OPTIMIZATION OF IMAGE QUALITY AND RADIATION DOSE IN DIGITAL MAMMOGRAPHY

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Main topics

- Optimization in mammography
- Digital mammography
- Image quality evaluation
- Technical quality/clinical quality
- Dose measurements

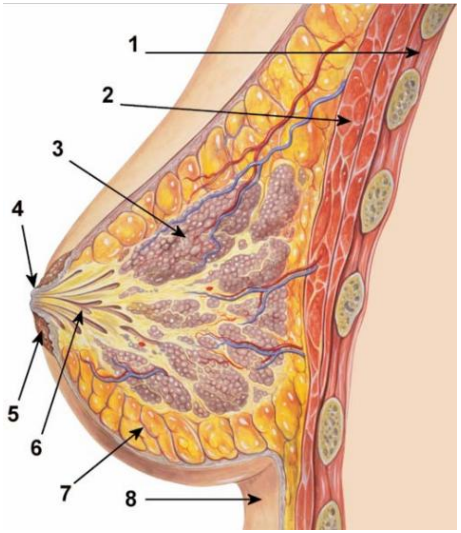
Optimization in mammography

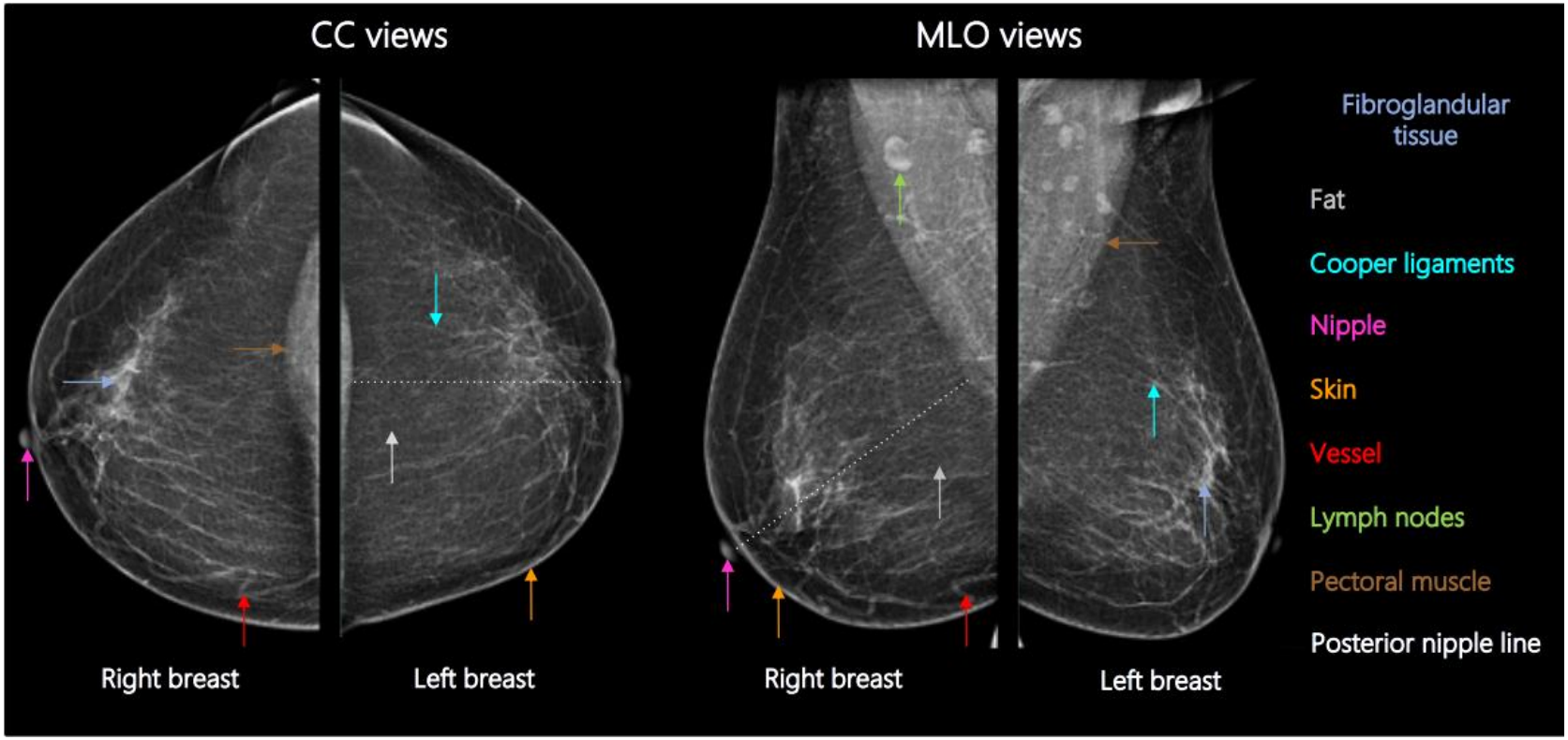
In general, radiological images should be acquired to ensure sufficient image quality to answer the diagnostic question while keeping the radiation dose as low as reasonably achievable (**ALARA principle**).

As will be discussed, achieving the optimal balance in mammography is **both particularly important and especially challenging**

Normal breast anatomy

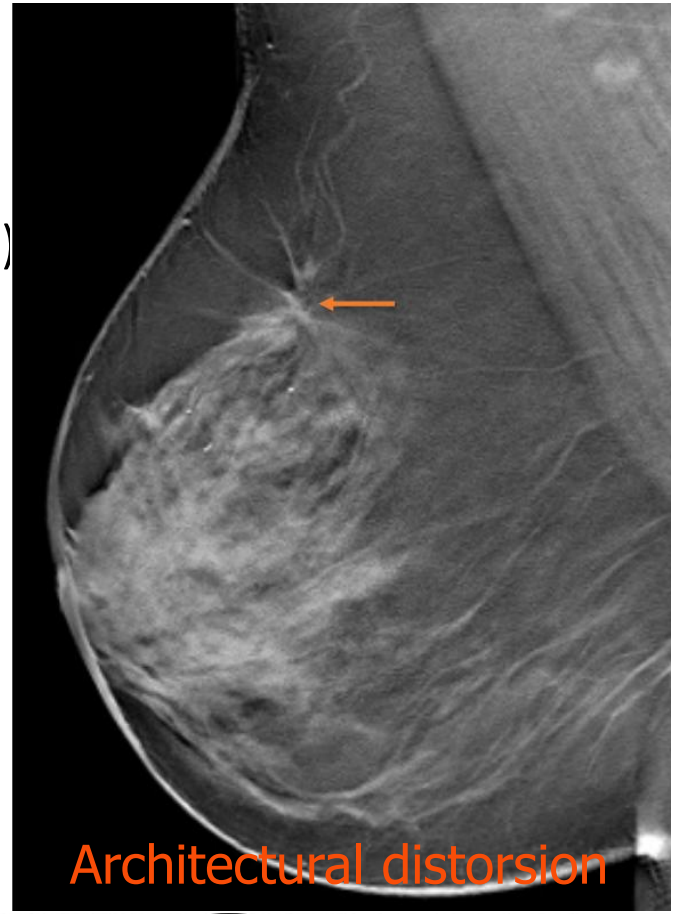
1. Chest wall
2. Pectoralis muscles
3. Lobules
4. Nipple
5. Areola
6. Duct
7. Fatty tissue
8. Skin

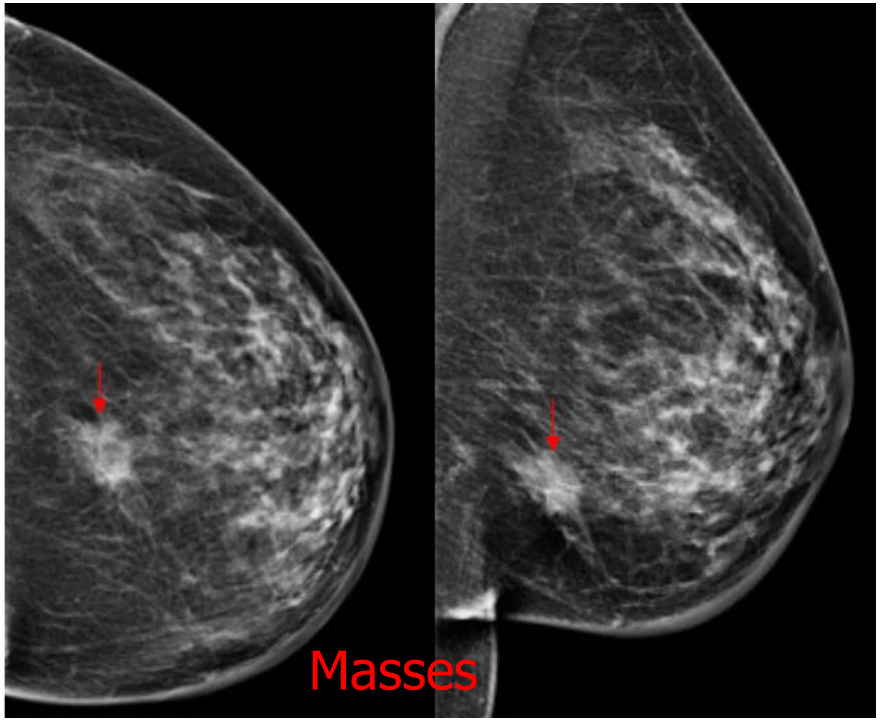


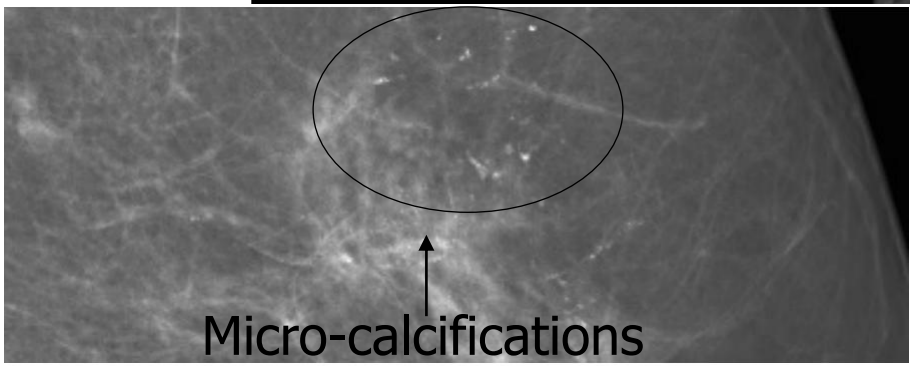


Features of Breast Cancer

- Increased Density (relative to previous exams)
- Architectural Distortion
- Micro-calcifications
- Masses







Optimization in mammography

The objective of optimization is to produce images that provide maximum visualization of breast anatomy and the signs of disease, which means:

- detect and characterize microcalcification cluster patterns and morphology
- visualize breast parenchyma and subtle architectural distortions
- detect soft tissue masses and assess shape, size, degree of local invasion

