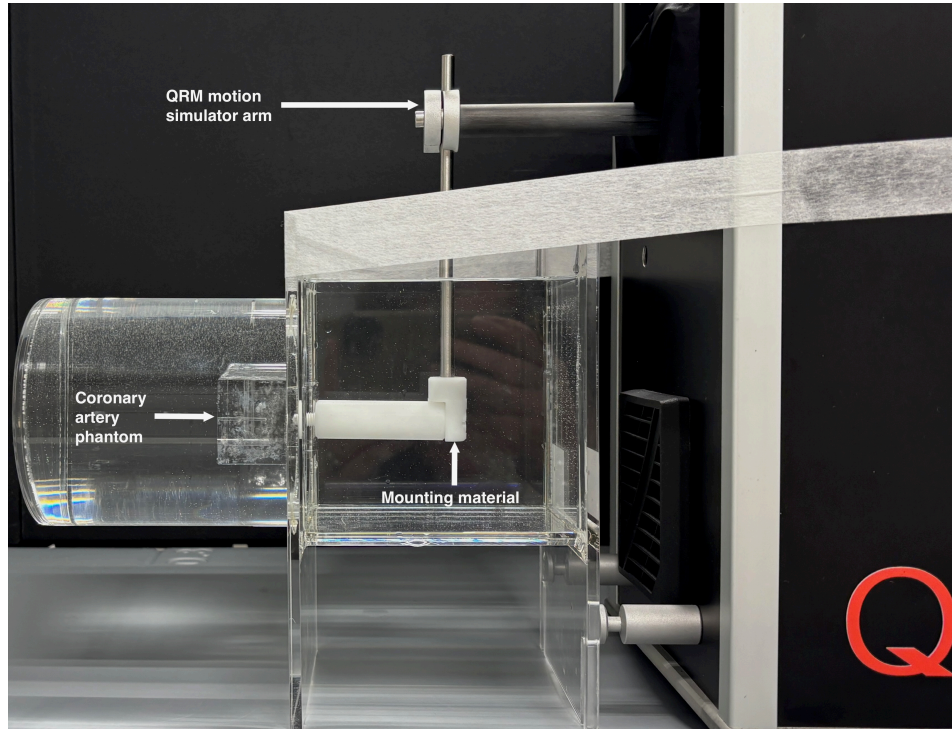


Figure 2.1: Overview of the experimental setup. The QRM motion simulator was positioned on the PCCT table, with the simulator arm extending toward the scan field. The phantom was mounted onto the simulator arm using custom mounting material and positioned inside a water tank.

For each target heart rate, the corresponding cycle duration was calculated as

$$T = \frac{60000}{\text{bpm}}, \quad (2.2)$$

where T is the cycle duration in milliseconds. The difference between the 500 ms reference cycle and the target cycle duration was defined as the total temporal shortening. To account for the non-uniform shortening of the cardiac cycle at higher heart rates, the cardiac cycle was divided into a systolic phase from 0 to 200 ms and a diastolic phase from 200 to 500 ms. Because the reduction in cardiac cycle duration at higher heart rates occurs predominantly during diastole, 60% of the total shortening was assigned to the diastolic phase and 40% to the systolic phase [28, 30]. This approach preserved the overall shape and amplitude of the reference trajectory while approximating the shorter diastolic interval expected at higher heart rates.

The resulting systolic and diastolic durations were used to construct a new time axis for each target heart rate. The original x-, y-, and z-displacement trajectories were then linearly resampled onto this new time axis. Finally, the first and last points of each generated profile were constrained to match, resulting in a closed periodic motion cycle.

The displacement of the phantom in the generated motion profiles for 120, 130, 140, and 150 bpm are shown in Figure 2.2. The z-direction displacement is shown as a representative motion component; remaining x- and y-displacement profiles and corresponding MATLAB code are provided in Appendix C.

The ECG signal generated by the motion simulator was provided to the PCCT system to enable ECG-synchronized image acquisition. This ensured a fixed phase reference within the motion cycle for all simulated heart rates.

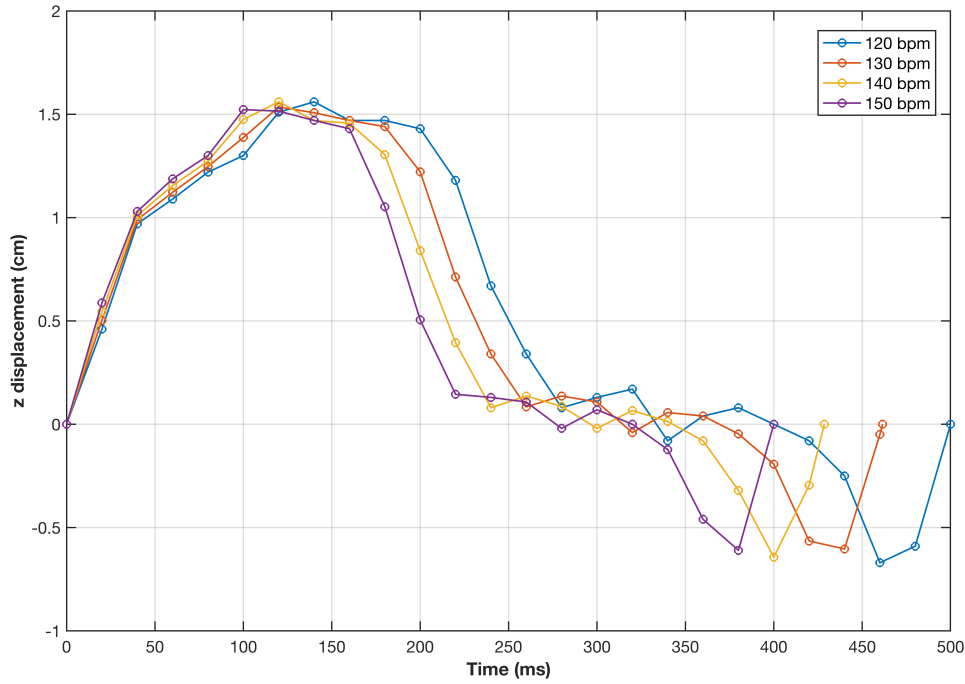


Figure 2.2: Comparison of the z-direction displacement of the generated motion profiles for the simulated neonatal heart rates. Cardiac cycle duration shortened with increasing heart rate while preserving the overall displacement pattern.

2.3. PCCT acquisition protocols and image reconstruction

Two ECG-synchronized PCCT acquisition modes were compared in this study:

1. **High-pitch flash ECG-triggered acquisition**
2. **Prospective sequential ECG-gated acquisition**

For each acquisition mode, images were reconstructed at multiple virtual monoenergetic energy levels. Each unique combination of acquisition mode and reconstruction setting was considered a separate protocol condition.

2.3.1. General scanner and reconstruction settings

All scans were performed on a photon-counting CT system (NAEOTOM Alpha, Siemens Healthineers GmbH). The acquisition protocol was based on the clinical scan settings currently used for neonatal coronary PCCT imaging, as the aim of this study was to evaluate scan quality under conditions representative of current clinical practice. Therefore, a tube potential of 70 kV and a CARE keV image quality (IQ) level of 60 were used for all acquisitions. CARE (Combined Applications to Reduce Exposure) keV IQ is an automatic exposure control setting that controls the target image quality and corresponding tube current selection for the acquisition [31]. ECG leads were connected from the cardiac motion simulator to the PCCT system to enable ECG-synchronized scanning.

The detector configuration was 144×0.4 mm. The scan range was standardized across all protocols to ensure identical z-axis coverage of the coronary phantom. The field of view, matrix size, slice thickness, and slice increment were kept constant across acquisitions.

All images were reconstructed using the vascular kernel Bv48 in combination with Quantum Iterative Reconstruction (QIR) at strength level 4, as this reconstruction approach represents the clinical standard for vascular imaging. The Bv48 kernel is optimized for vascular imaging and was used to preserve vessel edge sharpness, while QIR was applied to reduce image noise. Strength level 4 was selected because this combination reflects the clinical reconstruction standard for vascular imaging and was considered to provide an appropriate balance between coronary edge sharpness and image noise.

Virtual monoenergetic images were reconstructed according to the energy settings included in the final scan protocol. Flash acquisitions were reconstructed at 40, 47, and 53 keV, whereas prospective sequential acquisitions were reconstructed at 53 keV. Apart from the selected VMI energy level, all reconstructions used identical geometric reconstruction parameters. An overview of the acquisition and reconstruction settings of both protocols is provided in Table 2.1.

2.3.2. High-pitch flash ECG-triggered protocol

The flash protocol consisted of a high-pitch spiral acquisition intended to maximize temporal resolution within a short scan duration and thereby reduce motion artifacts. ECG triggering was performed at 160 ms after the R-peak, corresponding to the end-systolic phase of the cardiac cycle. Due to the high-pitch spiral acquisition, the effective reconstruction phase varied along the z-axis. Depending on the z-position of the acquisition window, the reconstruction window started between 32% and 43% of the R–R interval and ended between 45% and 57%. Acquisitions were performed at each simulated heart rate setting.

Protocol-specific acquisition and reconstruction parameters, including pitch and rotation time, are summarized in Table 2.1.

2.3.3. Prospective sequential ECG-gated protocol

The prospective sequential protocol consisted of ECG-gated acquisitions with image reconstruction for different cardiac phases. The prospective protocol was used for: static reference evaluation and dynamic phase-dependent motion analysis.

For the repeated measurements at 0 bpm, images were reconstructed at 160 ms of the cardiac cycle and at three virtual monoenergetic levels: 40, 47, and 53 keV. In addition, a single-phase prospective acquisition was performed at 120 bpm and VMI level of 53 keV for the dose comparison. This acquisition used the same prospective sequential acquisition and reconstruction approach as the static flash scan, matching the 160 ms trigger timing, with a selected 0% phase reconstruction. This scan was included to separate the dose contribution of a single-phase prospective acquisition from the phase-resolved dynamic prospective acquisitions.

The measurements using the dynamic phantom were performed at different heart rates, and the prospective scans were reconstructed at the reference virtual monoenergetic level of 53 keV. Reconstructions were generated from 25% to 80% of the R–R interval in 5% increments. The reconstruction range was determined by both technical and clinical considerations. Reconstructions were only available from 25% of the R–R interval because of the delay between the ECG trigger and the actual start of data acquisition. To limit the total amount of reconstructed data to a manageable level, cardiac phases were reconstructed up to 80% of the R–R interval. This upper limit was considered sufficient because previous research has shown that the clinically most relevant cardiac phases are generally contained within this part of the cardiac cycle [32].

For a reliable comparison, acquisition parameters were matched as closely as possible to those of the flash protocol, including tube potential, z-coverage, and reconstruction settings. The phantom fitted within a single stack, allowing the scan to be performed without table movement or additional stack acquisitions. Protocol-specific acquisition and reconstruction parameters are summarized in Table 2.1.

Table 2.1: Overview of the acquisition and reconstruction parameters for the evaluated PCCT protocols.

Parameter	Flash protocol	Prospective protocol ^a
Tube potential	70 kV	70 kV
Effective mAs	16	16
CARE keV IQ level	60	60
Rotation time	250 ms	250 ms
Pitch	3.2	n.a.
Collimation	144 × 0.4 mm	144 × 0.4 mm
ECG triggering	160 ms	Phase-resolved acquisition
Reconstructed slice thickness	0.4 mm	0.4 mm
Increment	0.2 mm	0.2 mm
Matrix	512 × 512	512 × 512
FOV	102 × 102 mm	102 × 102 mm
Kernel	Bv48	Bv48
QIR strength	4	4
VMI level	40, 47, 53 keV	0 bpm: 40, 47, 53 keV 120–150 bpm: 53 keV
Reconstruction phase	start: 32–43%, end: 45–59% of the R–R interval	0 bpm: 160 ms 120–150 bpm: 25–80% in 5% steps

^a Although multiple stacks for the prospective sequential acquisition were not required, the Zeefree reconstruction option was used, which reconstructs the data without slice overlap, reducing potential artifacts along the z-axis [33].

2.4. Image analysis

2.4.1. Comparison framework

To address the study aims, the quantitative analysis was structured around three predefined comparisons, each targeting a different aspect of tube visibility, geometric preservation, and motion sensitivity:

1. **Static inter-protocol comparison:** comparison between flash and prospective acquisitions at 0 bpm and virtual monoenergetic levels of 40, 47, and 53 keV to assess the influence of the virtual monoenergetic level in the absence of motion.
2. **Cardiac motion inter-protocol comparison:** comparison between flash and prospective acquisitions at 120, 130, 140, and 150 bpm with a virtual monoenergetic level of 53 keV. Two comparison strategies were used: a z-matched phase comparison and a best-phase comparison. For the z-matched comparison, the prospective reconstruction phase was selected to correspond as closely as possible to the midpoint of the z-dependent reconstruction window of the flash acquisition. For the best-phase comparison, one prospective reconstruction phase was selected for each heart-rate condition based on the quantitative image-quality metrics. Selection was primarily based on the highest Dice Similarity Coefficient (DSC), with edge rise distance (ERD), circularity, and contrast-to-noise ratio (CNR) used as supporting metrics.
3. **Within-protocol motion-robustness analysis:** separate analysis of the flash and prospective datasets across heart-rate conditions.

The inter-protocol comparison was limited to 53 keV, which was used as the reference virtual monoenergetic level in this study. This restriction was chosen to keep the number of reconstructions and quantitative analyses feasible, while still allowing the research questions regarding motion robustness and inter-protocol differences to be addressed. This framework therefore combined direct inter-protocol comparisons where appropriate with protocol-specific evaluation of motion-robustness.

2.4.2. Workflow

The scanner output consisted of a folder with Digital Imaging and Communications in Medicine (DICOM) files, containing data of the reconstructions for all acquisitions. To prepare the data for analysis, the files were uploaded to Orthanc [34], an open-source DICOM-server, which separated image series into distinct folders based on DICOM metadata. The separated series were exported using a Python script `simple_orthanc` implemented using `pydicom`, developed at Erasmus MC [35]. This preprocessing step

produced a structured dataset for quantitative image analysis.

For each protocol condition, the dynamic cardiac motion dataset was compared with the corresponding static reference dataset using the following workflow:

1. DICOM slices were grouped into image volumes.
2. The static and dynamic volumes were cropped to a common region of interest.
3. A 3D tube mask was generated for both datasets.
4. The dynamic scan was registered to the static reference and the corresponding tubes were matched between the two datasets.
5. Cross-sectional images were sampled for each tube, and quantitative metrics were extracted.

The corresponding `Matlab` code is provided in Appendix D.

Registration and segmentation

Tube masks were generated from the cropped volumes using HU-based thresholding between 200 and 1500 HU, followed by connected component filtering, morphological closing, and hole filling. The dynamic scan was aligned to the static reference using mask-based centroid registration. Since the motion simulator introduced translational displacement only, without rotational motion, the registration was restricted to translation. The resulting transform was applied to both the image volume and the binary tube mask. After registration, spatial overlap between the dynamic and static tube masks was quantified using the DSC, which was included as one of the quantitative image-quality metrics and is defined in Equation 2.9.

After registration, connected component analysis was used to identify the six tubes. For each tube, 35 cross-sectional slices were extracted along the 20 mm tube length, corresponding to a sampling interval of approximately 0.59 mm. This number of cross-sectional slices was chosen to provide sufficient coverage of the full tube segment while avoiding unnecessary oversampling of adjacent, highly similar slices. The sampling positions were distributed on both sides of the centroid to cover the full tube segment. Within each sampled slice, the local HU distribution was used to define a threshold corresponding to the transition between the iodinated tube and the surrounding background. This threshold was calculated as the midpoint between the median HU inside the initial tube mask and the median HU in a surrounding local background ring. A subpixel contour at this threshold was then extracted from the original HU image and constrained to the expected tube region to prevent inclusion of neighboring structures or background noise.

2.4.3. Quantitative image quality

For this study, motion-robustness was defined as the extent to which the acquisition protocol preserved the geometry, edge sharpness, and positional stability of the small vessel structures under cardiac motion.

Motion-robustness was assessed using segmentation-derived quantitative metrics describing geometry, shape, edge definition, attenuation, contrast, and spatial agreement. Geometry-, shape-, and edge-related metrics were considered indicators of structural preservation, while attenuation, contrast, and spatial agreement metrics were used as image-quality measures. The relevance of each metric depended on the image-quality question, and protocol interpretation was therefore task-based: contrast-related metrics were weighted most strongly for visual tube detection, while geometry-derived metrics were used to assess size preservation and segmentation stability. Metrics were computed for each sampled cross-section, and final tube-level values were obtained by taking the median across the 35 sampled cross-sections. The median rather than the mean was used to reduce the influence of local segmentation failures or artifacts near the end of the tubes, which were expected to occur in small moving structures near the spatial-resolution limit.

Quantitative metrics

The quantitative metrics used to evaluate the reconstructed tube structures are described below. Together, these metrics characterize tube visibility, geometric preservation, boundary sharpness, spatial overlap, and attenuation accuracy.

Equivalent diameter For each segmented tube cross-section with measured area A , the equivalent diameter (D_{eq}) was defined as:

$$D_{eq} = 2\sqrt{\frac{A}{\pi}}. \quad (2.3)$$

For dynamic acquisitions, the equivalent diameter error was computed relative to the corresponding static reference:

$$E_{D_{eq}} = |D_{eq,dyn} - D_{eq,ref}|, \quad (2.4)$$

where $D_{eq,dyn}$ is the equivalent diameter measured in the dynamic scan and $D_{eq,ref}$ is the equivalent diameter measured in the static reference scan.

Cross-sectional area error Cross-sectional area A was derived from the segmented tube contour. For dynamic acquisitions, the cross-sectional area error E_A was computed relative to the corresponding static reference:

$$E_A = |A_{dyn} - A_{ref}|, \quad (2.5)$$

where A_{dyn} and A_{ref} are the cross-sectional areas measured in the dynamic and static reference scans, respectively.

Circularity The circularity (C) of each segmented cross-section was defined as:

$$C = \frac{4\pi A}{P^2}, \quad (2.6)$$

where A is the segmented cross-sectional area and P is the contour perimeter. A value closer to 1 indicates a more circular cross-section.

Edge rise distance Edge sharpness was quantified using the 10–90% edge rise distance. For each radial intensity profile across the tube boundary, ERD was defined as the spatial distance between the 10% (x_{10}) and 90% (x_{90}) points of the edge transition:

$$ERD = x_{90} - x_{10}. \quad (2.7)$$

Attenuation bias Mean attenuation was measured within the segmented tube region. For dynamic acquisitions, attenuation bias B_{HU} was defined relative to the corresponding static reference:

$$B_{HU} = CT_{dyn} - CT_{ref}, \quad (2.8)$$

where CT_{dyn} and CT_{ref} are the mean attenuation values in the dynamic and static reference scans, respectively.

Dice similarity coefficient Spatial agreement between the segmented tube mask in the dynamic scan X and the static reference mask Y was quantified using the DSC:

$$DSC = \frac{2|X \cap Y|}{|X| + |Y|}. \quad (2.9)$$

Contrast-to-noise ratio The CNR between the tube and the background material was defined as:

$$CNR = \frac{|CT_{tube} - CT_{bg}|}{\sigma_{bg}}, \quad (2.10)$$

where CT_{tube} is the mean attenuation within the segmented tube, and CT_{bg} and σ_{bg} are the mean and standard deviation of the background attenuation. Background attenuation was extracted from circular regions with a radius of 10 pixels in three tube-free image slices.

Application to the analyses

The quantitative metrics were applied to four analysis categories: repeatability and quality control, static inter-protocol comparison, cardiac motion inter-protocol comparison and within-protocol motion-robustness analysis. The same metrics were used where applicable; however, the reference values used for geometric error metrics depended on the analysis. In the static comparison, the nominal values from the phantom design were used as reference. In the dynamic measurements, the corresponding static acquisition was used as reference, allowing motion-induced changes to be evaluated relative to the non-moving condition.

Repeatability and quality control To assess mechanical, positioning and image analysis repeatability, five repeated scans were acquired for both protocols under the reference dynamic condition of 120 bpm at 53 keV, using identical acquisition and reconstruction settings. For the prospective protocol, repeatability was evaluated for the available phase reconstructions within this condition. For each scan, the predefined quantitative metrics were extracted using the same analysis pipeline.

Repeated acquisitions were summarized as mean values to reduce the influence of random variation, but averaging only improves precision when variability is mainly random. It does not correct systematic effects related to spatial resolution, partial-volume averaging, segmentation thresholding, residual motion, or phase-dependent reconstruction differences. Therefore, mean values were interpreted together with the corresponding standard deviation and coefficient of variation. Lower SD and CV values were interpreted as better repeatability. CV was reported for circularity, ERD, DSC, and CNR, as these are relative or dimensionless metrics for which percentage variability is interpretable. For signed geometric error metrics, such as diameter and area error, CV was not reported because values may approach zero or change sign, making percentage variability unstable and less meaningful. The CV was calculated as:

$$CV = \frac{\sigma}{\mu} \times 100\%. \quad (2.11)$$

Normalized errors were reported for equivalent diameter and cross-sectional area because the same absolute deviation represents a larger proportional error in the 1.1 mm tubes than in the 2.0 mm tubes. Normalized errors were not calculated for circularity, ERD, DSC, and CNR, because these metrics either already have a normalized scale or do not have a reference value. The normalized errors were calculated from the absolute differences:

$$E_{D_{eq},rel} = \frac{E_{D_{eq}}}{D_{eq,ref}} \times 100\% \quad (2.12)$$

$$E_{A,rel} = \frac{E_A}{A_{ref}} \times 100\% \quad (2.13)$$

For attenuation, the attenuation bias was the primary measure to preserve the direction of the attenuation difference. When reported as a proportional measure, the relative attenuation bias was calculated from the signed attenuation difference relative to the corresponding static reference attenuation value:

$$B_{HU,rel} = \frac{B_{HU}}{CT_{ref}} \times 100\%. \quad (2.14)$$

In addition to these descriptive repeatability measures, the distribution of repeated measurements was visually assessed using repeatability plots. These plots were used to identify metric-specific variation across repeated scans, tube diameter, and tube orientation.

Repeatability was assessed under one reference dynamic condition only; therefore, the repeatability results were not treated as repeatability estimates for all scan settings. Instead, they were used as a guideline for the magnitude of scan-to-scan variation that may be expected in the study. To express this variation in a way that could be compared with differences observed in the remaining results, the repeatability coefficient (RC) was calculated [36, 37, 38].

For each diameter group and metric, five repeated scans were available for the x-, y-, and z-oriented tubes. Because these orientations may show systematic differences due to their orientation relative to the motion direction and imaging plane, orientations were not averaged directly before estimating repeatability. Instead, the SD across the five repeated scans was first calculated separately for each orientation within the same diameter group. The orientation-specific variances were then averaged within each tube diameter:

$$SD_{within,d} = \sqrt{\frac{SD_{x,d}^2 + SD_{y,d}^2 + SD_{z,d}^2}{3}}, \quad (2.15)$$

where $SD_{x,d}$, $SD_{y,d}$, and $SD_{z,d}$ are the standard deviations across the five repeated scans for the x-, y-, and z-oriented tubes, respectively, within diameter group d . This pooling estimates scan-to-scan variation separately for each diameter group, while preventing systematic differences between tube orientations from being treated as repeatability error.

The repeatability coefficient was then calculated separately for each diameter group as:

$$RC_d = 1.96 \times \sqrt{2} \times SD_{within,d}. \quad (2.16)$$

Here, $SD_{within,d}$ represents the variability of a single repeated scan within diameter group d . The factor $\sqrt{2}$ is included because the RC describes the expected difference between two repeated scan results acquired under identical conditions. This does not refer to one specific pair of scans in the dataset; rather, it represents the expected pairwise difference between any two scans from the same repeatability process. Since both repeated scans contribute measurement variation, the variance of their difference is $SD_{within,d}^2 + SD_{within,d}^2 = 2SD_{within,d}^2$. The standard deviation of this difference is therefore $\sqrt{2} \times SD_{within,d}$, which gives the $\sqrt{2}$ term in the RC formula [39].

The resulting RC represents the range within which approximately 95% of absolute differences between two repeated scan results under identical conditions are expected to fall.

The RC was used in addition to the statistical model results and answers a different question than a 95% confidence interval. A 95% confidence interval describes the uncertainty around an estimated mean or model coefficient, whereas the RC describes the expected measurement variation between repeated observations of the same condition. Therefore, the RC provided a reference for judging whether an observed difference was larger than the variation expected from repeatability alone. Differences smaller than the RC were considered possibly attributable to normal scan-to-scan variation, whereas differences larger than the RC were considered to exceed the expected repeatability variation.

Static inter-protocol comparison For the static inter-protocol comparison, the scan at each virtual monoenergetic level was registered individually. The static acquisitions reflected the actual imaged geometry of the tubes and were therefore used to evaluate baseline protocol differences in image quality rather than motion-induced deviations. This distinction prevented deviations caused by factors other than motion, including differences between the phantom design and actual phantom geometry, static scanner-related limitations, and segmentation or image-processing uncertainty, from being misclassified as motion-induced error in the dynamic analyses. For this static comparison, geometric accuracy was assessed relative to the nominal phantom dimensions.

For this comparison, equivalent diameter error and cross-sectional area error were defined as:

$$E_{D_{eq,nom}} = |D_{eq} - D_{nom}|, \quad (2.17)$$

$$E_{A,nom} = |A - A_{nom}|, \quad (2.18)$$

where D_{eq} and A are the measured equivalent diameter and cross-sectional area, and D_{nom} and A_{nom} are the corresponding nominal values.

The nominal area was calculated from the nominal tube diameter as:

$$A_{\text{nom}} = \pi \left(\frac{D_{\text{nom}}}{2} \right)^2. \quad (2.19)$$

Dynamic analyses For the dynamic analyses, including the cardiac motion inter-protocol comparison and the within-protocol motion-robustness analysis, equivalent diameter error, cross-sectional area error, attenuation bias, and DSC were calculated relative to the corresponding static reference scan. Circularity, ERD, and CNR were evaluated directly for each reconstructed dataset.

2.5. Radiation dose assessment

Dose optimization is essential in neonatal CT because of the increased radiosensitivity of neonates and the importance of minimizing unnecessary radiation exposure according to the ALARA principle. Scanner reported dose indicators included the Computed Tomography Dose Index Volume (CTDI_{vol}), the dose-length product (DLP) and the size-specific dose estimate (SSDE). CTDI_{vol} reflects standardized scanner output, whereas SSDE incorporates a correction for object size and is therefore more relevant for small phantom or neonatal-sized objects.

In addition, the effective scan length was estimated from the scanner-reported dose metrics as the ratio between DLP and CTDI_{vol}:

$$L_{\text{eff}} = \frac{DLP}{CTDI_{\text{vol}}}. \quad (2.20)$$

The effective scan length was included to allow comparison of the dose-related scan extent between acquisition protocols.

Dose values were reported descriptively because the protocols differed in acquisition and reconstruction choices. For both protocols, dose was recorded for the static acquisition, the repeated 120 bpm acquisitions, and the single acquisitions at 130, 140, and 150 bpm. For the prospective protocol, dose was also reported for a single-phase dynamic acquisition, which served as a baseline condition for prospective scanning with one selected reconstruction phase. This distinction was made because the phase-resolved prospective acquisitions allowed reconstructions from 25% to 80% of the R–R interval.

2.6. Statistical analysis

2.6.1. Endpoints and general approach

All statistical analyses were performed separately for each quantitative endpoint and followed the pre-defined comparison framework rather than a single model including all protocol factors simultaneously.

For comparative and inferential analyses, the mean across the repeated scans was used as the representative value for the 120 bpm, 53 keV condition, thereby preventing overrepresentation of this condition in the overall dataset.

Because only six independent tube structures were available, p-values were interpreted as supportive evidence rather than as the basis for protocol selection. Final interpretation was based on consistency across descriptive trends, model results, visual assessment, and physical plausibility. Statistical significance was assessed using the resulting p-values, with a significance level of $\alpha = 0.05$.

2.6.2. Descriptive analysis

Descriptive analyses of the results were used to evaluate trends in the tube-level metrics across protocol, heart rate, tube diameter, tube orientation, and, where applicable, reconstruction phase and virtual monoenergetic level. In addition, tube-specific observations in the x-, y-, and z-directions were inspected descriptively to assess whether geometric errors showed consistent spatial anisotropy.

For the prospective acquisitions with cardiac motion, phase curves were generated to visualize how each endpoint varied across the reconstructed cardiac phases.

2.6.3. Inferential analysis

Inferential statistical analyses were performed in MATLAB using the Statistics and Machine Learning Toolbox. Linear mixed-effects models were fitted using the `fitlme` function [40]. This approach was used because the same tubes were measured repeatedly across different scan conditions. Measurements from the same tubes were therefore not fully independent. To account for this repeated-measures structure, tube identity was included as a random intercept using the model term `(1|tube_id)`. This allowed each tube to have a tube-specific baseline level for the evaluated metric, while the fixed effects estimated the average effects of the scan conditions across the full set of tubes. The random intercept therefore captured residual between-tube variability not explicitly modeled by the fixed effects, including potential orientation-related differences or small local differences in tube positioning. This prevented repeated measurements from the same physical tube from being treated as statistically independent observations.

Models were fitted using restricted maximum likelihood estimation. The linear mixed-effects model was used as an inferential comparison framework to estimate differences between predefined scan conditions, rather than as a mechanistic model of the underlying motion or image-formation process. The suitability of the additive linear formulation was assessed using model diagnostics. Residual-versus-fitted and quantile-quantile plots were inspected for systematic patterns, and observed-versus-fitted plots were used to evaluate whether there was major systematic lack of fit. Exploratory models including quadratic terms were additionally fitted to assess whether relevant non-linear trends were present. Because these terms were close to zero and the diagnostic plots showed no major systematic lack of fit for most metrics, the linear mixed-effects formulation was considered adequate for the statistical comparisons. The diagnostic plots and exploratory model assessment are provided in Appendix F. If substantial deviations from model assumptions were observed, the corresponding inferential results were interpreted cautiously and descriptive analyses were given greater weight.

The general model structure was:

$$Y_{ij} = \beta_0 + \sum_{m=1}^p \beta_m X_{m,ij} + u_{tube,i} + \varepsilon_{ij}, \quad (2.21)$$

where Y_{ij} is the value of the endpoint for observation j of tube i , β_0 is the fixed intercept, β_m are the fixed-effect coefficients, $X_{m,ij}$ are the scan or reconstruction parameters included in the specific comparison, $u_{tube,i}$ is the random intercept for tube i , and ε_{ij} is the residual error term.

The fixed effects differed between comparison categories. For the static inter-protocol comparison, acquisition protocol, virtual monoenergetic level, and tube diameter were included. For the cardiac motion inter-protocol comparisons at 53 keV, acquisition protocol, heart rate, and tube diameter were included; separate models were fitted for the z-matched phase and best-phase comparisons. For the flash within-protocol motion-robustness analysis, virtual monoenergetic energy, heart rate, and tube diameter were included. For the prospective within-protocol motion-robustness analysis, heart rate, reconstruction phase, and tube diameter were included, with reconstruction phase treated as a categorical variable.

2.6.4. Interpretation of endpoints

For the interpretation of the equivalent diameter error, cross-sectional area error, and ERD, lower values indicated better performance. For circularity, CNR, and DSC, higher values indicated better shape preservation, contrast, or spatial agreement, respectively. Attenuation bias was analyzed as a signed quantity to distinguish overestimation from underestimation relative to the static reference.

For the static inter-protocol comparison, the directly measured static tube metrics were interpreted as measures of baseline comparability between acquisition modes rather than as motion-related error metrics.

2.6.5. Post-hoc comparison

The primary inferential results were based on fixed-effect tests from the mixed-effects models. Holm-corrected post-hoc pairwise comparisons were performed only for selected significant fixed effects with more than two levels. These included reconstruction energy within the flash within-protocol analysis, reconstruction phase within the prospective within-protocol analysis, and reconstruction energy within the static inter-protocol comparison. Acquisition protocol and tube diameter were interpreted directly from the mixed-model estimates and did not require post-hoc correction.

Fixed-effect p-values were reported for the mixed-effects models. Estimated differences and Holm-adjusted p-values were reported for selected post-hoc comparisons.

3

Results

3.1. Validation of the phantom and experimental setup

3.1.1. Visualization of the scans

Resulting PCCT images were included to relate the quantitative results to the visual appearance of the iodine-filled tubes (Figure 3.1). Although 53 keV was used as the reference reconstruction for the quantitative analysis, the images are shown at 40 keV because this energy provided the clearest visual depiction of the tubes. Corresponding images at 47 keV and 53 keV are provided in Appendix G.1.

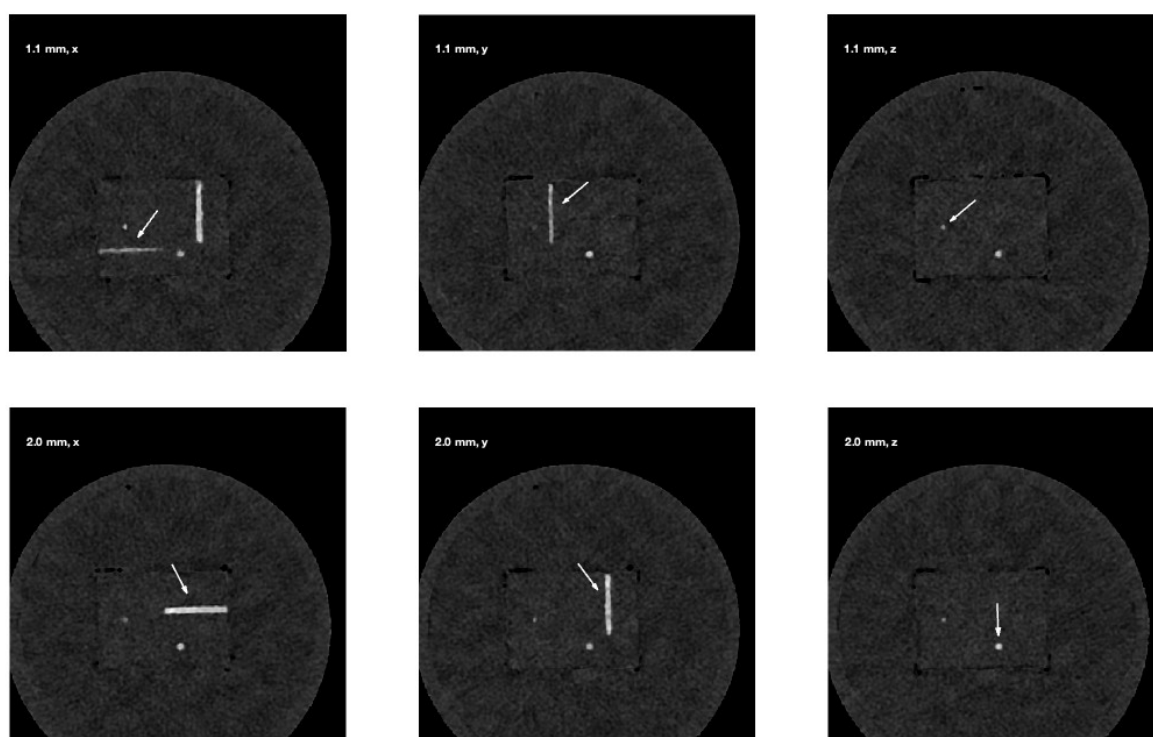


Figure 3.1: Representative axial PCCT images of the six iodine-filled tube configurations reconstructed at 40 keV. Images are shown for the 1.1 mm and 2.0 mm tubes in the x-, y-, and z-orientations. White arrows indicate the tube assessed in each panel. Although 53 keV was used as the quantitative reference reconstruction, 40 keV is shown here because it provided the clearest visual depiction of the tubes.

3.1.2. Visual validation of segmentation and registration

Representative segmentation and registration outputs are shown for the 120 bpm, 40 keV flash reconstruction. Figure 3.2 shows the segmented tubes before and after registration, illustrating the spatial correspondence between the static and registered cardiac motion scan segmentations.

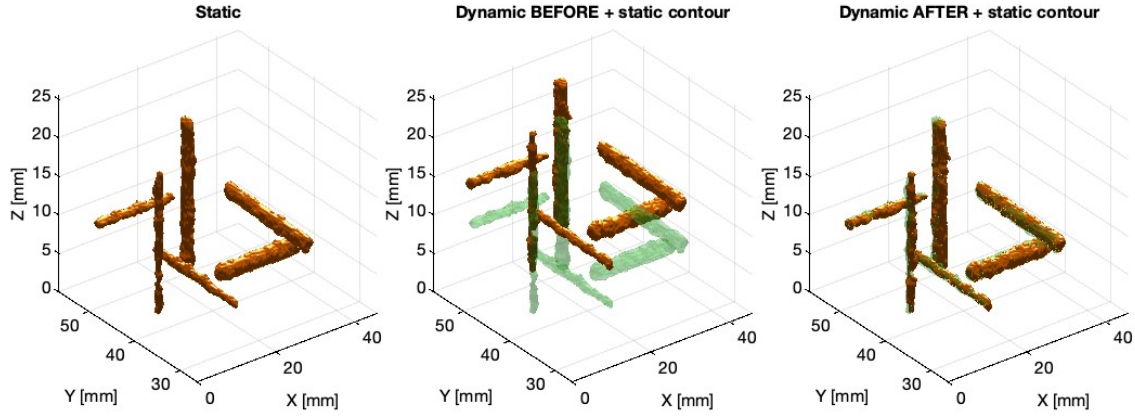


Figure 3.2: Three-dimensional visualization of the segmented tubes before and after registration for the 120 bpm, 40 keV reconstruction flash. The static segmentation and registered dynamic scan segmentation are shown to illustrate spatial correspondence after registration.

Figures 3.3 and 3.4 show representative examples of the segmentation-derived metric extraction. These visualizations illustrate how the segmented tube cross-sections and attenuation profiles were used to derive geometric and HU-based metrics.

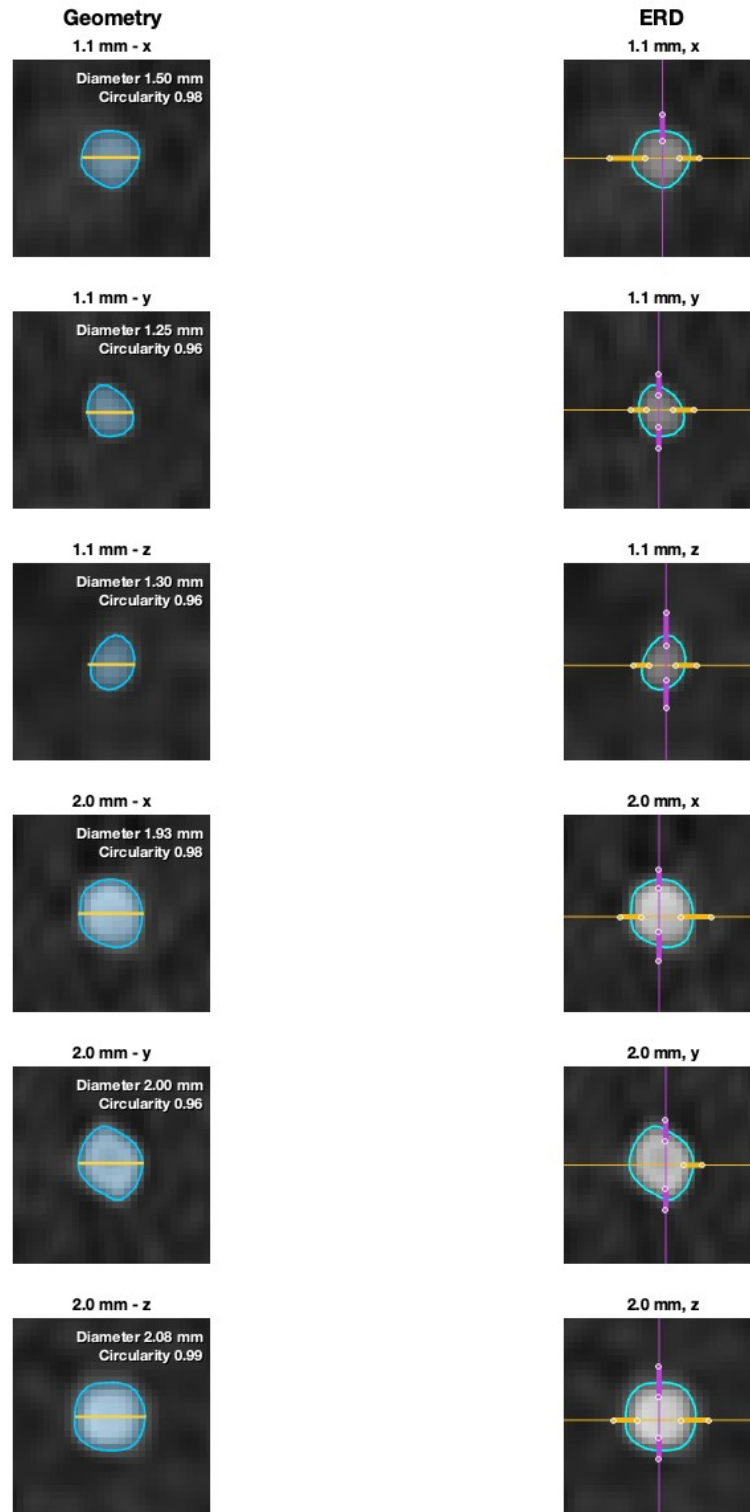


Figure 3.3: Representative visualization of segmentation-derived metric extraction for all six tubes. The left column shows the segmented tube cross-section, with the blue contour indicating the segmented boundary. The right column illustrates the determination of the ERD from horizontal and vertical edge-transitions. Horizontal profiles are shown in yellow and vertical profiles in purple.

