COMPREHENSIVE MOCK EXAM 10

Supplemental to: Clinical Ultrasound Board Review 2026-2027

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ARDMS SPI (Sonography Principles and Instrumentation) Examination, Inteleos — 110 questions, single-best-answer, four options (A-D), domain-weighted per official ARDMS SPI Content Outline

Board-Review Questions

1. A sonographer is scanning a first-trimester obstetric patient and is asked by the supervising physician to minimize acoustic exposure to the embryo while still obtaining diagnostic images. Which principle should most directly guide the sonographer's technique?

A. Increase overall gain to offset reduced power

B. Use the highest frequency transducer regardless of need

C. Apply ALARA — minimize acoustic output and scan time

D. Default to spectral Doppler for all cardiac views

2. During an echocardiographic exam, the sonographer notices the Thermal Index (TI) display value increases substantially when spectral Doppler is activated compared with 2D imaging alone. What is the most accurate explanation for this observed change?

A. TI reflects only mechanical effects, not thermal

B. Longer pulse duration raises tissue heating potential

C. The TI display must be malfunctioning

D. TI is calculated only from receiver gain

3. A high-level disinfectant is required after an endocavity transducer is used for a transvaginal exam, per the Spaulding classification. What is the primary rationale for classifying this device as semicritical rather than noncritical?

A. Its footprint is smaller than an abdominal probe

B. It contacts mucous membranes, raising infection risk

C. It operates at a different frequency range

D. Semicritical status applies only to obstetric patients

4. A sonographer reports recurring wrist and shoulder discomfort after performing several vascular duplex exams per day. Which ergonomic modification is most directly supported as a preventive measure for work-related musculoskeletal disorders in sonography?

A. Increase output power to shorten scan times

B. Switch exclusively to endocavity transducers

C. Reduce monitor brightness during scanning

D. Neutral wrist position, adjustable table height, scheduled breaks

5. A patient arrives for a renal ultrasound with a written order stating only 'abdominal pain.' Before beginning the exam, what action best reflects appropriate scope of practice and clinical history correlation for the sonographer?

A. Proceed with standard protocol, no further inquiry

B. Interview the patient and correlate history with scanning

C. Render a final diagnosis directly to the patient

D. Decline the exam for lacking a diagnosis code

6. During daily quality assurance, a sonographer scans a tissue-mimicking phantom and finds that vertical and horizontal distance measurements deviate outside the department's acceptable tolerance. What is the most appropriate next step?

A. Ignore small deviations under 5 mm

B. Increase patient volume to offset downtime

C. Document only, take no further action

D. Remove system from service, report for repair

7. A new screening test for a vascular condition correctly identifies 92 of 100 patients who truly have the disease, but also incorrectly flags 15 of 100 disease-free patients as positive. Which statement correctly characterizes this test?

A. High sensitivity and high specificity

B. Low sensitivity and low specificity

C. Low sensitivity and high specificity

D. High sensitivity, comparatively lower specificity

8. Tissue Doppler Imaging (TDI) differs from conventional color Doppler primarily in which respect?

A. Displays only arterial flow, never venous or tissue

B. Optimized for low-velocity, high-amplitude myocardial motion

C. Requires no wall filter adjustment at all

D. Cannot be color-encoded, spectral display only

9. The mechanism by which ultrasound contrast agents enhance signal is best described as:

A. Direct absorption of the beam by a dissolved dye

B. Selective filtering of low-frequency harmonics before reception

C. Gas microbubbles producing strong nonlinear harmonic backscatter

D. Cooling of insonated tissue, raising propagation speed

10. Elastography is used clinically to assess which tissue property?

A. Relative tissue stiffness via displacement or shear-wave speed

B. Absolute blood flow velocity within a vessel

C. Tissue attenuation coefficient at a fixed frequency

D. Precise anatomic distance between two structures

11. Fusion (hybrid) imaging, which co-registers a real-time ultrasound image with a previously acquired CT or MRI dataset, is most clinically useful for which purpose?

A. Eliminating the need for any acoustic output

B. Automatically calculating the Thermal Index

C. Guiding biopsy/ablation by correlating with prior CT/MRI

D. Replacing the need for patient positioning

12. A transducer operates at a frequency of 5 MHz. Assuming the standard soft-tissue propagation speed of 1.54 mm/microsecond, what is the approximate wavelength?

A. 0.15 mm

B. 0.62 mm

C. 0.31 mm

D. 1.54 mm

13. As an ultrasound beam travels through tissue, its intensity decreases due to absorption, scattering, and reflection combined. What is the general term for this total intensity loss with distance?

A. Attenuation

B. Amplification

C. Rarefaction

D. Refraction

14. Axial resolution refers to a system's ability to distinguish two closely spaced reflectors when they are positioned:

A. Side by side, perpendicular to beam travel

B. At different depths, different frames

C. Along the beam axis, one behind the other

D. Same depth, different scan planes

15. Axial resolution is most directly determined by which parameter?

A. The imaging frame rate value

B. The physical transducer diameter

C. The selected focal zone depth

D. One-half the spatial pulse length

16. Lateral resolution is generally best (i.e., numerically smallest/finest) at which location in the ultrasound beam?

A. At the transducer face itself

B. At the focal zone, the narrowest point

C. In the far field beyond focus

D. It does not vary with depth

17. A sonographer increases the number of focal zones from one to four during 2D imaging to improve lateral resolution over a longer depth range. What is the expected trade-off?

A. Improved frame rate

B. Decreased frame rate, more pulses needed per frame

C. No change, focal zones don't affect pulse count

D. Improved axial resolution, frame rate unchanged

18. Duty factor in pulsed ultrasound is best defined as:

A. The number of pulses transmitted per second

B. The physical length of the listening region

C. Fraction of time actively transmitting vs. total period

D. Percentage of echoes successfully processed

19. In conventional pulsed B-mode imaging, the typical duty factor is approximately:

A. Approximately 50%

B. Approximately 25%

C. Approximately 10%

D. Less than 1%

20. An ultrasound beam traveling through tissue encounters an oblique interface between two media with different propagation speeds. What physical phenomenon governs the resulting change in the beam's direction of travel?

A. An attenuation effect

B. An absorption effect

C. A reverberation effect

D. Refraction (Snell's law)

21. Ultrasound systems calculate depth and distance assuming a constant propagation speed of 1540 m/s in all soft tissue. When the beam passes through a region of fat, which propagates sound more slowly than the assumed average, what type of error results?

A. No error, the system auto-compensates by tissue type

B. Propagation speed error, deeper structures misplaced

C. Error only in color Doppler, never B-mode

D. Refraction-only artifact, no effect on depth

22. Acoustic impedance of a tissue is calculated as the product of which two properties?

A. Tissue density and propagation speed

B. Frequency and wavelength

C. Attenuation coefficient and depth

D. Amplitude and power

23. The strength of a reflected echo at a tissue interface is primarily determined by:

A. The absolute depth of the interface

B. The transducer's operating frequency alone

C. The acoustic impedance mismatch at the interface

D. The number of focal zones selected

24. Compared to soft tissue, cortical bone and calcified structures produce markedly different acoustic behavior. Which statement correctly describes this difference?