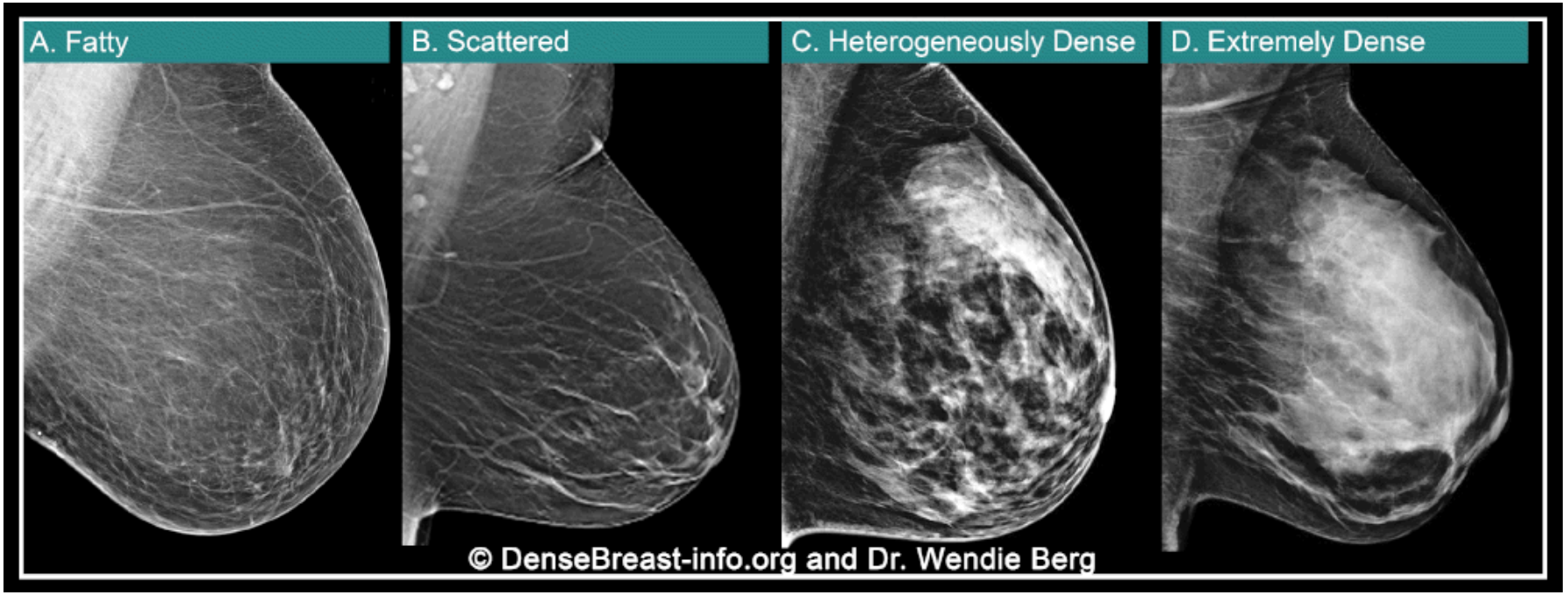
BUT

without exposing the breast to unnecessary radiation

(the breast is a highly radiosensitive organ)

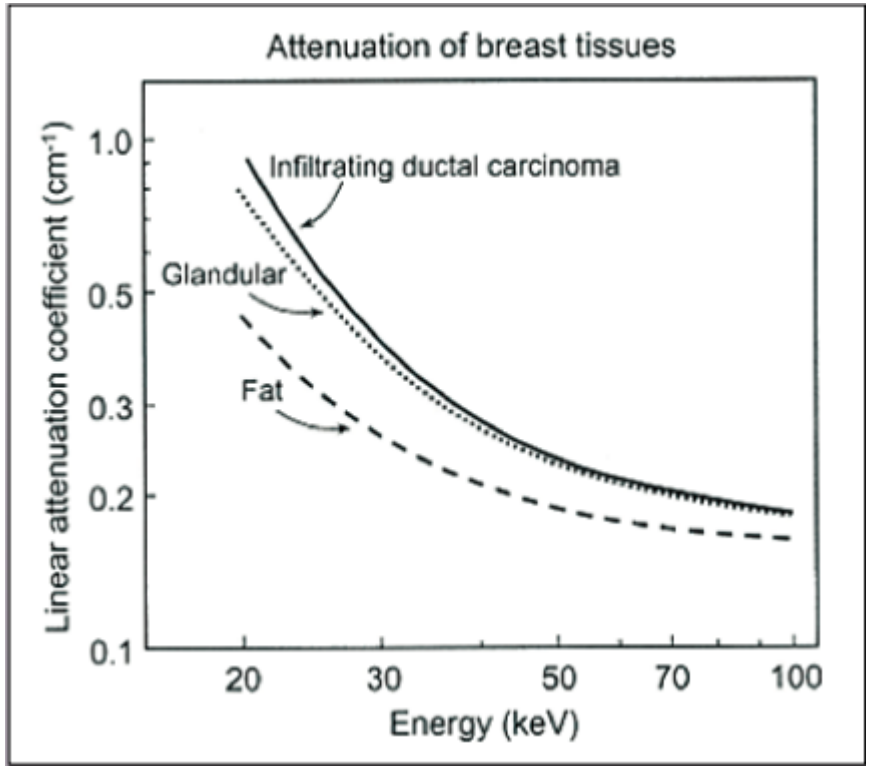
Challenges

Breast imaging presents various challenges. The density of the breast can vary depending on age, genetic factors, and the menstrual cycle (in premenopausal women). Although this variability is physiological, it has important implications for lesion detectability.



Challenges

Fibro-glandular and tumor tissue have similar attenuation properties, so **high contrast resolution is required**

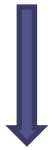


Source: Bushberg, The Essential Physics of Medical Imaging

Challenges

On the market, different mammography systems are available, differing in terms of:

- anode/ filters combinations
- detector (technology and specific proprieties)
- Automatic Exposure Control (they may operate in different modes)
- image processing



Device-specific optimization is required

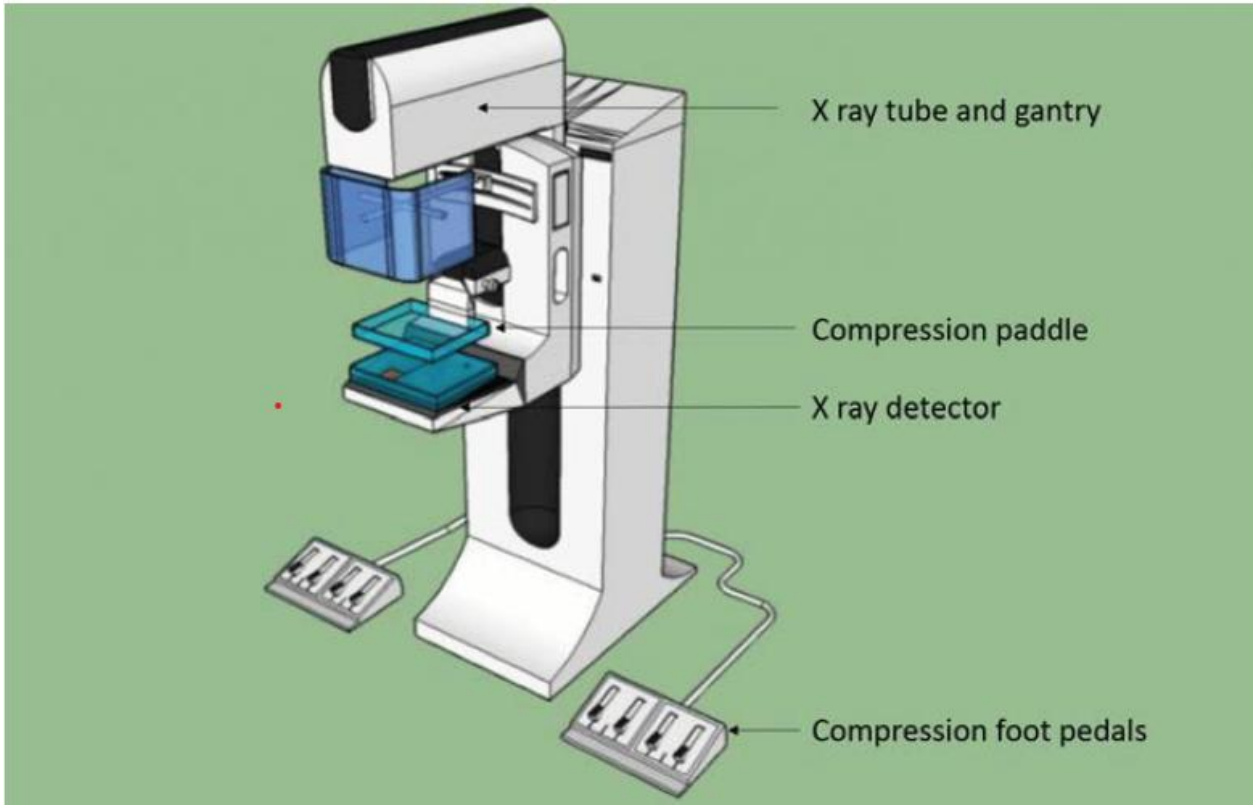
Main topics

- Optimization in mammography
- Digital mammography
- Image quality evaluation
- Technical quality/clinical quality
- Dose measurements

Digital mammography system

A complex chain that includes X-Ray tube, detector, software, PACS (Picture Archiving and communication System)

Digital mammography system



Digital mammography system

X ray generator	
Power	≥5.0 kW
Tube voltage range	24–35 kVp (maximum 1 kVp step)
Tube load range	At least 5–400 mAs
AEC	Manual and automated selection of kVp, mAs, filter
Exposure time range	At least 30 ms to 2 s
Anode and filter material	Materials need to allow for low dose/high penetration spectra, even for thick or dense breasts. Examples of appropriate target materials are, for X rays: molybdenum (Mo), rhodium (Rh), tungsten (W) and for beam filter(s): Mo, Rh, aluminium (Al), silver (Ag)
Focal spot	Two (approximately 0.3 mm and 0.1 mm); with both manual and automated selection of the focal spot available
Anode heat capacity	At least 200 000 HU

To enhance the visibility of low contrast details !!

To minimize blurring and provide visibility of the small calcifications

Digital mammography system

Suitable spectral shaping is selected on order to obtain a high contrast sensitivity

The first dedicated X-Ray tube used molybdenum both as anode material and beam filter.

Mo anode (Z = 42) produces characteristic X-ray peaks at 17.6 keV and 19.7 keV and the filter, with an attenuation K edge at 20 keV, absorbs most of the hard X-Rays. This combination produces an X-Ray spectrum with a peak around 20 keV, optimal to visualize low contrast details using a reasonable radiation dose to the patient for smaller breasts.