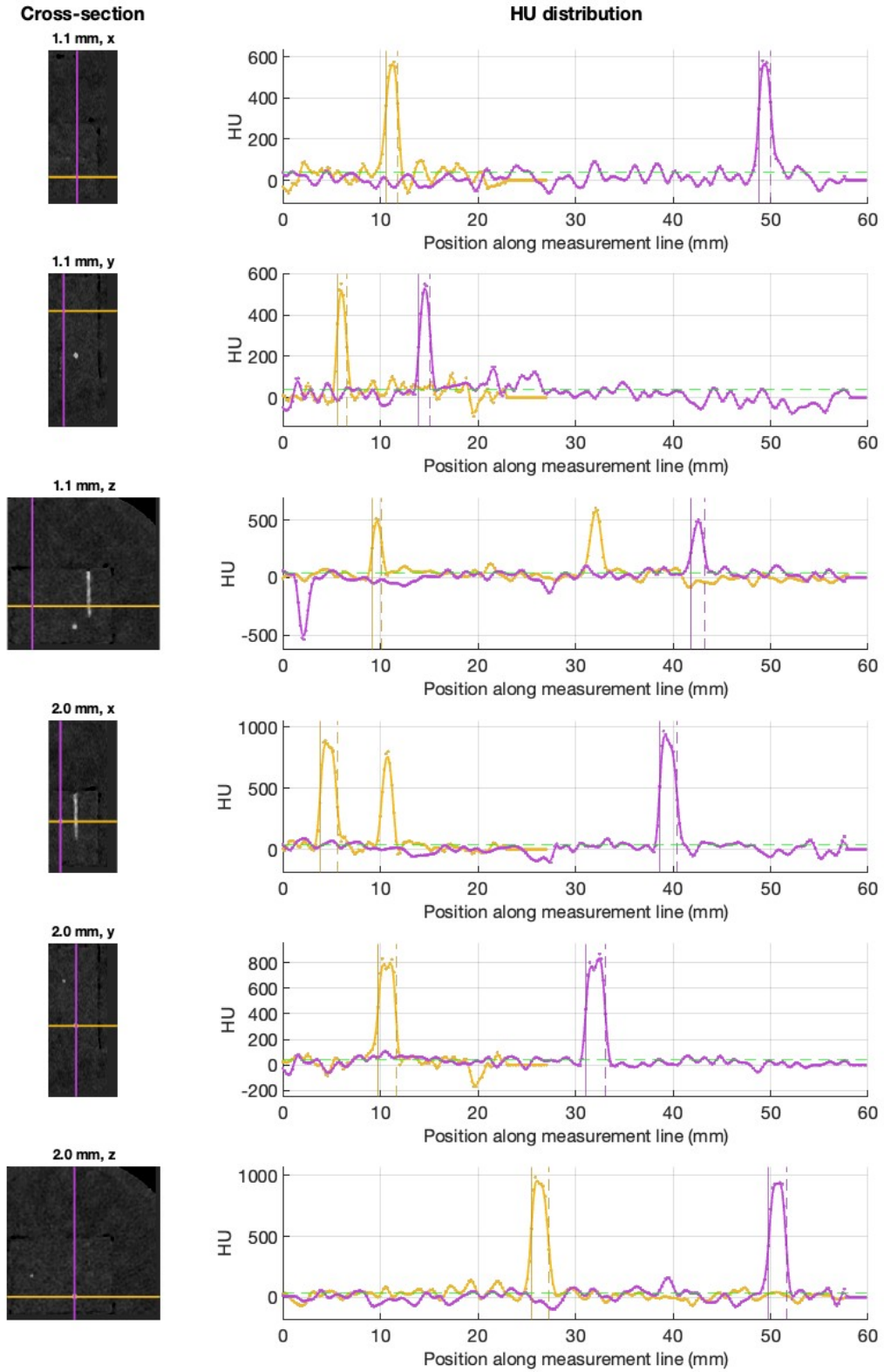


Figure 3.4: Representative visualization of segmentation-derived metric extraction for all six tubes. The left column shows the cross-section of the phantom used for the HU profiles. The right column shows the HU distribution within the cross-section. Horizontal profiles are shown in yellow and vertical profiles in purple; solid lines indicate smoothed HU values and dotted points indicate raw HU samples. Dashed and solid vertical lines mark the detected vessel boundaries, and the green dashed horizontal line in the HU plots indicates the mean PMMA background attenuation.

3.1.3. Repeatability of the experimental setup

Repeatability was evaluated using repeated acquisitions at 120 bpm and 53 keV. For each protocol, repeated measurements were assessed for DSC, CNR, ERD, and circularity.

Flash repeatability across the five repeated scans is shown in Figure 3.5 and summarized in Table 3.1. Normalized equivalent diameter and cross-sectional area errors were higher in the 1.1 mm tubes than in the 2.0 mm tubes. DSC showed greater variability in the CV than circularity and CNR, particularly in the smaller tubes.

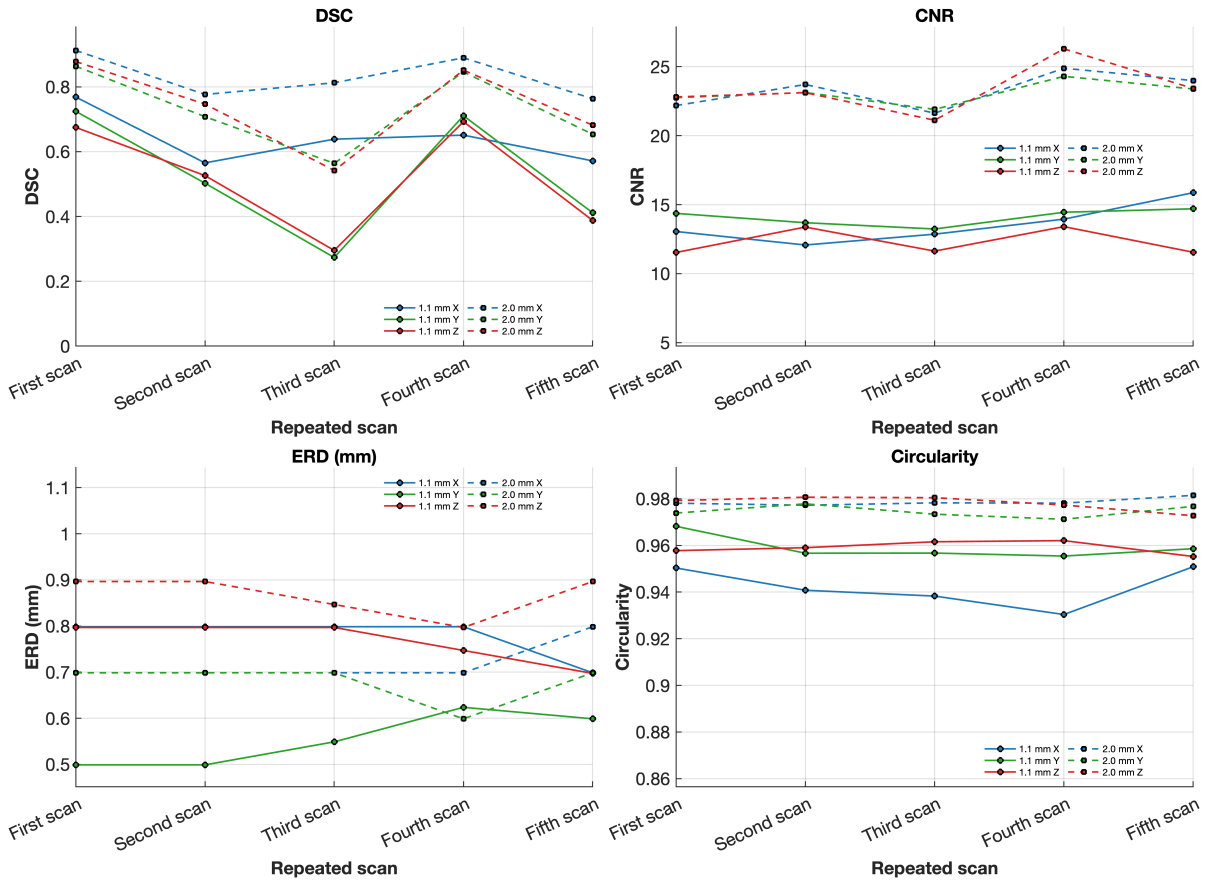


Figure 3.5: Repeatability of the flash protocol across five repeated scans at 120 bpm and 53 keV. The figure shows DSC, CNR, ERD, and circularity for the x-, y-, and z-oriented 1.1 mm and 2.0 mm tubes.

Table 3.1: Repeatability of segmentation-derived metrics for the six tubes across five repeated flash scans at 120 bpm and 53 keV. Values in the upper section are reported as mean $\pm$ standard deviation. The lower section provides the normalized errors and coefficient of variation values for the repeated measurements.

Flash							
	E_{Deq}	E_A	C	ERD	B_{HU}	DSC	CNR
Tube	$\mu \pm \sigma$ [mm]	$\mu \pm \sigma$ [mm ²]	$\mu \pm \sigma$	$\mu \pm \sigma$ [mm]	$\mu \pm \sigma$ [HU]	$\mu \pm \sigma$	$\mu \pm \sigma$
1.1 mm, x	0.095 $\pm$ 0.029	0.166 $\pm$ 0.054	0.942 $\pm$ 0.009	0.778 $\pm$ 0.045	-36.6 $\pm$ 26.1	0.64 $\pm$ 0.08	13.56 $\pm$ 1.45
1.1 mm, y	0.125 $\pm$ 0.011	0.221 $\pm$ 0.024	0.959 $\pm$ 0.005	0.554 $\pm$ 0.057	-0.1 $\pm$ 13.9	0.52 $\pm$ 0.19	14.09 $\pm$ 0.61
1.1 mm, z	0.110 $\pm$ 0.020	0.219 $\pm$ 0.051	0.959 $\pm$ 0.003	0.767 $\pm$ 0.045	-12.6 $\pm$ 12.1	0.52 $\pm$ 0.17	12.30 $\pm$ 1.00
2.0 mm, x	0.064 $\pm$ 0.020	0.185 $\pm$ 0.058	0.979 $\pm$ 0.002	0.719 $\pm$ 0.045	-4.8 $\pm$ 4.3	0.83 $\pm$ 0.07	23.28 $\pm$ 1.34
2.0 mm, y	0.066 $\pm$ 0.008	0.193 $\pm$ 0.022	0.975 $\pm$ 0.003	0.679 $\pm$ 0.045	-20.9 $\pm$ 11.1	0.73 $\pm$ 0.13	23.09 $\pm$ 0.88
2.0 mm, z	0.102 $\pm$ 0.030	0.293 $\pm$ 0.085	0.978 $\pm$ 0.003	0.867 $\pm$ 0.045	-3.6 $\pm$ 18.7	0.74 $\pm$ 0.14	23.35 $\pm$ 1.87
	E_{Deq}	E_A	C	ERD	B_{HU}	DSC	CNR
Tube	Norm. error (%)	Norm. error (%)	CV (%)	CV (%)	Norm. bias (%)	CV (%)	CV (%)
1.1 mm, x	8.7	17.7	0.9	5.7	-9.5	12.8	10.7
1.1 mm, y	11.2	22.5	0.5	10.3	-0.03	37.0	4.3
1.1 mm, z	9.0	18.3	0.3	5.8	-3.7	33.9	8.1
2.0 mm, x	3.4	6.7	0.2	6.2	-0.9	8.1	5.7
2.0 mm, y	3.6	7.3	0.3	6.6	-3.8	17.6	3.8
2.0 mm, z	5.6	13.0	0.3	5.1	-0.7	18.4	8.0

For the prospective protocol, repeatability of the separated reconstructed phases are shown in Figure 3.6. Individual repeated measurement results for the 50% and 75% reconstructions showed trends comparable to the flash repeatability analysis and are provided in Appendix G.2.1.

Prospective repeatability varied across reconstruction phases. At the 50% phase, repeated measurements showed larger variability than at the 75% phase for most evaluated metrics. Circularity showed only small absolute variation across repeated scans. The repeatability results for two representative phases are summarized in Table 3.2.

The full phase-resolved tube-level results and corresponding normalized errors and CV values are provided in Appendix G.2.3 and Appendix G.2.4.

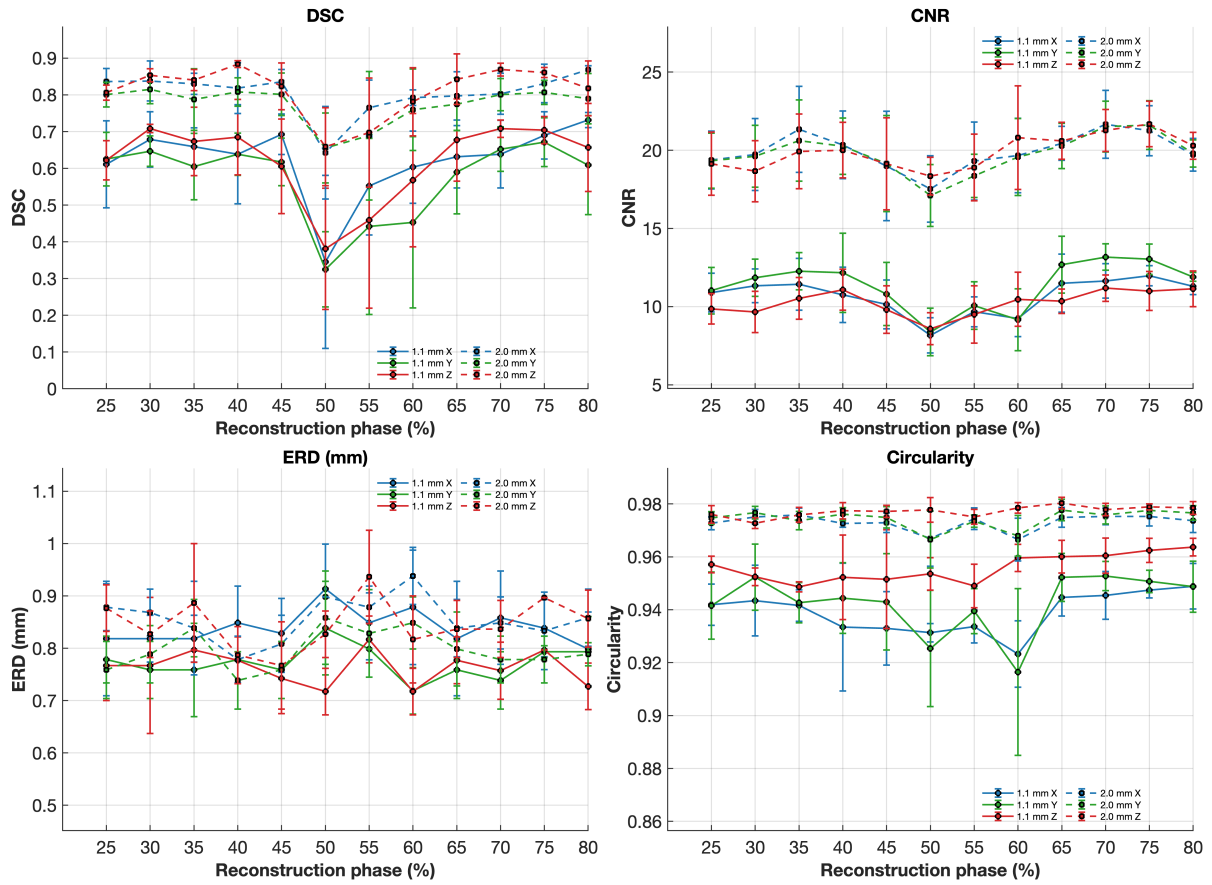


Figure 3.6: Phase-resolved repeatability of the prospective protocol at 120 bpm and 53 keV. The figure shows DSC, CNR, ERD, and circularity across reconstructed phases for the x-, y-, and z-oriented 1.1 mm and 2.0 mm tubes. Error bars indicate the standard deviation across the repeated measurements.

Table 3.2: Mean and standard deviation of segmentation-derived metrics for the six tubes across five repeated scans using the prospective protocol at 120 bpm and 53 keV. Equivalent diameter error (E_{Deq}) and edge rise distance (ERD) are reported in mm, cross-sectional area error (E_A) in mm^2 , attenuation bias (B_{HU}) in HU, and circularity (C), Dice similarity coefficient (DSC), and contrast-to-noise ratio (CNR) are dimensionless.

Prospective 50% phase							
	E_{Deq}	E_A	C	ERD	B_{HU}	DSC	CNR
Tube	$\mu \pm \sigma$ [mm]	$\mu \pm \sigma$ [mm^2]	$\mu \pm \sigma$	$\mu \pm \sigma$ [mm]	$\mu \pm \sigma$ [HU]	$\mu \pm \sigma$	$\mu \pm \sigma$
1.1 mm, x	0.158 ± 0.049	0.273 ± 0.068	0.931 ± 0.003	0.913 ± 0.086	-71.5 ± 16.7	0.35 ± 0.24	8.17 ± 1.13
1.1 mm, y	0.179 ± 0.068	0.315 ± 0.127	0.925 ± 0.022	0.838 ± 0.089	-64.4 ± 24.0	0.33 ± 0.10	8.38 ± 1.51
1.1 mm, z	0.110 ± 0.020	0.199 ± 0.038	0.954 ± 0.006	0.717 ± 0.045	-86.4 ± 26.0	0.38 ± 0.16	8.59 ± 1.03
2.0 mm, x	0.075 ± 0.041	0.217 ± 0.117	0.967 ± 0.010	0.898 ± 0.000	-76.6 ± 25.0	0.64 ± 0.13	17.52 ± 2.13
2.0 mm, y	0.127 ± 0.043	0.377 ± 0.126	0.967 ± 0.011	0.858 ± 0.089	-85.5 ± 24.2	0.66 ± 0.10	17.11 ± 1.98
2.0 mm, z	0.093 ± 0.024	0.268 ± 0.077	0.978 ± 0.005	0.827 ± 0.045	-25.1 ± 10.7	0.66 ± 0.11	18.34 ± 1.21
Prospective 75% phase							
	E_{Deq}	E_A	C	ERD	B_{HU}	DSC	CNR
Tube	$\mu \pm \sigma$ [mm]	$\mu \pm \sigma$ [mm^2]	$\mu \pm \sigma$	$\mu \pm \sigma$ [mm]	$\mu \pm \sigma$ [HU]	$\mu \pm \sigma$	$\mu \pm \sigma$
1.1 mm, x	0.124 ± 0.025	0.210 ± 0.042	0.947 ± 0.003	0.838 ± 0.055	-8.5 ± 12.1	0.69 ± 0.06	11.98 ± 0.65
1.1 mm, y	0.115 ± 0.007	0.198 ± 0.017	0.951 ± 0.004	0.793 ± 0.011	15.9 ± 12.5	0.67 ± 0.07	13.04 ± 0.97
1.1 mm, z	0.101 ± 0.025	0.187 ± 0.051	0.962 ± 0.005	0.797 ± 0.000	-55.4 ± 9.1	0.70 ± 0.04	11.00 ± 1.25
2.0 mm, x	0.059 ± 0.012	0.169 ± 0.036	0.975 ± 0.004	0.833 ± 0.074	-29.7 ± 15.8	0.83 ± 0.05	21.25 ± 1.61
2.0 mm, y	0.083 ± 0.012	0.242 ± 0.032	0.978 ± 0.001	0.778 ± 0.045	-22.3 ± 9.0	0.81 ± 0.03	21.61 ± 1.56
2.0 mm, z	0.092 ± 0.021	0.264 ± 0.060	0.979 ± 0.001	0.896 ± 0.000	11.4 ± 7.6	0.86 ± 0.01	21.68 ± 1.45

Based on the repeated measurements, repeatability coefficients (RCs) were calculated per protocol, tube diameter, and metric. For the prospective protocol, RCs were calculated for all reconstructed phases. To keep the main text focused, Table 3.3 reports the RC values for all flash metrics and for the 50% and 75% prospective phases, as these phases were most relevant for the subsequent interpretation of the quantitative results. The prospective RC values of all reconstructed phases are provided in Appendix G.2.5.

Table 3.3: Repeatability coefficients (RCs) for the repeated reference scans at 120 bpm and 53 keV. RC values are reported per protocol, tube diameter, and endpoint.

Protocol	Diameter	E_{Deq}	E_A	C	ERD	B_{HU}	DSC	CNR
Flash	1.1 mm	0.060	0.125	0.017	0.136	51.2	0.438	2.98
Flash	2.0 mm	0.059	0.169	0.007	0.124	35.5	0.317	3.94
Prospective 50%	1.1 mm	0.139	0.238	0.037	0.210	62.6	0.488	3.44
Prospective 50%	2.0 mm	0.102	0.301	0.025	0.160	58.2	0.305	5.03
Prospective 75%	1.1 mm	0.058	0.110	0.011	0.089	31.3	0.159	2.73
Prospective 75%	2.0 mm	0.043	0.124	0.006	0.138	31.5	0.101	4.27

3.2. Quantitative image analysis

3.2.1. Static inter-protocol comparison

Flash and prospective acquisitions obtained at 0 bpm were compared across virtual monoenergetic reconstruction levels of 40, 47, and 53 keV.

Figure 3.7 shows the static quantitative measurements across reconstruction energies for both protocols. The remaining cross-sectional area error to nominal area plot is provided in Appendix G.3.1.

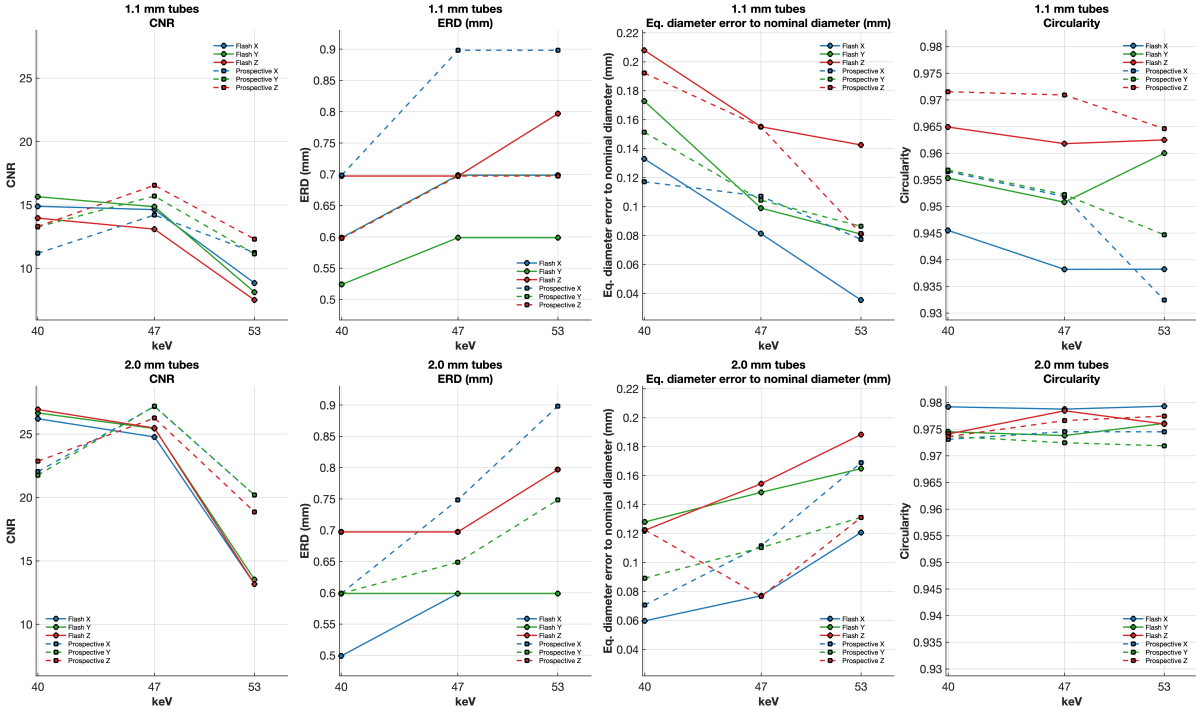


Figure 3.7: Static inter-protocol comparison between flash and prospective acquisitions at 0 bpm across reconstruction energies of 40, 47, and 53 keV.

The corresponding mixed-effects model results are summarized in Table 3.4.

Table 3.4: Linear mixed-effects model results for the static inter-protocol comparison at 0 bpm.

Endpoint	Effect	p-value	Statistical result
$E_{Deq,nom}$	Protocol	0.418	Not significant
$E_{Deq,nom}$	keV	0.566	Not significant
$E_{Deq,nom}$	Tube diameter	0.990	Not significant
$E_{A,nom}$	Protocol	0.394	Not significant
$E_{A,nom}$	keV	0.757	Not significant
$E_{A,nom}$	Tube diameter	0.005	Significant
C	Protocol	0.946	Not significant
C	keV	0.245	Not significant
C	Tube diameter	0.003	Significant
ERD	Protocol	0.002	Significant
ERD	keV	< 0.001	Significant
ERD	Tube diameter	0.751	Not significant
CNR	Protocol	0.190	Not significant
CNR	keV	< 0.001	Significant
CNR	Tube diameter	< 0.001	Significant

No significant protocol effect was observed for equivalent diameter error, cross-sectional area error, or circularity. Reconstruction energy did not significantly affect these geometry-related metrics.

For edge sharpness, ERD was significantly affected by both protocol and reconstruction energy. Relative to flash, prospective acquisition was associated with higher ERD values. Holm-corrected post-hoc comparisons showed that ERD was significantly higher at 47 keV than at 40 keV ($p_{\text{Holm}} = 0.022$) and at 53 keV than at 40 keV ($p_{\text{Holm}} < 0.001$). Selected Holm-corrected post-hoc comparisons are summarized in Table 3.9.

For contrast-related image quality, CNR varied significantly as a function of reconstruction energy and tube diameter. Holm-corrected post-hoc comparisons showed that CNR at 53 keV was significantly lower than at both 40 keV and 47 keV (both $p_{\text{Holm}} < 0.001$).

Tube diameter significantly affected absolute cross-sectional area error and circularity; the 1.1 mm tubes showed larger absolute cross-sectional area error, lower circularity, and lower CNR than the 2.0 mm tubes.

3.2.2. Cardiac motion inter-protocol comparison

Z-matched phase comparison

Flash and z-matched phase prospective acquisitions at 53 keV were compared across heart rates. The reconstructed cardiac phase of the flash protocol depended on the z-position within the scan range. This resulted in reconstruction windows of 32–45% at 120 bpm, 34–52% at 130 bpm, 37–55% at 140 bpm, and 43–57% at 150 bpm. The matched prospective phases were 40% at 120 bpm, 45% at 130 bpm, 45% at 140 bpm, and 50% at 150 bpm.

Figure 3.8 shows the direct comparison between flash and z-matched prospective acquisitions. Plots of the remaining metrics are provided in Appendix G.3.2.

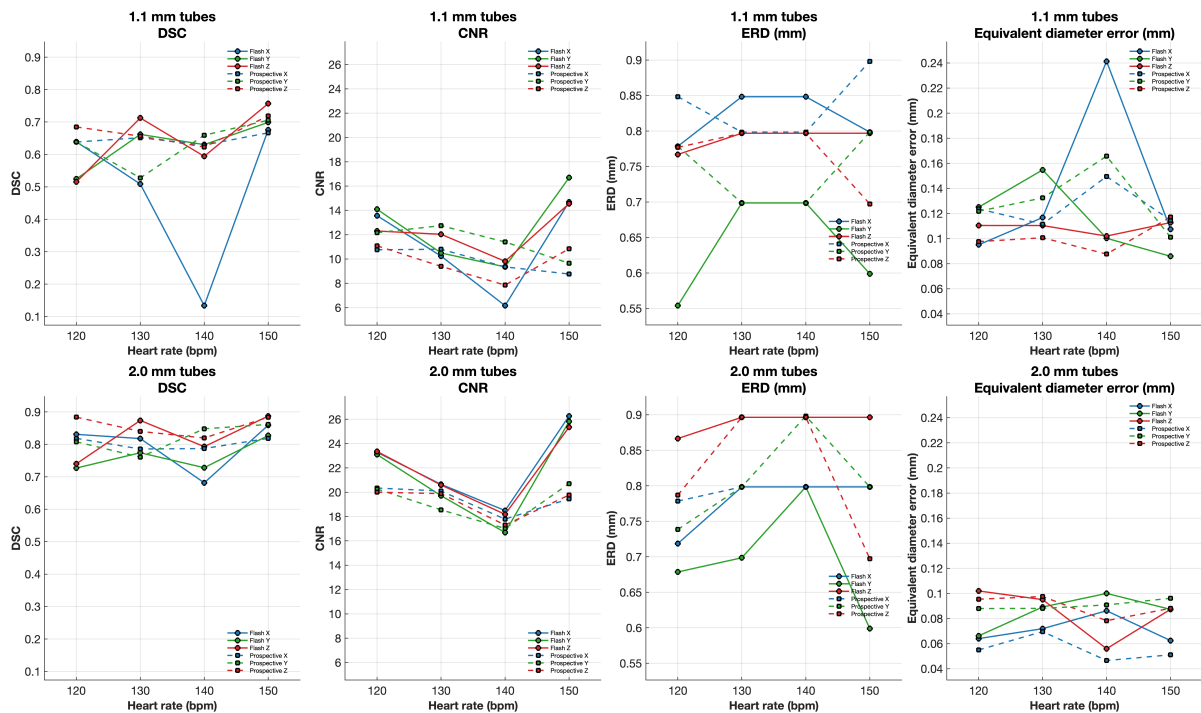


Figure 3.8: Direct comparison between flash and phase-matched prospective acquisitions at 53 keV. The figure shows representative differences in equivalent diameter error, ERD, DSC, and CNR across heart rates and tube diameters.

The mixed-effects model results are summarized in Table 3.5.

Table 3.5: Linear mixed-effects model results for the direct inter-protocol comparison at 53 keV using the z-matched phase.

Endpoint	Effect	p-value	Statistical result
$E_{D_{eq}}$	Protocol	0.716	Not significant
$E_{D_{eq}}$	Heart rate	0.364	Not significant
$E_{D_{eq}}$	Tube diameter	0.001	Significant
E_A	Protocol	0.762	Not significant
E_A	Heart rate	0.485	Not significant
E_A	Tube diameter	0.537	Not significant
C	Protocol	0.701	Not significant
C	Heart rate	0.470	Not significant
C	Tube diameter	0.003	Significant
ERD	Protocol	0.130	Not significant
ERD	Heart rate	0.113	Not significant
ERD	Tube diameter	0.530	Not significant
B_{HU}	Protocol	0.657	Not significant
B_{HU}	Heart rate	0.208	Not significant
B_{HU}	Tube diameter	0.468	Not significant
DSC	Protocol	0.062	Not significant
DSC	Heart rate	0.012	Significant
DSC	Tube diameter	< 0.001	Significant
CNR	Protocol	< 0.001	Significant
CNR	Heart rate	< 0.001	Significant
CNR	Tube diameter	< 0.001	Significant

No significant protocol effect was observed for equivalent diameter error, cross-sectional area error, circularity, ERD, attenuation bias, or DSC. Protocol significantly affected CNR. Heart rate significantly affected DSC and CNR, while tube diameter significantly affected equivalent diameter error, circularity, DSC, and CNR.

Tube-specific x-, y-, and z-directional deviations were present, but no single spatial axis dominated the directional differences.

Best-phase comparison

Flash acquisitions at 53 keV were compared with the best selected prospective phase for each heart-rate condition. The selected phases were 75% at 120 bpm, 70% at 130 bpm, 35% at 140 bpm, and 50% at 150 bpm. These selected phases represent the best phase according to the predefined multi-metric selection for each heart-rate condition.

Figure 3.9 shows the comparison between flash and the selected prospective phase for each heart rate. The corresponding mixed-effects model results are summarized in Table 3.6. Plots of the remaining metrics are provided in Appendix G.3.3.

In the best-phase comparison, protocol significantly affected DSC and CNR. Heart rate significantly affected DSC and CNR, while tube diameter significantly affected equivalent diameter error, circularity, DSC, and CNR.

The directional tube-specific plots showed only small x-, y-, and z-directional differences. No consistent protocol-dependent directional bias was apparent.

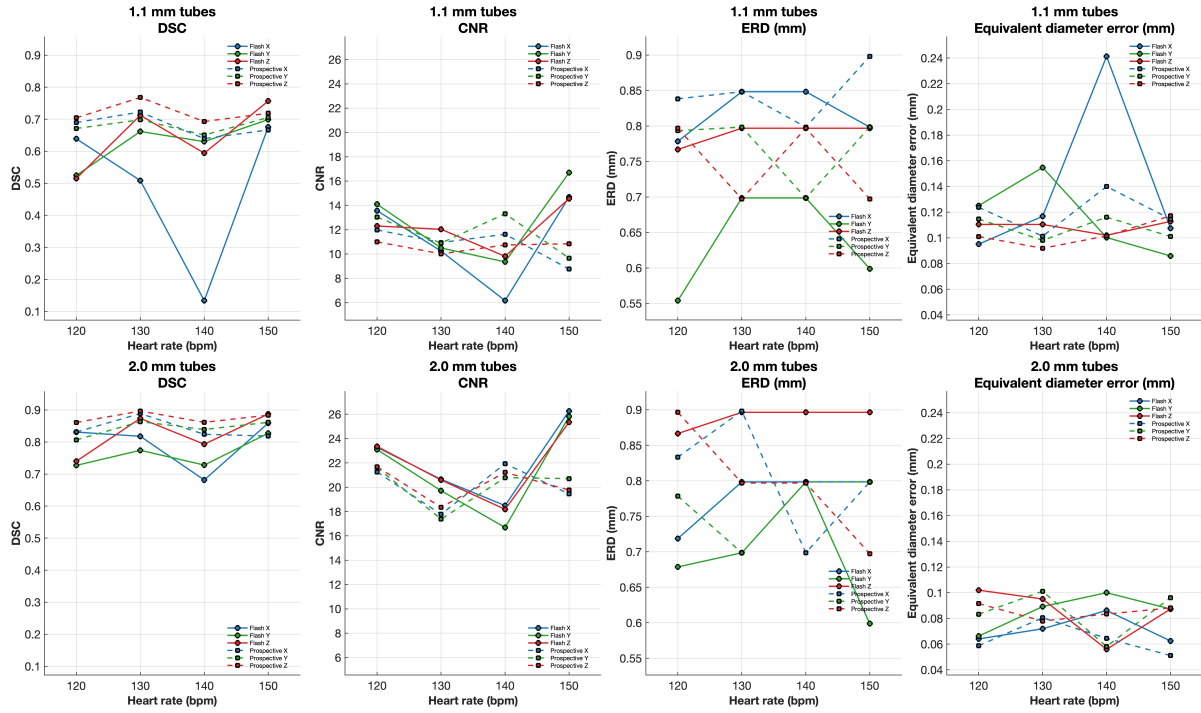


Figure 3.9: Direct comparison between flash and prospective acquisitions at 53 keV using the selected prospective phase for each heart-rate condition. The figure shows representative differences in equivalent diameter error, ERD, DSC, and CNR across heart rates and tube diameters.

Table 3.6: Linear mixed-effects model results for the direct flash versus prospective comparison at 53 keV using the best selected prospective phase for each heart-rate condition.

Endpoint	Effect	p-value	Statistical result
$E_{D_{eq}}$	Protocol	0.315	Not significant
$E_{D_{eq}}$	Heart rate	0.678	Not significant
$E_{D_{eq}}$	Tube diameter	< 0.001	Significant
E_A	Protocol	0.320	Not significant
E_A	Heart rate	0.785	Not significant
E_A	Tube diameter	0.150	Not significant
C	Protocol	0.294	Not significant
C	Heart rate	0.214	Not significant
C	Tube diameter	0.004	Significant
ERD	Protocol	0.273	Not significant
ERD	Heart rate	0.809	Not significant
ERD	Tube diameter	0.677	Not significant
B_{HU}	Protocol	0.223	Not significant
B_{HU}	Heart rate	0.396	Not significant
B_{HU}	Tube diameter	0.433	Not significant
DSC	Protocol	0.001	Significant
DSC	Heart rate	0.009	Significant
DSC	Tube diameter	< 0.001	Significant
CNR	Protocol	0.031	Significant
CNR	Heart rate	< 0.001	Significant
CNR	Tube diameter	< 0.001	Significant

3.2.3. Within-protocol motion-robustness analysis: flash

Within the flash protocol, reconstruction energy, heart rate, and tube diameter were evaluated. Differences across reconstruction energy, heart rate, and tube diameter are shown in Figure 3.10. Plots of the remaining metrics are provided in Appendix G.3.4.

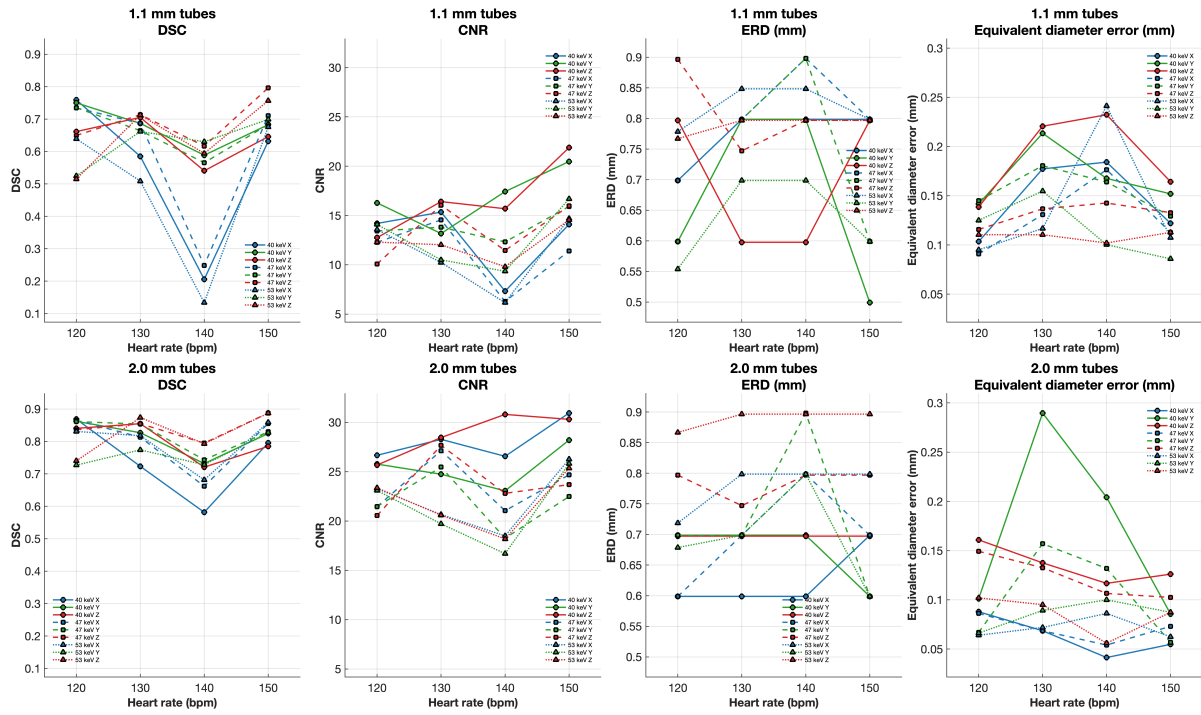


Figure 3.10: Flash within-protocol quantitative results. The figure shows representative trends in ERD, DSC, and CNR across reconstruction energy, heart rate, and tube diameter.

Figure 3.11 illustrates the local loss of segmented tube continuity observed at 140 bpm, corresponding to the quantitative deviations in the flash analysis.

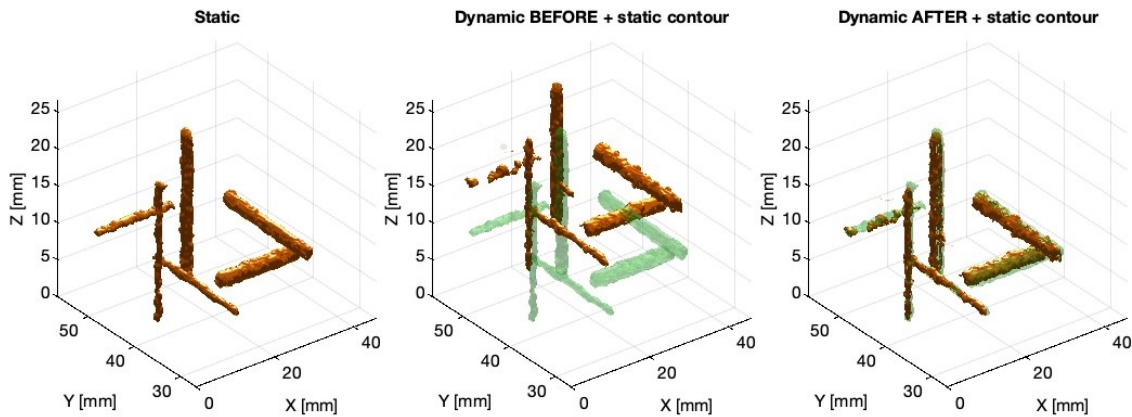


Figure 3.11: Representative three-dimensional visualization of the segmented tubes for the dynamic flash acquisition at 140 bpm reconstructed at 53 keV.

The mixed-effects model results are summarized in Table 3.7.

Table 3.7: Linear mixed-effects model results for the flash within-protocol analysis.

Endpoint	Effect	p-value	Statistical result
$E_{D_{eq}}$	keV	< 0.001	Significant
$E_{D_{eq}}$	Heart rate	0.003	Significant
$E_{D_{eq}}$	Tube diameter	0.013	Significant
E_A	keV	< 0.001	Significant
E_A	Heart rate	0.006	Significant
E_A	Tube diameter	0.638	Not significant
C	keV	0.926	Not significant
C	Heart rate	0.004	Significant
C	Tube diameter	0.125	Not significant
ERD	keV	0.002	Significant
ERD	Heart rate	0.018	Significant
ERD	Tube diameter	0.708	Not significant
B_{HU}	keV	0.210	Not significant
B_{HU}	Heart rate	< 0.001	Significant
B_{HU}	Tube diameter	0.751	Not significant
DSC	keV	0.169	Not significant
DSC	Heart rate	< 0.001	Significant
DSC	Tube diameter	< 0.001	Significant
CNR	keV	< 0.001	Significant
CNR	Heart rate	< 0.001	Significant
CNR	Tube diameter	< 0.001	Significant

Reconstruction energy significantly affected equivalent diameter error, cross-sectional area error, ERD, and CNR. Heart rate significantly affected equivalent diameter error, cross-sectional area error, ERD, DSC, circularity, attenuation bias, and CNR. Tube diameter significantly affected equivalent diameter error, DSC, and CNR.