A. Bone attenuates less, allowing excellent through-transmission

B. Bone attenuates far more, often producing shadowing

C. Bone and fluid attenuate identically

D. Calcification always produces enhancement, not shadowing

25. A sonographer notes multiple equally spaced, parallel bright echoes extending deep to a strongly reflective structure, such as a metallic surgical clip. What artifact is this?

A. Reverberation between two reflectors

B. A refraction artifact type

C. A range ambiguity artifact

D. A diffuse speckle pattern

26. A specific subtype of reverberation artifact, seen as a tapering, solid bright line deep to a small, strongly reflective structure like a metal fragment or air bubble, is called:

A. Comet-tail artifact

B. Mirror-image artifact

C. Grating lobe artifact

D. Twinkle artifact

27. Frame rate, the number of complete B-mode images displayed per second, is most directly limited by which combination of factors?

A. Transducer footprint size only

B. Depth, sector width/scan lines, and focal zones

C. Overall gain setting only

D. Body habitus alone, independent of settings

28. A sonographer must choose between a 3 MHz and a 9 MHz transducer for an exam. Which statement correctly describes the expected trade-off?

A. 9 MHz penetrates deeper but has lower resolution

B. Frequency affects neither penetration nor resolution

C. 3 MHz gives better resolution, poorer penetration

D. 9 MHz gives better resolution, poorer penetration

29. The piezoelectric effect, fundamental to ultrasound transducer function, refers to:

A. Absorption of sound energy converted directly into heat

B. Magnetic alignment of crystal domains in a field

C. Crystals generating voltage when deformed, and vice versa

D. A chemical reaction between gel and crystal surface

30. During routine preventive maintenance, a sonographer performs an element (crystal) integrity test on a linear array transducer and identifies several nonfunctioning elements. What clinical image artifact would this defect most likely produce?

A. An increased frame rate

B. An improved lateral resolution

C. A vertical dropout band artifact

D. Uniformly increased overall brightness

31. A curvilinear (convex) array transducer is typically the preferred choice for which type of examination?

A. General abdominal imaging, wide field of view

B. Cardiac imaging via narrow intercostal window

C. Superficial vascular access needing high near-field resolution

D. Transvaginal endocavity imaging

32. Phased array (sector) transducers are the preferred choice for adult transthoracic echocardiography primarily because:

A. The widest surface-level field of view

B. They operate only in M-mode display

C. Small footprint between ribs, steers electronically

D. They cannot be used with Doppler

33. When selecting an endocavity transducer for a transvaginal exam, which frequency range is generally most appropriate compared with a standard abdominal transducer?

A. Substantially lower, to penetrate the abdominal wall

B. Frequency selection is irrelevant here

C. Higher, since deep penetration is not needed

D. The same exact frequency, always

34. A 2D matrix array transducer, used for real-time volumetric (3D/4D) cardiac imaging, differs from a standard 1D linear or phased array primarily by:

A. A 2D element grid, steering in azimuth and elevation

B. Using one single very large crystal element

C. Eliminating any backing material

D. Having no electronic beam steering at all

35. The backing (damping) material behind the piezoelectric element in a transducer serves primarily to:

A. Increase the transducer's operating frequency

B. Electronically amplify the returning echo signal

C. Absorb backward energy, shorten ringing, aid resolution

D. Match impedance of the transducer to air

36. The matching layer, positioned between the piezoelectric element and the patient's skin, serves primarily to:

A. Absorb backward-directed acoustic energy

B. Generate the electrical pulse driving the crystal

C. Shorten the crystal's ring-down time

D. Bridge impedance between crystal and skin

37. A continuous-wave (CW), nonimaging 'pencil' Doppler probe is used clinically in which situation?

A. Standard grayscale anatomic B-mode imaging

B. Volumetric 3D/4D acquisition

C. Measuring very high-velocity jets, e.g. stenosis

D. Superficial soft-tissue elastography

38. A '1.5D' array transducer is best described as:

A. Identical to a standard 1D linear array

B. Elements in a single row, no elevational rows

C. Limited elevational subdivision beyond a fixed 1D array

D. A synonym for a full 2D matrix array

39. The thickness of a piezoelectric crystal element is the primary determinant of which transducer characteristic?

A. The transducer's field of view width

B. Center (resonant) frequency — thinner is higher

C. The number of focal zones available

D. Color Doppler scale setting

40. Bandwidth, in the context of transducer performance, refers to:

A. The physical width of the transducer footprint

B. The maximum depth of penetration achievable

C. The number of piezoelectric elements in the array

D. The range of frequencies within the transmitted pulse

41. A modern broadband (wide-bandwidth) transducer allows the sonographer to electronically select different operating frequencies (frequency-agile scanning) from the same physical probe. What is the primary clinical benefit of this capability?

A. Shift lower for penetration or higher for resolution, one probe

B. It eliminates the need for any focal zones

C. It automatically corrects propagation speed error

D. It removes the need for a matching layer

42. Harmonic imaging exploits which underlying physical phenomenon?

A. Nonlinear propagation generating harmonic frequencies with depth

B. Linear, undistorted propagation at the transmitted frequency

C. The piezoelectric effect occurring twice per pulse

D. Reflection exclusively at air-tissue interfaces

43. Compared with fundamental (conventional) imaging, tissue harmonic imaging generally provides which clinical benefit?

A. Worse contrast resolution in difficult patients

B. No difference in image quality

C. Better contrast, less near-field clutter/artifact

D. Elimination of the need for gain adjustment

44. The reduction in near-field artifact seen with harmonic imaging is primarily explained by which fact about harmonic signal generation?

A. Harmonic signal builds with depth, weak near the skin

B. A completely different transducer with no near field

C. Harmonic frequencies are strongest at the skin surface

D. The receiver is disabled for the first few cm

45. A transducer with a wide bandwidth (broadband, short spatial pulse length) compared with a narrowband transducer of the same center frequency will generally provide:

A. Worse axial resolution, more frequencies present

B. Better axial resolution, shorter pulse length

C. No change, resolution depends only on frequency

D. Improved lateral, but not axial, resolution

46. As a sonographer increases the imaging depth setting on the console, the system automatically adjusts pulse repetition frequency (PRF). What is the relationship?

A. PRF automatically decreases as depth increases

B. PRF automatically increases as depth increases

C. PRF is unaffected by depth setting

D. PRF is manual and never adjusts with depth

47. Time Gain Compensation (TGC), sometimes called depth gain compensation, functions by:

A. Depth-selective amplification, compensating deep attenuation

B. Uniformly amplifying every echo, regardless of depth

C. Reducing transmit power at deep focal zones

D. Automatically adjusting frequency with depth

48. Overall (system) gain, as distinct from TGC, functions by:

A. Amplifying only echoes from a specific depth range

B. Uniformly amplifying all echoes, regardless of depth

C. Increasing transmit power only, not the receiver

D. Adjusting focal zone position

49. A sonographer sets the overall gain excessively high while imaging the liver. What artifact is most likely to result?

A. Saturation, obscuring subtle grayscale texture

B. Complete loss of all signal, a black screen

C. Improved contrast resolution

D. Decreased frame rate

50. The number and position of focal zones selected during 2D imaging directly affects lateral resolution and, as previously noted, frame rate. Which additional trade-off should the sonographer weigh when using multiple focal zones for a deep structure?