Two anode targets (dual track X-Ray tube) and two filters, in Molybdenum and Rhodium respectively, were produced to improve image quality also for larger and denser breasts :

Mo target and 30 micron Mo filter, for fatty breast up to 4 cm thick

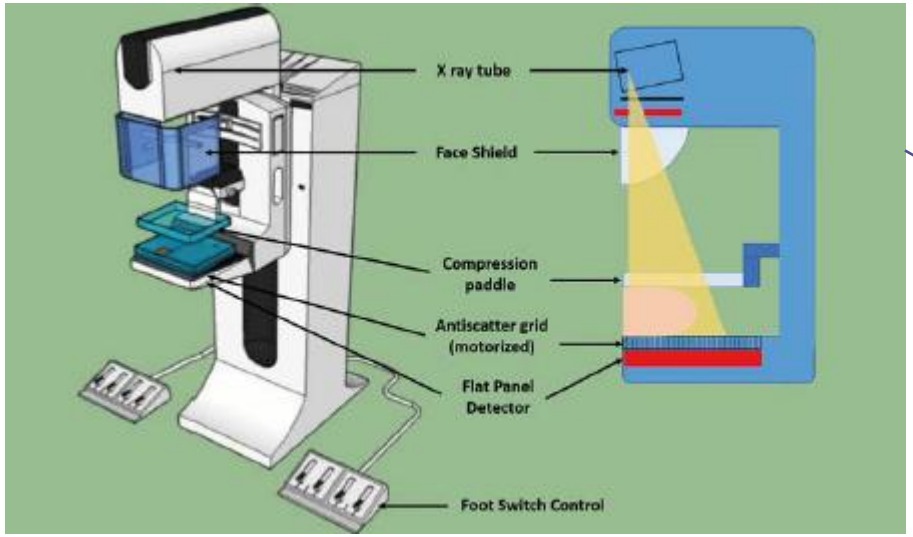
Mo target and 25 micron Rh filter, for glandular breast, from 5 to 7 cm thick

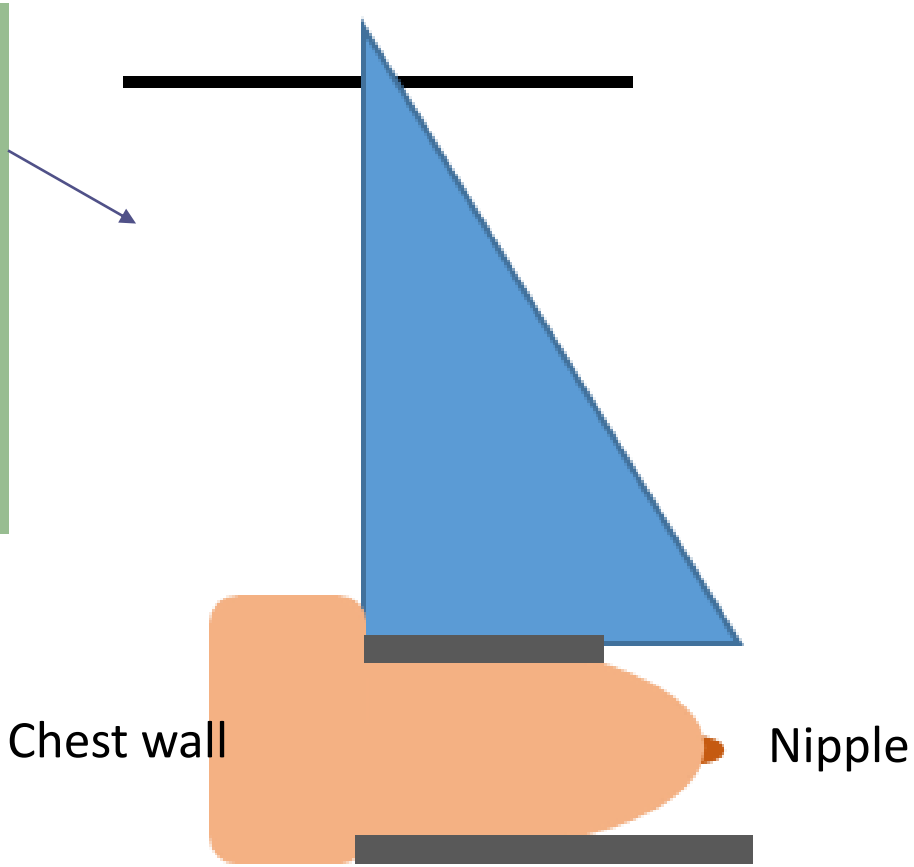
Rh target and 25 micron Rh filter, for breast thickness higher than 7 cm

The new equipments (coupled with flat panel detectors) use anode in Tungsten and filter materials are Rhodium, Silver and for special technique (as tomosynthesis) also Aluminium.

Digital mammography system

Only half beam is used to form the image. To obtain the full field coverage, target angle larger than 20° is needed.

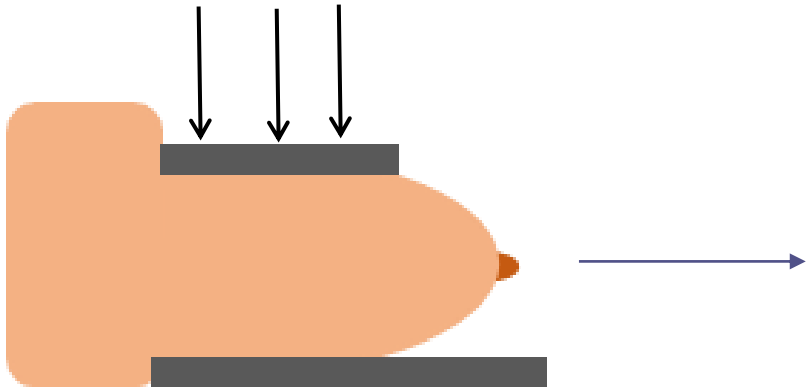




Digital mammography system

Compression is essential for optimizing image quality and reducing patient dose, but it is also uncomfortable and painfull for the patient.
Recommended specifications for compression:

- Manual and automated breast compression
- Maximum force (when automated: 150 N ÷ 200 N
- Emergency release
- Automatic decompression after exposure



Digital mammography system

Nowadays, almost only mammography systems equipped with **digital flat-panel detectors** are commercially available

Recommended minimum specifications	
Detector type	Preferably direct conversion flat panel (amorphous Selenium (a-Se), but also indirect conversion flat panel (CsI)
Effective field size	At least equivalent to 24 cm x 29 cm
Pixel size	≤ 100 µm
Dead pixel map	Provided
Image depth	At least 12 bit
System spatial resolution	≥ 7 lp/mm

Digital mammography system

Image processing and diagnosis workstation	
Workstation	System of the latest technology workstation class hardware (processor generation/type/speed, RAM, hard disk, storage systems, etc.)
Monitors	At least two diagnostic mammography approved monitors of at least 5 MP and 53.5 cm (21 inches), with automated self-calibration
Display graphic card	High end medical grade
Storage capacity	As required (e.g. at least 1.5 Terabyte (TB))
Workstation capabilities	— Display of multiple images and priors for comparison purposes Multimodality viewer capability for display of ultrasound, X ray, digital mammography, MRI, PET, CT on a third colour monitor
Mouse, keypad	Dedicated keypad for mammography, common keypad, mouse

Main topics

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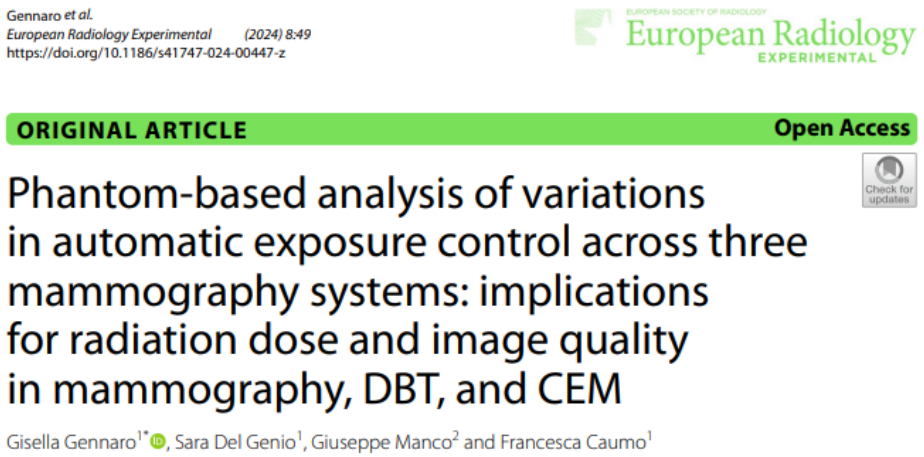
AEC



Gennaro *et al.*
European Radiology Experimental (2024) 8:49
https://doi.org/10.1186/s41747-024-00447-z

European Radiology EXPERIMENTAL

ORIGINAL ARTICLE Open Access

Phantom-based analysis of variations in automatic exposure control across three mammography systems: implications for radiation dose and image quality in mammography, DBT, and CEM

Gisella Gennaro[1*], Sara Del Genio[1], Giuseppe Manco[2] and Francesca Caumo[1]

AEC plays a crucial role in optimization, because it determines the the “optimal” exposure conditions needed to achieve a specific image quality.

In digital mammography the detector is used both for a short pre-exposure (few ms) and image acquisition.
The pre-exposure provides information on breast composition and identifies the most absorbent area.
The height of the compression paddle estimates the breast thickness

AEC

Depending on breast composition and breast thickness, the system selects

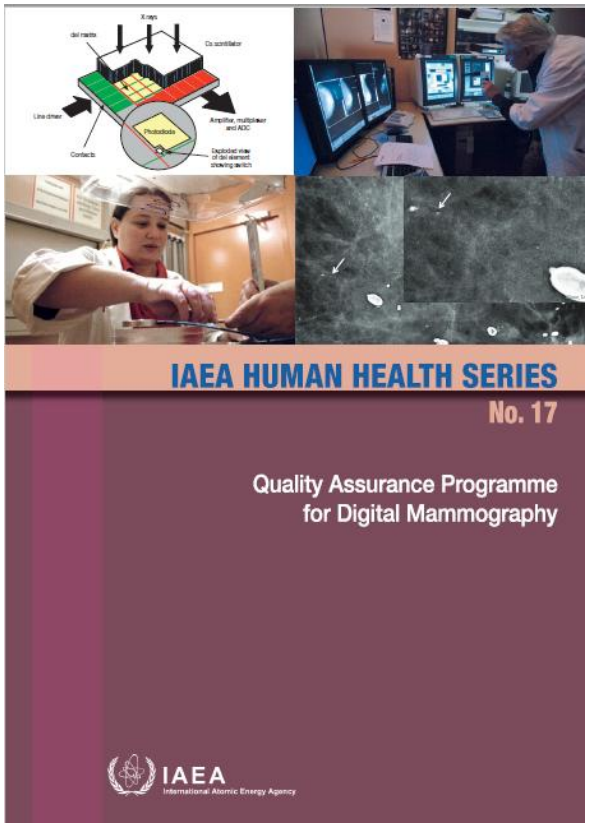
- anode/filter combination
- kV
- mAs

in order to obtain **the desired balance between image quality (IQ) and radiation dose.**

To design, calibrate and test the performance of a AEC system, simple and reproducible methods are commonly adopted, based on the use of:

- a simple phantom
- a easy to measure and reproducible metric

Protocols


IAEA HUMAN HEALTH SERIES
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Quality Assurance Programme
for Digital Mammography
IAEA
International Atomic Energy Agency


E
F
M
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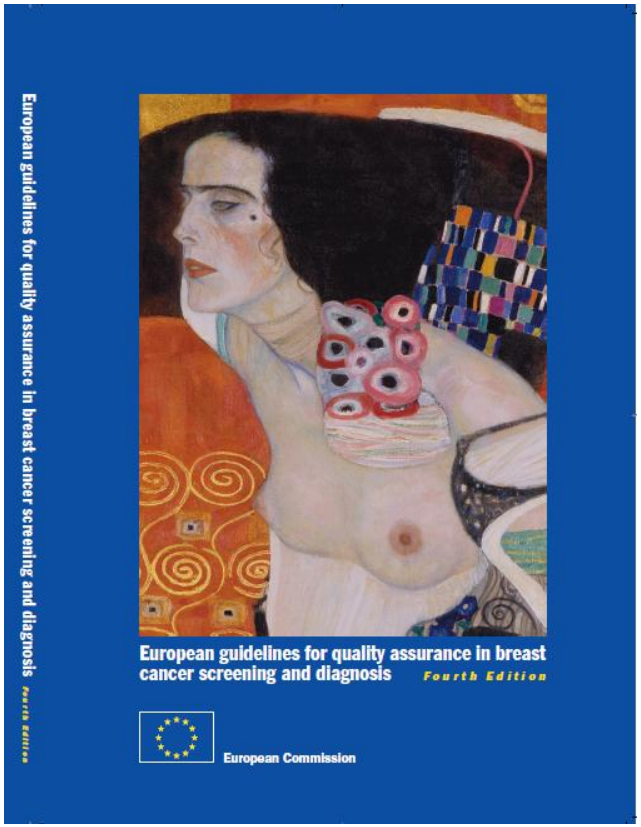
Mammo Protocol
EFOMP
Mammo Working Group Protocol, March 2015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European Commission

SDNR (EFOMP)

Signal-difference-to-noise ratio (SDNR)

$$SDNR = \frac{MPV_{bkg} - MPV_{Al}}{SD_{bkg}}$$

SDNR (EFOMP)

PMMA Slabs simulate breasts

An aluminium detail
0.2 mm thick - side 10 – 15 mm

Image of the phantom is acquired with **active** compression device and AEC (automatic exposure control)

DICOM FOR PROCESSING IMAGES are used for the analysis

Analysis

A software for image analysis is required.