Increasing reconstruction energy was associated with lower equivalent diameter error and lower cross-sectional area error. Holm-corrected post-hoc comparisons showed that equivalent diameter error was significantly lower at both 47 keV and 53 keV compared with 40 keV. Cross-sectional area error also decreased significantly with increasing reconstruction energy, with significant differences between 40 and 47 keV, 40 and 53 keV, and 47 and 53 keV.

In contrast, ERD was higher at 47 and 53 keV than at 40 keV. Holm-corrected post-hoc comparisons showed that ERD was significantly higher at 47 keV than at 40 keV and at 53 keV than at 40 keV, whereas no significant difference was observed between 47 and 53 keV. For CNR, 40 keV yielded significantly higher values than both 47 keV and 53 keV. Selected Holm-corrected post-hoc comparisons are summarized in Table 3.9.

Heart-rate effects were non-monotonic. The largest deviations were observed at 140 bpm, where DSC and circularity were lower and ERD was higher than at 120 bpm.

Within the flash protocol, directional differences remained small across reconstruction energies and heart rates. No x-, y-, or z-dominant error pattern was observed.

3.2.4. Within-protocol motion-robustness analysis: prospective

Within the prospective protocol, heart rate, reconstruction phase, and tube diameter were evaluated. Figure 3.12 shows the phase behavior of the prospective protocol across the reconstructed cardiac cycle. Plots of the remaining metrics are provided in Appendix G.3.5.

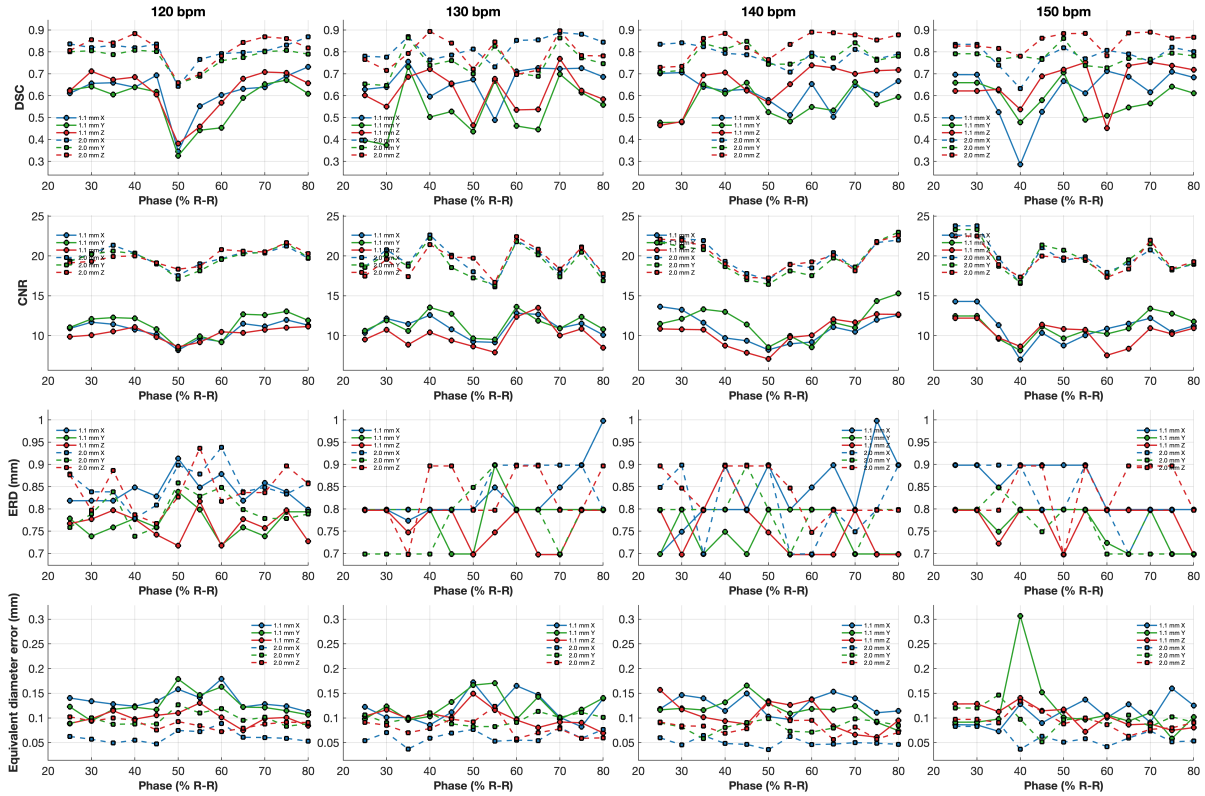


Figure 3.12: Prospective within-protocol quantitative results. The figure shows representative phase behavior for DSC, CNR, and attenuation bias across the reconstructed cardiac cycle.

Because the strongest prospective effect was the variation in the DSC for different reconstructed cardiac phases, representative 3D segmentations at unfavorable and favorable phases are shown in Figure 3.13 to illustrate the segmentation basis of this difference.

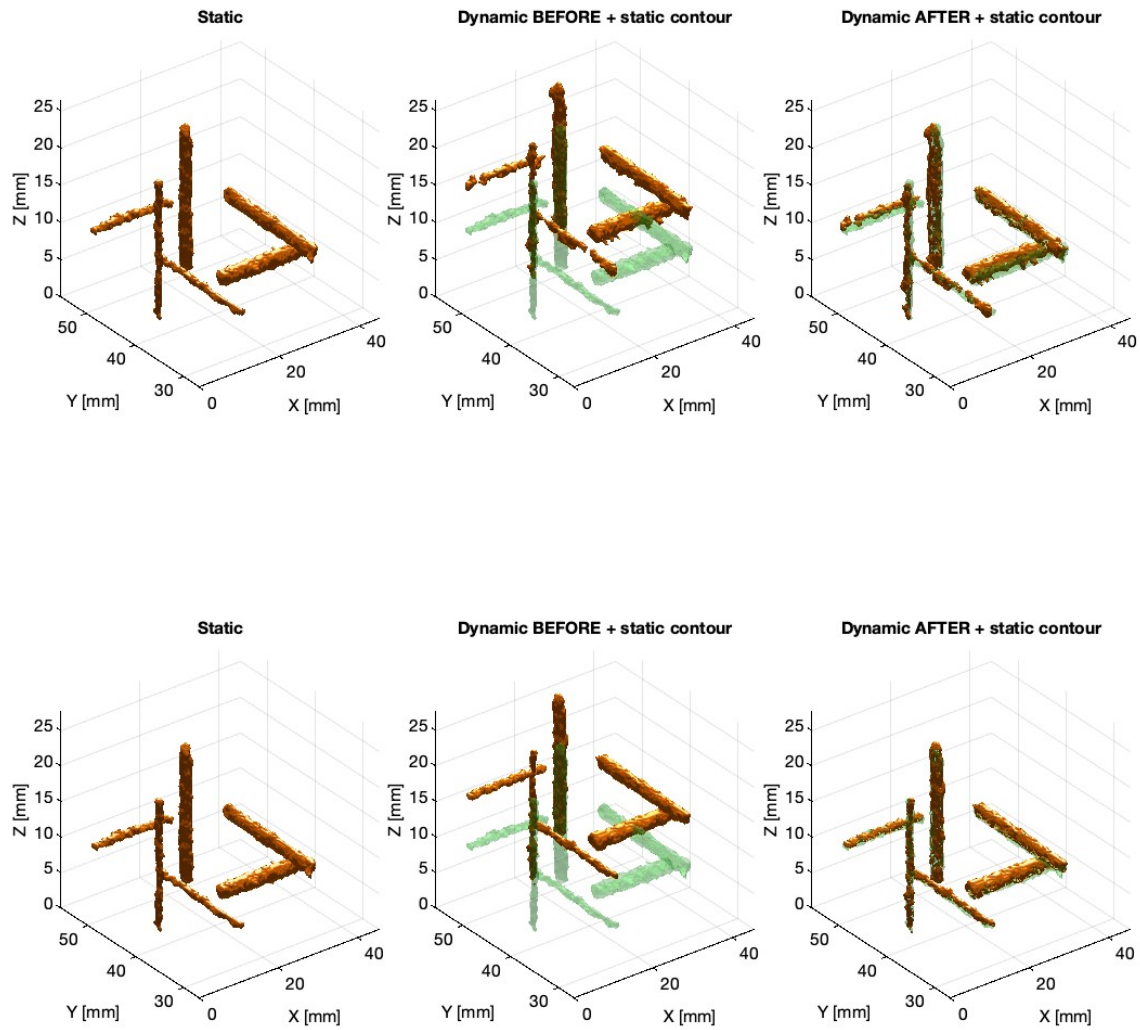


Figure 3.13: Representative three-dimensional visualizations of the segmented tubes in prospective acquisitions reconstructed at 150 bpm and 53 keV. The upper visualization shows the segmentation output at 40% of the R–R interval, and the lower visualization shows the segmentation output at 70% of the R–R interval. Each visualization includes the static segmentation, the dynamic segmentation before registration with the static contour, and the dynamic segmentation after registration with the static contour.

The mixed-effects model results are summarized in Table 3.8. Heart rate significantly affected circularity and attenuation bias. Reconstruction phase significantly affected equivalent diameter error, cross-sectional area error, DSC, circularity, attenuation bias, and CNR, while no significant phase effect was observed for ERD. Tube diameter significantly affected equivalent diameter error, DSC, circularity, and CNR.

The highest DSC values were observed in later reconstructed phases, particularly around 70–80% of the R–R interval. The lowest mean DSC occurred at 50% R–R. Holm-corrected post-hoc comparisons showed that DSC was significantly lower at 50% R–R than at 70% R–R ($p_{\text{Holm}} < 0.001$), 75% R–R ($p_{\text{Holm}} < 0.001$), and 80% R–R ($p_{\text{Holm}} = 0.001$).

In addition, Holm-corrected post-hoc comparisons showed significantly higher equivalent diameter error and cross-sectional area error at 50% R–R than at 75% R–R. Circularity also depended on reconstruction phase, with lower circularity at 40% R–R than at 80% R–R ($p_{\text{Holm}} = 0.024$).

Table 3.8: Linear mixed-effects model results for the prospective within-protocol analysis.

Endpoint	Effect	p-value	Statistical result
$E_{D_{eq}}$	Heart rate	0.261	Not significant
$E_{D_{eq}}$	Phase	0.011	Significant
$E_{D_{eq}}$	Tube diameter	0.005	Significant
E_A	Heart rate	0.064	Not significant
E_A	Phase	0.006	Significant
E_A	Tube diameter	0.684	Not significant
C	Heart rate	0.047	Significant
C	Phase	< 0.001	Significant
C	Tube diameter	< 0.001	Significant
ERD	Heart rate	0.251	Not significant
ERD	Phase	0.410	Not significant
ERD	Tube diameter	0.397	Not significant
B_{HU}	Heart rate	0.001	Significant
B_{HU}	Phase	< 0.001	Significant
B_{HU}	Tube diameter	0.964	Not significant
DSC	Heart rate	0.505	Not significant
DSC	Phase	< 0.001	Significant
DSC	Tube diameter	< 0.001	Significant
CNR	Heart rate	0.635	Not significant
CNR	Phase	< 0.001	Significant
CNR	Tube diameter	< 0.001	Significant

Attenuation bias differed significantly between 50% R–R and 70% R–R ($p_{\text{Holm}} < 0.001$). CNR was significantly lower at 50% R–R than at 25% R–R ($p_{\text{Holm}} < 0.001$), 30% R–R ($p_{\text{Holm}} < 0.001$), and 75% R–R ($p_{\text{Holm}} < 0.001$). Overall, reconstruction phase was the dominant variable within the prospective protocol, whereas heart rate showed fewer significant effects than phase. Selected Holm-corrected post-hoc comparisons for the prospective within-protocol analysis are summarized in Table 3.9.

3.2.5. Post-hoc comparisons

Selected Holm-corrected post-hoc comparisons for the main significant effects are summarized in Table 3.9. The complete set of significant Holm-corrected comparisons is provided in Appendix G.4.

Table 3.9: Selected Holm-corrected post-hoc comparisons illustrating the main significant effects. The estimated difference represents the model-estimated value of the first level minus the second level in each comparison.

Analysis	Metric	Comparison	Estimated difference	p_{Holm}
Flash	$E_{D_{eq}}$	40 vs 47 keV	0.027	0.020
Flash	$E_{D_{eq}}$	40 vs 53 keV	0.044	< 0.001
Flash	E_A	40 vs 47 keV	0.078	0.007
Flash	E_A	40 vs 53 keV	0.131	< 0.001
Flash	E_A	47 vs 53 keV	0.053	0.046
Flash	ERD	40 vs 47 keV	-0.066	0.008
Flash	ERD	40 vs 53 keV	-0.078	0.003
Flash	CNR	40 vs 47 keV	3.511	< 0.001
Flash	CNR	40 vs 53 keV	4.543	< 0.001
Prospective	$E_{D_{eq}}$	50 vs 75% R-R	0.025	0.010
Prospective	E_A	50 vs 75% R-R	0.053	0.007
Prospective	DSC	50 vs 70% R-R	-0.116	< 0.001
Prospective	DSC	50 vs 75% R-R	-0.099	< 0.001
Prospective	DSC	50 vs 80% R-R	-0.090	0.001
Prospective	C	40 vs 80% R-R	-0.008	0.024
Prospective	B_{HU}	50 vs 70% R-R	-36.821	< 0.001
Prospective	CNR	25 vs 50% R-R	2.515	< 0.001
Prospective	CNR	30 vs 50% R-R	3.110	< 0.001
Prospective	CNR	50 vs 75% R-R	-2.762	< 0.001
Static	ERD	40 vs 47 keV	-0.073	0.022
Static	ERD	40 vs 53 keV	-0.118	< 0.001
Static	CNR	40 vs 53 keV	5.870	< 0.001
Static	CNR	47 vs 53 keV	7.256	< 0.001

3.3. Radiation dose results

Scanner-reported $CTDI_{vol}$, DLP, SSDE, water-equivalent diameter (D_w), and effective scan length are summarized in Table 3.10. Here, D_w represents the water-equivalent patient size used for SSDE calculation. Values are reported as mean $\pm$ standard deviation for the repeated 120 bpm acquisitions and as single values for the remaining scan conditions.

For the static scans, flash and prospective acquisitions showed comparable DLP values, although the prospective scan had higher $CTDI_{vol}$ and SSDE. Because DLP reflects both dose per unit length and scan length, the comparable DLP with higher $CTDI_{vol}$ indicates a shorter effective scan length for the prospective acquisition. The single-phase prospective acquisition showed a similar dose level to the static prospective scan.

For the repeated dynamic acquisitions at 120 bpm, the prospective protocol resulted in substantially higher dose metrics than the flash protocol. $CTDI_{vol}$ was approximately 6.5 times higher for prospective than for flash, while DLP was approximately 3.3 times higher. This difference was accompanied by differences in effective scan length. Across the dynamic flash acquisitions, $CTDI_{vol}$ remained constant, whereas the prospective dynamic acquisitions showed slightly decreasing $CTDI_{vol}$, DLP, and SSDE values with increasing heart rate.

Table 3.10: Dose metrics for the evaluated PCCT protocols. Values are shown as mean $\pm$ standard deviation for the repeated 120 bpm scans and as single values for the static, single-phase prospective, 130 bpm, 140 bpm, and 150 bpm acquisitions.

Protocol	Condition	$CTDI_{vol}$ (mGy)	DLP (mGy·cm)	L_{eff} (cm)	$SSDE$ (mGy)	D_w (cm)
Flash	Static	0.22	2.51	11.4	0.56	10.5
Flash	120 bpm	0.22 ± 0	2.516 ± 0.005	11.4	0.558 ± 0.004	10.52 ± 0.04
Flash	130 bpm	0.22	2.51	11.4	0.55	10.5
Flash	140 bpm	0.22	2.51	11.4	0.55	10.5
Flash	150 bpm	0.22	2.51	11.4	0.55	10.6
Prospective	Static	0.38	2.17	5.7	1.01	9.82
Prospective	Single phase	0.39	2.23	5.7	1.04	9.79
Prospective	120 bpm	1.428 ± 0.008	8.22 ± 0.05	5.8	3.74 ± 0.02	9.786 ± 0.005
Prospective	130 bpm	1.33	7.67	5.8	3.49	9.82
Prospective	140 bpm	1.28	7.35	5.7	3.35	9.8
Prospective	150 bpm	1.17	6.74	5.8	3.08	9.8

4

Discussion

4.1. Summary of main findings

This study shows that the choice between flash and prospective ECG-synchronized PCCT cannot be based on a single image-quality metric. Under matched 53 keV dynamic conditions, both protocols showed broadly comparable geometric preservation, whereas the clearest matched difference was related to CNR. When the most favorable prospective phase was selected, prospective acquisition improved DSC, but this advantage did not extend consistently to the other endpoints. Therefore, the benefit of prospective acquisition appears to be metric-dependent rather than a uniform improvement in coronary artery visualization.

Within-protocol analyses showed that flash was primarily influenced by reconstruction energy and heart-rate-dependent acquisition timing, whereas prospective acquisition was primarily influenced by reconstruction phase. In the flash protocol, 40 keV provided the strongest iodine contrast and highest CNR, while 53 keV reduced equivalent diameter and cross-sectional area errors. In the prospective protocol, later reconstruction phases around 70–80% of the R–R interval were generally more favorable than mid-cycle phases, particularly around 50% of the R–R interval. Heart-rate effects were non-monotonic, indicating that image quality depended on the interaction between acquisition timing, motion phase, tube orientation, and z-position rather than on heart rate alone.

The 1.1 mm tubes consistently showed the largest deviations from the static measurements and should therefore be considered the limiting case for neonatal coronary origin visualization. Taken together, the findings support high-pitch flash acquisition as the preferred practical acquisition strategy in the evaluated setup. For the primary task of visual coronary origin assessment, 40 keV is favored as the main reconstruction because of its contrast and CNR advantage, while 53 keV remains useful as a complementary reconstruction for geometric or size-based assessment. Prospective acquisition may be useful when phase-resolved assessment is needed, but this should be weighed against the higher radiation dose and the additional acquisition and reconstruction complexity.

4.2. Interpretation of the repeatability analysis

The repeatability analysis was used to assess measurement stability and to provide a reference for interpreting the subsequent protocol comparisons. The repeatability coefficient was used to assess whether observed differences were large relative to scan-to-scan variation. This was particularly relevant for the prospective reconstruction phases, where some phase differences were small and therefore required cautious interpretation. In contrast, trends that were consistent across endpoints, tube diameters, orientations, and visual reconstructions were considered less likely to be explained by repeatability limitations alone.

Circularity was the most stable metric across repeated acquisitions, whereas DSC showed greater variability, especially for the 1.1 mm tubes and for mid-cycle prospective reconstruction phases. This difference is consistent with the way the metrics respond to small segmentation changes. Circularity describes the shape of each segmented cross-section and is therefore less affected by small positional shifts, provided that the segmented tube remains approximately round. In contrast, DSC depends on the spatial overlap with the reference segmentation and is therefore more sensitive to small mismatches between repeated acquisitions. The higher DSC variability around 50% of the R–R interval suggested that this reconstruction phase was less stable for repeated prospective measurements, possibly because residual motion or phase-timing differences had a larger effect on the spatial overlap.

Edge-rise distance was large relative to the investigated tube diameters. Across the analyzed conditions, mean ERD was approximately 0.7 mm for flash acquisition, corresponding to about 64% of the 1.1 mm tube diameter and 35% of the 2.0 mm tube diameter. For the prospective acquisition, mean ERD was higher, approximately 0.8 mm, while differences between reconstruction phases remained small. This suggests that edge blurring was present throughout the reconstructed phases rather than being limited to a specific reconstruction phase. Because the edge transition occupied a substantial part of the smallest tube cross-section, the 1.1 mm tubes were more sensitive to partial-volume averaging, motion-related blurring, segmentation variability, and local visibility loss than the 2.0 mm tubes. The observed size dependence across repeatability metrics therefore shows that measurement stability was not determined by acquisition protocol alone, but also by the relation between tube diameter and effective spatial resolution.

Within the prospective protocol, reconstruction phase affected repeatability. The mid-cycle region around 50% of the R–R interval showed greater variability, whereas later phases were generally more stable. This pattern is consistent with later reconstruction phases corresponding more closely to diastole, when cardiac motion is reduced and tube position is more reproducible between repeated acquisitions. However, because some differences between reconstructed phases were close to the observed repeatability range, small differences between individual phases should not be overinterpreted as evidence for a truly superior reconstruction phase. The repeatability results therefore support broader comparisons between mid-cycle and later phases rather than precise ranking of adjacent reconstruction phases.

Overall, the repeatability analysis shows that the limiting factor for the smallest structures was not a single reconstruction phase or metric, but the interaction between finite spatial resolution, edge blurring, residual motion sensitivity, and segmentation uncertainty. Consequently, subsequent protocol comparisons were considered robust only when the observed differences exceeded the repeatability-related variation and were supported by consistent trends across metrics, tube diameters, orientations, and visual reconstructions.

4.3. Interpretation of the static inter-protocol comparison

The static inter-protocol comparison was used to separate baseline acquisition and reconstruction effects from motion-induced artifacts. Because no motion was present, differences observed in the static scans cannot be attributed to cardiac motion and should instead be interpreted as baseline effects of acquisition mode, reconstruction energy, tube size, segmentation, and image analysis.

In the absence of motion, flash and prospective acquisition showed broadly comparable representation of tube diameter, cross-sectional area, circularity and contrast-to-noise ratio. This indicates that both protocols reproduced the phantom structures similarly under stationary conditions. Therefore, the static results do not suggest a clear baseline advantage of either protocol for the basic visualization of coronary-sized tubes.

The main static difference between protocols was observed for the sharpness of the transition between the contrast-filled tube and the surrounding background. In the raw data plots, flash and prospective ERD values overlapped considerably, in particular at 40 keV, where protocol differences were small for both tube diameters. At higher reconstruction energies, the separation became more visible, particularly for the 2.0 mm tubes at 53 keV, where mean prospective ERD was approximately 0.15 mm higher than flash. The mixed-effects model summarized this pattern as a significant protocol effect, indicating that prospective acquisition was associated with a wider static edge transition.

The repeatability coefficients support a cautious interpretation of this finding. For flash acquisition, the ERD RC was 0.136 mm for the 1.1 mm tubes and 0.124 mm for the 2.0 mm tubes, meaning that most mean static protocol differences were within the expected repeatability range. For prospective acquisition, the interpretation is less straightforward because the repeatability coefficient was phase-dependent; however, the RCs were generally of a similar or larger magnitude than the observed static protocol differences. The ERD difference should therefore be interpreted as a systematic but modest baseline shift between protocols.

Reconstruction energy affected CNR and ERD, consistent with the expected energy dependence of iodine attenuation. However, improved contrast did not consistently translate into more accurate geometric measurements, because segmentation-derived geometry depends not only on contrast but also on edge sharpness, image noise, partial-volume effects, and boundary localization. At lower VMI energies, increased iodine attenuation may improve tube visibility while also affecting noise and the apparent edge transition, which can shift the segmented contour. This distinction was especially relevant for the 1.1 mm tubes, where partial-volume effects and small boundary shifts have a relatively large influence on the measured tube size.

Descriptively, equivalent diameter error tended to decrease at higher keV for the 1.1 mm tubes, whereas the opposite trend was observed for the 2.0 mm tubes. For the smallest tubes, this pattern is consistent with lower-keV reconstructions increasing the apparent tube size through stronger iodine contrast, edge blooming, or threshold-dependent segmentation effects. In contrast, for the 2.0 mm tubes, equivalent diameter error increased at higher-keV, which is more consistent with the reduction in CNR at higher reconstruction energies. This indicates that the effect of reconstruction energy on geometric accuracy was size-dependent.

The mixed-effects model did not show a significant main effect of protocol for CNR, however the descriptive plots showed an energy-dependent reversal: CNR was higher for flash at 40 keV, whereas it was higher for prospective acquisition at 47 and 53 keV. These opposing directions may partly cancel when the additive model estimates one average protocol effect across all keV levels and tube diameters. Therefore, the absence of a significant main protocol effect should not be interpreted as evidence that both protocols behaved identically under all reconstruction conditions.

The RC analysis supports this interpretation. Within flash acquisition, the decrease in CNR for the 2.0 mm tubes from approximately 26 at 40 keV to approximately 13 at 53 keV exceeded the CNR repeatability coefficients for both tube diameters, indicating that this energy-dependent change exceeded expected scan-to-scan variation. In contrast, protocol differences at individual energy levels were more condition-dependent and were not consistently separated from the repeatability range. Overall, CNR appeared to be governed mainly by the interaction between reconstruction energy, acquisition mode, tube diameter, and noise behavior rather than by a single protocol effect.

These static findings provide the baseline for interpreting the dynamic analyses. Because the 0 bpm acquisitions were not affected by cardiac motion, differences observed in the static scans reflect baseline effects of acquisition mode and reconstruction energy. Therefore, when a difference is absent or small in the static condition but appears or increases during dynamic acquisitions, this additional degradation is more likely to be related to motion than to baseline protocol differences alone.

4.4. Interpretation of the dynamic inter-protocol comparison

The dynamic inter-protocol comparison consisted of two complementary analyses. The z-matched phase comparison evaluated the protocols under the most standardized conditions, whereas the best-phase comparison evaluated the additional optimization potential of prospective acquisition when favorable reconstruction phases were selected.

4.4.1. Z-matched phase inter-protocol comparison

The z-matched phase comparison represents the most standardized inter-protocol evaluation, because flash and prospective acquisitions were compared at the same reconstruction energy and at the closest available prospective phase. Under the matched 53 keV conditions, the protocols were broadly comparable for most geometric endpoints. Small differences visible in the raw data plots were generally within the corresponding repeatability ranges. This indicates that neither acquisition strategy showed

a robust advantage in size-based or shape-based geometric preservation under matched conditions.

The main matched inter-protocol difference was CNR. The mixed-effects model showed a significant protocol effect, with flash acquisition having higher CNR on average than prospective acquisition. However, the raw data plots show that this separation was condition-dependent rather than uniform across all heart rates and tube diameters. The flash advantage was most pronounced at 150 bpm, where the mean CNR difference between the protocols of approximately 5.5 for the 1.1 mm and approximately 6 for the 2.0 mm tubes, exceeded the RC range and difference for both tube diameters, whereas differences at 130 and 140 bpm were smaller and mostly within repeatability limits.

Heart-rate effects were non-monotonic. Several endpoints showed less favorable behavior around 140 bpm, whereas some measurements recovered at 150 bpm. For flash acquisition, the DSC repeatability coefficient was relatively large, particularly for the 1.1 mm tubes, and the mean increase in equivalent diameter error at 140 bpm remained within the corresponding repeatability coefficient. Therefore, the 140 bpm dip should not be interpreted as a definitive heart-rate ranking.

Nevertheless, the 140 bpm flash behavior remains relevant because the raw data plots showed a local deviation in the X-oriented 1.1 mm tube that exceeded the equivalent diameter error repeatability range and coincided with visual loss of local tube visibility in the dynamic flash reconstruction (Figure 3.11). This suggests a local instability in tube representation rather than an isolated change in the overlap metric. The possible mechanism, related to z-dependent flash timing and phase sampling, is discussed further in the within-protocol flash analysis in Section 4.5.1.

Overall, the matched comparison suggests that clinically relevant visualization cannot be judged from CNR alone. Under the matched phase conditions at 53 keV, flash may offer a visibility advantage in selected settings, but the matched data do not show a consistent overall superiority of either protocol for coronary-sized tube visualization.

4.4.2. Best-phase inter-protocol comparison

The main improvement in the best-phase comparison was observed for DSC, which was significantly higher for prospective acquisition. This suggests that phase selection improved overlap-based preservation and continuity of the segmented tube volume. However, the magnitude of this advantage should be interpreted cautiously in relation to repeatability. The flash DSC repeatability coefficient was relatively large, especially for the 1.1 mm tubes, meaning that the exact depth of the flash DSC dip around 140 bpm could change if the acquisition were repeated. In that case, the estimated DSC advantage of prospective acquisition might become smaller.

The DSC advantage did not translate into a broad geometric advantage. This distinction reflects the different meaning of the metrics. DSC penalizes displacement, discontinuity, and incomplete preservation of the segmented volume, whereas equivalent diameter and cross-sectional area can still be calculated from remaining visible cross-sections. Figure 3.13 shows this behavior: in less favorable phases, some tubes were locally interrupted or incompletely segmented, whereas later phases showed more coherent tube preservation. The DSC improvement should therefore be interpreted as improved overlap-based preservation rather than uniform improvement of all geometric endpoints.

For protocol selection, the DSC advantage of prospective reconstruction at the best phase should not be interpreted as general superiority of the protocol. The advantage was phase-dependent and mainly reflected improved spatial coherence. Clinically, this distinction is relevant because neonatal TGA coronary assessment depends on reliable visualization of very small coronary structures within a short and predictable acquisition. An improvement in DSC does not automatically mean that the coronary artery would be more consistently identifiable in clinical practice. Therefore, the potential benefit of prospective phase selection needs to be weighed against other clinically relevant factors, including contrast, geometric preservation, radiation exposure, acquisition simplicity, and the favorable iodine contrast of low-keV flash reconstructions.

4.5. Interpretation of the within-protocol analyses

4.5.1. Flash within-protocol motion-robustness analysis

Within the flash protocol, reconstruction energy introduced a trade-off between contrast and segmentation-based geometric accuracy. Lower-keV reconstructions increased iodine attenuation and tube-to-background separation, which was reflected by higher CNR and lower ERD at 40 keV. The contrast advantage of 40 keV was also robust relative to repeatability. In Figure 3.10, mean CNR over all heart rates decreased from approximately 15 at 40 keV to 12 at 53 keV in the 1.1 mm tubes and from 27 at 40 keV to 22 at 53 keV in the 2.0 mm tubes. These diameter-specific decreases exceeded the corresponding flash CNR repeatability coefficients of 2.98 and 3.94, supporting the practical relevance of the 40 keV contrast advantage.

However, stronger iodine signal can also broaden the visible high-attenuation region and shift threshold-based boundaries outward, especially in the 1.1 mm tubes, where the edge transition occupies a large fraction of the object diameter. This provides a plausible explanation for why 40 keV improved contrast and apparent edge definition, whereas 53 keV was associated with smaller equivalent diameter and cross-sectional area errors. In contrast to CNR, these geometric and ERD differences were statistically significant but generally smaller than, or close to, the corresponding repeatability coefficients. They should therefore be interpreted as systematic but modest shifts in segmentation behavior rather than large practical differences in true tube geometry.

Heart rate also influenced flash performance, but non-monotonically. The raw data plots showed less favorable behavior around 140 bpm and recovery at 150 bpm, especially for DSC, CNR, circularity, and attenuation bias. The model detected significant heart-rate effects for several endpoints, but the 140 bpm DSC reduction remained within the relatively large DSC repeatability range. Therefore, if the 140 bpm acquisition were repeated, the exact depth of the DSC dip could change, and the statistical strength of the heart-rate effect might be reduced. The 140 bpm acquisition remains interesting because it showed consistent signs of local tube degradation across several endpoints and visual assessment. This suggests timing-sensitive instability in flash imaging rather than random variation in one metric alone. This non-monotonic behavior is plausible because flash acquisition samples a specific part of the motion trajectory, and the effective reconstruction phase varies with z-position. Image quality therefore depends on the interaction between cardiac motion, acquisition timing, tube orientation, and the z-dependent reconstruction phase. A higher heart rate may therefore be less detrimental than an intermediate heart rate if the acquisition window coincides with a more quiescent part of the motion cycle. This interpretation is supported by the prospective phase data, in which the mid-cycle region around 50% R–R was generally less favorable than later phases around 70–80% R–R.

Tube diameter affected flash performance mainly through detectability and segmentation stability. The 2.0 mm tubes showed higher DSC and CNR than the 1.1 mm tubes, whereas the smallest tubes were more susceptible to partial-volume averaging, local visibility loss, and segmentation instability. This is clinically relevant because neonatal coronary assessment concerns structures close to the spatial resolution limit, where small absolute boundary deviations can have large proportional effects. The 1.1 mm tubes therefore represent the most demanding test condition and should weigh strongly in the interpretation of protocol robustness.

Overall, flash imaging did not show a single optimal reconstruction energy. The 40 keV reconstruction was more favorable for CNR and apparent edge definition, whereas 53 keV was more favorable for size-based geometric accuracy. The most robust practical advantage was contrast-related; several geometric differences were statistically detectable but remained close to repeatability limits.

4.5.2. Prospective within-protocol motion-robustness analysis

Within the prospective protocol, reconstruction phase had a stronger influence on image quality than heart rate. Phase was significantly associated with DSC, CNR, circularity and attenuation bias, whereas heart rate was not significant for most endpoints. This indicates that, under prospective acquisition, image quality was mainly determined by the selected cardiac phase rather than by heart rate alone.

However, the phase effects should be interpreted in relation to the repeatability coefficients. Several small differences between neighboring reconstruction phases were within the corresponding RC ranges, meaning that these differences could plausibly vary if the measurements were repeated.

Therefore, the significant phase effect does not justify a ranking of every individual phase. Instead, the more robust interpretation is that the mid-cycle region around 50% R–R was generally less favorable, with lower DSC and CNR and stronger attenuation deviations, whereas later phases around 70–80% R–R were generally more stable.

This phase dependence was most evident for DSC and CNR, which both changed significantly with reconstruction phase. However, comparison with the repeatability coefficients showed that not all individual phase-to-phase differences were robust. For DSC, several differences remained within the relatively large repeatability range, so the exact magnitude of the mid-cycle dip should not be overinterpreted. For CNR, the broader reduction around mid-cycle and recovery at later phases was more consistent and in some conditions exceeded the repeatability coefficient, but this was not uniform across all heart rates and tube diameters.

Tube diameter remained an important limiting factor. The 2.0 mm tubes showed consistently higher CNR than the 1.1 mm tubes, and this separation was generally larger than the CNR repeatability coefficients, supporting a robust diameter-related difference in visibility. DSC was also higher for the 2.0 mm tubes, although the magnitude of this difference should be interpreted more cautiously because the DSC repeatability range was relatively large, especially around less stable phases. Overall, the smaller 1.1 mm tubes were more sensitive to partial-volume averaging, local visibility loss, and boundary uncertainty. The prospective results therefore suggest that suitable phase selection can improve spatial coherence and visibility, but that the reliability of small differences between the reconstructed phases remains limited by repeatability and tube size.

Overall, prospective acquisition showed clear phase-dependent optimization potential. Later reconstruction phases generally improved overlap and contrast, whereas mid-cycle phases were more vulnerable to motion-related blurring and incomplete segmentation. This supports the selective value of prospective imaging when specific phase information is clinically needed.

4.6. Dose and protocol trade-off

The dose results demonstrated a clear difference between the two acquisition strategies. In the static condition and in the prospective single-phase acquisition, prospective imaging showed higher CTDI_{vol} and SSDE than flash, whereas DLP was slightly lower. In the dynamic acquisitions, however, prospective imaging resulted in substantially higher dose metrics than flash across all heart rates. This difference should be interpreted in relation to acquisition design rather than as a direct dose penalty associated with image quality alone.

The temporal flexibility is one of the strengths of prospective acquisition, but it is intrinsically linked to increased temporal sampling and therefore higher radiation exposure. Even when the smallest prospective acquisition window was used, the acquired data still permitted reconstruction over a wider range, of about 25% to 50% of the R–R interval, than flash under comparable conditions. In prospective sequential imaging, radiation exposure is determined by the selected cardiac phase window and, in clinical multi-stack acquisitions, also by the number of table positions required. In the present phantom setup, the object fitted within a single stack, but the wider temporal exposure window still resulted in higher dynamic dose metrics than flash.

The decrease in prospective dose metrics with increasing heart rate is consistent with the shorter absolute duration of the selected cardiac-cycle window at higher heart rates. In contrast, flash dose metrics remained low and nearly constant across heart rates, reflecting the more fixed acquisition design of the high-pitch scan. The relationship between DLP and CTDI_{vol} further supports this interpretation. The effective scan length for the flash acquisition resulted in 11.4 cm, while for the prospective static and single phase these were 5.7 cm. Although the nominal reconstructed anatomical range was kept comparable between protocols, the effective dose-relevant scan extent differed between acquisition modes. Therefore, comparable reconstructed coverage does not necessarily imply equivalent radiation burden.

4.7. Protocol recommendation

The results of this study indicate that protocol selection for neonatal coronary origin imaging should be guided by the balance between visibility, geometric preservation, radiation dose, and practical imple-

mentation. No single acquisition or reconstruction setting optimized all quantitative endpoints. However, when the findings are interpreted in relation to the primary clinical task, visual identification of the coronary origins in neonatal TGA, high-pitch ECG-triggered flash PCCT provides the most favorable overall strategy.

The acquisition recommendation is based on the matched 53 keV comparison and the dose results. Flash acquisition preserved geometry broadly comparably to prospective acquisition, while requiring substantially lower dynamic dose indices. The main protocol-related image-quality difference was observed for CNR, for which flash showed more favorable contrast behavior, particularly at higher heart rates. These findings support flash acquisition as the preferred default acquisition strategy in the evaluated phantom setup.

Prospective acquisition provides phase-selection flexibility and showed optimization potential for overlap-based preservation, particularly for DSC. However, this benefit was not consistent across all geometric endpoints and depended on selecting a favorable phase. In addition, prospective acquisition resulted in a higher radiation burden and greater workflow complexity. In neonates, such phase flexibility must provide a clear diagnostic advantage to be justified. Therefore, the additional dose burden does not support prospective acquisition as the routine default. Instead, prospective acquisition may be reserved for selected cases in which information from a specific cardiac phase is expected to improve interpretability, or as an optimization tool to identify favorable motion phases. The phase-resolved prospective results indicated that later phases, approximately 70–80% of the R–R interval, were generally more favorable than mid-cycle phases. This suggests that future optimization of the flash protocol should focus on ECG trigger timing and z-positioning, to determine whether the coronary origins can be placed closer to a favorable motion phase while preserving the low-dose advantage of high-pitch flash acquisition.

For reconstruction, 40 keV is recommended as the primary series for visual coronary origin assessment. This recommendation follows from the within-flash reconstruction energy analysis, where 40 keV provided the strongest iodine contrast and therefore the highest visibility of the tube structures. Because the clinical aim is primarily to identify the coronary origin rather than to perform precise lumen measurements, this contrast advantage is considered more relevant than small differences in segmentation-derived geometric accuracy. A 53 keV reconstruction should nevertheless be retained as a complementary series when geometric preservation, size-based assessment, or comparison with reference imaging is required.

Overall, the dose and image-quality balance favors low-keV flash acquisition for routine neonatal TGA coronary origin visualization, with prospective acquisition reserved for selected cases where phase-resolved information is expected to provide additional diagnostic value. The recommended protocol is summarized in Table 4.1.

Table 4.1: Recommended acquisition and reconstruction strategy derived from the study findings. The selected protocol and VMI energy level are shown in bold, while parameters kept constant are shown in regular font.

Parameter	Recommendation
→ Protocol	high-pitch flash prospective sequential
Tube potential	70 kV
Effective mAs	16
CARE keV IQ level	60
Rotation time	250 ms
Pitch	3.2
Collimation	144 × 0.4 mm
ECG triggering	160 ms
Reconstructed slice thickness	0.4 mm
Increment	0.2 mm
Matrix	512 × 512
FOV	102 × 102 mm
Kernel	Bv48
QIR strength	4
→ VMI level	40 47 53 keV

4.8. Limitations

Several limitations should be considered. These limitations do not invalidate the protocol comparison, but define the level at which the conclusions should be applied: the study provides evidence on protocol behavior under simulated neonatal cardiac motion, rather than direct evidence of clinical diagnostic performance.

First, this was a phantom study and therefore did not represent the full anatomical and interpretive complexity of in vivo neonatal cardiac PCCT. The tubes were simplified structures rather than full coronary branches embedded within realistic cardiac anatomy. The results should therefore be interpreted as a controlled assessment of protocol behavior under simulated motion, rather than as a direct measure of diagnostic performance.

Second, the number of independent tube structures was limited. Although mixed-effects models were used to account for repeated observations within tube diameter, the inferential strength remains constrained by the small number of structural units. For this reason, statistical results should be interpreted together with descriptive trends, effect directions, visual assessment, and technical plausibility.

Third, the investigated structures were small relative to the image sampling and effective spatial resolution. The reconstructed field of view was 102×102 mm with a 512×512 matrix, corresponding to an in-plane pixel size of approximately 0.20 mm. The 1.1 mm and 2.0 mm tubes were therefore represented by approximately 5.5 and 10 pixels across the diameter, respectively. Although pixel size does not determine the true spatial resolution by itself, this limited sampling means that small changes in apparent edge position, thresholding, interpolation, system blurring, motion, or partial-volume averaging could substantially affect equivalent diameter, cross-sectional area, circularity, ERD, and DSC. This limitation is particularly relevant for the 1.1 mm tubes, which are closest to the clinically relevant size range and most sensitive to small boundary deviations.

Fourth, the segmentation-derived metrics depended on the quality of the static reference and the ability to detect a coherent tube volume in the dynamic images. The static segmentation was not a perfect geometric ground truth, and some dynamic tubes were locally interrupted or incompletely visible. In these cases, quantitative metrics may reflect the feasibility of detection and segmentation as much as the underlying acquisition performance. Visual reconstructions, as shown in Figure 3.4, were therefore used to support the interpretation of metric deviations.

Fifth, the optimization spaces were not identical across protocols. The study provides stronger evidence for phase optimization within prospective imaging than for dynamic energy optimization within prospective imaging, because the dynamic prospective analysis was centered on phase behavior at 53 keV. Conversely, the role of reconstruction energy was demonstrated most clearly within flash and in the static comparison. Conclusions regarding protocol-specific optimization are therefore asymmetrical. In addition, best-phase comparison represents an estimate of optimization potential rather than direct prospective clinical performance. Because the favorable phase was selected after reconstruction, this analysis may overestimate the performance achievable when a single phase must be selected before image review.

Finally, no formal reader study was performed, and the contrast preparation was based on a small clinical reference set. The study can therefore support conclusions regarding quantitative motion robustness, segmentation-derived geometry, and attenuation-related behavior, but it cannot directly establish clinical preference, diagnostic confidence, or observer performance.