A. Multiple zones eliminate the need for TGC

B. Multiple zones always improve axial resolution

C. Better lateral resolution but slower frame rate, motion blur risk

D. Multiple zones have no image-quality trade-off

51. 'Write zoom,' applied before an image is frozen, differs from 'read zoom,' applied after freezing, in which key respect?

A. Write zoom re-scans at higher line density, true resolution gain

B. Write and read zoom are functionally identical

C. Read zoom improves resolution, write zoom does not

D. Write zoom works only in color Doppler mode

52. M-mode ('motion mode') display is best described as:

A. A 3D volumetric rendering updated in real time

B. A color map of flow velocity over the 2D field

C. A frequency-domain plot of Doppler shift vs. time

D. One scan line: depth on one axis, time on the other

53. Dynamic range in ultrasound imaging refers to:

A. The physical range of transducer frequencies available

B. The maximum depth of penetration achievable

C. The total number of focal zones selectable

D. The ratio of largest to smallest displayable echo amplitude

54. Narrowing the dynamic range setting on the display produces what visual effect on the image?

A. A smoother gradient with many subtle gray shades

B. No visible change to the image

C. Higher contrast, more black-and-white appearance

D. Improved penetration depth

55. Edge enhancement, a postprocessing feature available on most ultrasound systems, functions by:

A. Increasing contrast at echogenicity boundaries

B. Uniformly increasing brightness of the entire image

C. Reducing speckle throughout the image

D. Increasing penetration depth at deep boundaries

56. Persistence (frame averaging) improves image appearance primarily by:

A. Averaging frames, smoothing noise, some motion blur cost

B. Increasing frame rate substantially

C. Sharpening true edges more than any other tool

D. Eliminating the need for TGC

57. Spatial compounding forms a single displayed image by:

A. Averaging sequential frames from the same beam angle

B. Progressively increasing transmit frequency across the field

C. Combining frames from multiple beam angles into one

D. Applying only edge enhancement to a single frame

58. A recognized clinical benefit of spatial compound imaging is:

A. Increased speckle throughout the image

B. Worsened border definition on curved structures

C. Reduced angle-dependent artifacts and speckle

D. Elimination of the need for focal zones

59. Extended field of view (panoramic) imaging allows the sonographer to:

A. Acquire a true 3D volumetric dataset in one sweep

B. Stitch sequential frames into one elongated static image

C. Automatically increase penetration beyond system limits

D. Eliminate the need for a linear array transducer

60. Coded excitation is a transmit technique in which the system:

A. Sends a single-cycle uncoded pulse for axial resolution

B. Removes the need for a matching layer

C. Randomly varies transmit frequency, no defined purpose

D. Sends a longer encoded pulse, then decodes/compresses on receive

61. Real-time three-dimensional (3D) ultrasound acquisition, using a 2D matrix array, differs from earlier freehand or mechanical 3D methods primarily by:

A. Manual probe sweeping with a tracked position sensor

B. Acquiring only a single static 2D plane

C. Electronic beam steering through a volume, no probe motion

D. Being available only in Doppler mode

62. 4D ultrasound is best understood as:

A. A synonym for 2D imaging with color Doppler added

B. A single static 3D volume, no added information

C. Real-time, continuously updating 3D — time as the 4th dimension

D. A mode limited exclusively to fetal facial imaging

63. PACS (Picture Archiving and Communication System) primarily serves which function in a sonography department?

A. Performing the actual image acquisition at the bedside

B. Digitally storing, retrieving, and transmitting images

C. Calibrating transducer element integrity

D. Automatically generating the sonographer's clinical impression

64. Proper examination documentation standards in sonography require which practice regarding stored images?

A. Storing only the single best image of the exam

B. Omitting patient identifiers to protect privacy

C. Documenting measurements only, no images required

D. Accurate labeling: identifiers, laterality, anatomy, measurements

65. In conventional grayscale B-mode imaging, the brightness of a given pixel primarily corresponds to:

A. The distance of that reflector from the transducer

B. The amplitude of the returning echo, after processing

C. The propagation speed of sound at that location

D. The transducer's operating frequency

66. Range ambiguity artifact occurs when:

A. A strong reflector produces a bright comet-tail

B. Two adjacent scan lines interfere destructively

C. PRF too high for depth; deep echoes misplaced shallow

D. The system auto-compensates, so this cannot occur

67. Grating lobe artifact arises from which underlying cause?

A. Periodic array-element spacing creating secondary beams

B. Excessive overall system gain setting

C. A malfunctioning single crystal element

D. An improper TGC curve setting

68. Side lobes, distinct from grating lobes, are best described as:

A. Restricted to array transducers with periodic spacing

B. Present exclusively in Doppler spectral display

C. Off-axis secondary beams, even from a single element

D. A form of reverberation between transducer and skin

69. Mirror-image artifact is classically seen deep to which strongly reflective, curved structure?

A. The diaphragm, duplicating a nearby structure

B. A distended urinary bladder only

C. Only the endometrial stripe region

D. A superficial lymph node

70. Posterior acoustic enhancement is characteristically seen deep to which type of structure?

A. Dense cortical bone tissue

B. A simple fluid-filled cyst

C. A calcified gallstone structure

D. Bowel gas collection

71. Bowel gas classically produces which sonographic appearance?

A. Clean posterior shadowing, sharply defined, like bone

B. Posterior acoustic enhancement

C. 'Dirty shadowing' — indistinct low-level reverberation

D. No effect on the ultrasound beam whatsoever

72. A curved fluid-filled structure, such as the wall of a full urinary bladder or a cyst, may produce a specific artifact at its lateral margins due to the beam obliquely striking the curved surface. What is this artifact called?

A. Posterior acoustic enhancement

B. A comet-tail artifact

C. Refraction shadowing at margins

D. Grating lobe artifact

73. The fine, granular texture seen throughout solid organ parenchyma on ultrasound, which does not correspond to true individual anatomic structures, is called:

A. A true reverberation artifact

B. A grating lobe artifact

C. A range ambiguity artifact

D. Speckle, from scatterer interference

74. In spatial compound imaging, combining multiple angled frames into a single composite image generally has which effect on frame rate compared with conventional (non-compound) imaging?

A. Frame rate increases substantially

B. Frame rate is unaffected

C. Only affected in Doppler modes

D. Frame rate decreases before full display

75. Beyond reducing speckle, spatial compound imaging's multi-angle acquisition specifically helps reduce which additional category of artifact, since such artifacts tend to be angle-dependent and therefore diminish when views from several angles are averaged together?

A. Patient motion artifact from breathing

B. Electronic interference from nearby equipment

C. Angle-dependent refraction shadowing and reverberation

D. Aliasing on spectral Doppler

76. A sonographer notices excessive noise/clutter obscuring detail specifically in the far field of a difficult-to-image patient, while the near field looks acceptable. Which combination of adjustments is the most appropriate first troubleshooting step?

A. Decrease overall gain uniformly across the image

B. Increase far-field TGC, adjust persistence, depth-specific

C. Switch to a higher-frequency transducer

D. Reduce the number of focal zones to zero

77. The Doppler equation used to calculate blood flow velocity includes which variables?

A. Vessel diameter, blood viscosity, cardiac output

B. Transducer footprint size and patient BMI

C. Doppler shift, transmit frequency, speed, cosine of angle

D. Only transmitted frequency, no angle term needed

78. As the Doppler angle approaches 90 degrees (beam perpendicular to flow), what happens to the calculated Doppler shift and the resulting velocity measurement?