EFOMP protocol describes in detail how perform the analysis in ImageJ (free packages), but other solutions can be adopted-

MPV and SD of background should be measured in «square band» around the aluminium detail

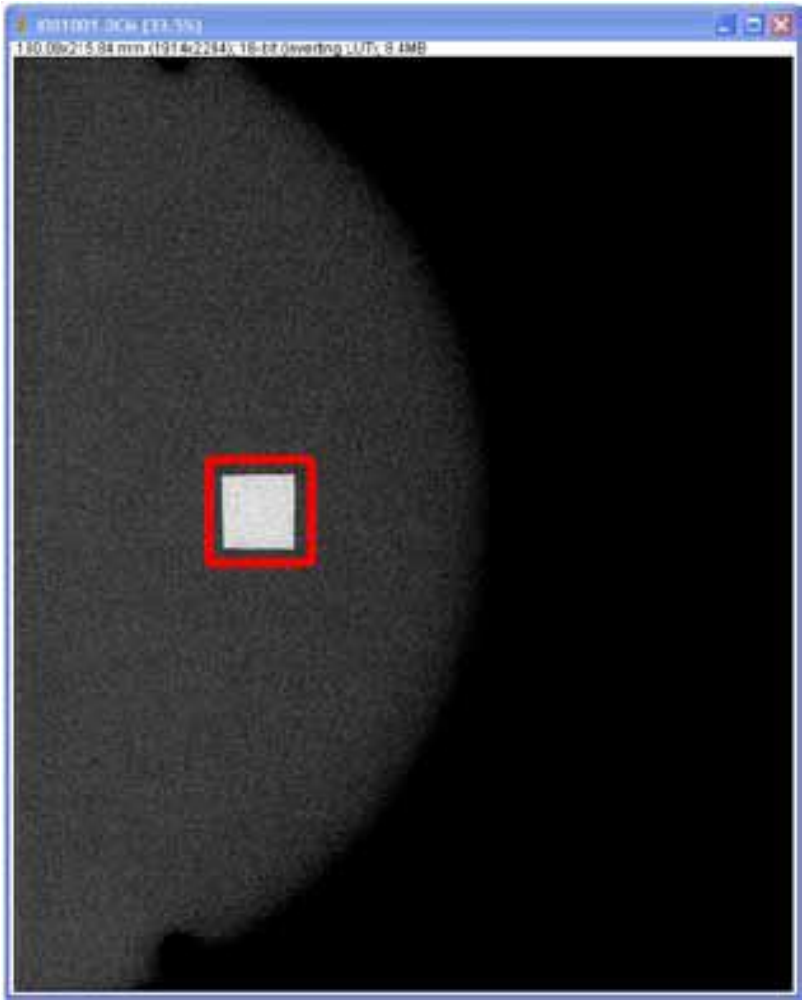


Image analysis

For each PMMA thickness the SDNR is calculated according to the definition.

SDNR variation respect to reference thickness 45 mm is evaluated, according to the following formula:

$$\Delta SDNR_{45mm} = (SDNR_i - SDNR_{45mm})/SDNR_{45mm} \times 100$$

The exposure parameters are recorded, in order to evaluate MGD (anode, filter, kV, mAs..)

SDNR

HOLOGIC (FFDM)				
PMMA (mm)	SDNR	$\Delta SDNR_{45mm}$ (%)	Limits $\Delta SDNR_{45mm}$	Test
20	8.56	30.8	*≥ 0%*	Passed
30	7.50	14.7	*≥ 0%*	Passed
40	6.84	4.5	*≥ 0%*	Passed
45	6.54	0.0	*0%*	Passed
50	5.67	-13.4	*≥ -15%*	Passed
60	5.24	-20.0	*≥ -30%*	Passed

Hologic					FFDM						
PMMA nominal thickness (mm)	PMMA measured thickness (mm)	HVL (mmAl)	mAs	µGy/mAs @ SDD	K_i (mGy)	g-fact	c-fact	s-fact	AGD (mGy)	AGD limit (mGy)	Test
20	21	0.534	40.0	24.14	1.027	0.549	0.915	1.042	**0.538**	**1.0**	Passed
30	32	0.546	63.0	26.50	1.833	0.416	0.952	1.042	**0.757**	**1.5**	Passed
40	45	0.566	83.0	31.59	2.974	0.320	1.035	1.042	**1.026**	**2.0**	Passed
45	53	0.575	100.0	34.34	3.959	0.283	1.090	1.042	**1.273**	**2.5**	Passed
50	60	0.588	92.0	40.23	4.338	0.255	1.135	1.042	**1.309**	**3.0**	Passed
60	75	0.606	92.0	53.61	5.977	0.212	1.206	1.042	**1.592**	**4.5**	Passed

SDNR

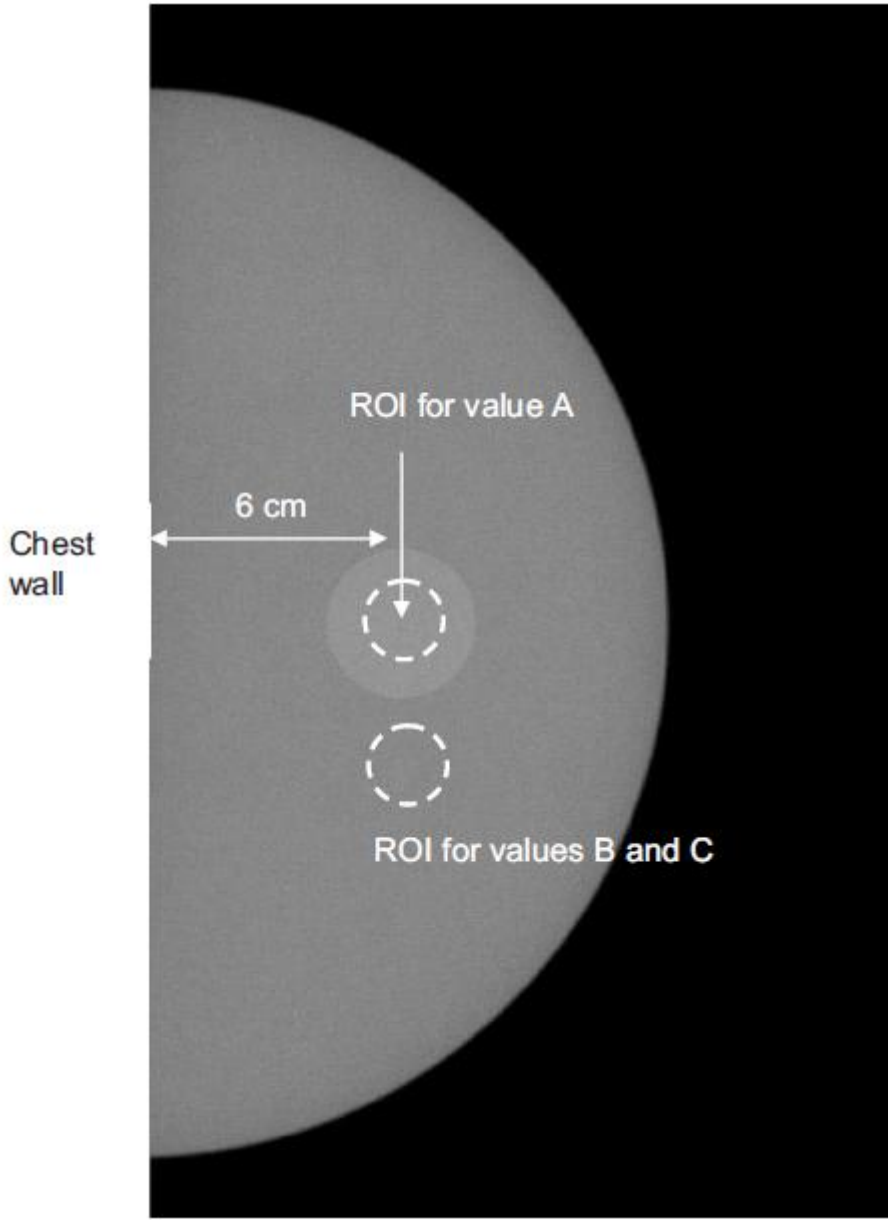
The SDNR is easy to measure and can be used to test the AEC performance.

There are significant differences in the SDNR produced by different mammography systems depending on:

- difference of x-ray source
- detector technology,
- read-out electronics
- the presence (in special case) of some type of pre-processing.

It is impossible to assess quality of any given digital mammography system on the basis of only the SDNR. It is not an absolute image quality index and other image quality parameters are necessary to perform inter-system comparison

SDNR (IAEA)



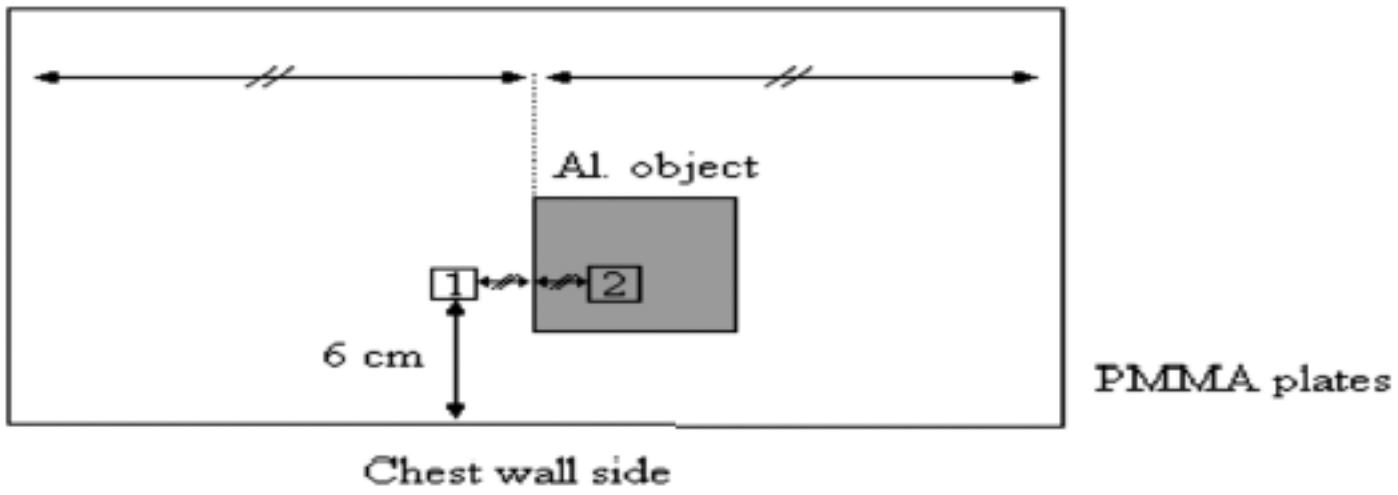
$$SDNR = |B - A|/C$$

A: Mean pixel Value
B: Mean pixel value
C: Standard deviation

CNR (EUREF)