Conclusion and future direction

This study investigated ECG-synchronized photon-counting CT strategies for neonatal coronary origin visualization in transposition of the great arteries. In the evaluated phantom setup, the findings support high-pitch ECG-triggered flash PCCT as the preferred practical acquisition strategy. The recommendation is based on the assumption that improved visibility and continuity of the small contrast-filled tubes in the phantom reflects improved visibility of the coronary origins and proximal coronary course in the clinical setting. In other words, when the full tube structure remains visible and traceable despite motion, the coronary origin is more likely to be identifiable. For this primary diagnostic task, 40 keV VMI is preferred because it increased iodine contrast and CNR, thereby improving tube conspicuity. A 53 keV reconstruction should be retained as a complementary series when geometric preservation, size-based assessment, or comparison with reference imaging is important.

The preference for flash acquisition is based on its favorable balance between image quality, radiation dose, and practical applicability. Under matched 53 keV conditions, flash and prospective acquisition showed broadly comparable geometric preservation, while flash resulted in substantially lower dynamic dose metrics. Prospective acquisition showed phase-dependent value, particularly through improved overlap-based preservation at favorable phases, but this benefit was not consistent across all endpoints and came with higher radiation dose, greater phase dependence, and increased acquisition complexity. Prospective acquisition should therefore be considered selectively, rather than as the default strategy.

Future work should focus on optimizing the selected flash protocol. Because high-pitch flash acquisition samples different z-positions at slightly different cardiac phases, ECG trigger timing and z-positioning should be investigated to determine whether the coronary origins can be placed closer to the favorable motion phase suggested by the prospective results, approximately 70–80% of the R–R interval. Further validation in anatomically realistic neonatal cardiac phantoms, patient data, and reader studies is needed to determine whether the quantitative advantage of 40 keV flash imaging translates into improved diagnostic confidence for coronary origin identification in TGA.

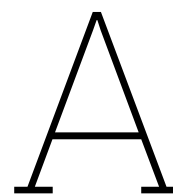
References

- [1] J. Xu, Q. Li, L. Deng, J. Xiong, Z. Cheng, and C. Ye. "Global, regional, and national epidemiology of congenital heart disease in children from 1990 to 2021". In: *Frontiers in Cardiovascular Medicine* Volume 12 - 2025 (2025). DOI: 10.3389/fcvm.2025.1522644. URL: <https://www.frontiersin.org/journals/cardiovascular-medicine/articles/10.3389/fcvm.2025.1522644>.
- [2] R. Sachdeva, A. K. Armstrong, R. Arnaout, L. Grosse-Wortmann, B. K. Han, L. Mertens, R. A. Moore, L. J. Olivieri, A. Parthiban, and A. J. Powell. "Novel Techniques in Imaging Congenital Heart Disease: JACC Scientific Statement". In: *Journal of the American College of Cardiology* 83.1 (2024), pp. 63–81. ISSN: 0735-1097. DOI: <https://doi.org/10.1016/j.jacc.2023.10.025>. URL: <https://www.sciencedirect.com/science/article/pii/S0735109723079214>.
- [3] E. J. Zucker. "Cardiac Computed Tomography in Congenital Heart Disease". In: *Radiologic Clinics of North America* 62.3 (2024). Advances and Innovations in Cardiovascular Imaging, pp. 435–452. ISSN: 0033-8389. DOI: <https://doi.org/10.1016/j.rcl.2023.12.015>. URL: <https://www.sciencedirect.com/science/article/pii/S003383892300235X>.
- [4] S. Moscatelli, M. Avesani, N. Borrelli, J. Sabatino, V. Pergola, I. Leo, C. Montanaro, F. V. Contini, G. Gaudieri, J. Ielapi, R. Motta, M. A. Merrone, and G. Di Salvo. "Complete Transposition of the Great Arteries in the Pediatric Field: A Multimodality Imaging Approach". In: *Children* 11.6 (2024), p. 626. DOI: 10.3390/children11060626.
- [5] V. Shukla, R. M. Freedom, and M. D. Black. "Single coronary artery and complete transposition of the great arteries: a technical challenge resolved?" In: *The Annals of Thoracic Surgery* 69.2 (2000), pp. 568–571. ISSN: 0003-4975. DOI: [https://doi.org/10.1016/S0003-4975\(99\)01324-7](https://doi.org/10.1016/S0003-4975(99)01324-7). URL: <https://www.sciencedirect.com/science/article/pii/S0003497599013247>.
- [6] T. R. Karl, A. Cochrane, and C. P. Brizard. "Arterial Switch Operation. Surgical Solutions to Complex Problems". In: *Texas Heart Institute Journal* 24.4 (1997), pp. 322–333.
- [7] P. Martins, V. Tran, G. Price, V. Tsang, and A. C. Cook. "Extending the surgical boundaries in the management of the left ventricular outflow tract obstruction in discordant ventriculo-arterial connections – a surgical and morphological study". In: *Cardiology in the Young* 18.2 (2008), pp. 124–134. DOI: 10.1017/S1047951108001947.
- [8] T. Flohr, B. Schmidt, S. Ulzheimer, and H. Alkadhi. "Cardiac imaging with photon counting CT". In: *British Journal of Radiology* 96.1152 (Oct. 2023), p. 20230407. ISSN: 0007-1285. DOI: 10.1259/bjr.20230407. URL: <https://doi.org/10.1259/bjr.20230407>.
- [9] C. H. McCollough, S. Leng, L. Yu, and J. G. Fletcher. "Dual- and Multi-Energy CT: Principles, Technical Approaches, and Clinical Applications". In: *Radiology* 276.3 (2015), pp. 637–653. DOI: 10.1148/radiol.2015142631. URL: <https://doi.org/10.1148/radiol.2015142631>.
- [10] M. Deleu, J. B. Maurice, L. Devos, M. Remy, and F. Dubus. "Image Quality Analysis of Photon-Counting CT Compared with Dual-Source CT: A Phantom Study for Chest CT Examinations". In: *Diagnostics* 13.7 (2023), p. 1325. DOI: 10.3390/diagnostics13071325.
- [11] F. Stalhammar, M. L. Aurumskjold, S. Meyer, M. Wiklund, P. Wingren, P. Liuba, and E. Hedstrom. "Photon-counting computed tomography for paediatric congenital heart defects yields images of high diagnostic quality with low radiation doses at both 70 kV and 90 kV". In: *Pediatric Radiology* 54.7 (Dec. 2024), pp. 1187–1196. DOI: 10.1007/s00247-024-05939-z. URL: <https://link.springer.com/article/10.1007/s00247-024-05939-z>.

- [12] K. Rajendran, M. Bruesewitz, J. Swicklik, A. Ferrero, J. Thorne, L. Yu, C. McCollough, and S. Leng. "Task-based automatic keV selection: leveraging routine virtual monoenergetic imaging for dose reduction on clinical photon-counting detector CT". In: *Physics in Medicine & Biology* 69.11 (May 2024), p. 115029. DOI: 10.1088/1361-6560/ad41b3. URL: <https://pubmed.ncbi.nlm.nih.gov/38648795/>.
- [13] A. Strassl, F. Lauriero, M. A. Rueda, C. Wassipaul, M. Weber, C. Loewe, D. Beitzke, and L. Beer. "High-pitch photon-counting detector computed tomography angiography of the coronary arteries: Qualitative and quantitative evaluation of monoenergetic image reconstructions". In: *European Journal of Radiology Open* 15 (2025), p. 100666. ISSN: 2352-0477. DOI: <https://doi.org/10.1016/j.ejro.2025.100666>. URL: <https://www.sciencedirect.com/science/article/pii/S2352047725000334>.
- [14] S. Tao, K. Rajendran, W. Zhou, J. G. Fletcher, C. H. McCollough, and S. Leng. "Improving iodine contrast to noise ratio using virtual monoenergetic imaging and prior-knowledge-aware iterative denoising (mono-PKAID)". In: *Physics in Medicine & Biology* 64.10 (May 2019), p. 105014. DOI: 10.1088/1361-6560/ab17fa. URL: <https://doi.org/10.1088/1361-6560/ab17fa>.
- [15] G. Jost, M. McDermott, R. Gutjahr, T. Nowak, B. Schmidt, and H. Pietsch. "New Contrast Media for K-Edge Imaging With Photon-Counting Detector CT". In: *Investigative Radiology* 58.7 (2023). URL: https://journals.lww.com/investigativeradiology/fulltext/2023/07000/new_contrast_media_for_k_edge_imaging_with.10.aspx.
- [16] S.-Y. Wang, W. Gao, Y.-M. Zhong, A.-M. Sun, Q. Wang, L.-W. HU, H.-S. Qiu, and J.-Y. Li. "Prospective ECG-triggering cardiac CT for infants with complex congenital heart disease using low-dose contrast medium, low tube voltage, and adaptive statistical iterative reconstruction". In: *Clinical Radiology* 72.6 (2017), pp. 502–507. ISSN: 0009-9260. DOI: <https://doi.org/10.1016/j.crad.2017.01.017>. URL: <https://www.sciencedirect.com/science/article/pii/S0009926017300582>.
- [17] M. Zheng, H. Zhao, J. Xu, Y. Wu, and J. Li. "Image quality of ultra-low-dose dual-source CT angiography using high-pitch spiral acquisition and iterative reconstruction in young children with congenital heart disease". In: *Journal of Cardiovascular Computed Tomography* 7.6 (2013), pp. 376–382. ISSN: 1934-5925. DOI: <https://doi.org/10.1016/j.jcct.2013.11.005>. URL: <https://www.sciencedirect.com/science/article/pii/S1934592513004760>.
- [18] D. Dey, P. Slomka, and D. Berman. "Achieving Very-Low-Dose Radiation Exposure in Cardiac Computed Tomography, Single-Photon Emission Computed Tomography, and Positron Emission Tomography". In: *Circulation. Cardiovascular imaging* 7 (July 2014), pp. 723–34. DOI: 10.1161/CIRCIMAGING.113.000897.
- [19] T. Vasavan, E. Ferraro, E. Ibrahim, P. Dixon, J. Gorelik, and C. Williamson. "Heart and bile acids – Clinical consequences of altered bile acid metabolism". In: *Biochimica et Biophysica Acta (BBA) - Molecular Basis of Disease* 1864.4, Part B (2018). CHOLANGIOCYTES IN HEALTH AND DISEASE: FROM BASIC SCIENCE TO NOVEL TREATMENTS, pp. 1345–1355. ISSN: 0925-4439. DOI: <https://doi.org/10.1016/j.bbadis.2017.12.039>. URL: <https://www.sciencedirect.com/science/article/pii/S0925443917304969>.
- [20] C. A. S. Dunning, K. Rajendran, A. Inoue, P. Rajiah, N. Weber, J. G. Fletcher, C. H. McCollough, and S. Leng. "Optimal Virtual Monoenergetic Photon Energy (keV) for Photon-Counting-Detector Computed Tomography Angiography". In: *Journal of Computer Assisted Tomography* 47.4 (July 2023), pp. 569–575. DOI: 10.1097/RCT.0000000000001450. URL: https://journals.lww.com/jcat/fulltext/2023/07000/optimal_virtual_monoenergetic_photon_energy__kev_.8.aspx.
- [21] Z. Ahmed, D. Campeau, H. Gong, K. Rajendran, P. Rajiah, C. McCollough, and S. Leng. "High-pitch, high temporal resolution, multi-energy cardiac imaging on a dual-source photon-counting-detector CT". In: *Medical Physics* 50.3 (2023), pp. 1428–1435. DOI: 10.1002/mp.16124.
- [22] Z. Ahmed, K. Rajendran, H. Gong, C. McCollough, and S. Leng. "Quantitative assessment of motion effects in dual-source dual-energy CT and dual-source photon-counting detector CT". In: *Proceedings of SPIE–the International Society for Optical Engineering*. Vol. 12031. 2022, 120311P. DOI: 10.1117/12.2611030.

- [23] R. Oberhoffer, D. Lang, and K. Feilen. "The diameter of coronary arteries in infants and children without heart disease". In: *European Journal of Pediatrics* 148.5 (1989), pp. 389–392. ISSN: 1432-1076. DOI: [10.1007/BF00595893](https://doi.org/10.1007/BF00595893). URL: <https://doi.org/10.1007/BF00595893>.
- [24] M. Fontos, Z. Nagy, Z. Prodán, L. Ablonczy, I. Kozma, M. Vértesaljai, É. Kis, and O. H. Bálint. "Coronary Artery Dimensions on Computed Tomography Following the Neonatal Arterial Switch Operation for the Complete Transposition of the Great Arteries". In: *Pediatric Cardiology* 46.8 (2025), pp. 2348–2359. ISSN: 1432-1971. DOI: [10.1007/s00246-024-03662-y](https://doi.org/10.1007/s00246-024-03662-y). URL: <https://doi.org/10.1007/s00246-024-03662-y>.
- [25] M. P. Crossett, M. Schneider-Kolsky, and J. Troupis. "Normal perfusion of the left ventricular myocardium using 320 MDCT". In: *Journal of Cardiovascular Computed Tomography* 5.6 (2011), pp. 406–411. ISSN: 1934-5925. DOI: <https://doi.org/10.1016/j.jcct.2011.10.003>. URL: <https://www.sciencedirect.com/science/article/pii/S1934592511003716>.
- [26] K. Nikolaou, A. Knez, S. Sagmeister, B. J. Wintersperger, P. Boekstegers, G. Steinbeck, M. F. Reiser, and C. R. Becker. "Assessment of Myocardial Infarctions Using Multidetector-Row Computed Tomography". In: *Journal of Computer Assisted Tomography* 28.2 (2004). ISSN: 0363-8715. URL: https://journals.lww.com/jcat/fulltext/2004/03000/assessment_of_myocardial_infarctions_using.21.aspx.
- [27] R. Booi, N. R. van der Werf, M. L. Dijkshoorn, A. van der Lugt, and M. van Straten. "Assessment of Iodine Contrast-To-Noise Ratio in Virtual Monoenergetic Images Reconstructed from Dual-Source Energy-Integrating CT and Photon-Counting CT Data". In: *Diagnostics* 12.6 (2022). ISSN: 2075-4418. DOI: [10.3390/diagnostics12061467](https://doi.org/10.3390/diagnostics12061467). URL: <https://www.mdpi.com/2075-4418/12/6/1467>.
- [28] K. de Waal, E. Petoello, E. Crendal, and N. Phad. "The effect of heart rate on systolic and diastolic time intervals and cardiac function in stable and sick preterm infants". In: *Early Human Development* 205 (2025), p. 106268. ISSN: 0378-3782. DOI: <https://doi.org/10.1016/j.earlhumdev.2025.106268>. URL: <https://www.sciencedirect.com/science/article/pii/S0378378225000787>.
- [29] E. K. K. ten Hoor. "Photon Counting CT Cardiac Protocol Optimization for Neonates: A Phantom Study: Focussed on the Coronary Arteries and Aortic Arch". Master's thesis. Delft University of Technology, 2024. URL: <https://resolver.tudelft.nl/uuid:eda4a55a-2f44-4dcb-af8a-445cd17670be>.
- [30] G. Heusch. "Heart rate in the pathophysiology of coronary blood flow and myocardial ischaemia: benefit from selective bradycardic agents". In: *British Journal of Pharmacology* 153.8 (2008), pp. 1589–1601. DOI: <https://doi.org/10.1038/sj.bjp.0707673>. URL: <https://bpspubs.onlinelibrary.wiley.com/doi/abs/10.1038/sj.bjp.0707673>.
- [31] K. Rajendran, M. Bruesewitz, J. Swicklik, A. Ferrero, J. Thorne, L. Yu, C. McCollough, and S. Leng. "Task-based automatic keV selection: Leveraging routine virtual monoenergetic imaging for dose reduction on clinical photon-counting detector CT". In: *Physics in Medicine and Biology* 69.11 (2024). DOI: [10.1088/1361-6560/ad41b3](https://doi.org/10.1088/1361-6560/ad41b3).
- [32] J. Le Roy, B. Azais, H. Zarqane, H. Vernhet Kovacsik, T. Mura, A. Lacampagne, and P. Amedro. "Selection of optimal cardiac phases for ECG-triggered coronary CT angiography in pediatrics". In: *Physica Medica* 81 (2021), pp. 155–161. ISSN: 1120-1797. DOI: <https://doi.org/10.1016/j.ejmp.2020.12.002>. URL: <https://www.sciencedirect.com/science/article/pii/S112017972030315X>.
- [33] T. Allmendinger, G. Boemio, and V. Karpatri. *Optimal Cardiac CT Imaging with ZeeFree*. White paper. Online only. Published by Siemens Healthineers AG. Forchheim, Germany: Siemens Healthineers AG, 2024. URL: https://marketing.webassets.siemens-healthineers.com/44f20d1545f9607d/050634dc0992/siemens-healthineers_CT_ZeeFree_Cardiac-Whitepaper.pdf.
- [34] S. Jodogne. "The Orthanc Ecosystem for Medical Imaging". In: *Journal of Digital Imaging* 31.3 (June 2018), pp. 341–352. ISSN: 1618-727X. DOI: [10.1007/s10278-018-0082-y](https://doi.org/10.1007/s10278-018-0082-y). URL: <https://doi.org/10.1007/s10278-018-0082-y>.
- [35] Radiology/Medical Physics. *Simple Orthanc*. <https://gitlab.com/radiology/medical-physics/simple-orthanc>. Computer software, GitLab repository. n.d.

- [36] D. Giavarina. "Understanding Bland Altman analysis". In: *Biochemia Medica* 25.2 (2015), pp. 141–151. DOI: 10.11613/BM.2015.015.
- [37] S. Vaz, T. Falkmer, A. E. Passmore, R. Parsons, and P. Andreou. "The Case for Using the Repeatability Coefficient When Calculating Test–Retest Reliability". In: *PLOS ONE* 8.9 (2013), e73990. DOI: 10.1371/journal.pone.0073990.
- [38] H. X. Barnhart and D. P. Barboriak. "Applications of the repeatability of quantitative imaging biomarkers: a review of statistical analysis of repeat data sets". In: *Translational Oncology* 2.4 (2009), pp. 231–235. DOI: 10.1593/tlo.09268.
- [39] J. Martin Bland and D. Altman. "STATISTICAL METHODS FOR ASSESSING AGREEMENT BETWEEN TWO METHODS OF CLINICAL MEASUREMENT". In: *The Lancet* 327.8476 (1986). Originally published as Volume 1, Issue 8476, pp. 307–310. ISSN: 0140-6736. DOI: [https://doi.org/10.1016/S0140-6736\(86\)90837-8](https://doi.org/10.1016/S0140-6736(86)90837-8). URL: <https://www.sciencedirect.com/science/article/pii/S0140673686908378>.
- [40] The MathWorks, Inc. *fitlme: Fit linear mixed-effects model*. MATLAB Statistics and Machine Learning Toolbox documentation. The MathWorks, Inc. 2026. URL: <https://www.mathworks.com/help/stats/fitlme.html>.



Literature Review

Photon-Counting CT in Neonatal Cardiac Imaging for Congenital Heart Disease: A Systematic Review

Abstract

Objective: Neonatal cardiac computed tomography (CT) for congenital heart disease (CHD) requires high diagnostic image quality at minimal radiation dose and is highly susceptible to motion. Photon-counting CT (PCCT) may improve dose efficiency, spatial resolution, and spectral imaging, but neonatal evidence and optimized protocols remain unclear. This review summarizes the current clinical evidence on PCCT for neonatal (or neonatal-inclusive) cardiac imaging in CHD and identifies which acquisition and reconstruction parameters have been evaluated to improve image quality, motion robustness, radiation dose, and diagnostic performance.

Methods: A PRISMA-guided search was performed on 10 December 2025 in Medline, Embase, Web of Science, and Scopus. Studies were included if they reported PCCT for neonatal (0–28 days) cardiac imaging in CHD or provided neonatal-inclusive pediatric cohorts relevant to neonatal protocols. Results on age, acquisition settings, reconstruction strategies, image quality assessment, diagnostic usability, and dose metrics were extracted. Due to heterogeneity and limited neonatal focused reporting, results were synthesized qualitatively.

Results: The search yielded 396 articles, of which 11 articles were included, containing seven clinical studies and four reviews. Neonatal outcomes were rarely reported as a distinct subgroup, therefore conclusions largely derive from neonatal-inclusive cohorts. PCCT examinations were consistently described as diagnostically usable, and one comparison reported a higher proportion of diagnostic scans with PCCT than with conventional energy-integrating CT. Protocols used high-pitch, free-breathing acquisition with short rotation times, typically without ECG gating, and relied on submillimeter collimation (0.2–0.4 mm) to support high spatial resolution. Parameter evaluation focused mainly on low-kV angiographic protocols (70–90 kV) compared with higher-kV approaches for spectral reconstructions and one tin-filtered approach. Reconstruction methods combined vascular kernels with quantum iterative reconstruction and, when used, low-keV virtual monoenergetic images.

Conclusion: Current evidence supports the feasibility of PCCT for neonatal CHD imaging and suggests the need for a neonatal protocol consisting of low-kV, high-pitch and free-breathing acquisition with vascular-oriented reconstruction. Robust neonatal protocol guidance will require neonatal-stratified studies with standardized image-quality, motion robustness, and dose endpoints.

Author

Sarah Stam
5876753

Supervisors

Marlies Goorden - TU Delft
Marcel van Straten - Erasmus MC
Thom van der Laan - Erasmus MC

27-01-2026



1. INTRODUCTION

Congenital heart disease (CHD) affects approximately one percent of newborns worldwide and is the leading cause of death in pediatric and neonatal patients [1].

The primary imaging modality for CHD is echocardiography. However, complex cardiac and extracardiac structures are often insufficiently visualized. In these cases, computed tomography (CT) is used to provide a more detailed visualization. In neonates, minimizing radiation dose while maintaining high diagnostic image quality is critical. However, conventional energy-integrating computed tomography (EID-CT) has inherent limitations, when imaging neonates with CHD[2].

An emerging CT detector technology, photon-counting CT (PCCT), offers the possibility to overcome some of these limitations by offering improved dose efficiency, higher spatial and contrast resolution, and noise reduction [3]. PCCT can improve preoperative anatomical assessment in complex congenital heart disease. Neonates with CHD often require detailed imaging to map the cardiac anatomy and related vascular structures. The ability to obtain high-detail images in a single, fast scan is relevant as many neonatal CHD patients can then be scanned without breath-hold and possibly without general anesthesia. These ultra-fast acquisitions are achieved using high-pitch helical scanning, implemented on conventional energy-integrating dual-source CT systems. The pitch reflects the table movement per gantry rotation relative to the total collimated beam width. Increasing the pitch shortens the acquisition time and therefore reduces motion sensitivity [4]. These systems provide the temporal resolution required for high-pitch (or flash) acquisitions, allowing cardiac imaging within a very short time interval. High-pitch energy-integrating CT has been shown to reduce motion artifacts while maintaining diagnostic image quality, and its speed supports free-breathing examinations [4, 5].

These features make dual-source PCCT promising for neonatal cardiac imaging, where there is a need to visualize small, fast-moving structures with as little radiation as possible and to minimize motion-related artifacts.

Despite these potential advantages, PCCT is a relatively new modality, and current evidence on its application in neonatal cardiac imaging for CHD is limited. Therefore, a structured overview of the available studies is needed to clarify how PCCT is currently used in neonatal cardiac imaging and which acquisition and reconstruction parameters have been investigated.

This review aims to address the following research question: *What evidence is currently available on the application of photon-counting CT in neonatal cardiac imaging for congenital heart disease, and which acquisition and reconstruction parameters have been evaluated or optimized to improve image quality, motion robustness, radiation dose, and diagnostic performance?*

1.1. Theoretical background on Photon-Counting CT

Conventional EID-CT systems are based on solid-state scintillator detectors in which incident X-ray photons are first converted into visible light and subsequently into an electrical signal by photodiodes [6], as represented in Figure 1 (a). Within a given integration time, the contributions of all detected photons are summed, yielding a signal that is approximately proportional to the total deposited X-ray energy, as shown in Figure 2 (a). Therefore, individual photon interactions are not resolved; instead, the detector integrates the total deposited energy over the integration time, meaning that higher-energy photons contribute proportionally more to the signal than lower-energy photons. As a result, the spectral information of the incident X-ray spectrum is lost and electronic noise is integrated with the X-ray signal, potentially reducing image quality, particularly at low dose levels [7].

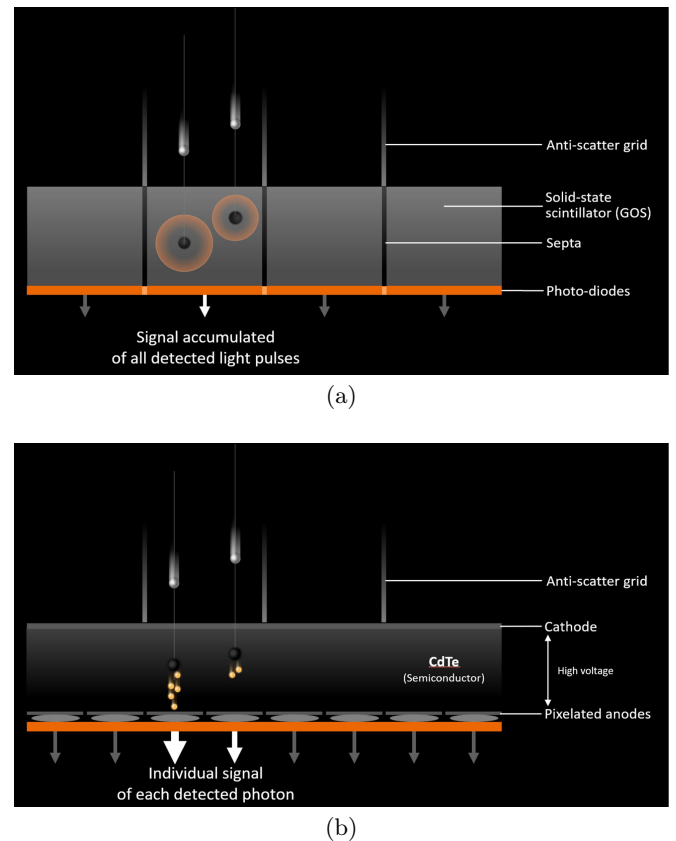


Figure 1. Schematic comparison of an energy-integrating detector (a) and a photon-counting detector (b). In EIDs, X-ray photons are converted to visible light in a scintillator and the resulting light pulses are integrated by photodiodes into a single accumulated signal. In PCDs, X-ray photons are directly converted in a semiconductor layer and each detected photon generates an individual electrical signal at the pixelated anodes, enabling photon counting and energy discrimination [8].

Photon-counting CT utilizes a different detector based on direct-conversion semiconductor materials, such as cadmium telluride (CdTe) or cadmium zinc telluride (CdZnTe) [7]. In these detectors, each incident X-ray photon is directly converted into electron-hole pairs, as shown in Figure 1 (b). The resulting charge cloud is collected by an array of electrodes and processed by fast, pixel-level readout electronics. Individual photon interactions are therefore detected as discrete pulses, represented in Figure 2 (b). The height of these pulses is proportional to the photon energy. By applying one or more energy thresholds, the detector can assign photons to predefined energy bins, enabling energy-resolved CT imaging [6].

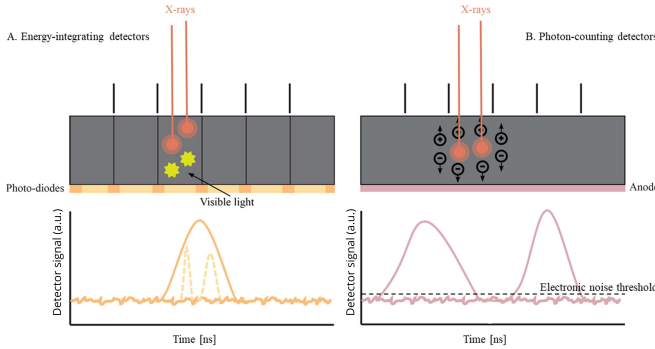


Figure 2. Temporal detector signals for an EID (a) and a PCD (b). In EID-CT, the X-ray signal from multiple photons arriving at the photo-diodes is summed over time together with electronic noise. In PCCT, individual photon interactions provide separate pulses, and an energy threshold can be applied to suppress the low-amplitude electronic noise. Adapted and modified from [9].

The use of photon-counting detectors (PCDs) offers several advantages over conventional energy-integrating detectors (EIDs). First, energy thresholds result in the reduction of electronic noise (Figure 2 (b)), since signals with lower intensities than the threshold, which are the result of noise, are excluded from the recorded data. This improves dose efficiency and facilitates image acquisition at reduced radiation dose while maintaining diagnostic image quality [10].

Second, the absence of light spread eliminates the need for septa, enabling the use of smaller detector pixels (as shown in Figure 1) and therefore higher intrinsic spatial resolution. In addition, the smaller pixel size reduces the partial-volume effects and improves delineation of small and complex anatomical structures. Moreover, as iodine attenuation is dominated by low-energy photons, improved detection and weighting of these photons can increase iodine-related contrast, which is beneficial for vascular and cardiac imaging [9, 11].

Third, sorting the photons into multiple energy bins enables advanced spectral imaging applications, including virtual monoenergetic images (VMI), material decomposition, and quantitative mapping of specific materials. Practically, low-keV VMI (40–60 keV) increases

iodine attenuation, therefore improving iodine contrast and CNR, which can support radiation dose reduction, whereas high-keV VMI (>100 keV) can reduce beam-hardening, metal-related artifacts and improve delineation near calcifications or stents [11, 12]. On clinical PCCT systems, task-based tools such as CARE keV image quality level can automatically select a VMI energy level based on the diagnostic task while targeting a predefined image quality (IQ) level [13].

Despite these advantages, PCDs present some technical challenges. For example, at high photon fluxes, pulse pile-up can occur when pulses corresponding to individual photons temporally overlap. The pile-up is shown schematically in Figure 3 and leads to count losses, non-linear detector response and energy misregistration [14]. However, clinical PCDs apply correction approaches, and indicating that the impact of these effects is limited under routine clinical scan conditions [11, 15].

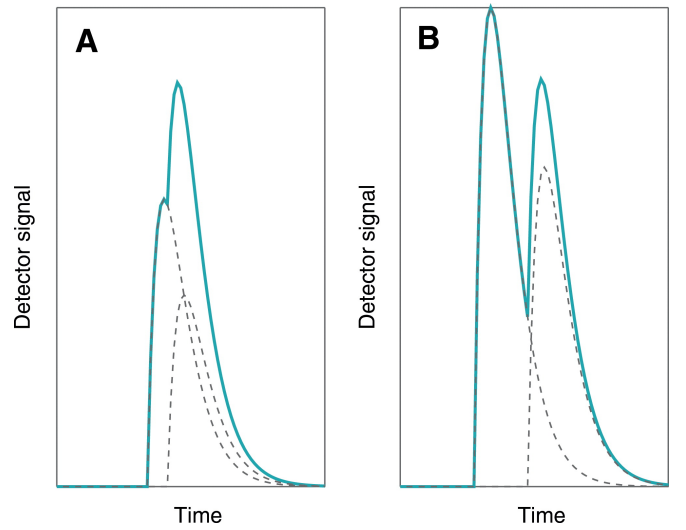


Figure 3. Conceptual illustration of pulse pile-up in a photon-counting CT detector. In (A), two photon interactions occur so close in time that the individual pulses (dashed lines) overlap and are registered as a single, higher-amplitude pulse (solid line). In (B), when the interactions are sufficiently separated relative to the resolving time, the pulses are recorded as two separate counts [16].

Photon-counting CT therefore provides a strong case for evaluation in neonatal cardiac imaging, as the combination of improved dose efficiency, higher spatial resolution, and energy-resolved reconstruction may result in more reliable visualization of small, rapidly moving cardiac structures at equal or lower radiation dose than conventional EID-CT.

2. METHODOLOGY

This systematic review was conducted in accordance with the Preferred Reporting Items for Systematic Review and Meta Analyses (PRISMA) statement [17]. All

literature regarding photon-counting CT for neonatal (or pediatric) cardiac imaging was systematically reviewed.

2.1. Literature Search

A systematic literature search was performed on 10 December 2025 using the following databases: Medline, Embase, Web of Science and Scopus. The full search strategy used for each database can be found in Appendix A.

2.1.1. Exclusion criteria

The articles found were evaluated in an initial screening. The screening was performed using Rayyan, a web based software designed to include or exclude articles from the literature search [18]. The articles were not deemed eligible if the article met any of the following exclusion criteria:

- The study did not involve photon-counting CT.
- The study did not focus on cardiac imaging.
- The study did not include neonatal imaging.
- The publication was a note or comment.
- The study was in a foreign language.

2.1.2. Full text screening and data extraction

Full-text articles were subsequently assessed for eligibility using the following exclusion criteria:

- The full text is not accessible through institutional access (Technical University of Delft or Erasmus Medical Center).
- The study was not relevant for photon-counting CT on neonates.
- The study did not address photon-counting CT for cardiac imaging on neonates.
- The publication was not a full research article and was not suitable for reference screening.

Reviews were not excluded from this review but rather used for reference screening (snowballing) and for the discussion section. With snowballing, additional articles were identified and included when deemed relevant [19].

2.2. Data extraction and synthesis

For each included study, information was extracted on study design and patient age group, together with PCCT acquisition parameters and image-quality evaluation (assessment approach and reported image-quality values).

A qualitative synthesis was conducted due to heterogeneity in study design, patient populations, and outcome reporting. To provide a transparent overview of the age distribution of the included evidence, studies were mapped to predefined age strata. Because age reporting and age-group definitions were highly inconsistent between the studies, these strata were defined pragmatically for this review as neonates (0–28 days), infants (1–12 months), and pediatric (1–18 years).

Findings were synthesized at two levels: (i) age-stratified synthesis when outcomes were reported somewhat separately for neonates, infants, or older pediatric patients; and (ii) cohort-level synthesis when studies reported mixed populations without separable age-specific outcomes. Where available, neonatal-specific evidence was prioritized.

3. RESULTS

The literature search yielded 396 articles. After the removal of duplicates, a total of 171 articles were screened based on title and abstract. Thirty-seven articles passed the criteria and underwent full-text assessment, resulting in 11 articles for this review. The resulting flowchart of the exclusion process can be found in Figure 4.

3.1. Study and patient characteristics

3.1.1. Study characteristics

In total, seven primary studies met the inclusion criteria and were included in the qualitative synthesis. All included primary studies were published from 2023 onward and reported early clinical experience with photon-counting CT on congenital heart disease imaging. Of the seven included studies, six employed a retrospective cohort design (five single-center and one multi-center), while one study used a prospective single-center design.

Most studies evaluated high-pitch PCCT in neonates, infants, or young children with suspected or confirmed CHD. Several studies included a comparison using conventional (dual-source) EID-CT, whereas others focused on protocol optimization within different PCCT protocols.

Four remaining review articles were identified and are discussed separately in Section 4 to compare the conclusions drawn from the primary studies with the conclusions and key statements reported in the reviews.

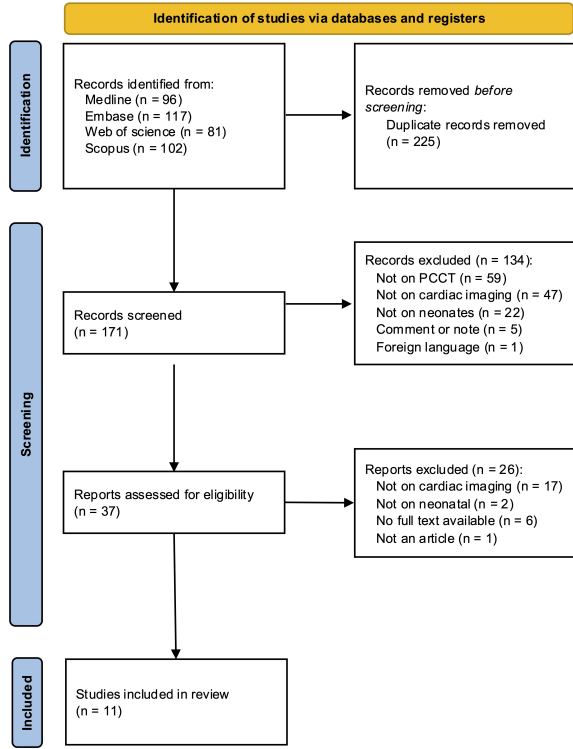


Figure 4. Flowchart of the inclusion process during the literature search

3.1.2. Patient characteristics

In the included primary studies, sample sizes ranged from small single-center series to larger cardiac CT cohorts, with PCCT sample sizes ranging from 30 to 4,772 patients [20, 21]. Several studies restricted inclusion to small patients, for example using a weight threshold of <10 kg ($N_{\text{PCCT}} = 58$) [22], whereas the remaining studies enrolled broader pediatric cohorts including patients <18 years. In mixed-modality cohorts, PCCT typically represented a subset of the total sample (e.g., 37/74, 100/200, 30/114 and 44/88) [20, 23–25].

Neonatal-specific reporting was limited across the included primary studies. Although the included cohorts largely comprised neonates and infants, several studies did not report outcomes separately for neonates. The reporting of the neonatal/infant subgroup outcomes in the broader pediatric cohorts was inconsistent. Overall, only three studies provided neonatal outcomes (reported with infants) in any form [21, 24, 26].

3.2. Acquisition and reconstruction parameters

An overview of the extracted variables on tube voltage, pitch, reconstruction kernels and iterative reconstruction, from the included studies, are provided in Table I.

3.2.1. Tube voltage and spectral mode

Across studies, two practical strategies were described:

- Low-kV protocols (70–90 kV): for dose-efficient angiographic imaging ([20–23, 26])
- Higher-kV protocols (100–120 kV): to enable additional spectral reconstructions ([22, 24, 25])

Neonatal evidence. Neonatal-inclusive studies generally favored low-kV protocols combined with virtual monoenergetic images (VMI) in the 45–55 keV range to increase visibility of small vessels and maintain image quality at low dose [21, 26]. These studies both compared 70 kV and 90 kV PCCT using similar reconstruction settings to investigate the effect of tube voltage and dose. Another study performed low-kV protocols on both PCCT and EID-CT [20].

Infant evidence. Infant-focused protocol comparisons resemble neonatal protocols, where low-kV approaches were evaluated [23] against EID-CT protocols or higher-kV PCCT approaches [22]. These studies highlight a practical trade-off in both neonatal and infant imaging: low-kV protocols can reduce dose while maintaining diagnostic vascular contrast, whereas higher-kV protocols allow additional spectral reconstructions but can increase dose with a similar contrast-to-noise ratio (CNR) [22–24].

Pediatric evidence. In broader pediatric cohorts, including neonates, protocol selection was more heterogeneous and appeared to be influenced by patient size and indication. Studies discussing neonatal, infant and overall pediatric application favor a low kV approach [21, 26] while some protocols favor higher-kV acquisition in older children when spectral outputs were desired [24, 25]. These spectral outputs, beyond VMIs, were generated using multi-energy post-processing methods resulting in virtual non-contrast images and iodine maps.

3.2.2. Pitch and gating

In all included cardiac PCCT studies, high-pitch acquisitions were the predominant approach in neonatal (or neonatal-inclusive) cohorts to minimize motion artifacts. High-pitch usage also results in the ability of imaging during free breathing without the need of ECG gating to synchronize the image acquisition with the cardiac cycle. ECG gating was used in a single study [21].

Infant-focused cohorts reported similar reliance on ungated high-pitch protocols in free-breathing infants [22, 23]. In studies including older children, breath-hold acquisitions were used when possible [21, 26].

All studies used typical parameters including short rotation times of 250 ms suitable for neonatal cardiac imaging.

Table I. Extracted study characteristics and relevant PCCT protocol parameters (tube voltage (kV), CARE keV IQ level, pitch, reconstruction kernel, Iterative Reconstruction (IR) algorithm and strength), with qualitative IQ assessment methods and dose outcomes for included primary studies.

Study	Year	Stratum*	Scanner	Pitch	CARE keV IQ level	Tube voltage (kV)	CTDI _{vol} (mGy)	SSDE (mGy)	DLP (mGy·cm)	Kernel	IR algorithm (strength) ^a	VMI level (keV)	IQ assess.
Habib Geryes et al. [21]	2025	N	PCCT	3.2	30	70–90	0.2 [0.2–0.3]	–	3.8 [2.9–5.3]	Bv44; Br66	QIR (4)	45	–
		I & P	PCCT	3.2	30	70–90	0.5 [0.3–1.0]	–	9.6 [4.7–23.3]	Bv44; Br66	QIR (4)	45	–
Dang et al. [22]	2025	N & I	PCCT	3.2	25	70	0.13±0.06	–	2.3±1.4	Bv48	QIR (3)	53	Likert
		N & I	PCCT	3.2	25	120	0.25±0.05	–	4.79±1.3	Bv48	QIR (3)	55	Likert
Hellms et al. [23]	2025	N, I & P	PCCT	3.2	45	70	0.20±0.12	–	4.06±3.22	Br40	QIR (2)	–	Likert
		N, I & P	EID-CT	3.0	n.a.	70	0.37±0.17	–	7.21±4.67	Bv40	ADMIRE (2)	–	Likert
Arguello Fletes et al. [24]	2025	N & I	PCCT	3.2	25	120	0.23 [0.21–0.27]	–	4.0 [3.26–5.17]	Bv48	QIR (3)	55	Likert
		N & I	EID-CT	3.4	n.a.	70; 80; 100 ^b	0.10 [0.10–0.12]	–	1.8 [1.5–2.2]	Br36	SAFIRE (3)	–	Likert
		P	PCCT	3.2	25	120	0.80 [0.50–2.01]	–	23.4 [10.23–50.87]	Bv48	QIR (3)	55	Likert
		P	EID-CT	3.4	n.a.	70; 80; 100 ^b	0.16 [0.14–0.50]	–	4.2 [2.8–11.2]	Br36	SAFIRE (3)	–	Likert
Stalhammar et al. [26]	2024	N & I	PCCT	3.2	70	70	0.17 [0.09–0.28]	0.40 [0.25–0.63]	3 [2–5] ^c	Bv44	QIR (4)	55	Likert-like
		N & I	PCCT	3.2	70	90	0.32 [0.23–0.59]	0.74 [0.58–1.35]	4 [3–8] ^c	Bv44; Bv40	QIR (2–4) ^d	55	Likert-like
		N, I & P	PCCT	3.2	70	70	0.32 [0.09–1.32]	0.68 [0.25–2.02]	5 [2–40] ^c	Bv44	QIR (4)	55	Likert-like
		N, I & P	PCCT	3.2	70	90	0.50 [0.23–1.96]	1.10 [0.58–2.71]	8 [4–53] ^c	Bv44; Bv40	QIR (2–4) ^d	55	Likert-like
Dirrichs et al. [20]	2023	N, I & P	PCCT	3.0	–	90	–	–	14.34±6.63	–	QIR (2)	–	5-pt scale
		N, I & P	EID-CT	3.0	–	70	–	–	16.14±9.26	–	–	–	5-pt scale
Krueger et al. [25]	2025	N, I & P	PCCT	3.2	14	Sn100 ^e	0.27 [0.12]	0.55 [0.20]	6.71 [4.29]	Br40	QIR (2)	–	Likert
		N, I & P	EID-CT	3.0	n.a.	Sn100 ^e	0.39 [0.21]	0.83 [0.31]	8.75 [7.60]	Br40	ADMIRE (3)	–	Likert

Note: n.a. = not applicable; – = not reported in the article

* *Stratum abbreviations*: N = neonates (0–28 days), I = infants (1–12 months), P = paediatric (1–18 years).