A. The shift approaches zero, cosine(90°) = 0

B. The Doppler shift becomes maximal

C. The angle has no effect on the calculated shift

D. Velocity is automatically corrected regardless of angle

79. For accurate spectral Doppler velocity measurement, professional guidelines generally recommend keeping the Doppler angle:

A. As close to 90 degrees as possible

B. Irrelevant, since modern systems eliminate angle-dependence

C. At or below about 60 degrees, ideally near 0

D. Always exactly 45 degrees for consistency

80. The pulse repetition frequency (PRF) directly sets the upper limit of a pulsed Doppler system's ability to unambiguously measure velocity, a limit known as:

A. The reflection coefficient

B. The attenuation limit

C. The acoustic impedance boundary

D. The Nyquist limit, one-half the PRF

81. Aliasing on pulsed (spectral or color) Doppler occurs when:

A. The Doppler angle exceeds 60 degrees

B. The wall filter is set too low

C. Overall gain is set too high

D. True shift exceeds the Nyquist limit (PRF/2)

82. Which combination of adjustments would most directly help resolve aliasing on a pulsed Doppler spectral waveform?

A. Decrease the Doppler angle to 0, nothing else

B. Raise scale/PRF, shift baseline, and/or lower frequency

C. Decrease overall gain

D. Switch off spectral Doppler, aliasing is unfixable

83. Continuous-wave (CW) Doppler, compared with pulsed-wave (PW) Doppler, has which key characteristic regarding aliasing and range resolution?

A. CW aliases at high velocities, exactly like PW

B. CW cannot alias, but has no range/depth resolution

C. CW has excellent range resolution but aliases easily

D. CW and PW are functionally identical in every way

84. Pulsed-wave (PW) Doppler achieves depth-specific measurement — unlike CW Doppler — through which mechanism?

A. Use of two separate, continuously active crystals

B. Random sampling of the entire beam path at once

C. Filtering based solely on echo amplitude, not timing

D. Range (depth) gating of the receiver window

85. The wall filter (high-pass filter) in Doppler processing serves to:

A. Amplify the weakest Doppler signals uniformly

B. Increase the Nyquist limit

C. Remove low-frequency, high-amplitude wall-motion clutter

D. Correct for an inappropriate Doppler angle

86. A sonographer sets the wall filter too high while attempting to assess slow venous flow. What is the likely consequence?

A. Improved visualization of the slow venous signal

B. No effect, wall filter affects only arterial signals

C. Genuine slow venous signal is filtered out too

D. The venous waveform will appear aliased

87. A low-resistance Doppler waveform, characterized by continuous forward flow throughout diastole, is classically expected in which vascular bed under normal conditions?

A. The normal external carotid artery

B. A resting, unstimulated extremity muscle artery

C. A renal transplant artery in acute rejection

D. The internal carotid artery, feeding the brain

88. A high-resistance, triphasic Doppler waveform (forward systolic peak, brief early-diastolic reversal, then a small late-diastolic forward component) is the normal expected pattern in which artery?

A. The internal carotid artery

B. The normal renal artery

C. The normal hepatic artery

D. A resting peripheral artery

89. The Resistive Index (RI), a common Doppler waveform measurement, is calculated as:

A. Diastolic velocity divided by systolic velocity

B. Mean velocity divided by peak systolic velocity

C. (Systolic minus diastolic velocity) divided by systolic velocity

D. Peak systolic velocity divided by vessel diameter

90. In color Doppler imaging, the displayed color at each pixel primarily encodes:

A. The exact anatomic tissue type present

B. Mean Doppler shift/velocity and direction, via color map

C. The acoustic impedance of the tissue

D. The attenuation coefficient at that depth

91. The standard color Doppler directional convention (often remembered by the mnemonic BART) states that:

A. Blue is always arterial, red always venous

B. Color assignment is entirely random by system

C. Red is always away, blue always toward, no exceptions

D. Blue Away, Red Toward — default map, reversible by the sonographer

92. Increasing the size of the color Doppler box (region of interest) while keeping other settings constant typically has what effect on color frame rate?

A. Color frame rate increases

B. Color frame rate decreases, more lines/pulses needed

C. Color frame rate is unaffected by box size

D. Color frame rate becomes infinite

93. 'Packet size' (also called ensemble length) in color Doppler refers to:

A. The physical size of the region of interest box

B. The number of piezoelectric elements per pulse

C. The maximum depth of the color box

D. Pulses per scan line; more improves accuracy, cuts frame rate

94. Power Doppler, compared with conventional (velocity-based) color Doppler, displays which parameter?

A. Signal amplitude (power), sensitive to slow/low-volume flow

B. Mean velocity only, no change from color Doppler

C. Flow direction only, same as conventional color

D. Tissue stiffness, similar to elastography

95. A key clinical advantage of power Doppler over conventional color Doppler is:

A. Power Doppler shows flow direction more accurately

B. Power Doppler is faster to acquire in every case

C. Power Doppler is less sensitive to slow flow

D. No aliasing, angle-independent, sensitive to slow flow

96. 'Flash artifact' on color or power Doppler refers to:

A. A persistent true vascular finding, unrelated to motion

B. A calibration error requiring service

C. A diffuse burst of color from motion, not true flow

D. An artifact seen only with spectral Doppler

97. 'Blooming' artifact on color Doppler is best described as:

A. A total absence of color signal in a patent vessel

B. An artifact unique to spectral Doppler tracing

C. Excess gain/power causing color to overspill true vessel size

D. A form of aliasing exclusive to CW Doppler

98. To optimize color Doppler gain for a given exam, the recommended technique is generally to:

A. Raise gain until background noise appears, then back off

B. Set color gain to the system's maximum, every exam

C. Set color gain to the system's minimum, every exam

D. Adjust only B-mode gain; color has no own setting

99. When assessing slow, low-velocity venous flow with color Doppler, which combination of settings is generally most appropriate?

A. High velocity scale/PRF and a high wall filter

B. Low velocity scale/PRF and a low wall filter

C. Maximum transmit power regardless of thermal limits

D. CW Doppler exclusively; PW cannot detect veins

100. CW Doppler's inability to localize the depth of a signal along the beam can, in certain circumstances, create ambiguity about which of two overlapping vessels a given signal originates from. This limitation is sometimes referred to as:

A. A Nyquist ambiguity

B. Range ambiguity of CW Doppler

C. A true aliasing effect

D. A spectral broadening effect

101. Spectral broadening, seen as a filling-in of the normally clear 'window' beneath the spectral Doppler waveform envelope, is most classically associated with:

A. Turbulent flow, e.g. distal to a significant stenosis

B. Perfectly laminar flow in a normal vessel

C. An excessively low wall filter setting

D. A Doppler angle of exactly 0 degrees

102. The size (length) of the pulsed Doppler sample volume along the beam axis is generally adjusted according to which principle?