Contrast-to-noise ratio (CNR)

$$CNR = \frac{\text{mean pixel value (signal)} - \text{mean pixel value (background)}}{\sqrt{\frac{\text{Standard deviation (signal)}^2 + \text{Standard deviation (background)}^2}{2}}}$$



MGD (or AGD)

Typical breasts simulated with PMMA

$$\mathbf{AGD = K \cdot g \cdot c \cdot s}$$

- **D** is the mean glandular dose
- **K** is the incident air kerma (without backscatter)
- The **factor g**, corresponds to a glandularity of 50% and it is tabulated for a range of HVL.
- The **factor c** corrects for the difference in composition of typical breasts from 50% glandularity and it is given for typical breasts in different age range
- The **factor s** corrects for differences due to the choice of X-ray spectrum

Standard phantom: 45 mm PMMA; Glandularity 50%

g-Factors for breasts simulated with PMMA

PMMA thickness (mm)	Equivalent breast thickness (mm)	Glandularity of equivalent breast (%)	g-factor (mGy/mGy) HVL (mm Al)										
			0.30	0.35	0.40	0.45	0.50	0.55	0.60	0.65	0.70	0.75	0.80
20	21	97	0.378	0.421	0.460	0.496	0.529	0.559	0.585	0.609	0.631	0.650	0.669
30	32	67	0.261	0.294	0.326	0.357	0.388	0.419	0.448	0.473	0.495	0.516	0.536
40	45	41	0.183	0.208	0.232	0.258	0.285	0.311	0.339	0.366	0.387	0.406	0.425
45	53	29	0.155	0.177	0.198	0.220	0.245	0.272	0.295	0.317	0.336	0.354	0.372
50	60	20	0.135	0.154	0.172	0.192	0.214	0.236	0.261	0.282	0.300	0.317	0.333
60	75	9	0.106	0.121	0.136	0.152	0.166	0.189	0.210	0.228	0.243	0.257	0.272
70	90	4	0.086	0.098	0.111	0.123	0.136	0.154	0.172	0.188	0.202	0.214	0.227
80	103	3	0.074	0.085	0.096	0.106	0.117	0.133	0.149	0.163	0.176	0.187	0.199

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The glandularity of typical breast in not 50%.
C- factor corrects for the difference in composition of typical breasts from 50% glandularity

c-Factors for breasts simulated with PMMA

PMMA thickness (mm)	Equivalent breast thickness (mm)	Glandularity of equivalent breast (%)	c-factor* HVL (mm Al)										
			0.30	0.35	0.40	0.45	0.50	0.55	0.60	0.65	0.70	0.75	0.80
20	21	97	0.889	0.895	0.903	0.908	0.912	0.917	0.921	0.924	0.928	0.933	0.937
30	32	67	0.940	0.943	0.945	0.946	0.949	0.952	0.953	0.956	0.959	0.961	0.964
40	45	41	1.043	1.041	1.040	1.039	1.037	1.035	1.034	1.032	1.030	1.028	1.026
45	53	29	1.109	1.105	1.102	1.099	1.096	1.091	1.088	1.082	1.078	1.073	1.068
50	60	20	1.164	1.160	1.151	1.150	1.144	1.139	1.134	1.124	1.117	1.111	1.103
60	75	9	1.254	1.245	1.235	1.231	1.225	1.217	1.207	1.196	1.186	1.175	1.164
70	90	4	1.299	1.292	1.282	1.275	1.270	1.260	1.249	1.236	1.225	1.213	1.200
80	103	3	1.307	1.299	1.292	1.287	1.283	1.273	1.262	1.249	1.238	1.226	1.213

* for typical breasts for women in the age group 50–64

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S-factor corrects for differences due to the choice of X-ray spectrum

Target material	Filter material	Filter thickness (µm)	s-factor
Mo	Mo	30	1.000
Mo	Rh	25	1.017
Rh	Rh	25	1.061
W	Rh	50–60	1.042
W	Ag	50–75	1.042

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New breast dosimetry

Received: 25 June 2023 | Revised: 26 September 2023 | Accepted: 10 November 2023

DOI: 10.1002/mp.16842

AAPM SCIENTIFIC REPORT

MEDICAL PHYSICS

Joint AAPM Task Group 282/EFOMP Working Group Report: Breast dosimetry for standard and contrast-enhanced mammography and breast tomosynthesis

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Image quality

Image quality is a complex concept. It depends on:

- contrast resolution
- spatial resolution
- noise
- dose level

How can image quality be evaluated?

Technical Image Quality (phantoms)	Clinical Image Quality (breasts)
Most phantoms are simplified models: provide images with uniform background displaying details in know locations for simple measurements. A few more complex phantoms mimic anatomical tissue however they still fail to reproduce the natural variability in human anatomy.	Wide variability of clinical images: different breast patterns and lesion types/characteristics/difficulties in both detection and characterization.
Physical evaluation, based on a priori knowledge of phantom contents: measurable parameters (like SDNR, MTF or many others indices) or assessement based on "visibility" of details. The mechanism of detecting an expected object whose spatial localization is known is significantly different than clinical detection.	Clinical evaluation: includes detection and characterization steps. Detection is necessary to localize a clinical feature; characterization is the decision made after the interpretation of clinical features. There is no a priori knowledge in routine clinical evaluation, and this change detection performance.
Detectability of image features in a uniform background is strongly correlated with radiation dose. Structured backgrounds, such as that generated by human anatomy due to the superimposition of normal tissue, pose a significantly different challenge and one that cannot be eliminated by increasing dose.	Detectability is mainly restricted by normal tissue superimposition, which produces structured background. Such "anatomical noise" cannot be reduced by adjusting dose. This is the main reason why "subtraction techniques" (tomosynthesis, dual energy, etc) have been introduced.
Technical image quality is normally evaluated by unprocessed images, the processing effect not being related to the physics of image formation.	Clinical image quality is normally evaluated on processed images. Radiologists do not use unprocessed images, which are usually not archived in the PACS.
Possible optimization methods are applied using unprocessed images, under the general principle that acceptable/achievable TIQ should be obtained while keeping radiation dose as low as possible.	Clinical image quality is normally evaluated using processed images. Post-processing algorithms might nullify differences between unprocessed images obtained at slightly different dose levels as determined by the optimization process.

Techincal Image Quality IQ e CIQ

Test objects (TO) or phantoms are reproducible and known objects.
Images of these objects can be analysed to evaluate image quality parameters, which describe the so-called technical image quality (TIQ)

Clinical image quality (CIQ) is usually evaluated within studies properly designed, including many clinical images and several radiologists, to consider the wide variability in human perception and decision criteria.

The relationship between TIQ and CIQ is complex and not easy to characterise.

TIQ e CIQ

In general, the improvements in technical image quality are likely to lead to improvements in clinical image quality,

….but it is very difficult (virtually impossible) to use absolute or relative thresholds for TIQ parameters to identify an acceptable level of diagnostic image quality

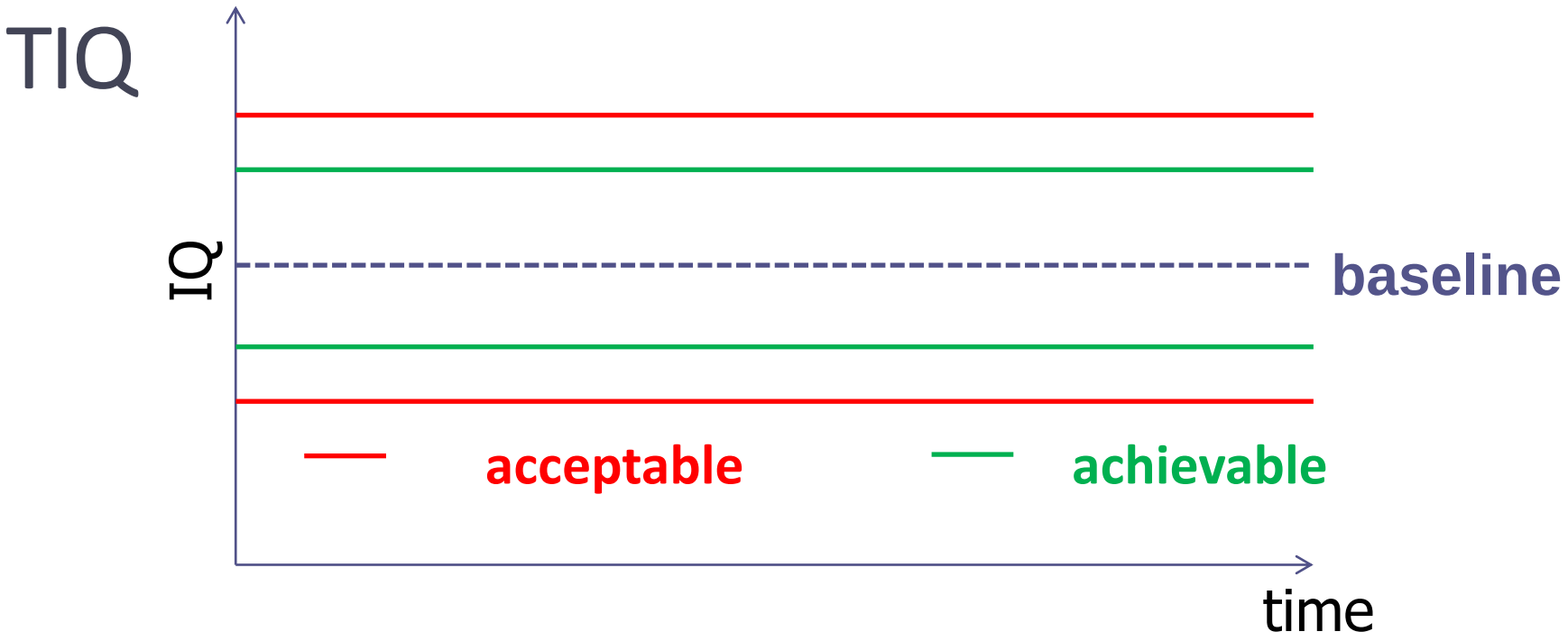
Evaluation of image quality using test objects and phantoms

Three objectives

1. During acceptance/commissioning test, the technical image quality is evaluated on images obtained **in AEC mode** (at the dose level suggested from the manufacturer).

   The level of IQ can be compared with the minum acceptable level, as defined for the adopted QC protocol, to verify the performance of the systems

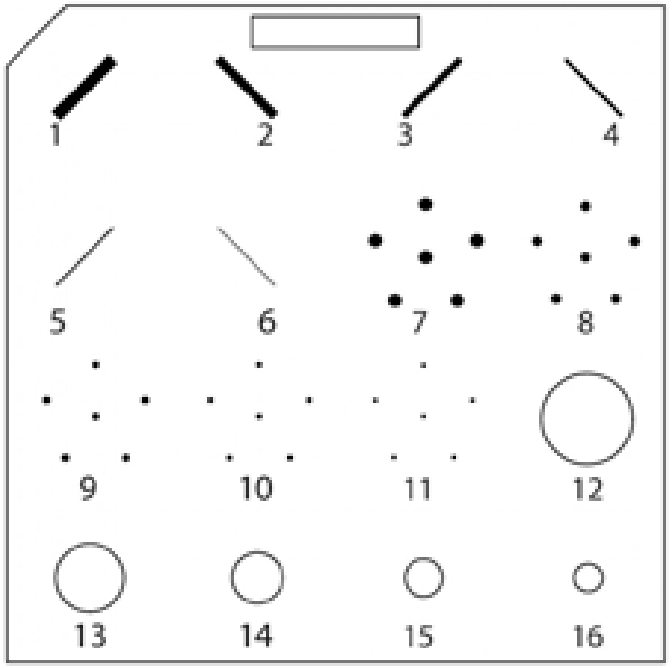
2. Define a baseline for one or more IQ indices and a “control interval” to be used as reference thereafter for longterm reproducibility tests.