^a The iterative reconstruction algorithm used for PCCT was Quantum Iterative Reconstruction (QIR). For EID-CT, alternative iterative reconstruction methods were applied (SAFIRE and ADMIRE).

^b For EID-CT, tube voltage was 70 kV ($n = 93$), 80 kV ($n = 5$), or 100 kV ($n = 2$).

^c Stalhammar et al. [26] reported DLP values both unmatched and matched for scan length, for clarity matched values are reported.

^d Stalhammar et al. [26] reported differently used values for QIR with level 2 (57 %), level 3 (17 %) and level 4 (26 %).

^e Sn100 = 100 kV with added tin filtration.

3.2.3. Spatial resolution and collimation

Across studies, high-resolution acquisition was enabled through thin detector collimation and small reconstructed slice thicknesses of 0.4; 0.6 or 1.0 mm. The reported detector configurations varied between scanners and protocols, however most studies used a 144-row configuration with a collimation of 0.4 mm (144×0.4) [20, 21, 23–26]. One study reported a thinner collimation of 120×0.4 [22]; however, this configuration is likely a reporting error, as the corresponding ultra-high-resolution acquisition mode is commonly described as 120×0.2 mm.

3.2.4. Reconstruction kernels and iterative reconstruction

Across studies, most protocols combined thin-slice cardiovascular reconstructions with vascular or soft-tissue kernels and quantum iterative reconstruction (QIR) to reduce noise. When reported, virtual monoenergetic images (VMI) were reconstructed in the low-keV range (45–55 keV) to enhance contrast.

Neonatal evidence. Neonatal-specific studies generally combined vascular kernels with higher QIR strengths to maintain diagnostic interpretability at very low dose [20, 21, 26]. The kernels generally mentioned are Bv or Br kernels referring to body imaging for vascular structures or regular/soft tissues. The sharpness of the kernel ranges from 40 to 66 with higher numbers for a higher resolution.

In one study, the CARE keV IQ level was set to 30 keV

and images were reconstructed at 45 keV VMI, using both Bv44 and Br66 kernels with QIR level 4 generating a high-resolution vascular image [21]. Another neonatal-inclusive cohort used vascular kernels (Bv44/Bv40) with reconstruction at 55 keV and applied QIR level 4 for the low-kV protocol, while the higher-kV protocol used QIR levels 2–4, for the protocol optimization across dose levels [26]. A further neonatal-inclusive cohort reported QIR level 2, with a more conservative denoising setting in routine practice [20].

Overall, neonatal reconstruction strategies prioritized vessel sharpness (Bv kernels) and noise reduction (higher QIR), with low-keV VMI used to preserve contrast in very small vessels.

Infant evidence. Studies including infants used similar reconstruction approaches, with VMI levels around 53–55 keV, mostly Bv48 kernels and QIR level 3 for PCCT [22, 24]. Other infant cohorts reported Br40 and/or QIR level 2 [20, 23].

Pediatric evidence. In broader pediatric cohorts, reconstruction approaches were more heterogeneous, ranging from Bv-kernels and high QIR strategies aligned with neonatal CHD priorities [21, 26] to Br40 with QIR level 2 in older children [25]. Where reported, reconstruction energies remained near 55 keV [24, 26].

3.3. Image quality and motion robustness

Neonatal and infant PCCT protocols, as discussed in the acquisition and reconstruction parameters sec-

tion, were comparable between studies. Mostly based on high-pitch scanning to minimize motion artifacts, without ECG gating and during free breathing. Because only a limited subset of studies reported neonatal-focused outcomes and several cohorts were mixed, quantitative image quality findings are synthesized primarily for neonates and infants combined with older pediatric results summarized separately when relevant.

An overview of the extracted variables on qualitative image quality assessment, from the primary included studies, are provided in Table I.

3.3.1. Quantitative image quality

Quantitative image quality was not reported in all studies, however when reported it was assessed using region-of-interest (ROI) based signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR). Two studies provided direct quantitative comparisons between PCCT and EID-CT, while others evaluated protocol choices within PCCT.

Neonate and infant evidence. Evidence was mixed and depending on EID-CT or PCCT protocol comparisons. In a study comparing PCCT to EID-CT both vascular SNR and CNR were higher with PCCT, supporting improved quantitative vascular performance [20]. PCCT protocol optimization showed that 70 kV achieved lower dose with similar image quality compared with 120 kV, indicating that dose-efficient low-kV imaging can maintain both quantitative and diagnostic quality [22]. However, when comparing PCCT with EID-CT, neonatal and infant outcomes were inconsistent between studies: one study found no differences [23], while another reported a lower CNR in the neonate and infant subgroup [24]. These comparisons are difficult to interpret without accounting for protocol matching. As summarized in Table I, when the tube voltage and spectral settings are similar, PCCT tends to achieve low dose levels in neonates/infants. When protocols differ between scanners, the observed differences in quantitative image-quality metrics may reflect protocol choices rather than detector technology alone.

Pediatric evidence. In older children (above one year), a cohort comparing PCCT to EID-CT reported a higher CNR in this population for PCCT [24]. However, another study found lower SNR with unchanged CNR for PCCT versus EID-CT [25].

3.3.2. Qualitative image quality and diagnostic confidence

Qualitative image quality has been reported in six out of seven studies; one study did not report any image quality assessment [21]. Most studies used a Likert scoring system [22–25], while remaining studies applied other rating scales comparable to the Likert scale [20, 26]. Overall, subjective findings were consistent across the evidence

base. PCCT provided diagnostically adequate-to-high subjective image quality for neonatal and pediatric CHD imaging, with only limited age-stratified reporting.

In studies including neonates and infants, subjective image quality ratings ranged from good to high. Low-kV studies reported that qualitative scores remained high when reducing tube voltage [22, 26]. Comparative studies against EID-CT typically did not find significant differences in qualitative image quality between PCCT and EID-CT under matched clinical conditions [23, 25]. Another comparison of PCCT and EID-CT resulted in higher qualitative scores for PCCT [20].

Only one study reported a clear age-stratified qualitative pattern: image quality was rated good in neonates and infants but excellent in pediatrics older than 12 months [24].

3.3.3. Motion robustness

Motion robustness was described in two ways in the studies: first through the technical rationale of PCCT protocols [21, 24], and second through subjective proxy measures such as motion artifact and visibility scores on Likert-type scales [20, 23, 25]. Two studies attributed motion mitigation to the high scan speed due to dual-source and high-pitch acquisition, which allows great coverage during free breathing and therefore supports diagnostic imaging without ECG gating [20, 21, 24, 26]. Where feasible, breath-hold was used as an additional strategy to mitigate motion in older children [20, 26].

Where motion was explicitly rated, some studies applied scoring systems to evaluate motion artifacts and/or visibility. In a comparison of PCCT to EID-CT, proximal coronary segments were evaluated separately for motion and visibility using a 5-point scale, showing no significant differences [23]. Similarly, a study comparing EID-CT and PCCT evaluated motion using Likert scoring, where no statistically significant differences were reported [25]. In contrast, another comparison with EID-CT where motion was one of seven image quality categories within a 5-point scale, PCCT achieved significantly higher scores [20].

3.4. Diagnostic quality and clinical usability

Diagnostic accuracy metrics were not always reported. Therefore, diagnostic usefulness was assessed based on (i) whether examinations were considered diagnostically adequate for CHD evaluation, and (ii) whether protocol or scanner comparisons resulted in any reported loss of interpretability or diagnostic confidence.

In studies that reported diagnostic quality, PCCT examinations were consistently considered diagnostic. PCCT protocol optimization showed that lowering tube voltage could increase image noise without compromising diagnostic usability: 70 kV images demonstrated higher

noise levels than 120 kV however all examinations remained of diagnostic quality [22]. Similarly, in an infant (including neonates) and pediatric study comparing 70 kV and 90 kV, both protocols produced high diagnostic quality ratings, with 70 kV achieving the most favorable subjective scores [26].

Scanner comparisons generally indicated comparable or improved diagnostic quality with PCCT. In a direct comparison with EID-CT, diagnostic quality was reported as comparable between PCCT and EID-CT [23], and in another all PCCT examinations were reported to be of diagnostic quality [25]. In contrast, in another comparison with EID-CT, PCCT achieved a higher proportion of diagnostic examinations (97% vs 77%), suggesting a reduction in non-diagnostic studies in that clinical setting [20].

3.5. Radiation dose

Radiation dose reporting varied between studies. The volume CT dose index (CTDI_{vol}) and dose-length product (DLP) were reported most consistently, while effective dose (ED) and/or size-specific dose estimates (SSDE) were provided less frequently. CTDI_{vol} and SSDE are the most directly interpretable indicators of scanner output and patient-size-adjusted dose, respectively, and are commonly used as inputs for dose conversion to ED.

An overview of the extracted variables on radiation dose (e.g., CTDI_{vol} , SSDE and DLP), from the primary included studies, are provided in Table I.

3.5.1. Neonatal and infant dose levels

Only one study reported neonatal dose values separately. Neonates had very low median dose metrics ($\text{CTDI}_{\text{vol}} = 0.2$ mGy; $\text{DLP} = 3.8$ mGy·cm), while the median of overall pediatric results show higher values ($\text{CTDI}_{\text{vol}} = 0.5$ mGy; $\text{DLP} = 9.6$ mGy·cm), consistent with patient size scaling and scan extent [21]. In the remaining studies, neonates and infants were typically included within the pediatric or infant population, limiting neonatal-specific dose comparisons.

3.5.2. Tube-voltage optimization within PCCT

Both within PCCT protocol comparisons showed lower dose at lower tube voltages. In an infant-inclusive cohort, 70 kV yielded approximately half the CTDI_{vol} , DLP, and ED compared with the 120 kV protocol (ED: 0.07 ± 0.044 vs 0.15 ± 0.041 mSv) [22]. Similarly, 70 kV produced lower CTDI_{vol} and (matched scan length) DLP than 90 kV in both the neonatal/infant subgroup and the full cohort, with SSDE and ED showing the same trend (ED: 0.22 [0.15–0.37] vs 0.30 [0.22–0.59] mSv for neonates/infants;

and 0.30 [0.15–0.82] vs 0.39 [0.22–1.51] mSv for the full cohort) [26].

3.5.3. PCCT versus EID-CT

Scanner comparisons with EID-CT provided mixed results. Two studies reported lower mean dose metrics for PCCT than for EID-CT, with reductions of approximately 25–45% in CTDI_{vol} and DLP [23, 25].

In contrast, one study reported higher median CTDI_{vol} , DLP and effective dose for PCCT than for EID-CT, particularly in the neonatal and infant subgroup. These values range from approximately two-fold up to several-fold higher for the age group beyond neonates and infants (ED: 0.09 [0.06–0.09] vs 0.03 [0.03–0.04] mSv), although only CTDI_{vol} reached statistical significance in that older subgroup [24].

In another comparison with EID-CT, the dose metrics were broadly comparable between scanners. No significant differences were observed for DLP and effective dose [20].

4. DISCUSSION

4.1. Main Findings and the Research Question

The current evidence on photon-counting CT for neonatal cardiac imaging in congenital heart disease is limited. The included studies were published from 2023 onward and were mostly retrospective, single center investigations, with small to moderate PCCT cohort sizes ranging from 30 to 100 patients. Neonatal outcomes were generally not reported as a separate subgroup, but studies combined neonates with infants or broader pediatric patient groups. As a result, most conclusions regarding neonatal imaging are based on neonatal-inclusive cohorts rather than neonate specific results.

However, despite these limitations, the available evidence indicates that PCCT is clinically feasible in neonatal CHD protocols and can produce images of diagnostic quality. Studies reporting diagnostic usability consistently describe PCCT scans as diagnostic, and one study demonstrated a higher proportion of diagnostic examinations with PCCT compared to conventional EID-CT [20]. This difference can be explained by the improved dose efficiency and noise suppression of PCCT due to energy-thresholding. In addition, the higher intrinsic spatial resolution from the smaller detector pixels, can increase vessel conspicuity at the low dose levels. Other studies, rated qualitative image quality as good-to-excellent [22, 23, 25, 26].

Addressing the second part of the research question: "Which acquisition and reconstruction parameters have been evaluated or optimized to improve image quality, motion robustness, radiation dose, and diagnostic performance?". Literature has evaluated and optimized a

defined set of acquisition and reconstruction parameters with respect to image quality, motion robustness, radiation dose, and diagnostic performance. The acquisition parameters tube potential and scan mode were compared the most, with two protocols: (i) low-kV protocols (70–90 kV) aimed at dose-efficient imaging [20, 21, 23, 26], and (ii) higher-kV approaches (120 kV) used when spectral reconstructions were desired [22, 24]. A non-contrast Tin-filtered (Sn100) approach was also used with quantum iterative reconstruction methods [25]. All studies used high-pitch acquisitions with very short rotation times of 250 ms, without ECG gating, reflecting on rapid free-breathing imaging in neonates [20–23, 25, 26].

With respect to the image reconstruction technique, studies varied vascular and soft-tissue kernels, quantum iterative reconstruction, and low-keV virtual monoenergetic images (often reconstructed around 45–55 keV) [21, 22, 26]. The studies mainly focused on tube voltage optimization and evaluation where low-kV protocols reduced CTDI_{vol} and DLP while maintaining diagnostic image quality [22, 26]. In contrast, findings on high-kV protocols were more mixed, with no consistent advantage in quantitative metrics such as CNR in the neonatal and infant subgroup [24].

Finally, the outcomes of the studies differed in a way that influences the interpretation of the evidence. The image quality was comparable between the PCCT and EID-CT scanners, while quantitative metrics depended on the way the protocols were matched and the reconstruction choices [23–25]. Motion robustness was achieved by high scan speed and high-pitch acquisition rather than quantified as an endpoint. An improvement in motion robustness was reported in one study for 90 kV PCCT compared to 70 kV EID-CT [20]. The conclusions of the radiation dose were protocol-dependent: PCCT showed a lower dose than EIDCT in some comparisons [23, 25], but not uniformly [24], highlighting that the choice of protocol is the basis of the resulting differences.

Taken together, the available evidence indicates that PCCT is feasible for neonatal CHD imaging. Most reported optimization findings are derived from studies employing low tube-voltage, high-pitch acquisitions combined with vascular-oriented kernels and iterative reconstruction, whereas neonatal evidence on high-kV (spectral) protocols and quantitatively assessed motion robustness remains limited. Importantly, interpretation across studies is limited by protocol heterogeneity, including differences in selected CARE keV IQ level and VMI settings, reconstruction kernels, and achieved dose levels, which contributes to variability in reported image quality and motion outcomes.

4.2. Protocol Implications

The current evidence supports a workflow in which protocol design is driven by the main constraints of neonatal imaging: (i) motion risk, (ii) visualization of

very small vessels, and (iii) dose minimization. Consequently, the most defensible strategy is to select acquisition settings that “outrun” motion and use reconstruction methods to recover vessel visibility at low dose, rather than attempting to maximize spectral outputs or spatial resolution.

In practice, a reasonable default neonatal protocol can be implemented as a free-breathing, high-pitch acquisition at low tube potential, combined with thin-slice vascular reconstructions. Vascular kernels and iterative reconstruction should be used to reduce noise while maintaining high resolution, and low-keV VMI can be used for a specific diagnostic task. Vascular kernel and QIR selection should balance edge sharpness against noise. The Bv44 kernel with a moderate iterative reconstruction (QIR 3) could be used as a default. However, if noise is limiting at very low dose, a smoother kernel (e.g., Bv40) and stronger QIR (QIR 3–4) might improve visualization, whereas a sharper kernel (e.g., Bv48) may be used for cases where maximal detail is required and should be paired with stronger denoising. High-kV protocols should be reserved for specific diagnostic indications, because neonatal benefit has not been demonstrated in the current evidence.

A further practical implication is that high spatial resolution in neonatal CHD imaging is commonly pursued through submillimeter collimation (0.2–0.4 mm) [20–26]. In neonates, this effectively shifts optimization from “how thin can we go” to “how do we control noise while staying thin,” reinforcing the neonatal strategy of combining fast high-pitch acquisition with low-kV imaging and appropriate iterative reconstruction.

4.2.1. Motion management as a workflow design choice

For neonates with high heart rates, motion robustness should be treated as a primary constraint which can be overcome by minimizing acquisition time and avoiding steps that increase motion risk. While the literature rarely quantifies motion outcomes directly, the consistent use of fast, high-pitch acquisitions in neonatal-inclusive cohorts supports these settings to reduce motion artifacts.

4.2.2. Spectral/high-kV options

Spectral and higher-kV protocols should only be used for specific diagnostic questions where the added spectral outputs are expected to influence interpretation. In neonates, the current evidence does not support a consistent advantage for spectral/high-kV strategies in quantitative image quality, and these approaches can introduce dose or noise trade-offs. Practically, this suggests a “spectral when needed” policy: keep the neonatal default in a low-kV angiographic mode, and escalate to spectral/high-

kV modes only when a predefined indication, where additional visualization of fine details is needed, exists.

4.3. Why Results Differ

In the literature, this specific PCCT application is implemented with different protocols and evaluated using varying endpoints; consequently, variation is expected. In addition, reported outcomes varied: different qualitative scales and different definitions of diagnostic outcome limit study comparability.

Differences in tube voltage, pitch, CNR, and reconstruction strength can shift the balance between dose and image quality, explaining why PCCT appears dose-lowering in some EIDCT comparisons but not in others. This is especially important in neonates, where small absolute changes in technique can produce large relative differences in noise and dose indices.

Neonatal cardiac CT operates near the limits of photon statistics and is therefore affected by changes in reconstruction and noise suppression. Quantitative image-quality metrics are sensitive to ROI definitions, kernel sharpness and iterative reconstruction. This contributes to findings where qualitative diagnostic adequacy remains high while quantitative metrics show mixed results, especially when comparing neonatal and infant subgroups to other pediatric groups.

Together, these factors imply that differences across studies are best explained by protocol matching choices, neonatal size sensitivity, and non-standardized endpoints rather than by contradictory evidence. This reinforces the need for future studies on neonatal CHD imaging using PCCT to define protocols based on the evaluated acquisition and reconstruction parameters.

4.4. Comparison with Existing Review Literature

Several recent reviews have examined PCCT in pediatric (including, but not specifically neonatal) imaging, and these observations provide a context for this neonate-focused review.

In general, the reviews [9, 27–29] emphasize that PCCT introduces improvements in spatial resolution, offers better CNR with the potential for radiation dose reduction, and spectral imaging capabilities that can add diagnostic information. In addition, the advantage of enhanced motion robustness, not as an intrinsic property of the detector, but as a result of short gantry rotation time and high-pitch dual-source scan acquisition without compromising image quality, is cited quite often. Similarly, all four reviews note the promise of dose reduction: since PCCT is more dose-efficient, neonates could benefit from any dose reductions. The spectral capabilities are highlighted as well, suggesting more clinically specific potential uses [28, 29]. At the same time, the reviews consistently acknowledge that overall pediatric clinical evi-

dence is still emerging and that neonatal-specific data are limited, with many statements derived from mixed pediatric cohorts or extrapolated from pediatric or adult experience [9, 27–29]. This limits the strength of neonatal-specific recommendations regarding the optimal balance between image acquisition and reconstruction parameters.

4.4.1. Similarities with prior reviews

Findings in this literature review are supported by the reviews’ core conclusion that PCCT is implementable in neonatal cardiovascular imaging and can provide diagnostically adequate examinations.

Reviews discuss improved motion robustness as a consequence of high-pitch, dual-source acquisition [9, 28, 29]. This review supports this as a feasibility and protocol observation: high-pitch free-breathing acquisition was the predominant approach and diagnostic examinations were achieved [20–26]. However, motion was rarely quantified using standardized metrics, and where motion-related scoring was reported, differences between scanners were small or non-significant, with one study reporting higher motion-related scores of PCCT compared to EID-CT [20, 23, 25]. Thus, this review aligns with the general expectation of motion benefit through scan speed, but highlights a need for neonatal studies to make motion robustness as a reportable outcome.

Regarding spectral imaging, a few studies discuss the potential [9, 27–29], however in neonatal CHD, these benefits remain theoretical so far. Where higher-kV acquisition was used to enable broader spectral outputs, results did not show a uniform advantage in CNR, and trade-offs related to dose and noise may occur [24]. This does not contradict the reviews’ description of spectral potential, but it does suggest that, in neonates, spectral imaging should currently be regarded as indication-driven rather than as a default in the protocol.

4.4.2. Nuance addition compared to prior reviews

The added value of this review is a different framework of two claims that are often generalized in other review literature: dose reduction and improved quantitative image quality. Reviews summarize PCCT as dose-efficient in pediatrics [9, 27–29]. In the included evidence, however, dose outcomes varied with protocol choices and with how comparisons were constructed: PCCT showed lower dose than EIDCT in some cohorts [23, 25] but higher in another study under the different clinical protocols [24]. This indicates that dose reduction should not be interpreted as a direct consequence of PCCT usage; rather, it is dependent on protocol parameter selection. The same applies to quantitative image quality metrics such as CNR: while reviews cite improved iodine contrast as a technical advantage [27, 29], the neonatal and infant sub-

group evidence includes mixed quantitative findings depending on protocol and reconstruction [24, 25]. Consequently, neonatal conclusions should prioritize observed qualitative diagnostic quality, while treating quantitative advantages as context-specific until more reporting is available.

In conclusion, this review supports the conclusions of prior literature regarding PCCT’s feasibility, high image quality, potential for dose reduction and fast scanning. It provides a focused examination of neonatal CHD imaging that stresses the dependency of these benefits on protocol selection and highlights neonatal-specific limitations that broader pediatric discussions have not yet fully addressed.

4.5. Limitations of the Review

As previously discussed, evidence for PCCT in neonatal CHD imaging remains early and limited. This review has several limitations that warrant acknowledgment. Given the acquired number of studies specific to the neonatal population, the review was necessarily limited and qualitative. Another limitation is that this review focused on studies that included neonates (often alongside infants or broader pediatrics); as such, some of the extrapolations to “neonatal use” come from mixed-population data where specific neonate-only data was not always available. This could mask certain issues unique to neonates.

Additionally, this review did not incorporate phantom studies. Although phantom studies are not directly clinical, it could have provided useful controlled insights into protocol optimization that are difficult to separate in heterogeneous patient cohorts.

Finally, as with any systematic review, there is always the possibility that some relevant data is missed. This was mitigated as much as possible by cross-referencing multiple sources and sticking to reported data for the conclusions. Concluding, while the findings of this review are considered well-founded, they should be interpreted in light of the early stage of evidence and limitations above.

4.6. Future Research Priorities

Future work should shift from feasibility reports to neonatal-specific, standardized evaluations that enable fair comparisons and clinically meaningful endpoints.

First, studies should mandate neonatal stratification and report results by weight or size metrics, as neonatal performance cannot be inferred from older pediatric averages.

Second, dose reporting should be more consistent across studies. Although $CTDI_{vol}$ and DLP were reported most frequently, SSDE was provided in only two studies. Including SSDE in addition to $CTDI_{vol}$ and DLP

would add value by accounting for patient size and improving comparability of dose estimates, particularly in neonates.

Third, image quality reporting should be done with predefined ROI definitions for quantitative metrics and a consistent qualitative scoring framework, ideally linked to specific diagnostic tasks.

Fourth, studies should specify diagnostic endpoints relevant for neonatal CHD and evaluate performance against a reference standard where feasible.

Finally, comparative studies should apply protocol-matching frameworks for PCCT versus EID-CT to separate technology effects from protocol differences. In addition, this should be done in multi-center cohorts to improve generalizability.

5. CONCLUSION

The clinical evidence on photon-counting CT for the imaging of neonatal congenital heart disease remains very limited and based on neonatal-inclusive cohorts rather than neonate-only reporting. Despite these constraints, the available studies indicate that PCCT is feasible for neonatal cardiac CT and can provide images of diagnostic quality.

In the included literature, the acquisition and reconstruction parameters most frequently evaluated was tube potential. This was combined with a high-pitch free-breathing scan mode, supported by vascular-oriented reconstruction (kernel choice, QIR, and low-keV VMI when used). Low-kV protocols (70–90 kV) showed the most consistent optimization potential, which seems to reduce dose indices while maintaining diagnostic image quality. However, these benefits must be interpreted with caution given the limited data and variability in current practices.

Evidence supporting routine benefit of spectral/high-kV strategies in neonates remains insufficient, and reported dose and quantitative image quality outcomes were strongly dependent on protocol matching and reporting methods.

Overall, PCCT appears promising for neonatal CHD imaging when implemented with low-kV, high-pitch protocols, but neonatal focused studies with standardized motion, image-quality, and dose endpoints are required to establish robust, evidence-based protocol guidance.

In particular, motion robustness is currently inferred from fast acquisition choices rather than being quantified. Future work should therefore measure motion-induced artifacts under neonatal-like heart rates using controlled phantom experiments and evaluate clinically relevant diagnostic tasks with advanced, task-based image-quality metrics and model observers. Such studies could identify parameter combinations that optimally balance motion robustness, spatial resolution, and radiation dose, and translate these findings into practical evidence based protocol recommendations for neonatal PCCT.

REFERENCES

- [1] Jiaoli Xu, Qinhong Li, Lili Deng, Jingxuan Xiong, Zugen Cheng, and Caixia Ye. Global, regional, and national epidemiology of congenital heart disease in children from 1990 to 2021. *Frontiers in Cardiovascular Medicine*, Volume 12 - 2025, 2025. doi:10.3389/fcvm.2025.1522644. URL <https://www.frontiersin.org/journals/cardiovascular-medicine/articles/10.3389/fcvm.2025.1522644>.
- [2] Evan J. Zucker. Cardiac computed tomography in congenital heart disease. *Radiologic Clinics of North America*, 62(3):435–452, 2024. ISSN 0033-8389. doi:https://doi.org/10.1016/j.rcl.2023.12.015. URL <https://www.sciencedirect.com/science/article/pii/S003383892300235X>. Advances and Innovations in Cardiovascular Imaging.
- [3] Thomas Flohr, Bernhard Schmidt, Stefan Ulzheimer, and Hatem Alkadhi. Cardiac imaging with photon counting CT. *British Journal of Radiology*, 96(1152):20230407, oct 2023. ISSN 0007-1285. doi:10.1259/bjr.20230407. URL <https://doi.org/10.1259/bjr.20230407>.
- [4] Minwen Zheng, Hongliang Zhao, Jian Xu, Yongjie Wu, and Jian Li. Image quality of ultra-low-dose dual-source ct angiography using high-pitch spiral acquisition and iterative reconstruction in young children with congenital heart disease. *Journal of Cardiovascular Computed Tomography*, 7(6):376–382, 2013. ISSN 1934-5925. doi:https://doi.org/10.1016/j.jcct.2013.11.005. URL <https://www.sciencedirect.com/science/article/pii/S1934592513004760>.
- [5] Dmitrij Kravchenko, Christopher Hart, Stephan Garbe, Julian A. Luetkens, Alexander Isaak, Narine Mesrobian, Mathieu Vergnat, Judith Leyens, Ulrike Attenberger, and Daniel Kuetting. Image quality and radiation dose of dual source high pitch computed tomography in pediatric congenital heart disease. *Scientific Reports*, 12(1):9934, June 2022. ISSN 2045-2322. doi:10.1038/s41598-022-13404-w. URL <https://doi.org/10.1038/s41598-022-13404-w>.
- [6] Shota Kondo Ikuo Kawashita Isao Takahashi Kazuo Awai Yuko Nakamura, Toru Higaki. An introduction to photon-counting detector ct (pcd ct) for radiologists. *Japanese Radiology*, 41:266–282, 2023. doi:https://doi.org/10.1007/s11604-022-01350-6. URL <https://link.springer.com/article/10.1007/s11604-022-01350-6>.
- [7] Bjoern Kreisler. Photon counting detectors: Concept, technical challenges, and clinical outlook. *European Journal of Radiology*, 149:110229, 2022. ISSN 0720-048X. doi:https://doi.org/10.1016/j.ejrad.2022.110229. URL <https://www.sciencedirect.com/science/article/pii/S0720048X22000791>.
- [8] Siemens Healthineers. Photon-counting ct: A quantum leap in computed tomography. <https://www.siemens-healthineers.com/nl/computed-tomography/ct-technologies-and-innovations/photon-counting-ct>.
- [9] Judith van der Bie, Thom van der Laan, Marcel van Straten, Ronald Booi, Daniel Bos, Marcel L. Dijkshoorn, Alexander Hirsch, Edwin H.G. Oei, and Ricardo P.J. Budde. Photon-counting ct: An updated review of clinical results. *European Journal of Radiology*, 190:112189, 2025. ISSN 0720-048X. doi:https://doi.org/10.1016/j.ejrad.2025.112189. URL <https://www.sciencedirect.com/science/article/pii/S0720048X2500275X>.
- [10] Avinash K. Nehra, Kishore Rajendran, Francis I. Baf-four, Achille Mileto, Prabhakar Shantha Rajiah, Kelly K. Horst, Akitoshi Inoue, Tucker F. Johnson, Felix E. Diehn, Katrina N. Glazebrook, Jamison E. Thorne, Nikkole M. Weber, Elisabeth R. Shanblatt, Hao Gong, Lifeng Yu, Shuai Leng, Cynthia H. McCollough, and Joel G. Fletcher. Seeing more with less: Clinical benefits of photon-counting detector ct. *RadioGraphics*, 43(5):e220158, 2023. doi:10.1148/rg.220158. URL <https://doi.org/10.1148/rg.220158>.
- [11] Maffei E. Positano V. Clemente A. De Gori C. Berti S. La Grutta L. Saba L. Bossone E. Mantini C. Cavaliere C. Punzo B. Celi S. Cademartiri F. Meloni, A. Technical principles, benefits, challenges, and applications of photon counting computed tomography in coronary imaging: a narrative review. *Cardiovascular diagnosis and therapy*, 14(4):698–724, 2024. doi:10.21037/cdt-24-52. URL <https://doi.org/10.21037/cdt-24-52>.
- [12] Bernhard PhD Flohr, Thomas PhD; Schmidt. Technical basics and clinical benefits of photon-counting ct. *Investigative Radiology*, 58(7):441–450, 2023. doi:10.1097/RLL.0000000000000980. URL https://journals.lww.com/investigativeradiology/fulltext/2023/07000/technical_basics_and_clinical_benefits_of.2.aspx.
- [13] Kishore Rajendran, Michael Bruesewitz, Joseph Swicklik, Andrea Ferrero, Jamison Thorne, Lifeng Yu, Cynthia McCollough, and Shuai Leng. Task-based automatic kev selection: leveraging routine virtual monoenergetic imaging for dose reduction on clinical photon-counting detector ct*. *Physics in Medicine Biology*, 69(11):115029, may 2024. doi:10.1088/1361-6560/ad41b3. URL <https://pubmed.ncbi.nlm.nih.gov/38648795/>.
- [14] Bernhard PhD Flohr, Thomas PhD; Schmidt. Approaches, advantages, and challenges to photon counting detector and multi-energy ct. *Abdom Radiol*, 49:3251–3260, 2024. doi:https://doi.org/10.1007/s00261-024-04357-x. URL <https://link.springer.com/article/10.1007/s00261-024-04357-x#Abs1>.
- [15] Charbel Mourad, Lucia Gallego Manzano, Anaïs Viry, Ronald Booi, Edwin H. G. Oei, Fabio Becce, and Patrick Omoumi. Chances and challenges of photon-counting CT in musculoskeletal imaging. *Skeletal Radiology*, 53(9):1889–1902, September 2024. ISSN 1432-2161. doi:10.1007/s00256-024-04622-6. URL <https://doi.org/10.1007/s00256-024-04622-6>.
- [16] Martin J. Willemink, Mats Persson, Amir Pourmorteza, Norbert J. Pelc, and Dominik Fleischmann. Photon-counting ct: Technical principles and clinical prospects. *Radiology*, 289(2):293–312, 2018. doi:10.1148/radiol.2018172656. URL <https://doi.org/10.1148/radiol.2018172656>.
- [17] Matthew J Page, Joanne E McKenzie, Patrick M Bossuyt, Isabelle Boutron, Tammy C Hoffmann, Cynthia D Mulrow, Larissa Shamseer, Jennifer M Tetzlaff, Elie A Akl, Sue E Brennan, Roger Chou, Julie Glanville, Jeremy M Grimshaw, Asbjørn Hróbjartsson, Manoj M Lalu, Tianjing Li, Elizabeth W Loder, Evan Mayo-Wilson, Steve McDonald, Luke A McGuinness, Lesley A Stewart, James Thomas, Andrea C Tricco, Vivian A Welch, Penny Whiting, and David Moher. The prisma

- 2020 statement: an updated guideline for reporting systematic reviews. *BMJ*, 372, 2021. doi:10.1136/bmj.n71. URL <https://www.bmj.com/content/372/bmj.n71>.
- [18] Fedorowicz Z El-magarmid A. Ouzzani M, Hammady H. Rayyan-a web and mobile app for systematic reviews. *Systematic Reviews*, Volume 5, article number 210, 2016. doi:10.1186/s13643-016-0384-4. URL <https://link.springer.com/article/10.1186/s13643-016-0384-4>.
- [19] Claes Wohlin. Guidelines for snowballing in systematic literature studies and a replication in software engineering. In *Proceedings of the 18th International Conference on Evaluation and Assessment in Software Engineering*, EASE '14, New York, NY, USA, 2014. Association for Computing Machinery. ISBN 9781450324762. doi:10.1145/2601248.2601268. URL <https://doi.org/10.1145/2601248.2601268>.
- [20] T. Dirrichs, E. Tietz, A. Ruffer, J. Hanten, T. D. Nguyen, E. Dethlefsen, and C. K. Kuhl. Photon-counting versus dual-source ct of congenital heart defects in neonates and infants: Initial experience. *Radiology*, 307(5):e223088, 12 2023. ISSN 1527-1315. doi:10.1148/radiol.223088. URL <https://pubs.rsna.org/doi/full/10.1148/radiol.223088>.
- [21] Bouchra Habib Geryes, Salma Moalla, Esteban De Saint Etienne, Aurélie Moussier, François Pontana, Nathalie Boddaert, Valérie Pontvianne, Volodia Dangouloff-Ros, Laureline Berteloot, Cécile Lozach, Klervie Loiselet, Kahina Belhous, Raphael Levy, Nathalie Lassau, and Corinne Balleyguier. Photon-counting ct in pediatric patients: A multi-institutional evaluation of protocols and radiation exposure in 4,772 patients. *Diagnostic and Interventional Imaging*, 2025. ISSN 2211-5684. doi:<https://doi.org/10.1016/j.diii.2025.11.010>. URL <https://www.sciencedirect.com/science/article/pii/S2211568425002153>.
- [22] N. H. Dang, W. Zhou, G. M. Arguello Fletes, J. P. Weinman, and L. J. Malone. Balancing radiation dose and image quality in infants on photon-counting ct for pediatric cardiac imaging: comparing 70 kv and 120 kv protocols. *Pediatr Radiol*, 11 2025. ISSN 1432-1998. doi:10.1007/s00247-025-06467-0. URL <https://link.springer.com/article/10.1007/s00247-025-06467-0#citeas>.
- [23] S. Hellms, T. Werncke, J. Bottcher, C. M. Happel, J. Eckstein, M. B. Krueger, C. Panknin, A. Pfeil, T. F. Kaireit, P. Beerbaum, J. Vogel-Claussen, F. Wacker, and D. M. Renz. Reduction of radiation exposure and preserved image quality using photon-counting detector cardiac computed tomography without electrocardiographic gating in children with congenital heart disease. *Eur Radiol*, page 324–333, 7 2025. ISSN 1432-1084. doi:10.1007/s00330-025-11719-6. URL <https://link.springer.com/article/10.1007/s00330-025-11719-6#citeas>.
- [24] G. M. Arguello Fletes, W. Zhou, L. J. Malone, A. I. Fuentealba Cargill, J. P. Weinman, and L. P. Browne. Comparative radiation dose analysis in pediatric high-pitch cardiac cta using photon-counting versus energy-integrating detector ct. *Pediatric Radiology*, 55(12):2579–2589, 11 2025. doi:10.1007/s00247-025-06336-w. URL <https://link.springer.com/article/10.1007/s00247-025-06336-w#citeas>.
- [25] M. B. Krueger, T. Werncke, M. Eicke, N. Schwerk, J. Eckstein, C. Huisinga, C. Panknin, H. O. Shin, F. J. Silchmuller, R. E. Schultze-Florey, G. Hansen, F. Wacker, S. Hellms, and D. M. Renz. Photon-counting detector ct enables pediatric low-dose chest imaging with further reduction of radiation exposure. *Investigative Radiology*, 9 2025. ISSN 1536-0210. doi:10.1097/rli.0000000000001234. URL http://journals.lww.com/investigativeradiology/abstract/9900/diagnostic_and_technological_advances_in_magnetic.340.aspx.
- [26] F. Stalhammar, M. L. Aurumskjold, S. Meyer, M. Wiklund, P. Wingren, P. Liuba, and E. Hedstrom. Photon-counting computed tomography for paediatric congenital heart defects yields images of high diagnostic quality with low radiation doses at both 70 kv and 90 kv. *Pediatric Radiology*, 54(7):1187–1196, 12 2024. doi:10.1007/s00247-024-05939-z. URL <https://link.springer.com/article/10.1007/s00247-024-05939-z>.
- [27] M. J. Siegel and J. C. Ramirez-Giraldo. Photon counting detector computed tomography in pediatric cardiothoracic ct imaging. *Radiology advances*, 1(2), 7 2024. ISSN 2976-9337. doi:10.1093/radadv/umae012. URL <https://pmc.ncbi.nlm.nih.gov/articles/PMC12429202/>.
- [28] A. S. Perera Molligoda Arachchige and Y. Verma. Role of photon-counting computed tomography in pediatric cardiovascular imaging. *World j. clin. pediatr.*, 14(1):99288, 3 2025. ISSN 2219-2808. doi:10.5409/wjcp.v14.i1.99288. URL <https://pmc.ncbi.nlm.nih.gov/articles/PMC11686577/>.
- [29] F. L. Pasadyn and A. M. El-Ali. Photon-counting ct in children: A literature review. *Journal of Computer Assisted Tomography*, 10 2025. ISSN 1532-3145. doi:10.1097/rct.0000000000001810. URL https://journals.lww.com/jcat/abstract/9900/photon_counting_ct_in_children_a_literature.503.aspx.