A. Always set to maximum length regardless of vessel size

B. It has no effect on measurement specificity

C. Sized to the lumen for specificity; oversized risks adjacent signal

D. Must always equal the full length of the color box

103. According to the Doppler equation, if transmitted frequency, propagation speed, and Doppler angle are all held constant, the relationship between true blood flow velocity and the resulting Doppler frequency shift is:

A. Directly proportional

B. Inversely proportional

C. No mathematical relationship exists

D. Exponential, not linear

104. For a fixed true blood flow velocity and Doppler angle, using a higher transmitted (transducer) frequency has what effect on the resulting Doppler shift and the likelihood of aliasing?

A. Smaller shift, reduced aliasing risk

B. Larger shift; Nyquist unchanged, so more aliasing risk

C. No effect on the resulting Doppler shift

D. Eliminates the possibility of aliasing entirely

105. For evaluating a very high-velocity jet, such as through a severely stenotic cardiac valve, which Doppler modality is generally preferred to obtain an accurate peak velocity measurement without aliasing?

A. Color Doppler alone

B. Standard PW Doppler with a small sample volume

C. Power Doppler alone

D. CW Doppler; no Nyquist limit, cannot alias

106. On a spectral Doppler tracing, aliasing characteristically appears as:

A. Complete absence of any waveform

B. A smooth, symmetric waveform, no distortion

C. A gradual linear rise in baseline noise only

D. Waveform peak cut off, wrapping to the opposite side

107. The Doppler 'reject' control, distinct from the wall filter, functions by:

A. Adjusting the Nyquist limit directly

B. Changing the transmitted frequency of the Doppler pulse

C. An amplitude threshold, unlike the filter's frequency threshold

D. Automatically correcting the Doppler angle

108. When positioning the pulsed Doppler sample volume within a vessel lumen to assess for a possible focal area of turbulence or a jet (such as at a suspected stenosis), the sonographer should generally:

A. Always place the sample volume against the wall

B. Place it outside the lumen for a baseline reading

C. Center it in the lumen, best captures peak jet velocity

D. Use the largest possible sample volume covering everything

109. The color Doppler 'variance map' display mode, an alternative to the standard velocity/direction color map, is used to depict:

A. Vessel wall thickness alone

B. Variation in velocity/direction (turbulence)

C. Tissue attenuation coefficient alone

D. Absolute peak systolic velocity

110. A sonographer obtains a spectral Doppler waveform from a vessel and notes the tracing appears symmetric above and below the baseline in a regular, repeating sawtooth pattern despite genuinely unidirectional true flow. Which explanation is most consistent with this appearance?

A. True bidirectional flow, alternating each cardiac cycle

B. Expected normal venous flow, no further consideration

C. Aliasing — peak exceeds Nyquist, wraps around baseline

D. Transducer hardware failure requiring replacement

Answers

1. C. C. ALARA (As Low As Reasonably Achievable) is the governing safety principle in diagnostic ultrasound: output power, dwell time, and use of higher-MI modes (like spectral or color Doppler) should all be minimized to whatever level still produces a diagnostic image, especially in first-trimester obstetric scanning where thermal sensitivity is highest. Raising gain (A) is a receiver-side adjustment and does not reduce embryonic exposure. Maximizing frequency (B) is not itself an ALARA control and trades away penetration unnecessarily. Doppler modes (D) generally raise, not lower, acoustic output and dwell time, so defaulting to them would work against ALARA rather than support it.

2. B. B. Thermal Index estimates the potential for tissue heating and rises with duty factor and dwell time at a fixed location — both of which are characteristically higher in spectral (PW) Doppler than in 2D B-mode imaging, where the beam sweeps continuously across the field rather than dwelling on one line. This is a real, expected physical difference, not a malfunction (C). TI reflects transmit-side acoustic output and exposure time, not receiver gain (D), and it specifically estimates thermal — not mechanical

— bioeffect potential, which is the Mechanical Index's role, not something TI is defined as excluding by mode (A).

3. B. B. The Spaulding classification sorts devices by infection risk based on what tissue they contact: semicritical devices contact mucous membranes or non-intact skin (as an endocavity probe does against vaginal mucosa) and require high-level disinfection, while noncritical devices contact only intact skin (like an abdominal probe) and need only low-level disinfection. Footprint size (A) and operating frequency (C) are unrelated to the Spaulding risk tier. The classification applies to the tissue contacted in any patient, not exclusively obstetric use (D).

4. D. D. Sonographer musculoskeletal injury is strongly associated with repetitive, sustained awkward postures — particularly wrist flexion/extension, shoulder abduction, and prolonged static gripping of the transducer. Ergonomic mitigation focuses on neutral joint positioning, adjustable equipment height, transducer grip technique, and scheduled rest breaks. Acoustic output (A) and monitor brightness (C) have no bearing on musculoskeletal strain, and endocavity transducers (B) are not a general ergonomic substitute — the correct probe is chosen by exam indication, not injury prevention.

5. B. B. Sonographers are expected to obtain a focused clinical history and tailor technique and additional imaging (e.g., extending the exam toward tenderness or a palpable finding) accordingly — this falls squarely within scope of practice and directly improves diagnostic yield. Diagnosing and communicating findings to the patient (C) exceeds sonographer scope, which is the interpreting physician's role. Proceeding with zero history-taking (A) forgoes information that legitimately shapes the exam, and declining for an incomplete diagnosis code (D) is not a recognized reason to withhold a clinically ordered study.

6. D. D. A tissue-mimicking phantom is used specifically to verify that a system's distance/depth calibration and resolution remain within specification; a deviation beyond tolerance indicates the equipment may be producing inaccurate measurements and should be taken out of clinical service and referred for service/calibration before further patient use. Documenting without action (C) or arbitrarily tolerating an out-of-spec result (A) both risk continued use of a miscalibrated system, and increasing patient volume on a failing system (B) is directly contrary to the purpose of QA testing.

7. D. D. Sensitivity is the proportion of true-positive patients correctly identified (92/100 = 92%, high), while specificity is the proportion of disease-free patients correctly identified as negative — here 85/100 = 85% specificity, since 15 were false positives. That combination is high sensitivity with comparatively lower specificity, not uniformly high (A), uniformly low (B), or inverted (C, which would require missing true-positive cases rather than over-flagging negative ones).

8. B. B. Blood flow produces relatively low-amplitude, high-velocity Doppler signals, while myocardial (and other tissue) motion produces relatively high-amplitude, low-velocity signals. TDI reconfigures the filter and gain settings to preferentially display that tissue-motion signal — most commonly used to assess myocardial wall velocities — rather than blood flow. It is not restricted to arterial-only display (A), a wall filter is still relevant (its threshold is simply reset for tissue rather than blood signals, not eliminated, contradicting C), and TDI can be shown in both color-encoded and spectral formats depending on the application (contradicting D).

9. C. C. Ultrasound contrast agents are encapsulated microbubbles, typically containing a low-solubility gas, that oscillate nonlinearly when insonated at diagnostic pressures; this nonlinear oscillation generates strong harmonic backscatter that is far greater, relative to surrounding tissue, than linear reflection alone — allowing contrast-specific or harmonic imaging to separate bubble signal from background. They are not dyes that absorb the beam (A), do not act as a harmonic filter (B), and their effect is acoustic/mechanical rather than thermal (D).