The typical tollerance interval is calculated as baseline ± 3 standard deviations.

3. Monitor the performance of the system during the time (long term reproducibility), through “constancy test”

ACR Accreditation phantom

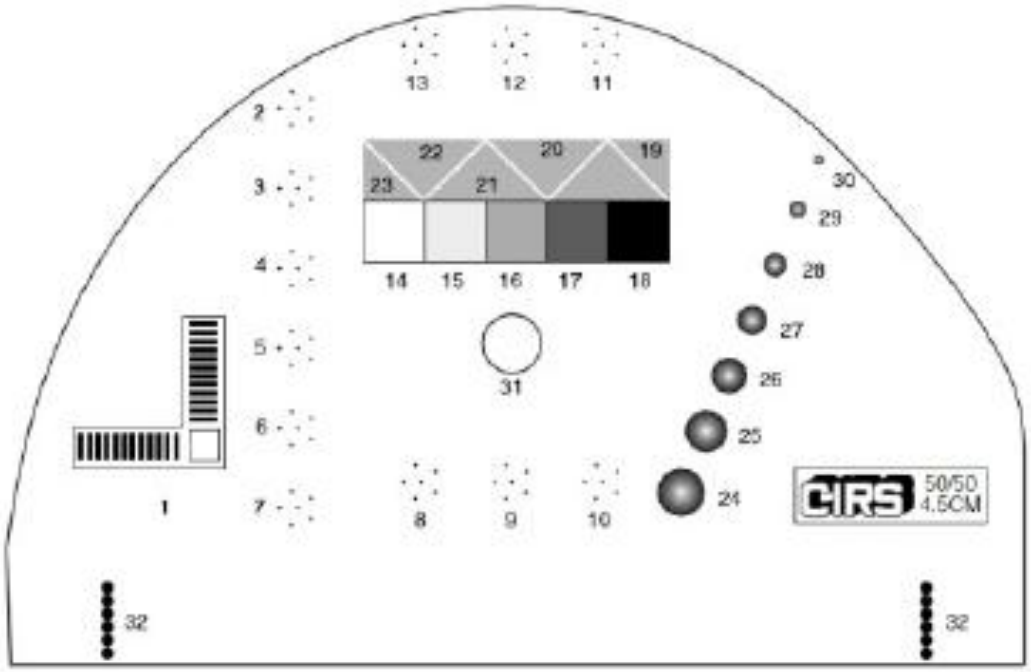


ACR Accreditation phantom

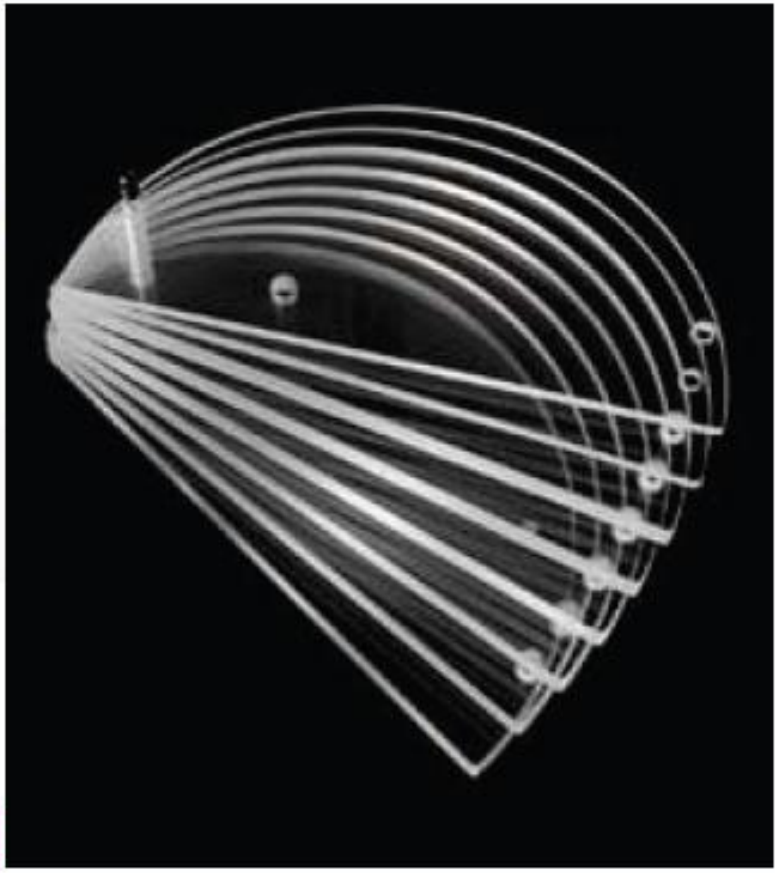
Nylon Fibers		Groups of specks		Masses	
1	ϕ = 1.56 mm	*7*	0.54 mm	*12*	ϕ = 2.00 mm
2	ϕ = 1.12 mm	*8*	0.40 mm	*13*	ϕ = 1.00 mm
3	ϕ = 0.89 mm	*9*	0.30 mm	*14*	ϕ = 0.75 mm
4	ϕ = 0.75 mm	*10*	0.24 mm	*15*	ϕ = 0.50 mm
5	ϕ = 0.54 mm	*11*	0.16 mm	*16*	ϕ = 0.25 mm
6	ϕ = 0.40 mm	-	-	-	

In US, the use of an ACR accredited phantom is mandatory. Each mammography facility needs to be accredited by an authorized accreditation body. Among the criteria, the submission of an ACR phantom image of acceptable quality is included. Furthermore, to maintain accreditation an image of the ACR phantom must be imaged weekly and these images must be recorded.

CIRS phantom



Leeds TOR MAS/TOR MAX



b)

Leeds TOR MAS/TOR MAX

- **Sensitometry**
(Ten-step grey-scale plus two points for Sensitometric measurements)

- **High Resolution limit**
(1.0 to 20.0 LP/mm)
x1 for TOR MAS
x2 for TOR MAX

- **Low Contrast Resolution**
(1.8 to 5 line pairs/mm, representing filamentary structures)

- **Low-contrast large-detail detectability**
(12 details, 5.6mm diameter)

- **High-contrast small-detail detectability**
(11 details, 0.5 and 0.25mm diameter)

- **Microcalcifications**
(Particles representing micro-calcifications on a step-wedge background, median sizes 125, 225, 325 microns)