APPENDIX

Appendix A

- The following search query was used to perform the literature search in medline: (Photons / AND Tomography, X-Ray Computed / OR ((photon* ADJ3 count* ADJ6 comput* ADJ3 tomogra*) OR (photon* ADJ3 count* ADJ6 (ct OR cta OR angiogra*)) OR NAEOTOM-Alpha).ab,ti,kw. OR (photon-count*).ti.) AND (exp Child/ OR Pediatrics / OR exp Intensive Care Units, Pediatric / OR exp Heart Defects, Congenital / OR (newborn* OR neonat* OR child* OR pediater* OR paediatr* OR ((congenital* OR malform*) ADJ3 (heart* OR cardiac*))).ab,ti,kw.)
- The following search query was used to perform the literature search in embase: ('photon counting computed tomography'/de OR 'naeotom alpha'/de OR ((photon* NEAR/3 count* NEAR/6 comput* NEAR/3 tomogra*) OR (photon* NEAR/3 count* NEAR/6 (ct OR cta OR angiogra*)) OR NAEOTOM-Alpha):ab,ti,kw OR (photon-count*):ti) AND (child/exp OR pediatrics/de OR 'pediatric intensive care unit'/exp OR 'congenital heart disease'/exp OR (newborn* OR neonat* OR child* OR pediater* OR paediatr* OR ((congenital* OR malform*) NEAR/3 (heart* OR cardiac*))).ab,ti,kw)
- The following search query was used to perform the literature search in Web of science: (TS=((photon* NEAR/2 count* NEAR/5 comput* NEAR/2 tomogra*) OR (photon* NEAR/2 count* NEAR/5 (ct OR cta OR angiogra*)) OR NAEOTOM-Alpha) OR TI=(photon-count*)) AND (TS=(newborn* OR neonat* OR child* OR pediater* OR paediatr* OR ((congenital* OR malform*) NEAR/2 (heart* OR cardiac*))))
- The following search query was used to perform the literature search in Scopus: (TITLE-ABS-KEY((photon* W/2 count* W/5 comput* W/2 tomogra*) OR (photon* W/2 count* W/5 (ct OR cta OR angiogra*)) OR NAEOTOM-Alpha) OR TITLE(photon-count*)) AND (TITLE-ABS-KEY(newborn* OR neonat* OR child* OR pediater* OR paediatr* OR ((congenital* OR malform*) W/2 (heart* OR cardiac*))))

B

Phantom design

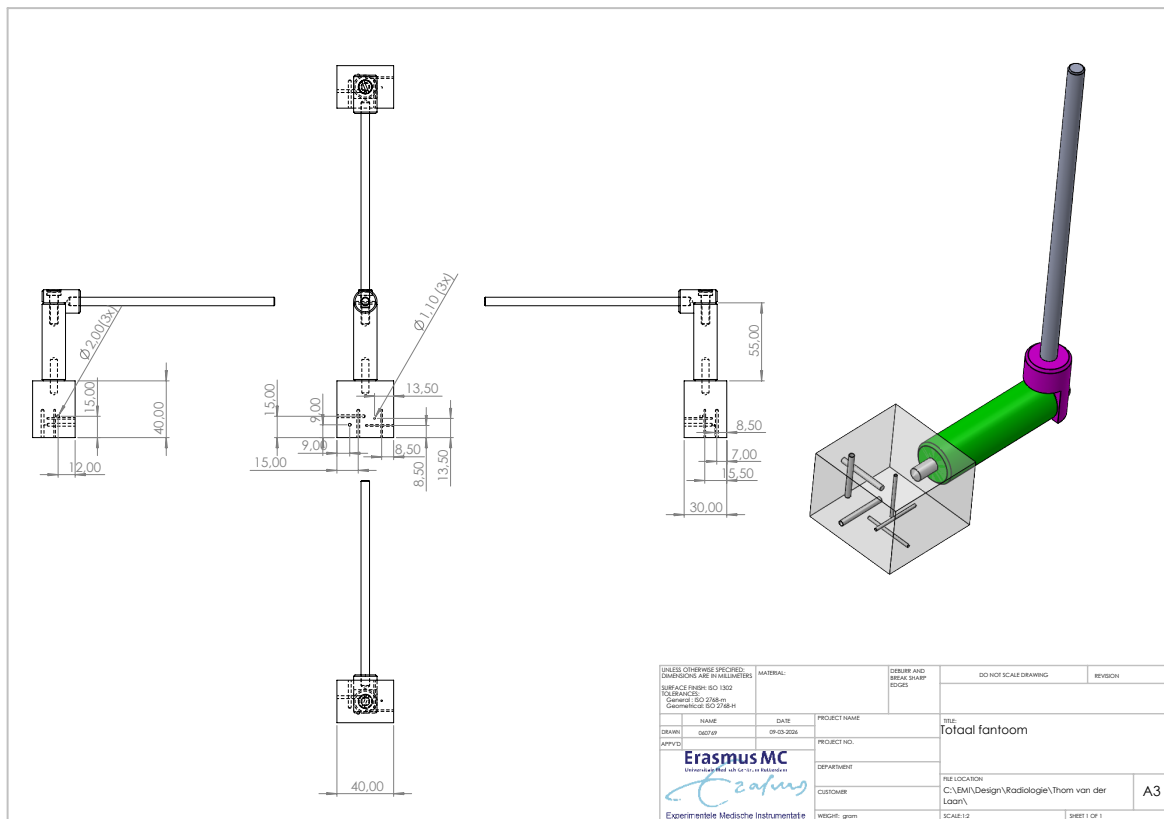
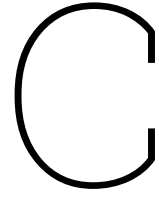


Figure B.1: Final phantom draft including the attachment piece



Cardiac motion profiles generation

Figure C.1 show the generated cardiac motion profiles for the x- and y-directions.

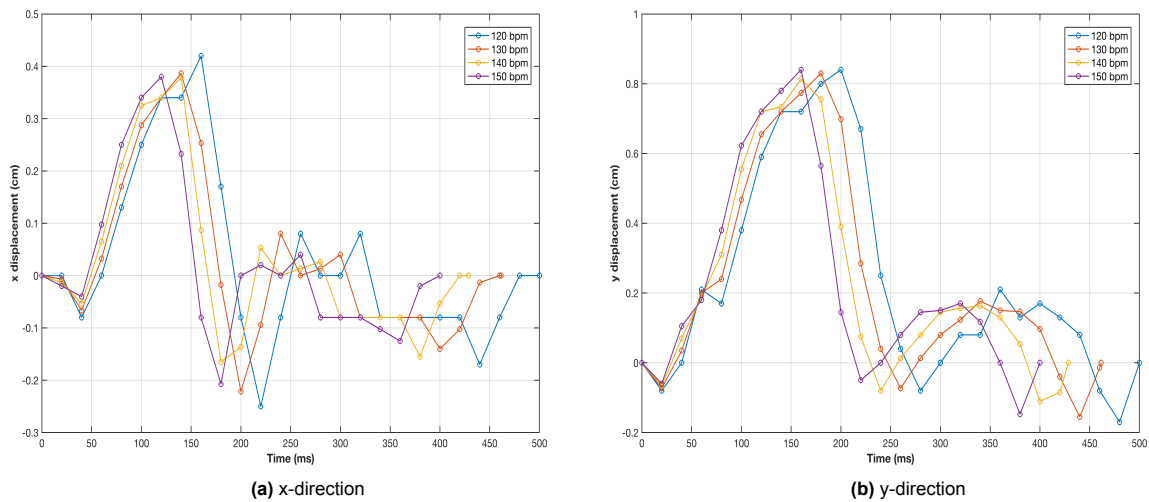


Figure C.1: Comparison of the a) x-direction displacement and b) y-direction displacement of the generated cardiac motion profiles for the simulated neonatal heart rates. Cardiac cycle duration shortened with increasing heart rate while preserving the overall displacement pattern.

The motion profiles for the higher heart rates were generated from a reference profile at 120 bpm, provided in Table C.1. This reference profile describes the displacement in the x -, y -, and z -directions over one cardiac cycle of 500 ms.

The reference profile was stored in MATLAB as a four-column matrix containing time and the three displacement components. The full reference profile is shown in Table C.1. In the script, the columns were extracted as separate vectors for interpolation:

```
ref_bpm = 120;  
T_ref = 60000 / ref_bpm;  
  
t_ref = ref(:,1);  
x_ref = ref(:,2);  
y_ref = ref(:,3);  
z_ref = ref(:,4);
```

Motion profiles for 130, 140, and 150 bpm were generated by temporally compressing the 120 bpm reference profile. The reference cycle was divided into a first phase from 0 to 200 ms and a second phase from 200 to 500 ms. The first phase represented the systolic or high-motion part of the cycle,

Table C.1: Reference motion profile at 120 bpm.

Time (ms)	x (cm)	y (cm)	z (cm)
0	0.00	0.00	0.00
20	0.00	-0.08	0.46
40	-0.08	0.00	0.97
60	0.00	0.21	1.09
80	0.13	0.17	1.22
100	0.25	0.38	1.30
120	0.34	0.59	1.51
140	0.34	0.72	1.56
160	0.42	0.72	1.47
180	0.17	0.80	1.47
200	-0.08	0.84	1.43
220	-0.25	0.67	1.18
240	-0.08	0.25	0.67
260	0.08	0.04	0.34
280	0.00	-0.08	0.08
300	0.00	0.00	0.13
320	0.08	0.08	0.17
340	-0.08	0.08	-0.08
360	-0.08	0.21	0.04
380	-0.08	0.13	0.08
400	-0.08	0.17	0.00
420	-0.08	0.13	-0.08
440	-0.17	0.08	-0.25
460	-0.08	-0.08	-0.67
480	0.00	-0.17	-0.59
500	0.00	0.00	0.00

while the second phase represented the diastolic or lower-motion part. To approximate the non-uniform shortening of the cardiac cycle at higher heart rates, 40% of the total shortening was assigned to the first phase and 60% to the second phase.

```
S_ref = 200;
D_ref = T_ref - S_ref;

frac_shortening_phase1 = 0.4;
frac_shortening_phase2 = 0.6;

T_new = 60000 / bpm;
deltaT = T_ref - T_new;

deltaS = frac_shortening_phase1 * deltaT;
deltaD = frac_shortening_phase2 * deltaT;

S_new = S_ref - deltaS;
D_new = D_ref - deltaD;
```

For each target heart rate, a new time vector was created using approximately 20 ms spacing. If the target cycle duration was not an exact multiple of 20 ms, the exact endpoint was added to ensure that the generated profile covered one complete cardiac cycle.

```
t_new = (0:20:floor(T_new/20)*20).';

if abs(t_new(end) - T_new) > 1e-9
t_new = [t_new; T_new];
end
```

The new time points were then mapped back to equivalent time points in the original 120 bpm profile. This was done separately for the two phases, so that the first phase of the shortened profile corresponded to the first 200 ms of the reference profile and the second phase corresponded to the remaining 300 ms. The MATLAB implementation was:

```
t_old = zeros(size(t_new));

idx_phase1 = t_new <= S_new;
idx_phase2 = t_new > S_new;

t_old(idx_phase1) = t_new(idx_phase1) * (S_ref / S_new);

t_old(idx_phase2) = S_ref + ...
(t_new(idx_phase2) - S_new) * (D_ref / D_new);
```

Equivalently, the time mapping can be written as:

$$t_{\text{old}} = \begin{cases} t_{\text{new}} \cdot \frac{S_{\text{ref}}}{S_{\text{new}}}, & t_{\text{new}} \leq S_{\text{new}}, \\ S_{\text{ref}} + (t_{\text{new}} - S_{\text{new}}) \cdot \frac{D_{\text{ref}}}{D_{\text{new}}}, & t_{\text{new}} > S_{\text{new}}. \end{cases} \quad (\text{C.1})$$

The displacement components for the generated profile were obtained by linear interpolation of the original reference profile at the mapped time points:

```
x_new = interp1(t_ref, x_ref, t_old, 'linear');
y_new = interp1(t_ref, y_ref, t_old, 'linear');
z_new = interp1(t_ref, z_ref, t_old, 'linear');
```

Finally, the first and last displacement values were set equal to those of the reference profile, ensuring that each profile started and ended at the neutral position:

```
x_new(1) = x_ref(1);  
y_new(1) = y_ref(1);  
z_new(1) = z_ref(1);  
  
x_new(end) = x_ref(end);  
y_new(end) = y_ref(end);  
z_new(end) = z_ref(end);
```

The final profile for each target heart rate was stored as a four-column table containing time, (x)-, (y)- and (z)-displacement, and exported as a CSV file for use in the cardiac motion simulator.

```
profiles(k).table = [t_new, x_new, y_new, z_new];
```

This procedure generated heart-rate-specific profiles with shorter cardiac cycle durations while preserving the displacement pattern and amplitude of the original 120 bpm reference trajectory.

D

Segmentation and registration code

D.1. Selected MATLAB functions used for image analysis

The full MATLAB script used for DICOM import, registration, segmentation, and metric extraction was not included in this thesis because of its length and the large number of auxiliary visualization and export routines. Instead, this appendix presents selected function fragments that illustrate the main structure of the analysis pipeline.

The complete MATLAB scripts used for DICOM preprocessing, segmentation, registration, metric extraction, statistical analysis, and figure generation were retained together with the processed phantom-derived metric tables. These materials were archived with the project files and can be made available for reproducibility of the quantitative analyses, subject to institutional data-management requirements.

D.1.1. Recursive DICOM import and metadata extraction

The first step of the pipeline consisted of recursively scanning the selected folder and extracting the DICOM metadata required for volume reconstruction.

```
function dicomFiles = scanDicomFilesRecursive(rootFolder)
    fileList = dir(fullfile(rootFolder, '**', '*'));
    fileList = fileList(~[fileList.isdir]);

    for k = 1:numel(fileList)
        fullPath = fullfile(fileList(k).folder, fileList(k).name);
        info = dicominfo(fullPath);
        entry.instanceNumber = info.InstanceNumber;
        entry.imagePositionPatient = info.ImagePositionPatient;
        entry.imageOrientationPatient = info.ImageOrientationPatient;
        dicomFiles(end+1) = entry;
    end
end
```

This function was used to automatically identify all readable DICOM files and extract the metadata necessary for sorting and reconstructing the image series. A recursive approach was chosen because the exported datasets contained multiple nested folders and manually sorting all files would have been inefficient and prone to error.

D.1.2. Slice position calculation

To reconstruct a correct 3D volume, slice order was determined from spatial metadata rather than relying only on instance number.

```
function slicePos = computeSlicePosition(imagePositionPatient, imageOrientationPatient)
    rowDir = imageOrientationPatient(1:3);
```

```

    colDir = imageOrientationPatient(4:6);
    sliceNormal = cross(rowDir, colDir);
    sliceNormal = sliceNormal / norm(sliceNormal);
    slicePos = dot(imagePositionPatient(1:3), sliceNormal);
end

```

This function calculates the physical slice position from the DICOM image position and orientation vectors. This method was used because instance numbers are not always sufficient for reliable sorting when multiple reconstructions or repeated series are present. The use of geometry-based positioning improved the robustness of volume reconstruction.

D.1.3. HU-based volume construction

After sorting, the selected image slices were stacked into a 3D volume and converted to Hounsfield Units using the DICOM rescale parameters.

```

I = double(dicomread(filePath));
HU = I * info.RescaleSlope + info.RescaleIntercept;
volHU(:,:,k) = single(HU);

```

This conversion was necessary because quantitative CT analysis should be performed in Hounsfield Units rather than raw pixel values. Using HU values ensured that attenuation-based metrics, such as mean tube attenuation and contrast-to-noise ratio, were physically interpretable and comparable across reconstructions.

D.1.4. Initial tube mask generation

A first segmentation of the tubes was obtained by thresholding the reconstructed HU volume.

```

mask = (volHU >= huMin) & (volHU <= huMax);
mask = bwareaopen(mask, minComponentSize);
mask = imfill(mask, 'holes');

```

This function was used to generate an initial binary tube mask by selecting a predefined HU range corresponding to the high-attenuation tube material. Small isolated components were removed and holes were filled to obtain more coherent objects. This step provided the basis for both the registration procedure and the later component-based metric extraction.

D.1.5. Centroid-based alignment

Before comparing the two volumes, an initial rigid alignment was estimated from the centroids of the segmented tube masks.

```

fixedCentroid = computeMaskCentroidMM(fixedMask, dx, dy, dz);
movingCentroid = computeMaskCentroidMM(movingMask, dx, dy, dz);
translationVec = fixedCentroid - movingCentroid;

```

This translation-based prealignment was used to compensate for small global offsets between the fixed and moving datasets. A centroid-based approach was chosen because it is computationally simple, robust, and appropriate for phantom data with a limited number of well-defined high-contrast structures.

D.1.6. DSC-based overlap analysis

To quantify spatial agreement between the segmented tubes before and after registration, the Dice similarity coefficient was calculated.

```

d = 2 * nnz(A & B) / (nnz(A) + nnz(B));

```

The Dice coefficient was used as an overlap metric because it directly quantifies the spatial similarity between two binary masks. In this study, both strict and tolerant Dice scores were calculated, allowing assessment of exact overlap as well as overlap within a small spatial margin.

D.1.7. Cross-sectional metric extraction

For each tube component, local cross-sectional properties were measured at several offsets along the tube axis.

```
CrossSectionArea_mm2 = aggregateSliceMetric(sliceRows.CrossSectionArea_mm2, 'median');  
MeasuredDiameter_mm  = aggregateSliceMetric(sliceRows.MeasuredDiameter_mm, 'median');  
Circularity           = aggregateSliceMetric(sliceRows.Circularity, 'median');
```

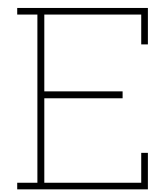
This approach was used because a single slice may be sensitive to local irregularities, noise, or partial volume effects. By sampling multiple cross-sections and aggregating the results using the median, the resulting tube metrics were made more robust and representative of the overall tube geometry.

D.1.8. PMMA-based contrast reference

To estimate contrast relative to the phantom background, a PMMA reference region was automatically identified.

```
cnr = abs(meanHU - backgroundMeanHU) / backgroundStdHU;
```

This reference was used to calculate the contrast-to-noise ratio of the tube relative to the surrounding PMMA material. Using a local phantom-specific reference was preferred over a global image background, because it better reflected the actual material context in which the tube structures were embedded.



Statistical analysis

E.1. Selected MATLAB functions used for statistical analysis

The full MATLAB scripts used for the statistical analysis were not included in this thesis because they mainly consisted of data cleaning, model fitting, figure generation, and export routines. Instead, this appendix presents selected code fragments that illustrate the main statistical workflow. For each fragment, a brief explanation is given together with the methodological rationale for its use in the present study. The complete statistical analysis scripts and processed metric tables were retained with the project archive and can be made available to reproduce the reported models, descriptive summaries, post-hoc comparisons, and diagnostic plots.

Two types of statistical analyses were performed. First, an analysis was carried out for the within-protocol and inter-protocol comparisons under motion conditions. Second, a separate static inter-protocol analysis was performed at 0 bpm to assess baseline comparability between the acquisition protocols in the absence of motion. The same general workflow was used in both scripts: data import, cleaning, standardization of variables, selection of analysis-specific subsets, descriptive summaries, linear mixed-effects modelling, and export of results.

E.1.1. Reading and cleaning workbook sheets

The statistical workflow started by reading the relevant worksheet tabs from the Excel workbook and removing rows that did not contain actual tube-level data.

```
T = readtable(inputFile, 'Sheet', sheetName, ...  
    'VariableNamingRule', 'preserve');  
T = T(keep, :);  
T = T(lower(strtrim(v)) ~= "tube_id", :);
```

This step was used to import the metric tables while preserving the original variable names. Empty rows, repeated header rows, and note rows were removed to ensure that only valid observations were retained for analysis. This cleaning step was necessary because the workbook structure contained intermediate labels and annotations that were not part of the quantitative dataset.

E.1.2. Standardization of variable names

To make the different workbook tabs directly comparable, variable names were first standardized.

```
names = lower(T.Properties.VariableNames);  
names = regexp(names, '[^a-z0-9]+', '_');  
names = matlab.lang.makeUniqueStrings(names);
```

This procedure converted variable names to a consistent lowercase format and replaced spaces or special characters by underscores. Standardization was necessary because the original workbook tabs did not always use identical naming conventions. Harmonizing the variable names reduced the

risk of downstream coding errors and allowed the same statistical functions to be applied across all comparison datasets.

E.1.3. Preparation of categorical analysis variables

After import, several variables were explicitly recoded as categorical analysis factors.

```
T.tube_id = categorical(string(T.tube_id));
T.protocol = categorical(normalizeProtocolName(T.protocol));
T.kev = categorical(string(T.kev));
T.heart_rate_bpm = categorical(string(T.heart_rate_bpm));
```

These conversions ensured that protocol, reconstruction energy, heart rate, and tube identity were treated as categorical factors rather than continuous numeric variables. This was appropriate because the study design consisted of predefined experimental conditions rather than continuously sampled values.

E.1.4. Derivation of tube-specific grouping variables

Additional grouping variables were derived from the tube identifiers.

```
T.diameter_group = inferDiameterGroup(T.tube_id);
T.orientation = inferOrientation(T.tube_id);
```

These derived variables were used to distinguish between the 1 mm and 2 mm tubes and between the three spatial tube orientations. Including these factors allowed the statistical analysis to account for known structural differences in imaging difficulty between tube sizes and orientations.

E.1.5. Selection of comparison-specific subsets

Separate subsets were created for each predefined analysis.

```
keep = str2double(string(T.kev)) == dynamicKeV & ...
    ismember(str2double(string(T.heart_rate_bpm)), dynamicHeartRates);
T = T(keep, :);
```

This filtering step was used to isolate the observations relevant to each analysis: within-flash comparisons, within-prospective comparisons, matched-phase inter-protocol comparisons, and best-phase inter-protocol comparisons. Restricting each analysis to a predefined subset ensured that the models directly addressed the relevant research questions and avoided mixing fundamentally different comparison types.

For the separate static baseline analysis, the data were filtered to the 0 bpm acquisitions at 40, 47, and 53 keV.

```
keep = ismember(str2double(string(Tin.kev)), staticKeV) & ...
    str2double(string(Tin.heart_rate_bpm)) == staticHeartRate;
T = Tin(keep, :);
```

This filtering step isolated the true static baseline condition. Restricting the analysis to 0 bpm ensured that the inter-protocol comparison was not influenced by motion and could therefore be interpreted as a baseline comparison of image quality and geometric fidelity.

E.1.6. Definition of the analyzed metrics

The statistical scripts used a predefined map of supported quantitative endpoints.

```
metricMap = {
    'eqdiametererror_mm', 'Equivalent diameter error (mm)';
    'areaerror_mm2',      'Area error (mm^2)';
    'ererror_mm',         'Edge rise distance error (mm)';
    'dice3d',             '3D Dice similarity coefficient';
    'circularity',         'Circularity';
    'attenuation_bias_hu', 'Attenuation bias (HU)';
```

```

        'cnr',                                'Contrast-to-noise ratio'
    };

```

This structure ensured that only supported metrics were included in the formal statistical analyses and that each variable could be linked to a clear label. In the static analysis, this set was expanded where relevant to include directly measured static quantities such as cross-sectional area and equivalent diameter error relative to the nominal tube diameter.

```

'eqdiametererror_nominal_mm', ...
'Equivalent diameter error to nominal diameter (mm)';
'crosssectionarea_mm2', ...
'Cross-sectional area (mm2)';

```

These additional metrics were included because the static analysis focused on baseline protocol comparability rather than motion-related deviation from a reference reconstruction.

E.1.7. Descriptive statistical summary

Before model fitting, descriptive statistics were calculated for each comparison and metric.

```

G.mean = splitapply(@(v) mean(v, 'omitnan'), x, Gid);
G.std = splitapply(@(v) std(v, 0, 'omitnan'), x, Gid);
G.median = splitapply(@(v) median(v, 'omitnan'), x, Gid);
G.iqr = splitapply(@(v) iqr(v(~isnan(v))), x, Gid);

```

Descriptive summaries were used to provide an overview of the central tendency and spread of each image-quality metric within each protocol condition. The inclusion of both mean/standard deviation and median/interquartile range allowed the distributions to be described robustly, even when some metrics were skewed or affected by outliers.

For the dynamic analyses, the grouping structure depended on the comparison type.

```
groupVars = {'protocol','kev','heart_rate_bpm','diameter_group'};
```

For the prospective within-protocol analysis, reconstruction phase was additionally included.

```
groupVars{end+1} = 'phase_percent';
```

For the static baseline analysis, the descriptive summaries were grouped by protocol, keV level, and diameter group.

```
groupVars = {'protocol','kev','diameter_group'};
```

These grouping structures were chosen so that the descriptive tables directly reflected the underlying experimental design.

E.1.8. Mixed-effects modelling

For inferential analysis, linear mixed-effects models were fitted separately for each metric and comparison.

```
mdl = fitlme(Ti, formula, 'FitMethod', 'REML');
```

Mixed-effects models were used because repeated measurements were available from the same tube structures across multiple protocol conditions. By including tube identity as a random intercept, the analysis accounted for within-tube dependence while estimating the effects of protocol-related factors such as energy, heart rate, phase, or acquisition type.

E.1.9. Model structure for the dynamic analyses

Different fixed effects were specified depending on the type of dynamic comparison being tested.

For the flash within-protocol motion analysis, the following model structure was used:

```
metric ~ kev + heart_rate_bpm + diameter_group + (1|tube_id)
```

This model was used to test whether image-quality metrics varied across reconstruction energy, heart rate, and tube diameter group within the flash acquisitions.

For the prospective within-protocol motion analysis, reconstruction phase replaced keV as the main varying acquisition factor at fixed energy:

```
metric ~ heart_rate_bpm + phase_percent_cat + diameter_group + (1|tube_id)
```

This model was appropriate because all prospective motion analyses were performed at 53 keV, while multiple reconstruction phases were available and therefore represented the principal additional factor of interest.

For the inter-protocol comparisons at 53 keV, the main comparison of interest was acquisition protocol:

```
metric ~ protocol + heart_rate_bpm + diameter_group + (1|tube_id)
```

This structure was used for both the matched-phase and best-phase comparisons, allowing the effect of protocol to be assessed while accounting for heart rate and tube diameter group.

E.1.10. Model structure for the static baseline analysis

For the static 0 bpm inter-protocol baseline analysis, a separate mixed-effects model was used.

```
metric ~ protocol + kev + diameter_group + (1|tube_id)
```

This model was used to estimate the effects of acquisition protocol and reconstruction energy under static conditions while accounting for repeated observations from the same tube by including tube identity as a random intercept. The mixed-effects approach was retained for consistency with the dynamic analyses and because observations from the same physical tube structure were not statistically independent.

The script also allowed inclusion of an optional protocol-by-energy interaction term:

```
metric ~ protocol * kev + diameter_group + (1|tube_id)
```

This interaction term was included in the script to test whether the difference between flash and prospective imaging depended on the selected virtual monoenergetic reconstruction level. In the present analysis, this interaction was disabled by default, such that the main-effects model was used unless explicit exploration of a protocol-dependent keV effect was required.

E.1.11. Model estimability checks

Before fitting the models, the script verified whether sufficient variation was present in the fixed effects.

```
tf = nLevels(T.kev) > 1 && ...  
    nLevels(T.heart_rate_bpm) > 1 && ...  
    nLevels(T.diameter_group) > 1;
```

This check was included to prevent fitting models to degenerate datasets in which one or more predictors had only a single level. Such models would not yield interpretable estimates for the corresponding fixed effects.

For the static analysis, an analogous check was performed for protocol, keV level, and diameter group.

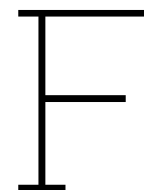
```
tf = nLevels(T.protocol) > 1 && ...  
    nLevels(T.kev) > 1 && ...  
    nLevels(T.diameter_group) > 1;
```

E.1.12. Extraction of coefficient estimates and omnibus tests

After fitting, the coefficient estimates and omnibus statistical tests were extracted and stored in standardized output tables.

```
coefTable = standardizeCoefficientTable mdl.Coefficients;  
anovaTable = standardizeAnovaTable(anova mdl);
```

These outputs were used to summarize both the estimated effect sizes and the overall statistical significance of each model term. Coefficient tables supported interpretation of the direction and magnitude of the effects, whereas the ANOVA tables provided a global test of whether each predictor contributed significantly to variation in the image-quality metric.



Model diagnostics

F.1. Residual diagnostics of representative mixed-effects models

To assess whether the linear mixed-effects models provided a reasonable description of the data, residual diagnostics were evaluated visually for the fitted models. For each model, a quantile–quantile (QQ) plot of the residuals and a residuals-versus-fitted plot were generated. The QQ plots were used to assess approximate normality of the residuals, whereas the residuals-versus-fitted plots were used to assess whether the residuals were distributed approximately randomly around zero without strong systematic patterns.

Figures F.1–F.5 show representative diagnostic plots for the main analysis types included in this thesis: flash within-protocol analysis, prospective within-protocol analysis, z-matched inter-protocol comparison, best-phase inter-protocol comparison, and static inter-protocol comparison. The selected examples focus on contrast-to-noise ratio (CNR), Dice similarity coefficient (DSC), and edge rise distance (ERD), because these metrics represent complementary aspects of image quality, segmentation overlap, and edge sharpness.

Across the full set of inspected models, the residual diagnostics were broadly similar. Most QQ plots were approximately linear in the central range, with deviations mainly occurring in the tails or in a small number of outlying observations. The residuals-versus-fitted plots generally showed scatter around zero without a strong systematic pattern. The ERD models showed somewhat more structured clustering and larger residual spread than CNR or DSC, consistent with the lower observed-versus-fitted agreement for this metric. However, these patterns were not severe enough to indicate that the mixed-effects models were inappropriate for the supportive inferential analyses.

Overall, the diagnostics showed some deviation from ideal normality and constant variance, particularly in the tails and in a limited number of outlying observations, but no severe or consistent violations were observed that would preclude the use of the mixed-effects models. The model results were therefore retained and interpreted together with the descriptive trends and effect estimates.

F.2. Observed-versus-fitted assessment

Observed-versus-fitted plots were used as a complementary check of model fit. In these figures, the left panel shows observed and fitted values ordered by fitted value, and the right panel compares observed values directly with fitted values using the identity line as a reference. These plots were not used as standalone evidence of model validity, but helped assess whether the fitted values captured the main level and ordering of the measurements.

The selected fitted-value plots mirror the representative residual diagnostics above. Across the inspected models, the fitted values generally captured the main ordering of the observations. The clearest agreement was observed for CNR, which followed strong systematic differences across energy, protocol, and motion settings. DSC also showed reasonable agreement, although with more observation-level scatter. ERD showed a less tight observed-versus-fitted relationship, reflecting a different pattern

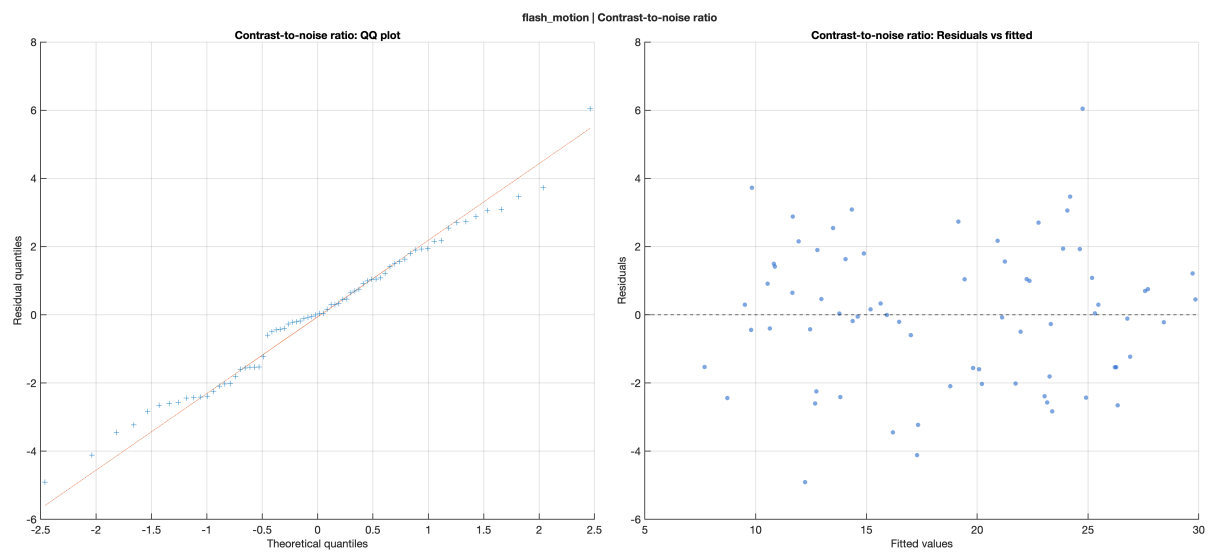


Figure F.1: Residual diagnostics for the flash within-protocol mixed-effects model of CNR. The QQ plot was approximately linear in the central range with some deviation in the tails, while the residuals-versus-fitted plot showed no marked systematic pattern.

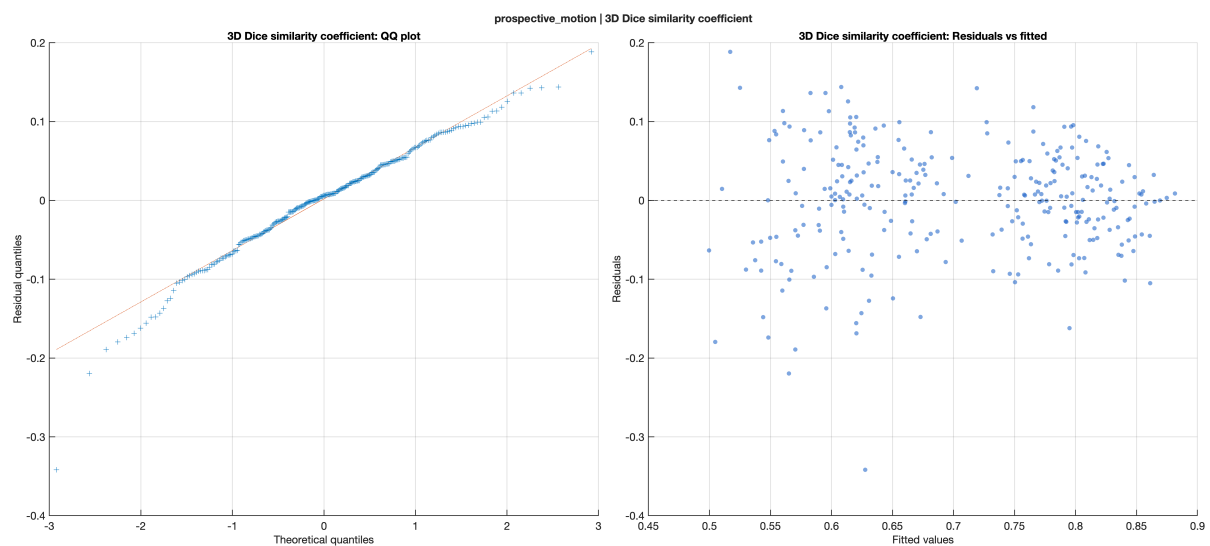


Figure F.2: Residual diagnostics for the prospective within-protocol mixed-effects model of DSC. The QQ plot showed approximate linearity with some deviation in the lower tail, and the residuals-versus-fitted plot showed moderate clustering but no pronounced funnel shape or strong nonlinear pattern.

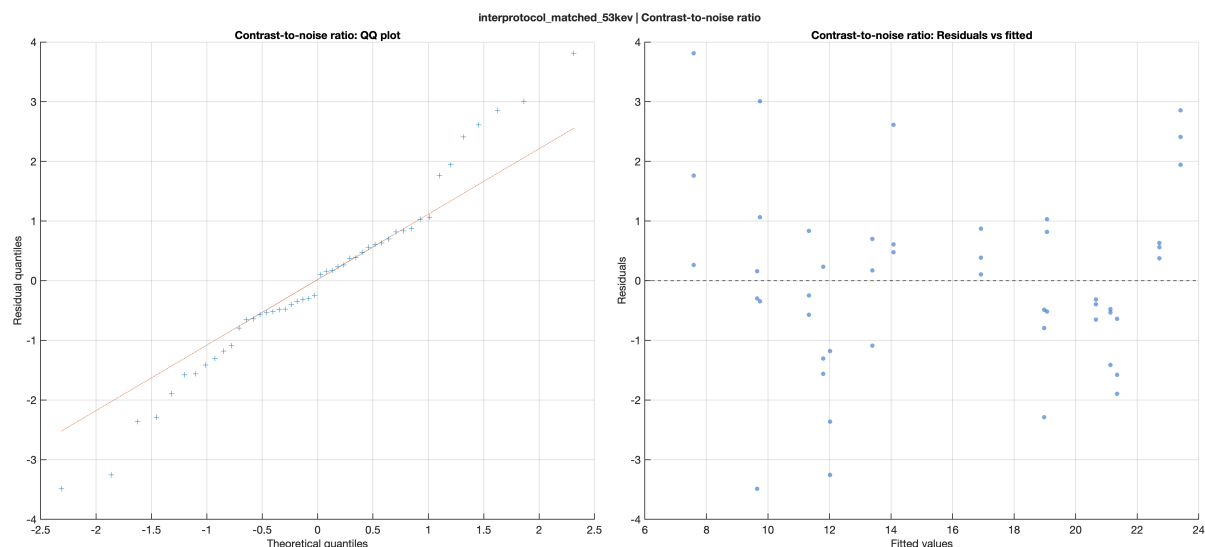


Figure F.3: Residual diagnostics for the matched-phase inter-protocol mixed-effects model of CNR at 53 keV. The QQ plot showed moderate tail deviation, while the residuals-versus-fitted plot showed no clear systematic structure, supporting cautious use of the model for inferential interpretation.

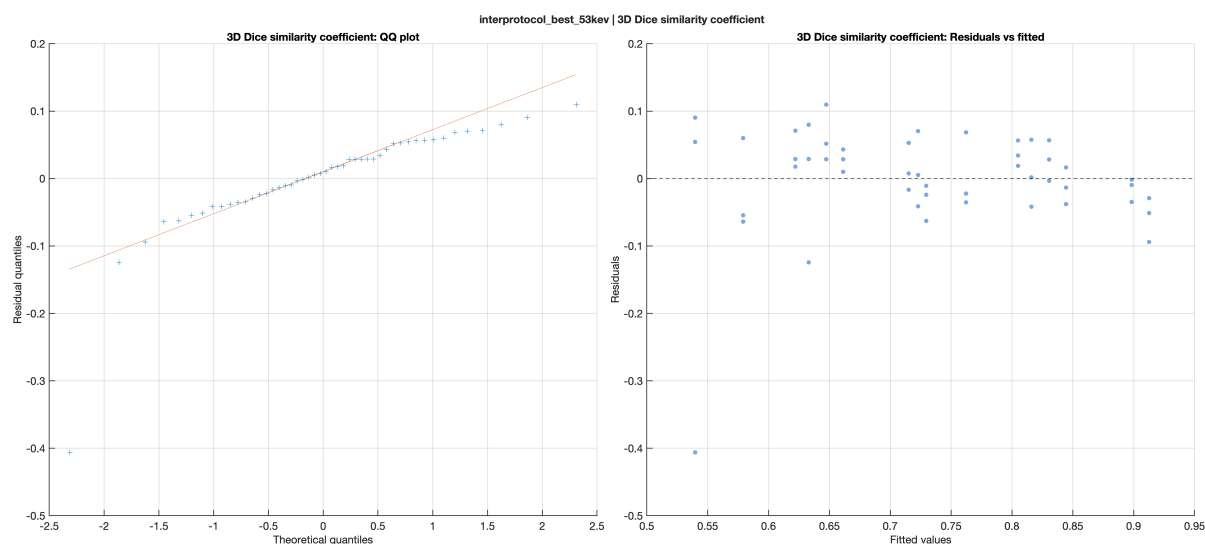


Figure F.4: Residual diagnostics for the best-phase inter-protocol mixed-effects model of DSC at 53 keV. The QQ plot showed approximate linearity with one more pronounced lower-tail deviation, and the residuals-versus-fitted plot showed generally random scatter around zero with a limited number of outlying residuals.

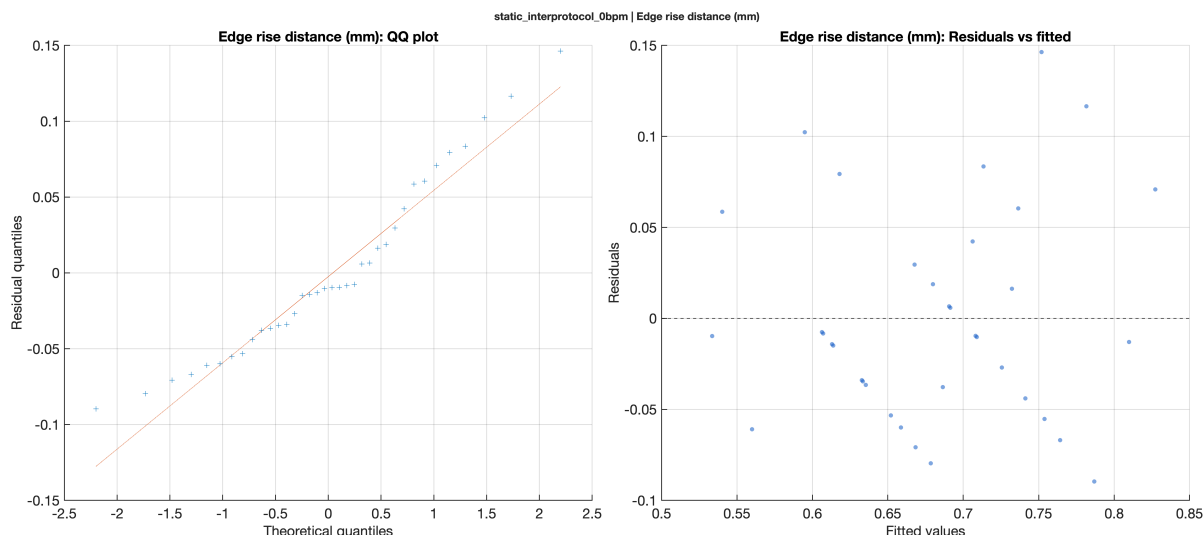


Figure F.5: Residual diagnostics for the static inter-protocol mixed-effects model of ERD at 0 bpm. The QQ plot and residuals-versus-fitted plot showed no major systematic deviations, supporting use of the model as a baseline comparability analysis under non-moving conditions.

in the data points and greater sensitivity to local edge definition. This weaker fit was taken into account by interpreting the ERD model results cautiously and together with the descriptive trends.

Thus, the observed-versus-fitted assessment supported the use of the mixed-effects models as supportive inferential analyses, while also showing that model fit varied between metrics. In particular, CNR was described best by the fixed effects included in the model, whereas ERD retained more unexplained variability.

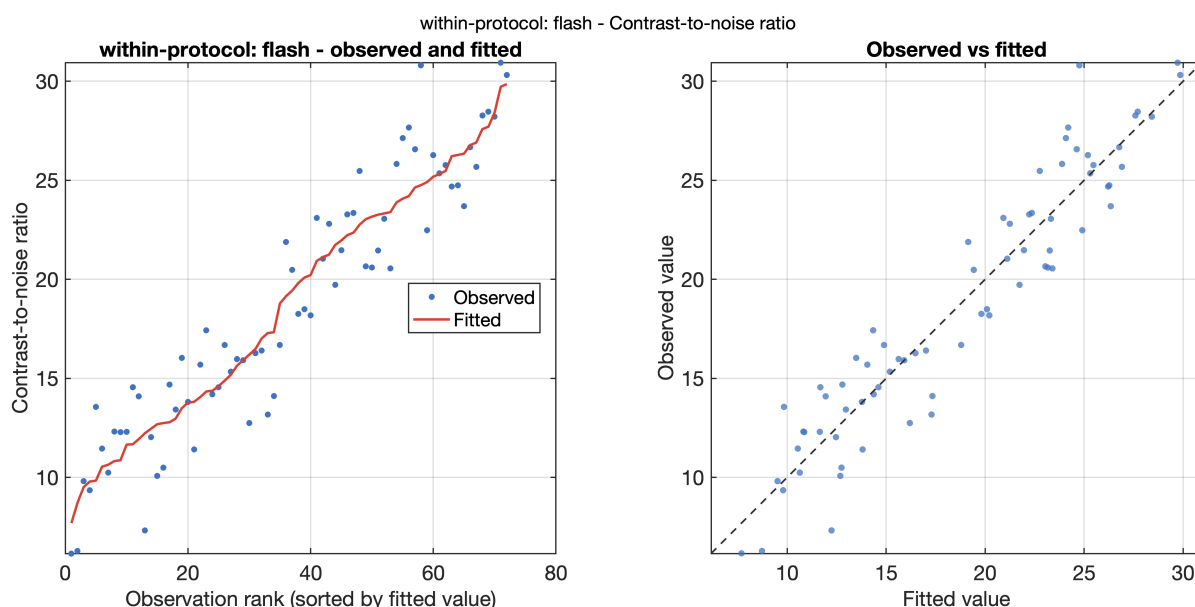


Figure F.6: Observed-versus-fitted assessment for the FLASH within-protocol CNR model. The fitted values captured the main ordering of the observations, with an observed-versus-fitted R^2 of 0.90.