10. A. A. Elastography measures how much a tissue deforms (strain elastography, via manual or physiologic compression) or how fast a shear wave propagates through it (shear-wave elastography) as a surrogate for stiffness — softer tissue deforms more or transmits shear waves more slowly than stiffer tissue. It is not a blood flow velocity measurement (B, that is Doppler's role), an attenuation-coefficient measurement (C), or a distance measurement (D, that is standard B-mode caliper function).

11. C. C. Fusion imaging spatially aligns live ultrasound with prior CT/MRI (often using electromagnetic or optical tracking), which is particularly valuable for localizing and targeting a lesion — for example during percutaneous biopsy or tumor ablation — that may be better seen or was originally characterized on the other modality. It does not eliminate acoustic output (A, ultrasound is still actively transmitting), has no relationship to automatic TI calculation (B), and does not remove the need for correct patient positioning (D).

12. C. C. Wavelength = propagation speed / frequency = 1.54 mm/μs ÷ 5 MHz (5 cycles/μs) = 0.308 mm ≈ 0.31 mm. Option D repeats the propagation speed constant itself rather than dividing by frequency. Option B is roughly double the correct value and A is roughly half, both consistent with a factor-of-2 arithmetic slip rather than the correct division.

13. A. A. Attenuation is the umbrella term for the progressive loss of beam intensity/amplitude as it travels through tissue, arising from absorption (conversion to heat), scattering, and reflection. Amplification (B) is the opposite process (a receiver-side gain function). Rarefaction (C) is the low-pressure half of a sound wave cycle, not an attenuation mechanism. Refraction (D) is a beam-bending phenomenon at an interface, not intensity loss per se.

14. C. C. Axial resolution is the minimum distance along the beam's path (parallel to beam travel, i.e., front-to-back) at which two reflectors can be seen as separate rather than one. Distinguishing reflectors side by side perpendicular to the beam (A) describes lateral resolution instead. Options B and D describe unrelated scan-plane/frame scenarios rather than a resolution definition.

15. D. D. Axial resolution equals one-half the spatial pulse length (SPL), where SPL = number of cycles per pulse × wavelength; shorter pulses (fewer cycles and/or higher frequency, shorter wavelength) improve axial resolution. Frame rate (A) affects temporal, not axial, resolution. Transducer diameter (B) and focal depth (C) are primarily lateral-resolution factors, governing beam width, not the along-the-beam SPL relationship that defines axial resolution.

16. B. B. Lateral resolution depends on beam width — the narrower the beam, the better (smaller) the lateral resolution value. Beam width is narrowest at the focal zone (the natural or electronically steered focus), making that the depth of best lateral resolution. It is not simply at the transducer face without regard to focusing (A), the far field beyond focus is where the beam diverges and widens again, worsening lateral resolution (C), and lateral resolution absolutely varies with depth relative to the focal zone (contradicting D).

17. B. B. Each additional focal zone requires the system to transmit and receive a separate set of scan lines focused at that specific depth, increasing the total number of pulses needed to build one complete frame — which lengthens the time to construct a frame and lowers frame rate (worsening temporal resolution). It therefore reduces, not improves, frame rate (contradicting A and C), and focal zone count affects lateral resolution and frame rate, not axial resolution directly (contradicting D).

18. C. C. Duty factor = pulse duration ÷ pulse repetition period, expressed as a fraction (or percentage) of total time the transducer spends actively transmitting versus the full cycle of transmit-plus-listen. Pulses per second (A) describes PRF, not duty factor. A physical listening-region length (B) describes something closer to sample volume or receive gating, not a time-based duty factor. Echo-processing success rate (D) is unrelated to the transmit-time-fraction definition of duty factor.

19. D. D. Conventional pulsed B-mode imaging uses very short pulses (only a few cycles) followed by a comparatively long listening period to allow deep echoes to return before the next pulse is sent, giving a duty factor typically well under 1%. This is far lower than 10% (C), 25% (B), or 50% (A), all of which would describe near-continuous-wave-like operation, not standard pulsed imaging.

20. D. D. Refraction is the bending of the beam's path as it crosses an interface obliquely between media of differing propagation speed, and the amount of bending is described quantitatively by Snell's law. Attenuation (A) and absorption (B) describe intensity loss, not directional change. Reverberation (C) is an artifact from multiple reflections bouncing between two strong reflectors, unrelated to oblique-angle beam bending.

21. B. B. Because the system assumes a fixed average propagation speed to convert echo return time into a displayed depth, any tissue that propagates sound at a different actual speed than 1540 m/s (like fat, which is slower) causes structures beyond it to be placed at an incorrect depth on the image — a propagation speed error. Systems do not automatically detect and correct for regional speed differences (contradicting A), this error affects B-mode/anatomic depth placement broadly, not Doppler exclusively (contradicting C), and it is a depth-placement error distinct from — though sometimes co-occurring with — refraction (contradicting D).

22. A. A. Acoustic impedance (Z) = tissue density (ρ) × propagation speed of sound (c) in that tissue. It is this impedance value, not frequency/wavelength (B), attenuation/depth (C), or amplitude/power (D), that determines how much reflection occurs at a tissue interface when paired with the impedance of the adjacent tissue.

23. C. C. Reflection strength at an interface depends on the acoustic impedance mismatch between the two adjoining tissues — a large mismatch (e.g., soft tissue to bone or soft tissue to air) produces a strong reflection, while a small mismatch produces a weak one. Depth alone (A) affects overall signal attenuation en route, not the reflection coefficient at the interface itself. Frequency alone (B) affects resolution/penetration, not the impedance-based reflection strength. Focal zone count (D) is unrelated to reflection physics at an interface.

24. B. B. Bone (and other calcified structures) has a very high attenuation coefficient relative to soft tissue, absorbing and reflecting most of the incident beam and leaving little energy to continue deeper — producing the classic posterior acoustic shadow. It does not attenuate less than soft tissue (contradicting A), it behaves very differently from fluid, which attenuates minimally and produces enhancement rather than shadowing (contradicting C), and shadowing — not enhancement — is the expected posterior finding deep to dense calcification (contradicting D).

25. A. A. Reverberation occurs when sound reflects repeatedly between two strongly reflective, closely spaced surfaces (such as the transducer face and a metallic clip, or two parallel strong interfaces); each round trip is displayed as a progressively deeper, equally spaced echo, producing the characteristic ladder-like pattern of parallel bright lines. Refraction (B) produces lateral displacement/duplication of an image, not a series of equally spaced parallel lines. Range ambiguity (C) is a Doppler PRF-related artifact, not a B-mode reflector artifact. Speckle (D) is a diffuse granular texture from coherent interference, not discrete parallel lines.

26. A. A. Comet-tail artifact is a specific reverberation pattern seen with very small, strongly reflective structures (metallic fragments, air bubbles, cholesterol crystals) that produces a narrow, tapering, closely packed series of reverberation echoes appearing as a solid bright tail. Mirror-image artifact (B) produces a duplicated structure symmetrically reflected across a strong specular reflector such as the diaphragm, a different mechanism. Grating lobe artifact (C) arises from array element spacing, and twinkle artifact (D) is a color Doppler phenomenon behind certain reflective structures — neither describes the tapering B-mode comet-tail appearance.