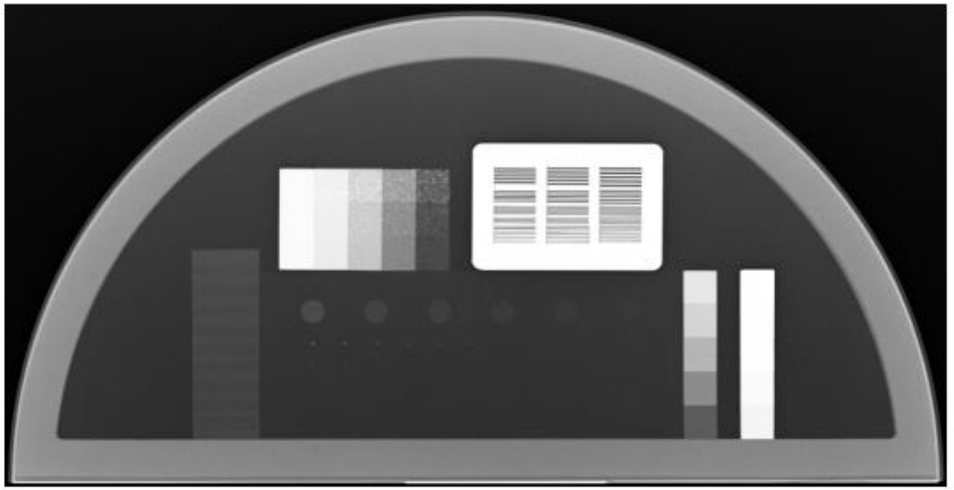


fig. 1 TOR MAS X-ray image

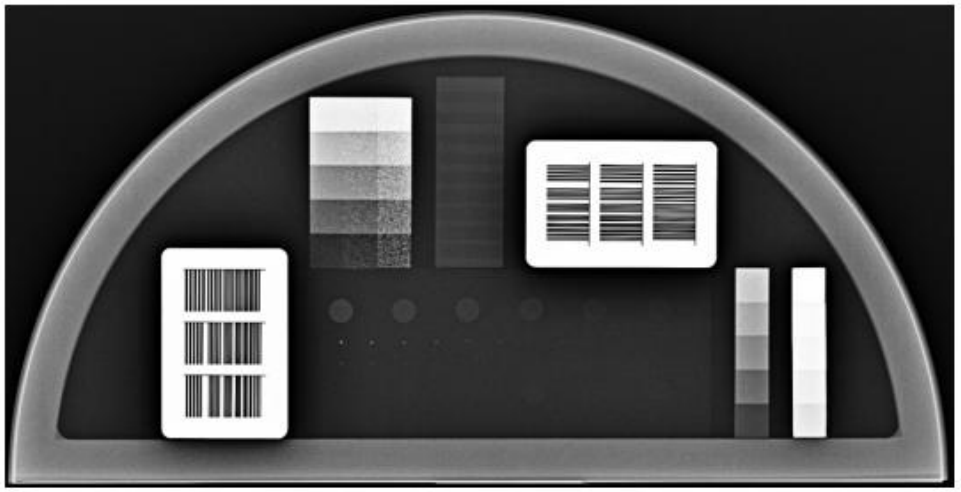
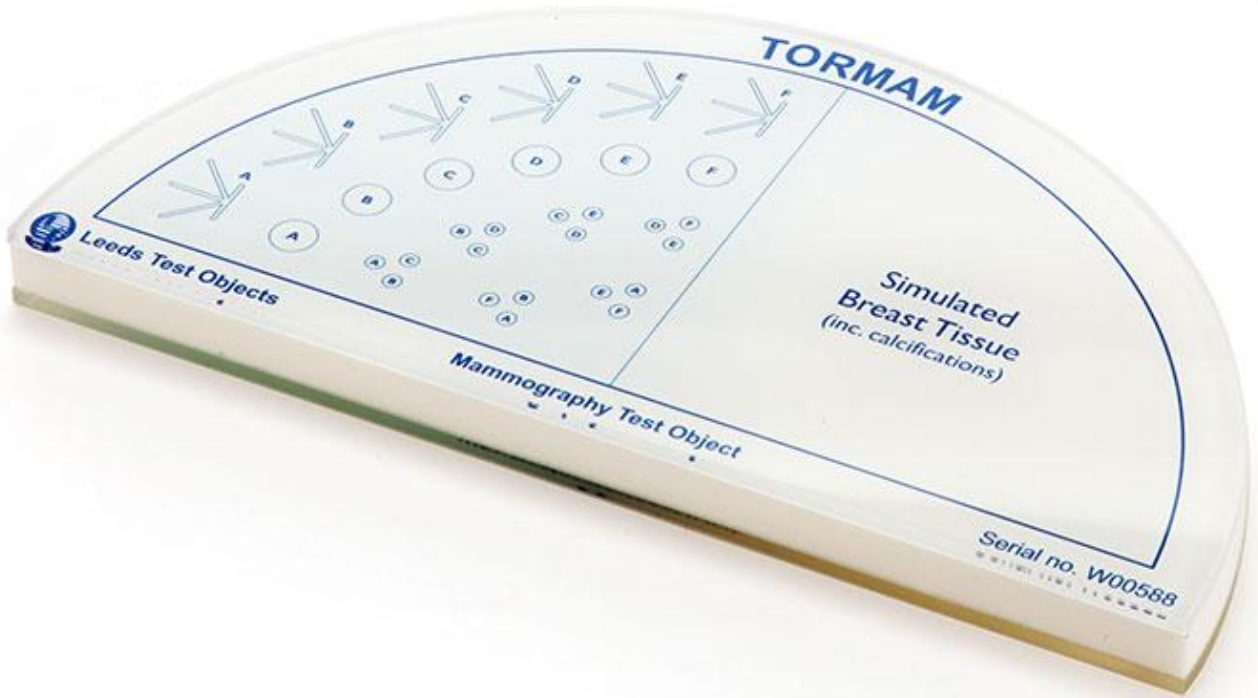
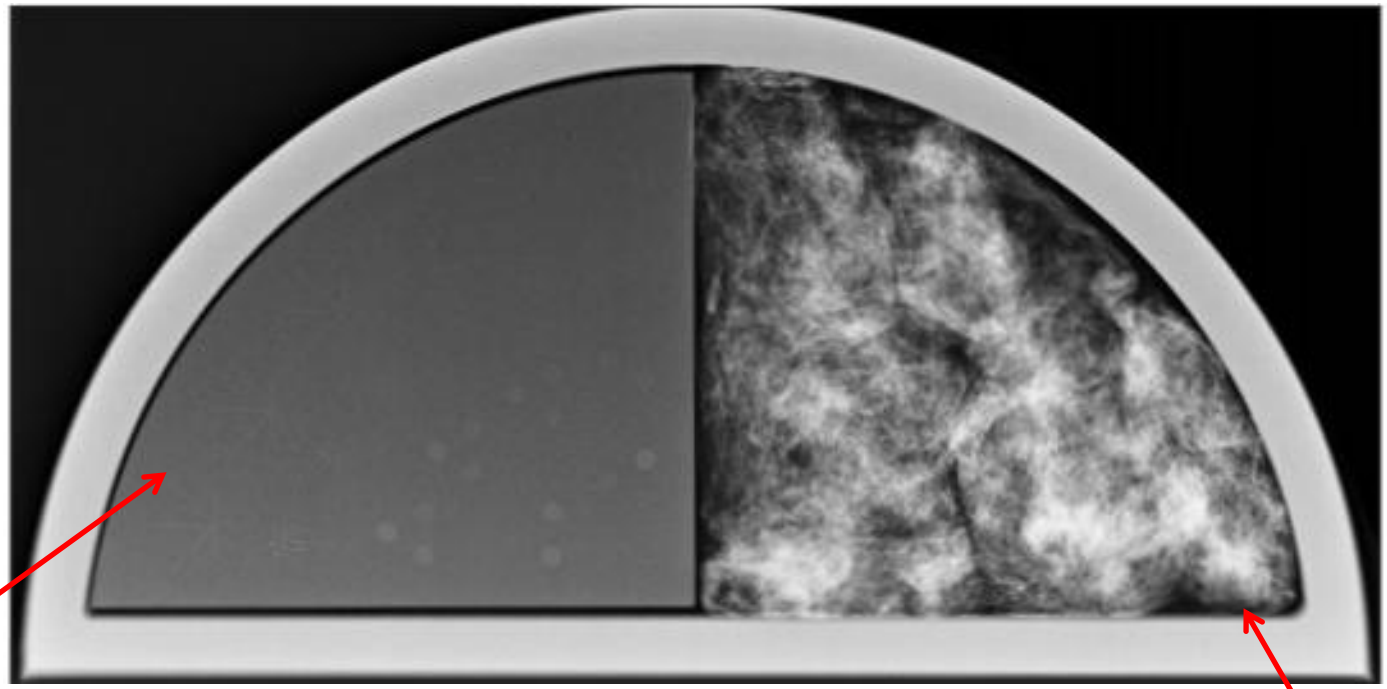


fig. 2 TOR MAX X-ray image

TOR MAM


TORMAM
Leeds Test Objects
Mammography Test Object
Simulated
Breast Tissue
(inc. calcifications)
Serial no. W00588

TOR MAM



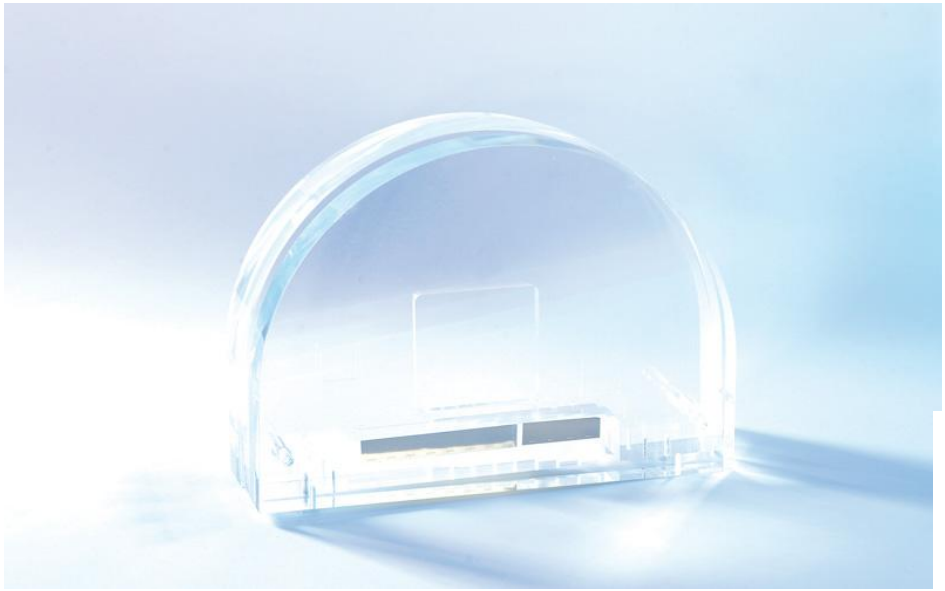
- Filaments
(6 groups of multi-directional filaments)

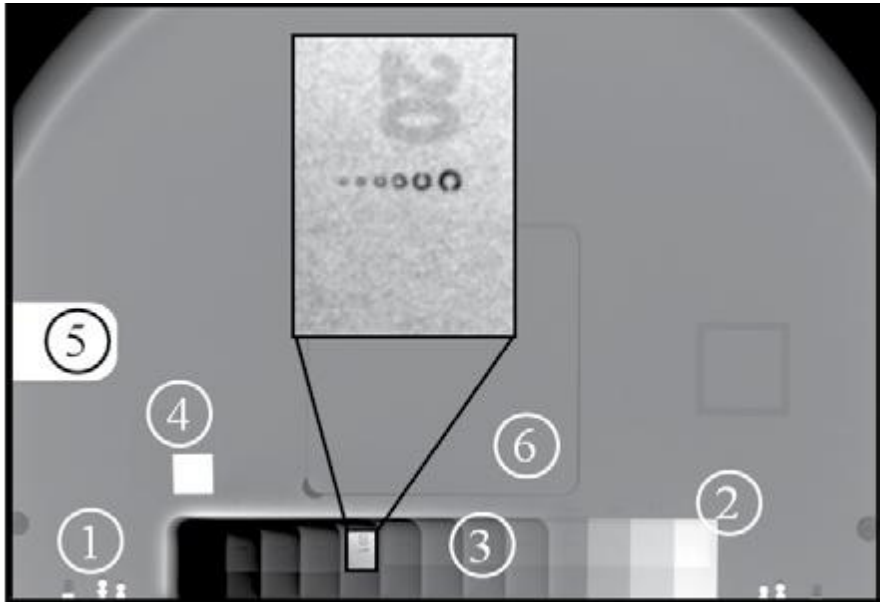
- Micro-calcifications
(6 groups of micro-calcifications in ranges of 354-224, 283-180, 226-150, 177-106, 141-90, 106-93)

- Threshold Contrast Details
(6 groups of 3, low contrast details groups)

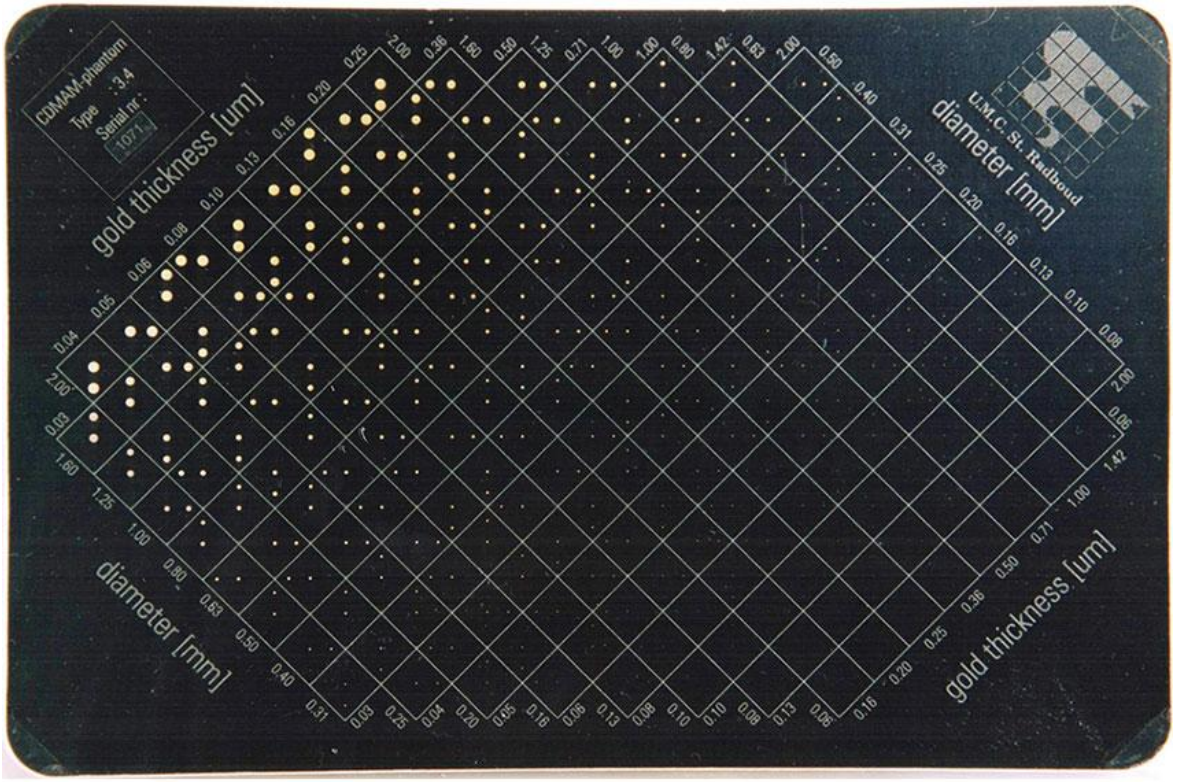
- Clincially Realistic Breast Tissue
(6 groups of micro-calcifications (as in the above) with a clinically realistic breast tissue feature)

Quart MAM/DIGI





CDMAM


CDMAM-phantom
Type : 3.4
Serial nr. : 1071
gold thickness [um]
diameter [mm]
U.M.C. St. Radboud
diameter [mm]
gold thickness [um]

CDMAM

CDMAM phantom consists of an aluminum base with gold discs of various thickness and diameter . The aluminium base is attached to a Plexiglas (PMMA) cover.