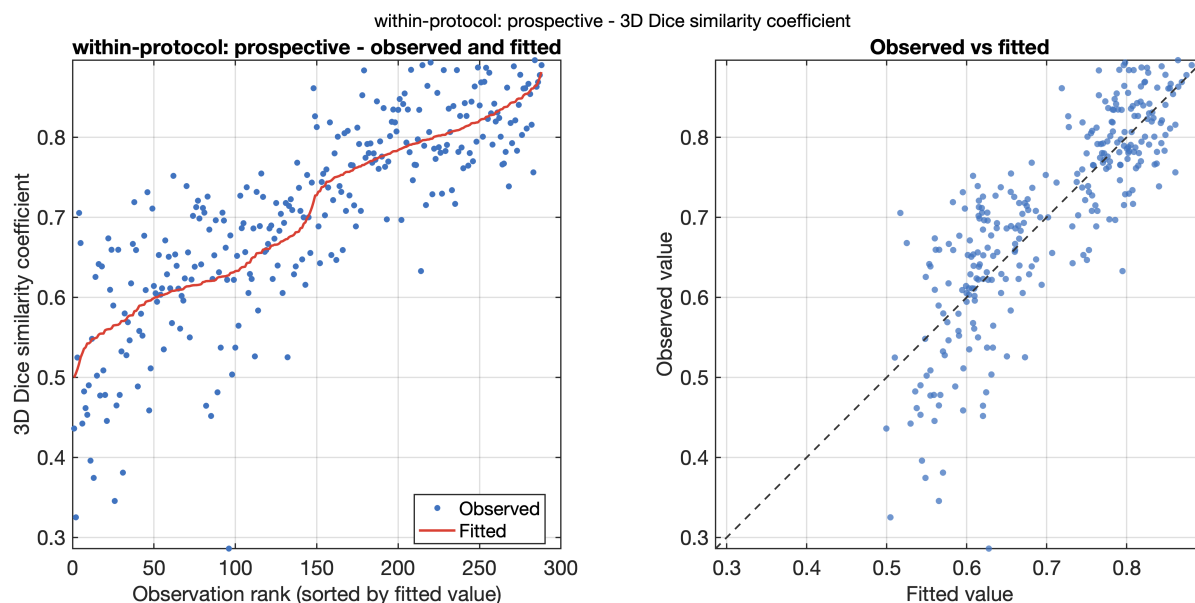


Figure F.7: Observed-versus-fitted assessment for the Prospective within-protocol DSC model. The fitted values captured the broad variation between settings, with remaining scatter consistent with the residual diagnostics.

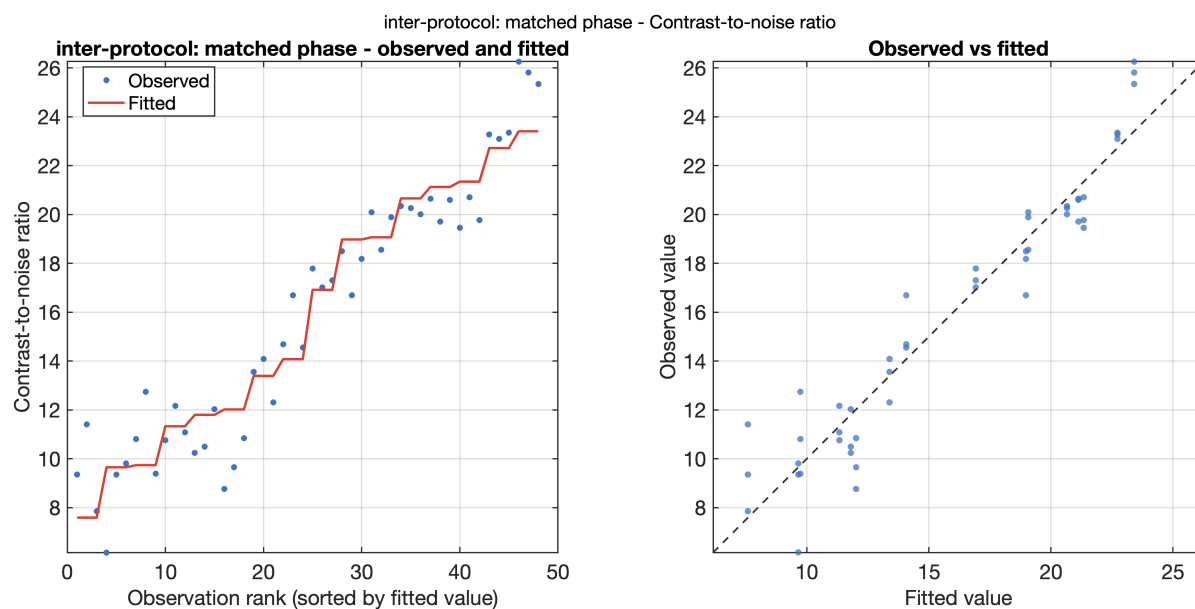


Figure F.8: Observed-versus-fitted assessment for the matched-phase inter-protocol CNR model at 53 keV. The model reproduced the main CNR differences between fitted conditions, with an observed-versus-fitted R^2 of 0.92.

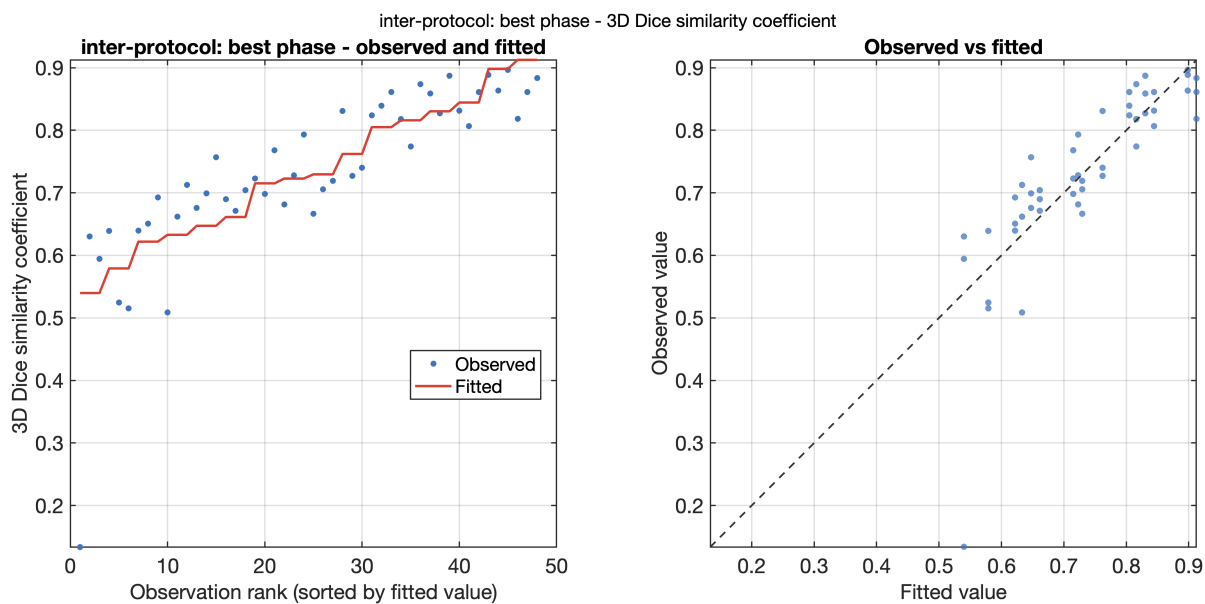


Figure F.9: Observed-versus-fitted assessment for the best-phase inter-protocol DSC model at 53 keV. The model captured the general trend in segmentation overlap while retaining visible observation-level scatter.

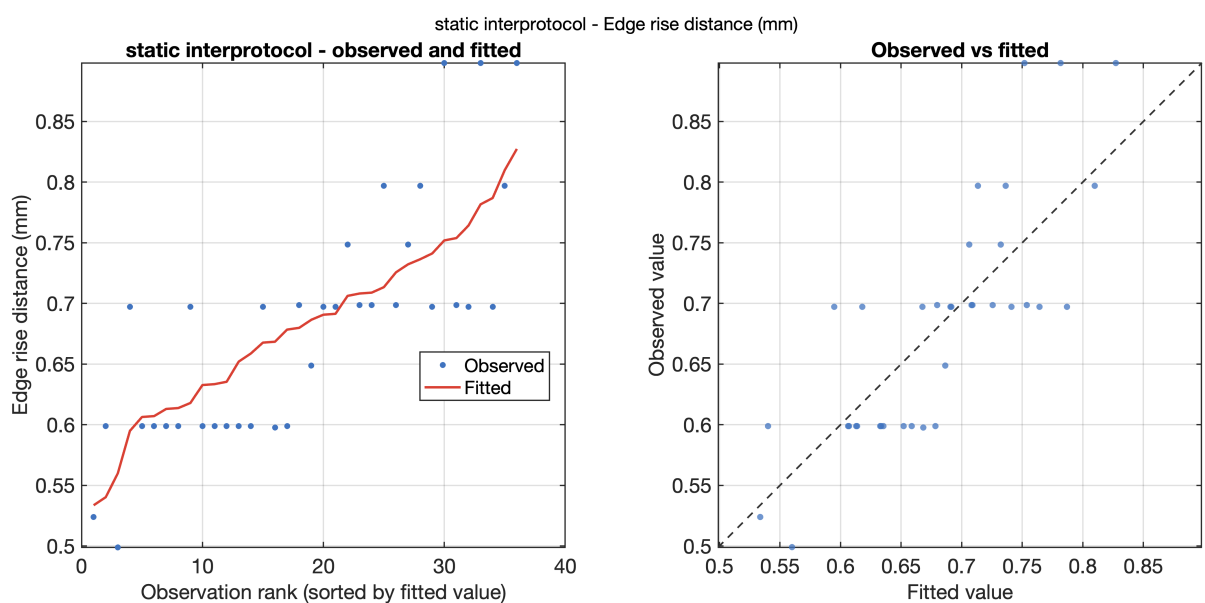


Figure F.10: Observed-versus-fitted assessment for the static inter-protocol ERD model at 0 bpm. The plot supports the use of the static model as a baseline comparability check, with an observed-versus-fitted R^2 of 0.64.

F.3. Exploratory assessment of quadratic terms

As a sensitivity analysis, exploratory mixed-effects models with quadratic terms were fitted for the ordered acquisition settings. For the flash within-protocol models, centered reconstruction energy and centered heart rate were included with squared terms. For the prospective within-protocol models, centered heart rate and centered phase were included with squared terms. For the inter-protocol models at 53 keV, centered heart rate was included with a squared term. The static 0 bpm comparison was not included in this sensitivity analysis, because it did not contain a motion or phase range over which curvature could be assessed.

The purpose of this analysis was not to interpret the quadratic coefficients as primary effects, but to check whether the main-effects mixed-effects models were sufficient for the thesis analyses. Therefore, the comparison focused on model performance: root-mean-square error (RMSE) and observed-versus-fitted R^2 . If the quadratic terms captured important curvature, the quadratic models would be expected to show lower RMSE and higher observed-versus-fitted R^2 than the main-effects models.

Tables F.1–F.4 show the complete linear-versus-quadratic comparison for all fitted outcomes in the flash within-protocol, prospective within-protocol, matched-phase inter-protocol, and best-phase inter-protocol analyses. Across these model families, the quadratic models did not provide a consistent improvement. RMSE was generally similar or slightly higher after adding the quadratic terms, and observed-versus-fitted R^2 was generally similar or lower. The linear mixed-effects models were therefore retained for the primary analysis because they provided a more parsimonious and interpretable description of the data without a meaningful loss of fit.

Table F.1: Linear-versus-quadratic model performance for the flash within-protocol mixed-effects models.

Metric	Linear RMSE	Quadratic RMSE	Linear R^2	Quadratic R^2	ΔR^2
Equivalent diameter error (mm)	0.033	0.033	0.548	0.546	-0.002
Area error (mm ²)	0.083	0.083	0.510	0.508	-0.002
Edge rise distance (mm)	0.072	0.073	0.457	0.441	-0.016
3D Dice similarity coefficient	0.078	0.103	0.714	0.506	-0.208
Circularity	0.024	0.025	0.507	0.452	-0.054
Attenuation bias (HU)	46.916	49.773	0.536	0.478	-0.058
Contrast-to-noise ratio	2.054	2.454	0.897	0.853	-0.044

Table F.2: Linear-versus-quadratic model performance for the prospective within-protocol mixed-effects models.

Metric	Linear RMSE	Quadratic RMSE	Linear R^2	Quadratic R^2	ΔR^2
Equivalent diameter error (mm)	0.022	0.022	0.528	0.509	-0.018
Area error (mm ²)	0.045	0.046	0.424	0.394	-0.029
Edge rise distance (mm)	0.053	0.054	0.317	0.289	-0.029
3D Dice similarity coefficient	0.069	0.074	0.684	0.637	-0.047
Circularity	0.008	0.008	0.803	0.792	-0.011
Attenuation bias (HU)	17.362	18.725	0.632	0.572	-0.060
Contrast-to-noise ratio	1.386	1.539	0.914	0.894	-0.020

Table F.3: Linear-versus-quadratic model performance for the matched-phase inter-protocol mixed-effects models at 53 keV.

Metric	Linear RMSE	Quadratic RMSE	Linear R^2	Quadratic R^2	ΔR^2
Equivalent diameter error (mm)	0.022	0.022	0.540	0.536	-0.004
Area error (mm ²)	0.045	0.045	0.315	0.314	-0.001
Edge rise distance (mm)	0.054	0.054	0.542	0.538	-0.003
3D Dice similarity coefficient	0.079	0.083	0.649	0.609	-0.039
Circularity	0.011	0.011	0.720	0.716	-0.005
Attenuation bias (HU)	26.743	26.911	0.358	0.350	-0.008
Contrast-to-noise ratio	1.511	1.708	0.919	0.896	-0.023

Table F.4: Linear-versus-quadratic model performance for the best-phase inter-protocol mixed-effects models at 53 keV.

Metric	Linear RMSE	Quadratic RMSE	Linear R^2	Quadratic R^2	ΔR^2
Equivalent diameter error (mm)	0.023	0.023	0.408	0.404	-0.004
Area error (mm ²)	0.049	0.049	0.091	0.091	-0.001
Edge rise distance (mm)	0.061	0.061	0.423	0.423	0.000
3D Dice similarity coefficient	0.077	0.086	0.668	0.584	-0.085
Circularity	0.011	0.011	0.704	0.699	-0.005
Attenuation bias (HU)	27.297	27.757	0.300	0.278	-0.023
Contrast-to-noise ratio	1.853	1.854	0.877	0.877	0.000

G

Results

G.1. Visualization of the scans

This section provides representative PCCT images of the iodine-filled tube configurations used in the quantitative analysis. Figures G.1 and G.2 are included to illustrate the difference in appearance of the 1.1 mm and 2.0 mm tubes in the different orientations and reconstruction energies.

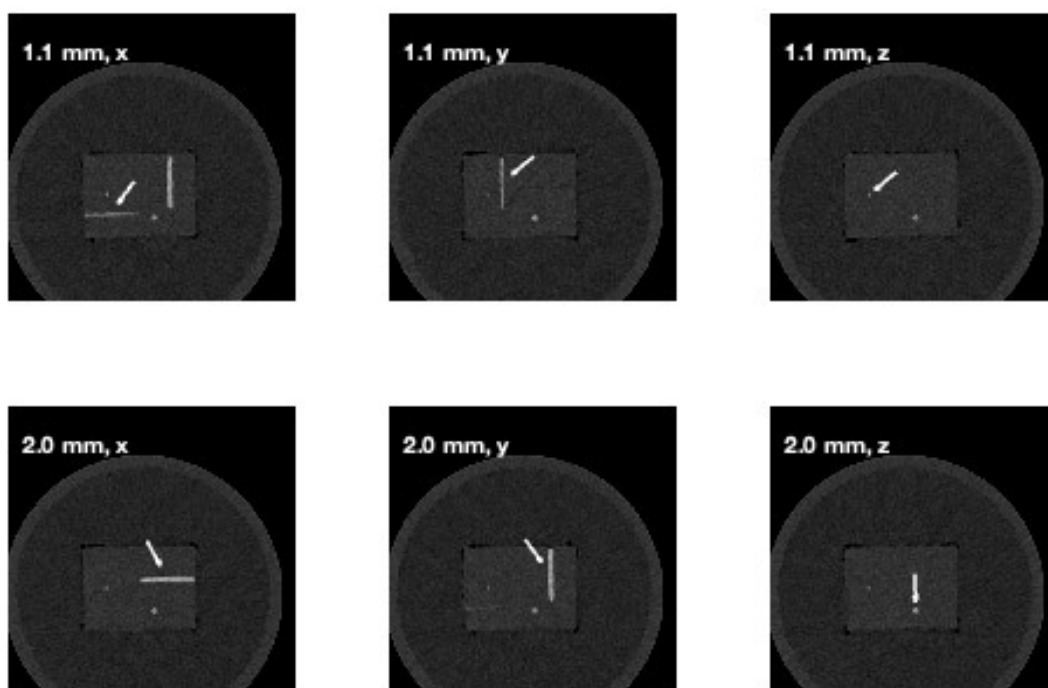


Figure G.1: Representative axial PCCT images of the six iodine-filled tube configurations reconstructed at 47 keV. White arrows indicate the tube assessed in each panel.

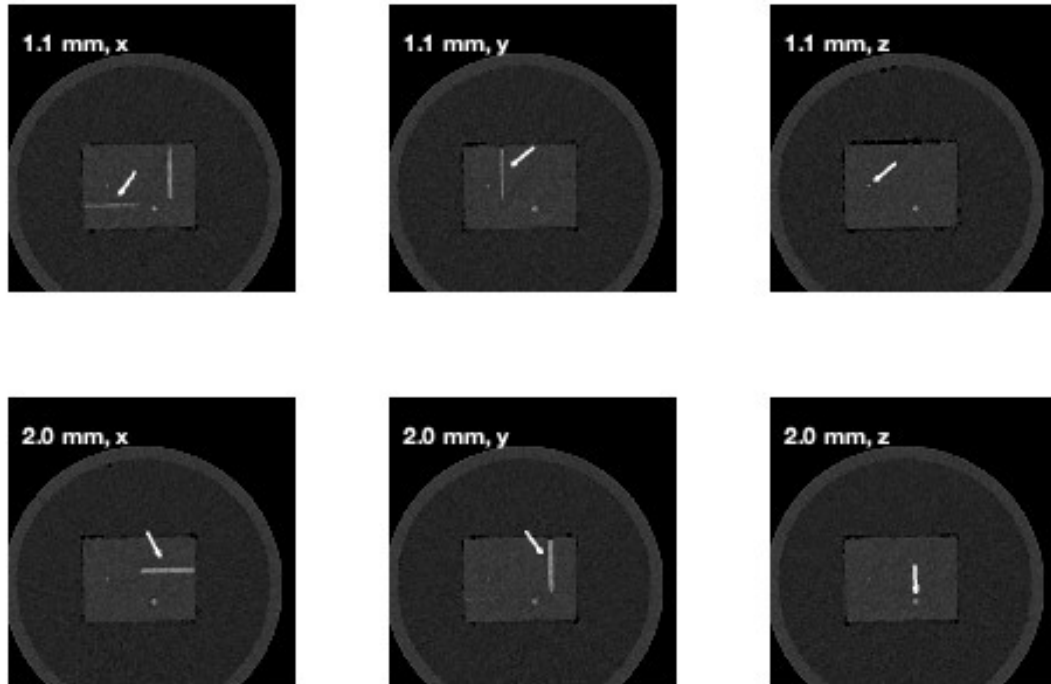


Figure G.2: Representative axial PCCT images of the six iodine-filled tube configurations reconstructed at 53 keV. White arrows indicate the tube assessed in each panel.

G.2. Repeatability and quality control

This section provides additional repeatability results supporting the repeatability analysis described in the main text. The figures and tables include repeated-measurement plots, phase-resolved prospective repeatability summaries, coefficient of variation values, and repeatability coefficients.

G.2.1. Prospective repeated measurements

Individual repeated measurements for the 50% and 75% reconstructions are shown separately in Figures G.3 and G.4. These phases represent a mid-cycle phase and a later-phase reconstruction, respectively.

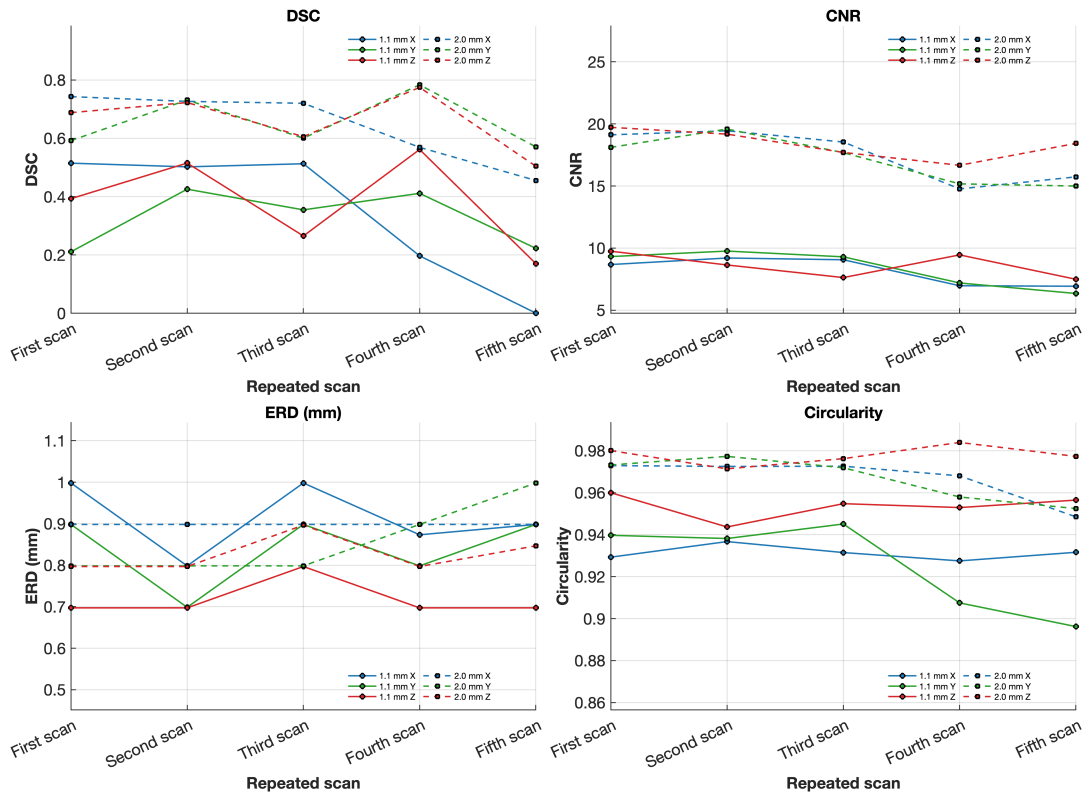


Figure G.3: Repeatability of the prospective protocol across five repeated scans at the 50% reconstructed phase. The figure shows DSC, CNR, ERD, and circularity for the x-, y-, and z-oriented 1.1 mm and 2.0 mm tubes.

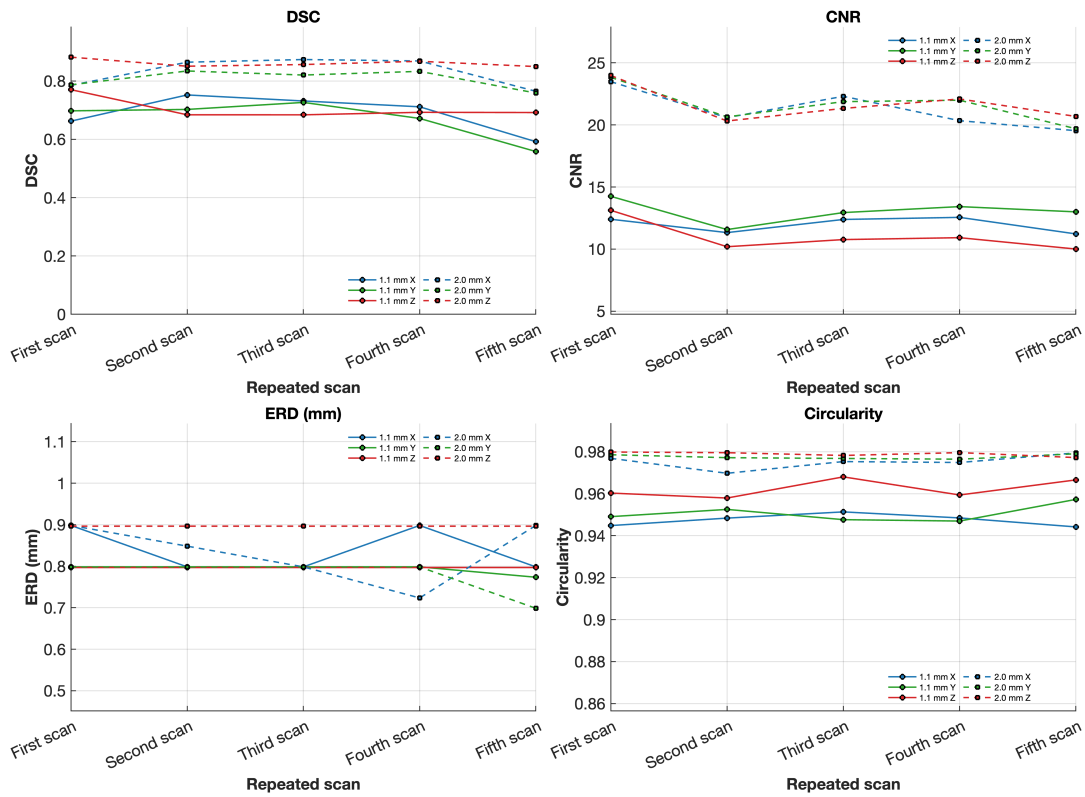


Figure G.4: Repeatability of the prospective protocol across five repeated scans at the 75% reconstructed phase. The figure shows DSC, CNR, ERD, and circularity for the x-, y-, and z-oriented 1.1 mm and 2.0 mm tubes.

G.2.2. Additional repeatability plots

Additional repeatability plots for equivalent diameter error, cross-sectional area error, and attenuation bias are provided in Figures G.5 and G.6. These plots complement the main repeatability figures, which focused on DSC, CNR, ERD, and circularity.

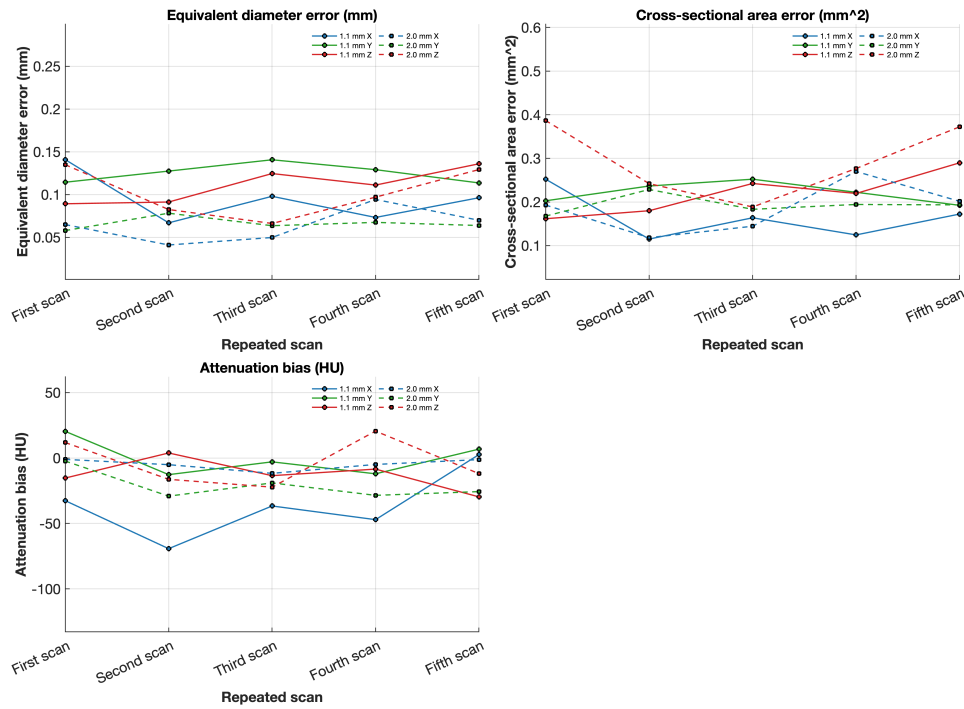


Figure G.5: Additional repeatability plots for the flash protocol across five repeated scans at 120 bpm and 53 keV. The figure shows equivalent diameter error, cross-sectional area error, and attenuation bias for the x-, y-, and z-oriented 1.1 mm and 2.0 mm tubes.

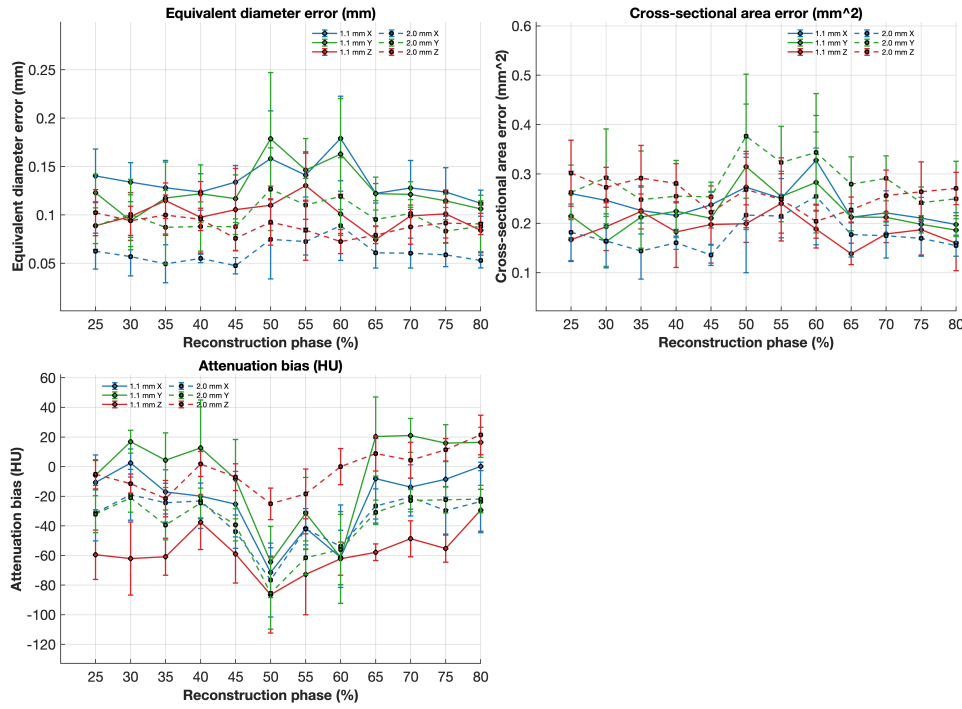


Figure G.6: Additional phase-resolved repeatability plots for the prospective protocol at 120 bpm and 53 keV. The figure shows equivalent diameter error, cross-sectional area error, and attenuation bias across reconstructed phases for the x-, y-, and z-oriented 1.1 mm and 2.0 mm tubes. Error bars indicate the standard deviation across the repeated measurements.

G.2.3. Prospective repeatability

The full prospective repeatability results for all reconstructed phases are provided in Table G.1. This table reports the mean and standard deviation of each endpoint for all reconstructed phases and tube orientations.

Table G.1: Mean (μ) and standard deviation (SD, σ) of diameter error, area error, circularity, edge rise distance (ERD), attenuation bias, Dice similarity coefficient (DSC), and contrast-to-noise ratio (CNR) for the six tubes across five repeated scans using the prospective protocol at different reconstruction phases.

Prospective repeatability							
Tube	E_{Deq}	E_A	C	ERD	B_{HU}	DSC	CNR
	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$
Prospective 25%							
1 mm, x	0.140 $\pm$ 0.028	0.260 $\pm$ 0.045	0.942 $\pm$ 0.008	0.818 $\pm$ 0.109	-10.6 $\pm$ 18.5	0.61 $\pm$ 0.12	10.90 $\pm$ 1.24
1 mm, y	0.123 $\pm$ 0.019	0.215 $\pm$ 0.044	0.942 $\pm$ 0.013	0.778 $\pm$ 0.045	-5.8 $\pm$ 9.8	0.63 $\pm$ 0.07	11.03 $\pm$ 1.49
1 mm, z	0.089 $\pm$ 0.025	0.167 $\pm$ 0.044	0.957 $\pm$ 0.003	0.767 $\pm$ 0.067	-59.5 $\pm$ 16.7	0.62 $\pm$ 0.05	9.86 $\pm$ 0.97
2 mm, x	0.063 $\pm$ 0.019	0.182 $\pm$ 0.057	0.973 $\pm$ 0.002	0.878 $\pm$ 0.045	-31.3 $\pm$ 18.8	0.84 $\pm$ 0.04	19.37 $\pm$ 1.85
2 mm, y	0.089 $\pm$ 0.019	0.263 $\pm$ 0.055	0.975 $\pm$ 0.002	0.759 $\pm$ 0.055	-32.2 $\pm$ 12.5	0.80 $\pm$ 0.03	19.33 $\pm$ 1.76
2 mm, z	0.102 $\pm$ 0.024	0.301 $\pm$ 0.067	0.976 $\pm$ 0.003	0.877 $\pm$ 0.045	-5.2 $\pm$ 9.8	0.81 $\pm$ 0.02	19.14 $\pm$ 2.02
Prospective 30%							
1 mm, x	0.134 $\pm$ 0.020	0.246 $\pm$ 0.048	0.947 $\pm$ 0.008	0.818 $\pm$ 0.045	2.9 $\pm$ 9.4	0.66 $\pm$ 0.07	11.68 $\pm$ 1.67
1 mm, y	0.094 $\pm$ 0.027	0.163 $\pm$ 0.051	0.954 $\pm$ 0.009	0.739 $\pm$ 0.055	15.3 $\pm$ 10.6	0.64 $\pm$ 0.04	12.08 $\pm$ 1.47
1 mm, z	0.098 $\pm$ 0.025	0.193 $\pm$ 0.048	0.952 $\pm$ 0.003	0.777 $\pm$ 0.130	-59.3 $\pm$ 24.8	0.71 $\pm$ 0.02	10.06 $\pm$ 1.84
2 mm, x	0.057 $\pm$ 0.020	0.164 $\pm$ 0.055	0.975 $\pm$ 0.003	0.838 $\pm$ 0.089	-21.7 $\pm$ 13.5	0.82 $\pm$ 0.04	20.28 $\pm$ 3.37
2 mm, y	0.100 $\pm$ 0.036	0.292 $\pm$ 0.099	0.977 $\pm$ 0.002	0.788 $\pm$ 0.055	-22.7 $\pm$ 9.2	0.81 $\pm$ 0.04	20.20 $\pm$ 3.05
2 mm, z	0.094 $\pm$ 0.015	0.273 $\pm$ 0.041	0.972 $\pm$ 0.002	0.797 $\pm$ 0.070	-10.9 $\pm$ 6.3	0.86 $\pm$ 0.02	19.32 $\pm$ 3.05
Prospective 35%							
1 mm, x	0.128 $\pm$ 0.028	0.226 $\pm$ 0.047	0.942 $\pm$ 0.006	0.818 $\pm$ 0.045	-17.0 $\pm$ 14.9	0.66 $\pm$ 0.05	11.43 $\pm$ 1.65
1 mm, y	0.118 $\pm$ 0.037	0.213 $\pm$ 0.064	0.943 $\pm$ 0.008	0.759 $\pm$ 0.089	4.3 $\pm$ 18.5	0.60 $\pm$ 0.09	12.26 $\pm$ 1.19
1 mm, z	0.115 $\pm$ 0.018	0.224 $\pm$ 0.035	0.949 $\pm$ 0.002	0.797 $\pm$ 0.000	-60.9 $\pm$ 12.5	0.67 $\pm$ 0.09	10.52 $\pm$ 1.33
2 mm, x	0.049 $\pm$ 0.020	0.144 $\pm$ 0.057	0.976 $\pm$ 0.003	0.838 $\pm$ 0.089	-24.4 $\pm$ 13.1	0.83 $\pm$ 0.03	21.34 $\pm$ 2.75

Continued on next page

Table G.1: Mean and standard deviation (SD) of diameter error, area error, circularity, edge rise distance (ERD), attenuation bias, Dice similarity coefficient (DSC), and contrast-to-noise ratio (CNR) for the six tubes across five repeated scans using the prospective protocol at different reconstruction phases (continued).