27. B. B. Frame rate is fundamentally limited by how long it takes to construct one complete frame, which depends on imaging depth (deeper imaging requires longer listening time per line), the number of scan lines needed to cover the field of view (wider sector = more lines), and the number of focal zones (each adds a full extra set of lines) — increasing any of these lengthens frame-construction time and lowers frame rate. Transducer footprint alone (A) does not determine frame rate. Gain (C) is a receiver amplification setting with no bearing on the number of pulses sent, and while body habitus (D) may prompt a sonographer to change depth or gain, it is not itself a direct determinant of frame rate independent of the system settings.

28. D. D. Higher-frequency sound is attenuated more quickly by tissue, limiting penetration depth, but its shorter wavelength improves resolution; lower-frequency sound penetrates deeper at the cost of coarser resolution. This makes the 9 MHz transducer the better-resolution/lower-penetration choice and the 3 MHz the deeper-penetration/lower-resolution choice — the reverse of options A and C, and directly contradicting the claim in B that frequency has no effect.

29. C. C. Piezoelectricity is a two-way property of certain crystals (natural or synthetic, e.g., PZT): mechanical deformation (from a returning echo) generates a voltage (receive function), and an applied voltage causes the crystal to deform/vibrate (transmit function) — this dual conversion between mechanical and electrical energy is the basis for both pulse generation and echo detection. It is not a heat-conversion process (A), a magnetic phenomenon (B), or a chemical gel interaction (D).

30. C. C. Each array element contributes to a portion of the scan line/aperture; when one or more elements fail (crystal dropout), the region of the image corresponding to that element's contribution shows a visible dark vertical band or streak, since that segment of the aperture is no longer transmitting/receiving properly. This is a defect, not a benefit, so it would not increase frame rate (A) or improve resolution (B), and it produces a localized dark defect rather than a uniform brightness change (D).

31. A. A. The curvilinear array's curved footprint produces a wide, trapezoidal field of view that expands with depth — well suited to abdominal imaging, where organs are distributed over a broad area at moderate depth. A narrow acoustic window for cardiac imaging (B) favors a phased/sector array's small footprint instead. High-resolution superficial vascular work (C) favors a linear array. Endocavity imaging (D) uses a dedicated small-footprint endocavity probe, not a standard curvilinear abdominal probe.

32. C. C. Cardiac imaging must be performed through the narrow intercostal spaces between ribs, so a transducer with a very small physical footprint is required; phased array transducers achieve this small footprint while using electronic beam steering (via timed element firing) to still sweep a wide sector-shaped field of view once the beam passes beyond the ribs. They do not have the widest surface-level field of view (A, that describes curvilinear/linear geometry instead), are not restricted to M-mode only (B), and are fully compatible with spectral, color, and other Doppler modes (contradicting D).

33. C. C. Because an endocavity transducer is placed close to the structures of interest (uterus, ovaries) without needing to penetrate the abdominal wall or deep pelvic soft tissue, it can use a higher operating frequency than a transabdominal probe, trading the now-unnecessary extra penetration for improved near-field resolution. A lower frequency (A) would sacrifice resolution without the added penetration being needed, frequency selection is very much relevant (contradicting B), and endocavity and abdominal probes are not interchangeable at the same frequency by default (contradicting D).

34. A. A. A conventional 1D array has elements arranged in a single row, allowing electronic steering/focusing only within the scan plane (azimuth); a 2D matrix array arranges elements in rows and columns (a true grid), allowing electronic steering and focusing in both azimuth and elevation simultaneously, which is what enables real-time volumetric acquisition without physically sweeping the probe. It is not a single large crystal (B, matrix arrays use many small elements), backing material is still required for damping (contradicting C), and matrix arrays rely heavily on electronic steering, not the absence of it (contradicting D).

35. C. C. The backing material sits behind the piezoelectric element and absorbs energy directed backward, while also damping (shortening) the crystal's ringing after each excitation — a shorter pulse (fewer cycles, shorter SPL) directly improves axial resolution. It does not alter operating frequency, which is set by crystal thickness (contradicting A), it is a passive damping component rather than an electronic amplifier (contradicting B), and impedance matching to the patient's skin is the role of the separate matching layer in front of the crystal, not the backing material (contradicting D).

36. D. D. Without a matching layer, the large impedance mismatch between the piezoelectric crystal and skin would reflect most of the transmitted energy back into the transducer rather than into the patient; the matching layer's impedance is designed to sit between the two, acting as a transitional step (ideally close to the geometric mean of crystal and tissue impedance) that improves the efficiency of energy transmission into the body. Absorbing backward energy (A) and shortening ring-down (C) are functions of the backing material, not the matching layer, and pulse generation (B) is an electrical, not acoustic-layer, function.

37. C. C. A nonimaging CW pencil probe transmits and receives continuously (rather than in pulses), which removes the Nyquist velocity limit that constrains pulsed Doppler — making it the tool of choice for accurately measuring very high velocities (e.g., severe aortic stenosis jets) that would alias badly with pulsed techniques, at the cost of having no imaging or range-resolution capability at all. It is not used for B-mode anatomic imaging (A) or volumetric acquisition (B), both of which require an imaging array, and it has no specific elastography application (D).

38. C. C. A 1.5D array adds a modest number of elevational element rows (more than the single row of a true 1D array, but far fewer than a full 2D matrix), enabling some electronic control of elevational focusing/slice thickness that a standard 1D array cannot achieve, while remaining simpler and less expensive than a full matrix array. This makes it clearly distinct from a standard 1D array (contradicting A and B) and clearly less complex than a true 2D matrix array (contradicting D).

39. B. B. A piezoelectric crystal vibrates most efficiently (resonates) at a frequency determined by its thickness: a thinner crystal has a higher resonant frequency, and a thicker crystal has a lower one, which is why higher-frequency transducers use physically thinner elements. Field of view (A) is governed by array geometry/footprint, not crystal thickness. Number of focal zones (C) is a user-selected imaging parameter, and color scale (D) is a Doppler PRF setting — neither is determined by crystal thickness.

40. D. D. Bandwidth describes the spread of frequencies present in the pulse (centered around the transducer's nominal/center frequency); a broadband transducer contains a wide range of frequencies in each pulse, while a narrowband transducer's pulse is more frequency-limited. It is unrelated to physical footprint width (A), maximum penetration depth (B), or element count (C), each of which is a separate transducer characteristic.

41. A. A. Because a broadband pulse contains a wide range of frequencies, the system can electronically emphasize a lower or higher portion of that range depending on the clinical need — improving penetration in a technically difficult patient by shifting lower, or resolution in a superficial target by shifting higher — all from a single physical probe rather than requiring a probe swap. Frequency agility does not eliminate the need for focal zones (B), correct propagation speed error (C), or remove the need for a matching layer (D), all of which are separate, unrelated system features.

42. A. A. As an ultrasound beam propagates through tissue, nonlinear distortion of the waveform progressively generates energy at harmonic frequencies (chiefly the second harmonic, twice the fundamental transmitted frequency); harmonic imaging selectively receives and displays this harmonic signal rather than the fundamental. This is explicitly a nonlinear, not linear, propagation effect (contradicting B), is unrelated to the piezoelectric effect occurring 'twice' (C, which is not how harmonics are generated), and is not limited to air-tissue interfaces (D).