The gold discs are arranged in a matrix of 16 rows by 16 columns. Within a row the disc diameter is constant, with (partly) logarithmic increasing thickness and within a column the thickness of the discs is constant and the diameter increases logarithmic. Each square contains two identical discs (same thickness, same diameter), one in the center and one in a randomly chosen corner.

Easily recognizable patterns have been avoided. The total matrix is rotated by 45 degrees and the corners of the matrix are skipped.

Image quality evaluation

Human-based scores – **Counting "visible" details**

A common and easy image quality assessment (derived from "analog world") consists in counting which details in the phantom image are "visible", according to different "visibility" criteria.

The evaluation can be done by a certain number of observers (expert and not-expert) to consider inter-observer variability

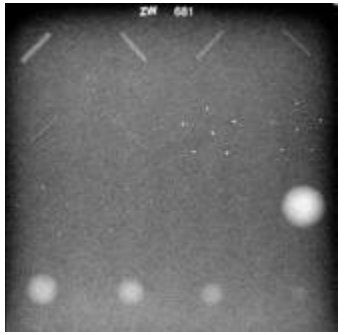
The following score can be assigned :

1	detail is well/totally visible,
0.5	detail t is not-well/partially visible,
0	detail is not visible at all

Image quality evaluation

Counting "visible" details is usually done with the ACR accreditation phantom or with the CIRS Model 011A. The sum of scores associated to all detail visibility is taken as overall IQ index, and a minimum threshold can be set for the total score and, separately, per each type of detail.

Detail type	Visibility criteria	Limiting value
Fibers	• Full=1 • Partial=0.5 • None=0	*4*
Calcs	• Five calcs=1 • 1 < calcs < 5=0.5 • None=0	*3 (groups)*
Masses	• Full=1 • Partial=0.5 • None=0	*3*



Total minimum acceptable score: 4 fibers + 3 groups of specks + 3 masses = 10 details

Image quality evaluation

Independently of the scale used to score phantom images, these human-based methods have inherent limitations.

Intra- and inter-observer variability unavoidably affects the image quality evaluation process.

Training of observers can reduce, but not eliminate the variability

The process of counting details or assessing visibility scores can be performed automatically by software.

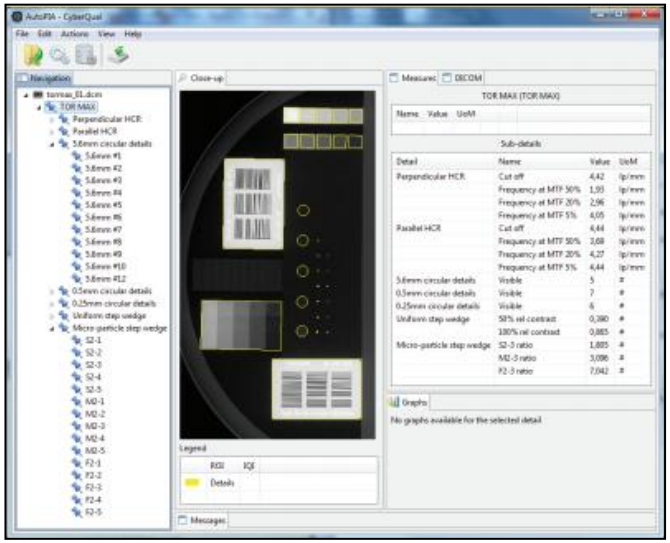
The uncertainty associated to intra- and inter-observer variability are eliminated. In general, this software do not apply observer models to mimic human visual system and often count or “see more” than the “mean observer”

CyberQual

Leeds Test Objects™
medical imaging phantoms

AutoPIA

AutoPIA (**A**utomatic **P**hantom **I**mage **A**nalysis) is a software for automatic analysis of Leeds Test Objects images used to evaluate image quality.



AutoPIA describes the algorithms used to calculate the quality indices and enables the user to verify how the quality index was calculated, for every image and every detail. The user always has control over all indices.

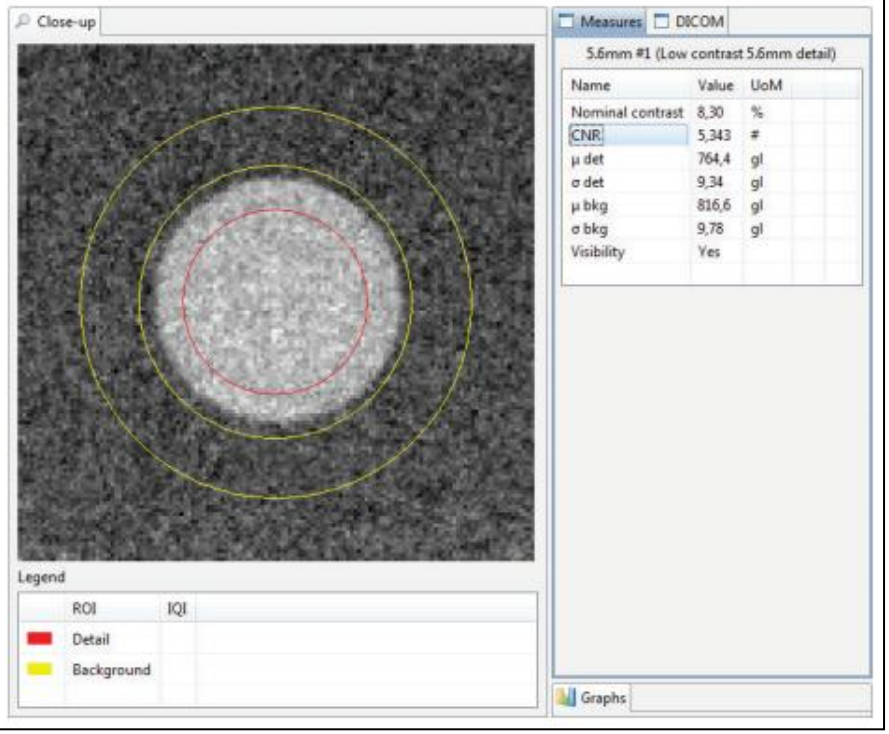
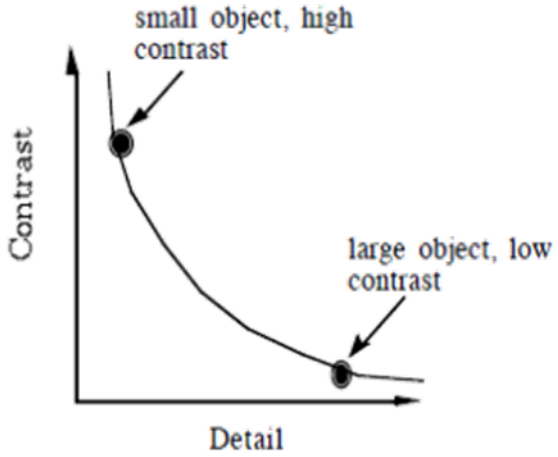


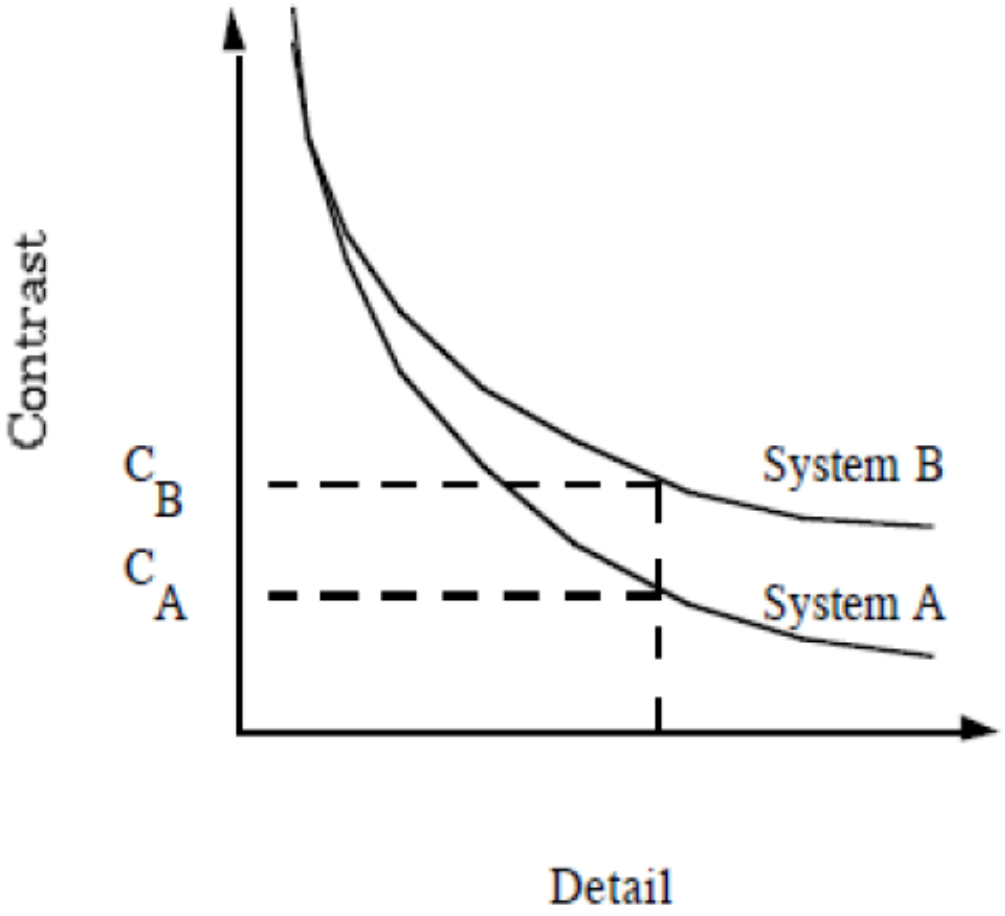
Image quality evaluation

Human-based scores – **“Visibility” threshold**

The routine assessment and control of IQ is often evaluated in terms of threshold contrast visibility of details varying in diameter and contrast.

Contrast detail curves, displaying the minimum detected contrast for each detail diameter, are commonly used for reporting.





The observer is able to distinguish a lower contrast in A system, compared to B system

Image quality evaluation

Human readings of images produced by C-D phantoms have the same limitations of intra- and inter-observer variability we mentioned for counting “visible” details methods.

Software packages have been developed in order to perform automated quantitative analysis of images produced with the CDMAM test object.
For the final presentation of the results is done using a global index (Inverse Image Quality Figure):

$$\mathbf{IQF_{inv} = 100 / \Sigma_i(Diameter_i * threshold\ thickness_i)}$$

Quantitative image quality evaluation

Digital images allow the calculation of a large number of quantitative parameters, using suitable software.
The only issue is to define appropriate IQIs for each given phantom, in order to show differences in image quality.

Quantitative IQIs are:

- objective
- reproducible
- more sensitive than human-based methods in assessing IQ differences

FOM

In assessing the performance of digital mammography systems the concept of the figure-of-merit (FOM) was introduced.

FOM are used:

- to compare techniques and exposure factors in the optimisation of mammography systems
- to compare the performance of similar digital mammography systems employing different spectra
- to compare different digital mammography systems

FOM

In the literature, the term SdNR has been used interchangeably with the CNR

Different formulation for the FOM

Samei et al. $$FOM = \frac{CNR^2}{E}$$

Kanga et al. $$FOM = \frac{CNR^2}{MGD}$$