Prospective repeatability (continued)							
	E_{Deq}	E_A	C	ERD	B_{HU}	DSC	CNR
Tube	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$
2 mm, y	0.087 $\pm$ 0.038	0.248 $\pm$ 0.099	0.974 $\pm$ 0.003	0.838 $\pm$ 0.055	-39.4 $\pm$ 10.2	0.79 $\pm$ 0.08	20.62 $\pm$ 2.60
2 mm, z	0.100 $\pm$ 0.021	0.292 $\pm$ 0.066	0.976 $\pm$ 0.003	0.887 $\pm$ 0.114	-21.6 $\pm$ 12.3	0.84 $\pm$ 0.03	19.93 $\pm$ 2.39
Prospective 40%							
1 mm, x	0.124 $\pm$ 0.019	0.216 $\pm$ 0.028	0.933 $\pm$ 0.024	0.848 $\pm$ 0.071	-19.9 $\pm$ 21.8	0.64 $\pm$ 0.13	10.76 $\pm$ 1.77
1 mm, y	0.122 $\pm$ 0.030	0.224 $\pm$ 0.053	0.944 $\pm$ 0.013	0.778 $\pm$ 0.045	12.7 $\pm$ 32.4	0.64 $\pm$ 0.06	12.16 $\pm$ 2.53
1 mm, z	0.098 $\pm$ 0.037	0.183 $\pm$ 0.072	0.952 $\pm$ 0.016	0.777 $\pm$ 0.045	-37.6 $\pm$ 18.4	0.68 $\pm$ 0.10	11.08 $\pm$ 1.31
2 mm, x	0.055 $\pm$ 0.004	0.161 $\pm$ 0.013	0.973 $\pm$ 0.001	0.778 $\pm$ 0.045	-23.2 $\pm$ 12.1	0.82 $\pm$ 0.07	20.34 $\pm$ 2.16
2 mm, y	0.088 $\pm$ 0.025	0.255 $\pm$ 0.072	0.976 $\pm$ 0.002	0.739 $\pm$ 0.055	-24.5 $\pm$ 10.0	0.81 $\pm$ 0.04	20.26 $\pm$ 1.80
2 mm, z	0.095 $\pm$ 0.012	0.281 $\pm$ 0.040	0.977 $\pm$ 0.003	0.787 $\pm$ 0.055	1.7 $\pm$ 8.4	0.88 $\pm$ 0.01	20.01 $\pm$ 1.77
Prospective 45%							
1 mm, x	0.134 $\pm$ 0.017	0.237 $\pm$ 0.026	0.933 $\pm$ 0.014	0.828 $\pm$ 0.067	-25.4 $\pm$ 22.1	0.69 $\pm$ 0.06	10.15 $\pm$ 1.56
1 mm, y	0.117 $\pm$ 0.025	0.210 $\pm$ 0.054	0.943 $\pm$ 0.018	0.759 $\pm$ 0.055	-8.4 $\pm$ 26.7	0.62 $\pm$ 0.06	10.81 $\pm$ 2.02
1 mm, z	0.105 $\pm$ 0.043	0.197 $\pm$ 0.078	0.952 $\pm$ 0.019	0.742 $\pm$ 0.067	-58.9 $\pm$ 19.7	0.61 $\pm$ 0.13	9.81 $\pm$ 1.52
2 mm, x	0.047 $\pm$ 0.008	0.136 $\pm$ 0.021	0.973 $\pm$ 0.004	0.808 $\pm$ 0.055	-43.9 $\pm$ 11.3	0.84 $\pm$ 0.03	18.99 $\pm$ 3.50
2 mm, y	0.088 $\pm$ 0.008	0.254 $\pm$ 0.030	0.975 $\pm$ 0.004	0.759 $\pm$ 0.055	-39.3 $\pm$ 10.9	0.80 $\pm$ 0.06	19.15 $\pm$ 3.07
2 mm, z	0.076 $\pm$ 0.013	0.222 $\pm$ 0.032	0.977 $\pm$ 0.002	0.767 $\pm$ 0.083	-7.1 $\pm$ 9.0	0.82 $\pm$ 0.06	19.13 $\pm$ 2.95
Prospective 50%							
1 mm, x	0.158 $\pm$ 0.049	0.273 $\pm$ 0.068	0.931 $\pm$ 0.003	0.913 $\pm$ 0.086	-71.5 $\pm$ 16.7	0.35 $\pm$ 0.24	8.17 $\pm$ 1.13
1 mm, y	0.179 $\pm$ 0.068	0.315 $\pm$ 0.127	0.925 $\pm$ 0.022	0.838 $\pm$ 0.089	-64.4 $\pm$ 24.0	0.33 $\pm$ 0.10	8.38 $\pm$ 1.51
1 mm, z	0.110 $\pm$ 0.020	0.199 $\pm$ 0.038	0.954 $\pm$ 0.006	0.717 $\pm$ 0.045	-86.4 $\pm$ 26.0	0.38 $\pm$ 0.16	8.59 $\pm$ 1.03
2 mm, x	0.075 $\pm$ 0.041	0.217 $\pm$ 0.117	0.967 $\pm$ 0.010	0.898 $\pm$ 0.000	-76.6 $\pm$ 25.0	0.64 $\pm$ 0.13	17.52 $\pm$ 2.13
2 mm, y	0.127 $\pm$ 0.043	0.377 $\pm$ 0.126	0.967 $\pm$ 0.011	0.858 $\pm$ 0.089	-85.5 $\pm$ 24.2	0.66 $\pm$ 0.10	17.11 $\pm$ 1.98
2 mm, z	0.093 $\pm$ 0.024	0.268 $\pm$ 0.077	0.978 $\pm$ 0.005	0.827 $\pm$ 0.045	-25.1 $\pm$ 10.7	0.66 $\pm$ 0.11	18.34 $\pm$ 1.21
Prospective 55%							
1 mm, x	0.141 $\pm$ 0.023	0.250 $\pm$ 0.040	0.934 $\pm$ 0.006	0.848 $\pm$ 0.071	-38.8 $\pm$ 5.8	0.55 $\pm$ 0.13	9.66 $\pm$ 0.97
1 mm, y	0.146 $\pm$ 0.033	0.254 $\pm$ 0.051	0.939 $\pm$ 0.009	0.798 $\pm$ 0.000	-31.5 $\pm$ 24.1	0.44 $\pm$ 0.24	9.93 $\pm$ 1.56
1 mm, z	0.130 $\pm$ 0.035	0.240 $\pm$ 0.060	0.949 $\pm$ 0.008	0.817 $\pm$ 0.045	-77.3 $\pm$ 30.9	0.46 $\pm$ 0.24	9.16 $\pm$ 2.03
2 mm, x	0.072 $\pm$ 0.014	0.215 $\pm$ 0.044	0.974 $\pm$ 0.004	0.878 $\pm$ 0.027	-43.2 $\pm$ 13.9	0.77 $\pm$ 0.08	19.02 $\pm$ 2.48
2 mm, y	0.110 $\pm$ 0.028	0.323 $\pm$ 0.073	0.974 $\pm$ 0.002	0.828 $\pm$ 0.084	-61.3 $\pm$ 11.2	0.69 $\pm$ 0.18	18.15 $\pm$ 1.38
2 mm, z	0.084 $\pm$ 0.031	0.248 $\pm$ 0.084	0.975 $\pm$ 0.003	0.936 $\pm$ 0.089	-18.5 $\pm$ 16.9	0.70 $\pm$ 0.15	18.67 $\pm$ 2.20
Prospective 60%							
1 mm, x	0.179 $\pm$ 0.044	0.328 $\pm$ 0.091	0.923 $\pm$ 0.013	0.878 $\pm$ 0.109	-61.3 $\pm$ 18.4	0.60 $\pm$ 0.10	9.24 $\pm$ 1.15
1 mm, y	0.163 $\pm$ 0.057	0.283 $\pm$ 0.102	0.916 $\pm$ 0.032	0.719 $\pm$ 0.045	-61.3 $\pm$ 31.0	0.45 $\pm$ 0.23	9.16 $\pm$ 1.98
1 mm, z	0.101 $\pm$ 0.020	0.188 $\pm$ 0.038	0.960 $\pm$ 0.005	0.717 $\pm$ 0.045	-62.2 $\pm$ 11.2	0.57 $\pm$ 0.18	10.47 $\pm$ 1.73
2 mm, x	0.089 $\pm$ 0.036	0.255 $\pm$ 0.098	0.967 $\pm$ 0.008	0.938 $\pm$ 0.055	-53.7 $\pm$ 27.9	0.79 $\pm$ 0.02	19.66 $\pm$ 2.38
2 mm, y	0.119 $\pm$ 0.040	0.344 $\pm$ 0.119	0.968 $\pm$ 0.008	0.848 $\pm$ 0.050	-56.0 $\pm$ 23.8	0.76 $\pm$ 0.11	19.57 $\pm$ 2.47
2 mm, z	0.072 $\pm$ 0.012	0.204 $\pm$ 0.034	0.979 $\pm$ 0.002	0.817 $\pm$ 0.083	0.0 $\pm$ 12.2	0.78 $\pm$ 0.09	20.81 $\pm$ 3.31
Prospective 65%							
1 mm, x	0.122 $\pm$ 0.010	0.212 $\pm$ 0.010	0.945 $\pm$ 0.007	0.818 $\pm$ 0.109	-8.2 $\pm$ 27.2	0.63 $\pm$ 0.08	11.50 $\pm$ 1.86
1 mm, y	0.122 $\pm$ 0.017	0.212 $\pm$ 0.027	0.952 $\pm$ 0.009	0.759 $\pm$ 0.055	20.4 $\pm$ 26.7	0.59 $\pm$ 0.11	12.69 $\pm$ 1.82
1 mm, z	0.075 $\pm$ 0.014	0.138 $\pm$ 0.022	0.960 $\pm$ 0.006	0.777 $\pm$ 0.045	-57.8 $\pm$ 5.6	0.68 $\pm$ 0.11	10.35 $\pm$ 0.78
2 mm, x	0.061 $\pm$ 0.016	0.177 $\pm$ 0.045	0.975 $\pm$ 0.004	0.838 $\pm$ 0.055	-26.6 $\pm$ 11.5	0.80 $\pm$ 0.07	20.44 $\pm$ 1.09
2 mm, y	0.095 $\pm$ 0.019	0.279 $\pm$ 0.055	0.978 $\pm$ 0.004	0.798 $\pm$ 0.071	-30.8 $\pm$ 8.2	0.77 $\pm$ 0.07	20.27 $\pm$ 1.45
2 mm, z	0.079 $\pm$ 0.009	0.227 $\pm$ 0.026	0.980 $\pm$ 0.002	0.837 $\pm$ 0.055	8.8 $\pm$ 11.7	0.84 $\pm$ 0.07	20.60 $\pm$ 1.18
Prospective 70%							
1 mm, x	0.128 $\pm$ 0.028	0.221 $\pm$ 0.045	0.945 $\pm$ 0.009	0.858 $\pm$ 0.089	-10.3 $\pm$ 15.4	0.64 $\pm$ 0.09	11.15 $\pm$ 0.71
1 mm, y	0.121 $\pm$ 0.013	0.212 $\pm$ 0.010	0.953 $\pm$ 0.006	0.739 $\pm$ 0.055	23.3 $\pm$ 7.9	0.65 $\pm$ 0.06	12.58 $\pm$ 0.86
1 mm, z	0.099 $\pm$ 0.011	0.178 $\pm$ 0.018	0.960 $\pm$ 0.007	0.757 $\pm$ 0.055	-45.0 $\pm$ 10.3	0.71 $\pm$ 0.02	10.74 $\pm$ 0.65
2 mm, x	0.060 $\pm$ 0.016	0.175 $\pm$ 0.045	0.975 $\pm$ 0.003	0.848 $\pm$ 0.050	-23.0 $\pm$ 11.2	0.80 $\pm$ 0.06	20.38 $\pm$ 0.95
2 mm, y	0.102 $\pm$ 0.014	0.291 $\pm$ 0.046	0.976 $\pm$ 0.003	0.778 $\pm$ 0.045	-21.0 $\pm$ 6.3	0.80 $\pm$ 0.04	20.47 $\pm$ 0.95
2 mm, z	0.088 $\pm$ 0.018	0.256 $\pm$ 0.052	0.978 $\pm$ 0.002	0.837 $\pm$ 0.055	12.2 $\pm$ 12.8	0.87 $\pm$ 0.02	20.51 $\pm$ 0.49
Prospective 75%							
1 mm, x	0.124 $\pm$ 0.025	0.210 $\pm$ 0.042	0.947 $\pm$ 0.003	0.838 $\pm$ 0.055	-8.5 $\pm$ 12.1	0.69 $\pm$ 0.06	11.98 $\pm$ 0.65
1 mm, y	0.115 $\pm$ 0.007	0.198 $\pm$ 0.017	0.951 $\pm$ 0.004	0.793 $\pm$ 0.011	15.9 $\pm$ 12.5	0.67 $\pm$ 0.07	13.04 $\pm$ 0.97

Continued on next page

Table G.1: Mean and standard deviation (SD) of diameter error, area error, circularity, edge rise distance (ERD), attenuation bias, Dice similarity coefficient (DSC), and contrast-to-noise ratio (CNR) for the six tubes across five repeated scans using the prospective protocol at different reconstruction phases (continued).

Prospective repeatability (continued)							
	E_{Deq}	E_A	C	ERD	B_{HU}	DSC	CNR
Tube	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$	$\mu \pm \sigma$
1 mm, z	0.101 $\pm$ 0.025	0.187 $\pm$ 0.051	0.962 $\pm$ 0.005	0.797 $\pm$ 0.000	-55.4 $\pm$ 9.1	0.70 $\pm$ 0.04	11.00 $\pm$ 1.25
2 mm, x	0.059 $\pm$ 0.012	0.169 $\pm$ 0.036	0.975 $\pm$ 0.004	0.833 $\pm$ 0.074	-29.7 $\pm$ 15.8	0.83 $\pm$ 0.05	21.25 $\pm$ 1.61
2 mm, y	0.083 $\pm$ 0.012	0.242 $\pm$ 0.032	0.978 $\pm$ 0.001	0.778 $\pm$ 0.045	-22.3 $\pm$ 9.0	0.81 $\pm$ 0.03	21.61 $\pm$ 1.56
2 mm, z	0.092 $\pm$ 0.021	0.264 $\pm$ 0.060	0.979 $\pm$ 0.001	0.896 $\pm$ 0.000	11.4 $\pm$ 7.6	0.86 $\pm$ 0.01	21.68 $\pm$ 1.45
Prospective 80%							
1 mm, x	0.112 $\pm$ 0.013	0.197 $\pm$ 0.024	0.949 $\pm$ 0.009	0.798 $\pm$ 0.071	0.2 $\pm$ 2.8	0.73 $\pm$ 0.02	11.31 $\pm$ 0.53
1 mm, y	0.106 $\pm$ 0.014	0.186 $\pm$ 0.023	0.949 $\pm$ 0.010	0.793 $\pm$ 0.067	16.4 $\pm$ 10.2	0.61 $\pm$ 0.13	11.90 $\pm$ 0.29
1 mm, z	0.084 $\pm$ 0.026	0.160 $\pm$ 0.056	0.964 $\pm$ 0.003	0.727 $\pm$ 0.045	-29.5 $\pm$ 14.2	0.66 $\pm$ 0.12	11.14 $\pm$ 1.14
2 mm, x	0.053 $\pm$ 0.008	0.155 $\pm$ 0.021	0.974 $\pm$ 0.004	0.858 $\pm$ 0.055	-23.6 $\pm$ 20.9	0.87 $\pm$ 0.01	19.71 $\pm$ 1.05
2 mm, y	0.088 $\pm$ 0.026	0.249 $\pm$ 0.076	0.977 $\pm$ 0.003	0.788 $\pm$ 0.022	-21.9 $\pm$ 9.3	0.79 $\pm$ 0.07	19.79 $\pm$ 0.87
2 mm, z	0.091 $\pm$ 0.011	0.271 $\pm$ 0.033	0.978 $\pm$ 0.002	0.857 $\pm$ 0.055	21.4 $\pm$ 13.3	0.82 $\pm$ 0.07	20.29 $\pm$ 0.86

G.2.4. Prospective repeatability: coefficient of variation

Normalized errors and coefficient of variation values for the prospective repeatability measurements are provided in Table G.2. These values complement the mean and standard deviation results by expressing variation relative to the measured value or reference geometry where applicable.

Table G.2: Normalized equivalent diameter error, cross-sectional area error, the coefficient of variation (CV) of circularity, edge rise distance (ERD), DSC, and contrast-to-noise ratio (CNR), and normalized attenuation bias for the six tubes across five repeated scans using the prospective protocol at different reconstruction phases.

Prospective repeatability: normalized errors and coefficient of variation							
	$E_{Deq,rel}$	$E_{A,rel}$	C	ERD	$B_{HU,rel}$	DSC	CNR
Tube	Norm. error (%)	Norm. error (%)	CV (%)	CV (%)	Norm. bias (%)	CV (%)	CV (%)
Prospective 25%							
1 mm, x	12.28	25.37	0.83	13.36	-3.04	19.40	11.37
1 mm, y	11.33	23.27	1.35	5.73	-1.68	11.65	13.49
1 mm, z	7.61	15.52	0.33	8.71	-15.85	8.57	9.88
2 mm, x	3.42	6.90	0.25	5.08	-5.54	4.27	9.53
2 mm, y	4.75	9.57	0.24	7.21	-5.69	4.18	9.09
2 mm, z	5.48	10.99	0.34	5.08	-0.97	2.55	10.54
Prospective 30%							
1 mm, x	11.71	24.02	0.85	5.45	0.84	9.94	14.33
1 mm, y	8.67	17.68	0.96	7.40	4.40	6.45	12.17
1 mm, z	8.41	17.98	0.34	16.72	-15.79	2.17	18.32
2 mm, x	3.11	6.24	0.28	10.65	-3.84	5.18	16.63
2 mm, y	5.37	10.65	0.21	6.93	-4.02	4.69	15.09
2 mm, z	5.03	9.95	0.17	8.84	-2.04	2.21	15.76
Prospective 35%							
1 mm, x	11.21	22.03	0.63	5.45	-4.86	7.80	14.45
1 mm, y	10.84	23.03	0.81	11.77	1.24	14.98	9.68
1 mm, z	9.81	20.85	0.20	0.00	-16.21	13.80	12.68
2 mm, x	2.70	5.46	0.26	10.65	-4.31	3.47	12.89
2 mm, y	4.67	9.04	0.36	6.52	-6.97	10.64	12.59
2 mm, z	5.34	10.63	0.29	12.81	-4.05	3.40	11.99
Prospective 40%							
1 mm, x	10.83	21.05	2.58	8.32	-5.67	21.13	16.43
1 mm, y	11.24	24.31	1.41	5.73	3.65	9.01	20.81
1 mm, z	8.35	17.00	1.67	5.73	-10.01	14.98	11.81
2 mm, x	3.00	6.10	0.15	5.73	-4.10	8.48	10.61
2 mm, y	4.71	9.30	0.25	7.40	-4.33	4.80	8.90
2 mm, z	5.11	10.24	0.32	6.93	0.33	1.04	8.86

Continued on next page

Table G.2: Normalized equivalent diameter error, cross-sectional area error, the CV of circularity, ERD, DSC, and CNR, and normalized attenuation bias for the prospective protocol (continued).

Prospective repeatability: normalized errors and coefficient of variation (continued)							
Tube	$E_{D_{eq},rel}$	$E_{A,rel}$	C	ERD	$B_{HU,rel}$	DSC	CNR
	Norm. error (%)	Norm. error (%)	CV (%)	CV (%)	Norm. bias (%)	CV (%)	CV (%)
Prospective 45%							
1 mm, x	11.71	23.15	1.49	8.08	−7.25	7.99	15.41
1 mm, y	10.77	22.78	1.93	7.21	−2.43	10.45	18.67
1 mm, z	9.01	18.39	1.96	9.00	−15.69	21.29	15.50
2 mm, x	2.59	5.16	0.38	6.76	−7.77	4.03	18.42
2 mm, y	4.70	9.25	0.42	7.21	−6.95	7.20	16.03
2 mm, z	4.06	8.11	0.23	10.87	−1.33	7.73	15.41
Prospective 50%							
1 mm, x	13.84	26.65	0.37	9.39	−20.42	68.20	13.82
1 mm, y	16.49	34.10	2.37	10.65	−18.53	31.48	18.07
1 mm, z	9.42	18.56	0.64	6.21	−23.02	43.23	11.97
2 mm, x	4.08	8.23	1.08	0.00	−13.55	19.61	12.14
2 mm, y	6.78	13.73	1.11	10.40	−15.14	14.55	11.57
2 mm, z	4.95	9.78	0.48	5.39	−4.72	16.10	6.58
Prospective 55%							
1 mm, x	12.35	24.42	0.67	8.32	−11.09	24.19	10.01
1 mm, y	13.51	27.54	0.91	0.00	−9.05	54.20	15.74
1 mm, z	11.15	22.37	0.87	5.45	−20.58	52.17	22.13
2 mm, x	3.96	8.15	0.42	3.11	−7.65	9.80	13.02
2 mm, y	5.89	11.78	0.24	10.08	−10.85	25.41	7.62
2 mm, z	4.50	9.05	0.29	9.52	−3.47	21.32	11.79
Prospective 60%							
1 mm, x	15.67	31.94	1.36	12.45	−17.52	16.32	12.49
1 mm, y	15.03	30.65	3.44	6.21	−17.64	51.38	21.66
1 mm, z	8.64	17.52	0.53	6.21	−16.55	31.91	16.54
2 mm, x	4.85	9.67	0.84	5.83	−9.51	2.67	12.11
2 mm, y	6.37	12.53	0.80	5.88	−9.91	14.62	12.60
2 mm, z	3.88	7.45	0.21	10.20	0.00	11.87	15.92
Prospective 65%							
1 mm, x	10.71	20.71	0.75	13.36	−2.34	13.44	16.19
1 mm, y	11.26	23.03	0.95	7.21	5.86	19.28	14.38
1 mm, z	6.38	12.86	0.64	5.73	−15.40	16.51	7.51
2 mm, x	3.32	6.72	0.39	6.52	−4.70	8.24	5.31
2 mm, y	5.10	10.18	0.41	8.84	−5.46	9.25	7.15
2 mm, z	4.24	8.28	0.22	6.52	1.66	8.16	5.73
Prospective 70%							
1 mm, x	11.19	21.55	0.95	10.40	−2.95	14.39	6.39
1 mm, y	11.18	23.00	0.58	7.40	6.71	9.28	6.81
1 mm, z	8.47	16.61	0.69	7.21	−11.97	3.30	6.04
2 mm, x	3.30	6.63	0.32	5.88	−4.07	6.95	4.68
2 mm, y	5.44	10.61	0.35	5.73	−3.72	5.44	4.62
2 mm, z	4.69	9.34	0.24	6.52	2.28	2.03	2.39
Prospective 75%							
1 mm, x	10.83	20.52	0.31	6.52	−2.44	9.29	5.40
1 mm, y	10.58	21.46	0.45	1.41	4.57	9.86	7.45
1 mm, z	8.64	17.42	0.47	0.00	−14.74	5.24	11.32
2 mm, x	3.21	6.43	0.36	8.88	−5.26	6.27	7.56
2 mm, y	4.46	8.82	0.12	5.73	−3.95	4.08	7.23
2 mm, z	4.90	9.62	0.12	0.00	2.13	1.57	6.70
Prospective 80%							
1 mm, x	9.80	19.25	0.91	8.84	0.06	2.78	4.73
1 mm, y	9.81	20.18	1.02	8.44	4.73	22.16	2.41
1 mm, z	7.19	14.87	0.35	6.13	−7.85	18.21	10.22
2 mm, x	2.89	5.88	0.46	6.37	−4.18	1.24	5.34
2 mm, y	4.72	9.09	0.28	2.83	−3.88	8.69	4.40

Continued on next page

Table G.2: Normalized equivalent diameter error, cross-sectional area error, the CV of circularity, ERD, DSC, and CNR, and normalized attenuation bias for the prospective protocol (continued).

Prospective repeatability: normalized errors and coefficient of variation (continued)							
Tube	$E_{Deq,rel}$	$E_{A,rel}$	C	ERD	$B_{HU,rel}$	DSC	CNR
	Norm. error (%)	Norm. error (%)	CV (%)	CV (%)	Norm. bias (%)	CV (%)	CV (%)
2 mm, z	4.86	9.87	0.25	6.37	4.02	9.09	4.24

G.2.5. Phase-resolved prospective repeatability coefficients

For the prospective protocol, repeatability coefficients were calculated for all reconstructed phases available under the repeated reference condition of 120 bpm and 53 keV. These full phase-resolved RC values are reported in Table G.3. In the main results, only the 50% and 75% phases are shown to support the interpretation of the quantitative analyses.

Table G.3: Phase-resolved repeatability coefficients (RCs) for the prospective protocol at 120 bpm and 53 keV. RC values are reported per reconstructed phase, tube diameter, and endpoint.

Phase	Diameter	E_{Deq}	E_A	C	ERD	B_{HU}	DSC	CNR
25%	1.1 mm	0.067	0.123	0.024	0.217	42.9	0.238	3.47
25%	2.0 mm	0.057	0.166	0.008	0.134	39.3	0.085	5.20
30%	1.1 mm	0.067	0.135	0.030	0.237	44.1	0.139	3.33
30%	2.0 mm	0.071	0.192	0.007	0.134	33.2	0.111	5.76
35%	1.1 mm	0.080	0.139	0.016	0.160	42.9	0.223	3.89
35%	2.0 mm	0.076	0.210	0.008	0.247	33.0	0.149	7.16
40%	1.1 mm	0.082	0.150	0.051	0.151	69.1	0.286	5.37
40%	2.0 mm	0.045	0.134	0.007	0.143	28.5	0.128	5.32
45%	1.1 mm	0.083	0.158	0.047	0.175	63.8	0.247	4.76
45%	2.0 mm	0.027	0.077	0.010	0.182	28.9	0.148	8.82
50%	1.1 mm	0.139	0.238	0.037	0.210	62.6	0.488	3.44
50%	2.0 mm	0.102	0.301	0.025	0.160	58.2	0.305	5.03
55%	1.1 mm	0.085	0.142	0.021	0.134	61.0	0.583	4.12
55%	2.0 mm	0.070	0.192	0.009	0.200	39.3	0.387	5.69
60%	1.1 mm	0.120	0.227	0.055	0.202	60.3	0.498	4.60
60%	2.0 mm	0.088	0.253	0.018	0.178	61.8	0.234	7.63
65%	1.1 mm	0.039	0.058	0.021	0.208	61.6	0.289	4.35
65%	2.0 mm	0.042	0.122	0.010	0.167	29.4	0.191	3.46
70%	1.1 mm	0.053	0.079	0.020	0.189	35.9	0.180	2.61
70%	2.0 mm	0.044	0.132	0.008	0.138	30.6	0.117	4.84
75%	1.1 mm	0.058	0.110	0.011	0.089	31.3	0.159	2.73
75%	2.0 mm	0.043	0.124	0.006	0.138	31.5	0.101	4.27
80%	1.1 mm	0.052	0.104	0.021	0.171	28.3	0.290	2.07
80%	2.0 mm	0.047	0.137	0.009	0.129	42.4	0.163	2.58

G.3. Additional quantitative data plots

This section provides additional plots of the results for quantitative metrics that were not shown in the main text. The plots are organized according to the same analysis structure as the main results and were used as visual support for the statistical analyses.

G.3.1. Static inter-protocol analysis

Additional plots for the static inter-protocol analysis are provided in Figure G.7. These plots show the remaining quantitative metrics for both protocols under static conditions.

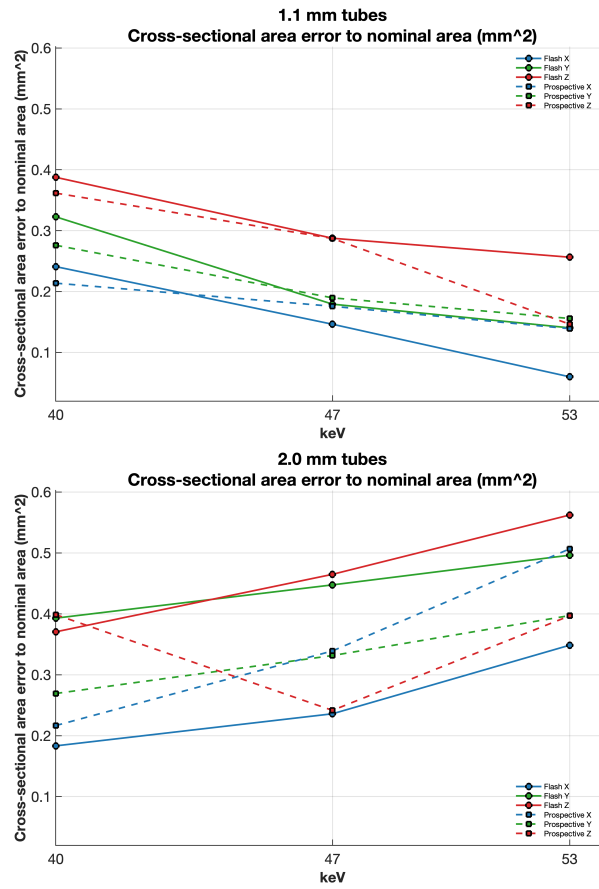


Figure G.7: Additional plots for the static inter-protocol analysis. The figure shows quantitative metrics not shown in the main text.

G.3.2. Z-matched phase inter-protocol analysis

Additional plots for the z-matched phase inter-protocol analysis are provided in Figure G.8. These plots show the remaining quantitative metrics for flash and prospective z-matched phase reconstructions.

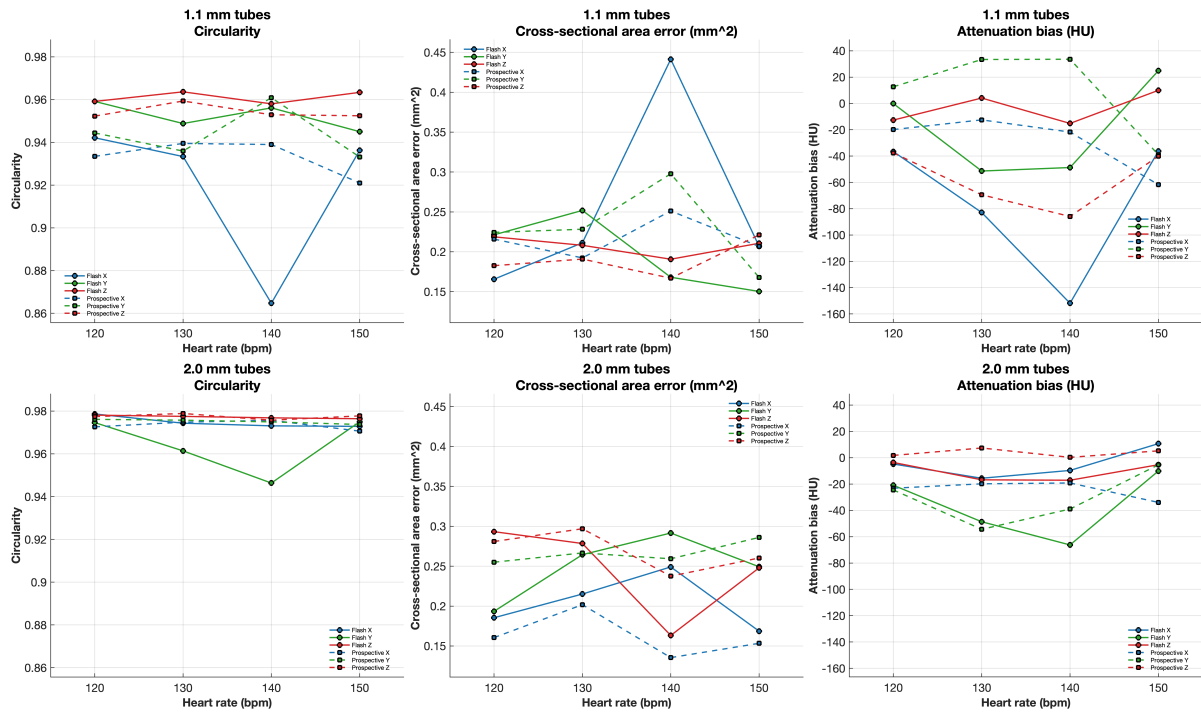


Figure G.8: Additional plots for the z-matched phase inter-protocol analysis. The figure shows quantitative metrics not shown in the main text.

G.3.3. Best phase inter-protocol analysis

Additional plots for the best phase inter-protocol analysis are provided in Figure G.9. These plots show the remaining quantitative metrics for flash and the prospective best-phase reconstructions.

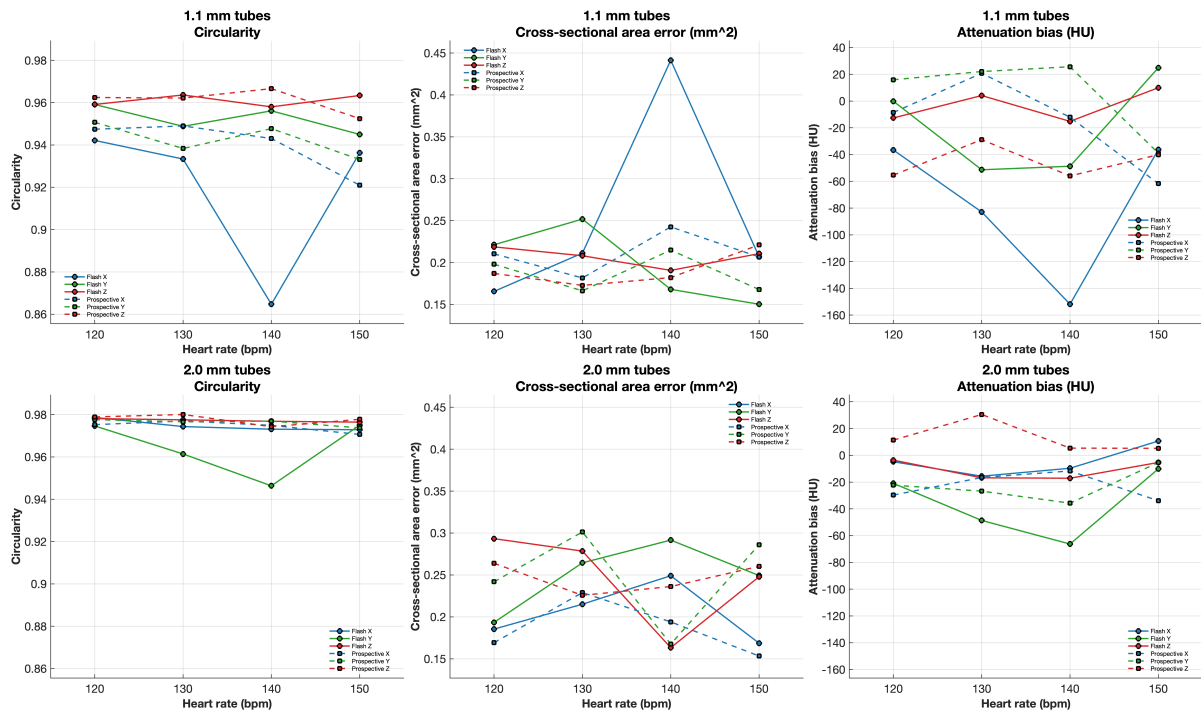


Figure G.9: Additional plots for the best phase inter-protocol analysis. The figure shows quantitative metrics not shown in the main text.

G.3.4. Flash within-protocol analysis

Additional plots for the flash acquisition analysis are provided in Figure G.10. These plots show the remaining quantitative metrics across heart rates and reconstruction energies.

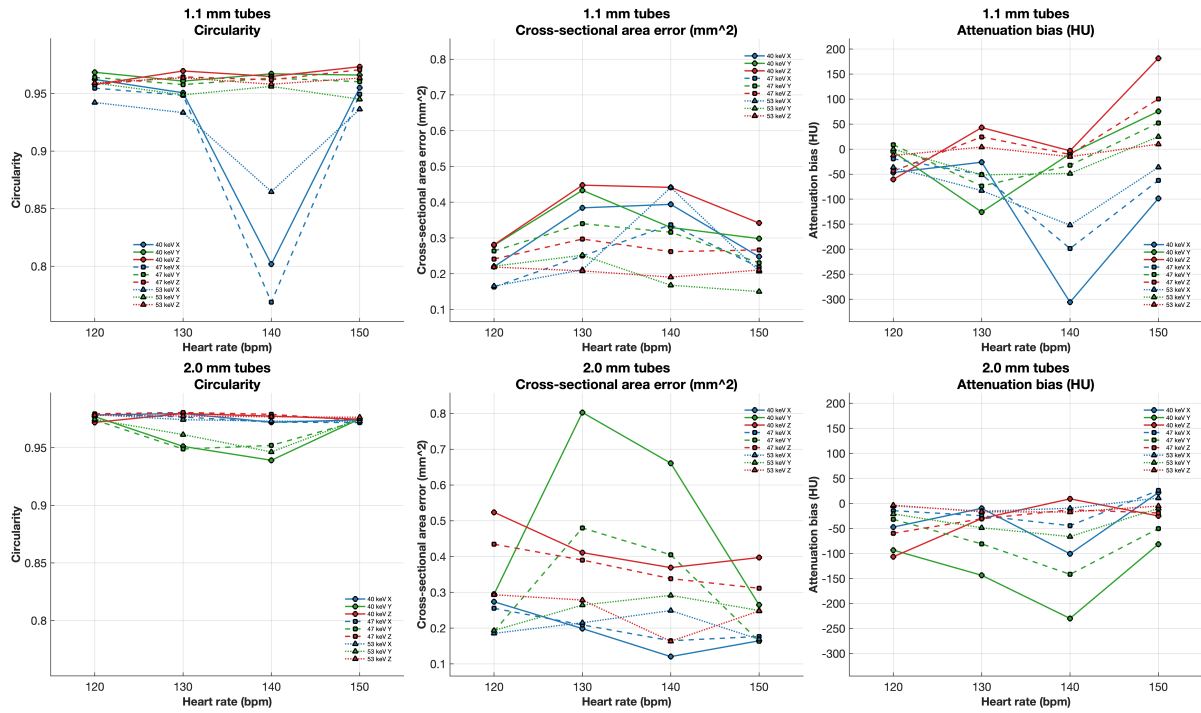


Figure G.10: Additional plots for the flash motion analysis. The figure shows quantitative metrics not shown in the main text.

G.3.5. Prospective within-protocol analysis

Additional plots for the prospective phase analysis are provided in Figure G.11. These plots show the remaining quantitative metrics across reconstructed phases and heart rates.

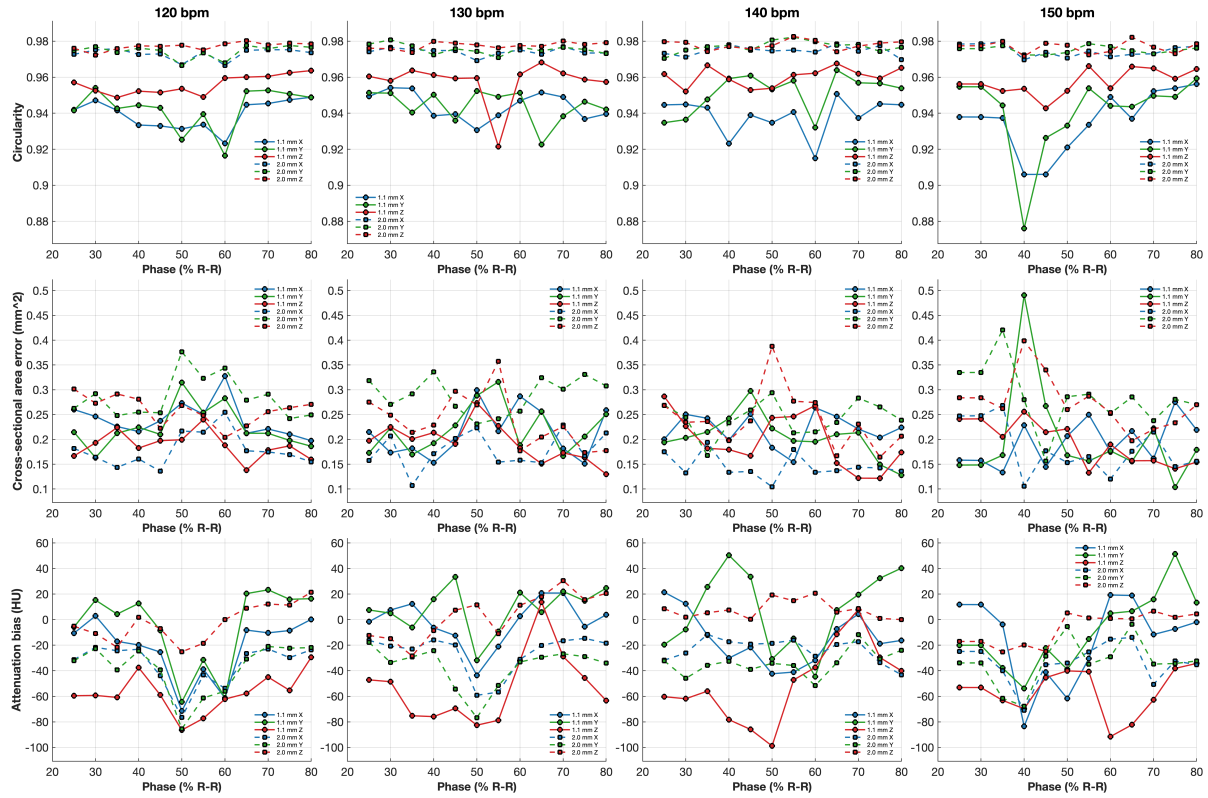


Figure G.11: Additional plots for the prospective phase analysis. The figure shows quantitative metrics not shown in the main text.

G.4. Post-hoc analysis

Table G.4: Holm-corrected post-hoc pairwise comparisons for significant fixed effects with more than two levels. Only comparisons remaining significant after Holm correction are shown.

Analysis	Metric	Comparison	Estimate diff.	p_{Holm}
Flash	E_{Deq}	40 vs 47 keV	0.027	0.020
Flash	E_{Deq}	40 vs 53 keV	0.044	< 0.001
Flash	E_A	40 vs 47 keV	0.078	0.007
Flash	E_A	40 vs 53 keV	0.131	< 0.001
Flash	E_A	47 vs 53 keV	0.053	0.046
Flash	ERD	40 vs 47 keV	-0.066	0.008
Flash	ERD	40 vs 53 keV	-0.078	0.003
Flash	CNR	40 vs 47 keV	3.511	< 0.001
Flash	CNR	40 vs 53 keV	4.543	< 0.001
Prospective	E_{Deq}	50 vs 75% R-R	0.025	0.010
Prospective	E_A	50 vs 75% R-R	0.053	0.007
Prospective	C	40 vs 65% R-R	-0.008	0.048
Prospective	C	40 vs 70% R-R	-0.008	0.046
Prospective	C	40 vs 80% R-R	-0.008	0.024
Prospective	B_{HU}	25 vs 50% R-R	24.546	< 0.001
Prospective	B_{HU}	30 vs 50% R-R	25.256	< 0.001
Prospective	B_{HU}	35 vs 50% R-R	18.665	0.021
Prospective	B_{HU}	35 vs 70% R-R	-18.157	0.028
Prospective	B_{HU}	40 vs 65% R-R	-18.898	0.018
Prospective	B_{HU}	40 vs 70% R-R	-20.117	0.008
Prospective	B_{HU}	45 vs 50% R-R	18.532	0.022
Prospective	B_{HU}	45 vs 70% R-R	-18.289	0.026
Prospective	B_{HU}	50 vs 60% R-R	-18.009	0.030
Prospective	B_{HU}	50 vs 65% R-R	-35.602	< 0.001
Prospective	B_{HU}	50 vs 70% R-R	-36.821	< 0.001
Prospective	B_{HU}	50 vs 75% R-R	-32.546	< 0.001

Continued on next page

Table G.4: Holm-corrected post-hoc pairwise comparisons for significant fixed effects with more than two levels (continued).

Analysis	Metric	Comparison	Estimate diff.	p_{Holm}
Prospective	B_{HU}	50 vs 80% R-R	-33.312	< 0.001
Prospective	B_{HU}	55 vs 65% R-R	-23.595	< 0.001
Prospective	B_{HU}	55 vs 70% R-R	-24.814	< 0.001
Prospective	B_{HU}	55 vs 75% R-R	-20.539	0.006
Prospective	B_{HU}	55 vs 80% R-R	-21.305	0.003
Prospective	B_{HU}	60 vs 65% R-R	-17.593	0.039
Prospective	B_{HU}	60 vs 70% R-R	-18.812	0.019
Prospective	DSC	25 vs 70% R-R	-0.072	0.034
Prospective	DSC	35 vs 50% R-R	0.095	< 0.001
Prospective	DSC	35 vs 55% R-R	0.070	0.046
Prospective	DSC	45 vs 50% R-R	0.073	0.029
Prospective	DSC	50 vs 70% R-R	-0.116	< 0.001
Prospective	DSC	50 vs 75% R-R	-0.099	< 0.001
Prospective	DSC	50 vs 80% R-R	-0.090	0.001
Prospective	DSC	55 vs 70% R-R	-0.091	0.001
Prospective	DSC	55 vs 75% R-R	-0.074	0.024
Prospective	DSC	60 vs 70% R-R	-0.078	0.011
Prospective	CNR	25 vs 50% R-R	2.515	< 0.001
Prospective	CNR	25 vs 55% R-R	2.057	< 0.001
Prospective	CNR	30 vs 40% R-R	1.617	0.006
Prospective	CNR	30 vs 45% R-R	1.814	< 0.001
Prospective	CNR	30 vs 50% R-R	3.110	< 0.001
Prospective	CNR	30 vs 55% R-R	2.652	< 0.001
Prospective	CNR	30 vs 60% R-R	1.702	0.003
Prospective	CNR	35 vs 50% R-R	1.893	< 0.001
Prospective	CNR	35 vs 55% R-R	1.435	0.029
Prospective	CNR	40 vs 50% R-R	1.493	0.019
Prospective	CNR	45 vs 75% R-R	-1.466	0.023
Prospective	CNR	50 vs 60% R-R	-1.407	0.036
Prospective	CNR	50 vs 65% R-R	-2.243	< 0.001
Prospective	CNR	50 vs 70% R-R	-1.911	< 0.001
Prospective	CNR	50 vs 75% R-R	-2.762	< 0.001
Prospective	CNR	50 vs 80% R-R	-2.109	< 0.001
Prospective	CNR	55 vs 65% R-R	-1.785	0.001
Prospective	CNR	55 vs 70% R-R	-1.453	0.026
Prospective	CNR	55 vs 75% R-R	-2.304	< 0.001
Prospective	CNR	55 vs 80% R-R	-1.651	0.005
Static	ERD	40 vs 47 keV	-0.073	0.022
Static	ERD	40 vs 53 keV	-0.118	< 0.001
Static	CNR	40 vs 53 keV	5.870	< 0.001
Static	CNR	47 vs 53 keV	7.256	< 0.001