43. C. C. Because harmonic signal builds up gradually with depth (it is weak or absent immediately at the transducer face and strongest a short distance into tissue), harmonic imaging is comparatively less affected by near-field reverberation, side-lobe, and clutter artifacts than fundamental imaging, and generally provides improved contrast resolution, particularly benefiting technically difficult body habitus patients. It improves rather than worsens image quality in these patients (contradicting A and B) and does not remove the ongoing need for operator gain adjustment (contradicting D).

44. A. A. Harmonic content builds cumulatively as the fundamental beam propagates and distorts nonlinearly through tissue, meaning very little harmonic energy exists right at the skin/near-field region where reverberation and clutter artifacts are typically generated; because the system displays mainly the harmonic signal, that near-field clutter — generated at the fundamental frequency — is largely filtered out. It is not that a different transducer or near-field-free geometry is used (B), harmonic signal is weakest, not strongest, near the skin (contradicting C), and the receiver is not simply disabled near-field (D) — rather, the fundamental-frequency clutter is filtered from the received signal.

45. B. B. Bandwidth and pulse length are inversely related: a wider bandwidth (broader range of frequencies in the pulse) corresponds to a shorter, more compressed pulse in time — a shorter spatial pulse length. Since axial resolution equals one-half of SPL, a shorter SPL from a broader bandwidth directly improves (numerically decreases, i.e., sharpens) axial resolution. This is the opposite of option A, contradicts the claim in C that bandwidth is irrelevant, and axial (not just lateral) resolution is the property directly affected here, contradicting D.

46. A. A. To avoid range ambiguity, a pulse must be allowed to travel to the deepest imaged structure and return before the next pulse is sent; as imaging depth increases, this round-trip travel time increases, forcing PRF to decrease (fewer pulses per second) so each pulse has time to fully return. The relationship is inverse, not direct (contradicting B), and is not independent of depth (contradicting C) —

while sonographers select imaging depth, PRF adjustment for range-ambiguity avoidance is handled automatically by the system as a consequence (contradicting D).

47. A. A. Because deeper-originating echoes have traveled farther and experienced more attenuation than shallow echoes, TGC selectively boosts amplification as a function of echo return time (depth), compensating for that depth-dependent signal loss so that similar reflectors appear similarly bright regardless of depth. Uniform amplification regardless of depth (B) describes overall gain, a different control. TGC is a receiver-side amplification adjustment, not a transmit-power reduction (C) or an automatic frequency change (D).

48. B. B. Overall gain applies a uniform amplification to all returning echoes across the whole image, brightening (or darkening) the entire display equally regardless of depth — this is what distinguishes it from TGC, which applies depth-selective amplification. It does not target a specific depth range only (A, that is TGC's function), it is a receiver-side, not transmit-power, adjustment (contradicting C), and it does not reposition focal zones (D).

49. A. A. Excessive gain drives even weak or moderate-amplitude echoes to the top of the display's dynamic range, causing them to appear as bright white indistinguishable from truly strong reflectors — a saturation effect that washes out the subtle grayscale differences needed to appreciate tissue texture and pathology. It does not produce a black screen (B, which would instead result from too little gain/signal), it worsens rather than improves contrast resolution (contradicting C), and gain has no direct effect on frame rate (contradicting D).

50. C. C. Adding focal zones improves lateral resolution across more of the imaged depth range (since each zone is individually focused), but because each additional zone requires its own full set of transmitted/received scan lines, frame rate necessarily drops — which can introduce motion blur when imaging a moving structure (e.g., cardiac or vascular pulsatility). Focal zones do not eliminate the need for TGC (A, an unrelated receiver function), do not affect axial resolution (contradicting B, which is an SPL-based property), and the frame-rate trade-off described above is a real cost, contradicting D.

51. A. A. Write zoom is applied before the frame is frozen, prompting the system to re-acquire the selected region using a higher density of scan lines (and often a smaller field of view), which genuinely increases spatial resolution in that region. Read zoom, applied to an already-frozen or already-acquired image, simply enlarges (interpolates) the existing pixel data — making the displayed image bigger without adding any new acquired information, so it does not improve true resolution. This is the reverse of the claim in C, contradicts B's claim of equivalence, and write zoom is not restricted to color Doppler mode (contradicting D).

52. D. D. M-mode transmits along a single fixed scan line and displays the depth of reflectors along that line on one axis, plotted against time on the other axis, producing a scrolling trace that shows motion (e.g., valve leaflet opening/closing, fetal heart wall motion) with excellent temporal resolution along that single line. It is not a 3D volumetric mode (A), not a 2D color flow map (B, that describes color

Doppler), and not a spectral Doppler frequency-versus-time plot (C, which is a different display entirely despite sharing a time axis).

53. D. D. Dynamic range describes the span between the weakest and strongest echo signal amplitudes a system can process and display as distinguishable grayscale values — a wide dynamic range allows more subtle amplitude differences to be shown as distinct shades. It is unrelated to the range of selectable transducer frequencies (A), maximum penetration depth (B), or focal zone count (C), each a separate system parameter.

54. C. C. Narrowing dynamic range compresses the number of distinguishable gray shades between the weakest and strongest signal, producing a starker, higher-contrast image that appears more black-and-white with fewer subtle gradations — useful in some contexts (e.g., emphasizing a border) but at the cost of subtle texture detail. This is the opposite of a smoother gradient (contradicting A), the visual change is very real (contradicting B), and dynamic range is a display/processing parameter, not a penetration-depth parameter (contradicting D).

55. A. A. Edge enhancement is an image-processing algorithm that detects and accentuates the contrast difference at tissue boundaries (where echogenicity changes abruptly), making anatomic borders and interfaces appear crisper and easier to delineate. It is not a uniform whole-image brightness adjustment (B, that is closer to gain), it is not a speckle-reduction tool (C, that is more the role of compounding or persistence), and it has no effect on penetration depth (D).

56. A. A. Persistence blends several consecutive frames together, which tends to average out random speckle/noise (which varies frame-to-frame) while true anatomic structures (which are consistent frame-to-frame) are reinforced — producing a smoother-looking image, but because it blends temporally separate frames, it can blur fast-moving structures and effectively reduces temporal resolution. It does not increase frame rate (B, if anything it works against high effective temporal resolution), it is not specifically an edge-sharpening tool relative to others (contradicting C), and it has no relationship to TGC (D).

57. C. C. Spatial compounding electronically steers the beam through several different angles across the same anatomic region in rapid succession and combines (averages) those multiple angled views into a single composite frame, in contrast to persistence, which averages frames over time from the same angle (contradicting A). It does not involve progressively changing transmit frequency across the field of view (B), and it is a real-time multi-angle acquisition technique, not a single-frame postprocessing step (contradicting D).

58. C. C. Because artifacts like refraction-related edge shadowing, reverberation, and speckle are angle-dependent phenomena that appear differently (or are absent) depending on the insonation angle, averaging multiple angled views together in spatial compounding tends to cancel out or reduce these angle-specific artifacts while reinforcing consistent true anatomic structures — improving overall image quality and often border definition, the opposite of B. It reduces, not increases, speckle (contradicting A), and does not remove the ongoing role of focal zones (D).

59. B. B. Extended field of view works by tracking the transducer's motion as it is swept along the skin surface and using frame-to-frame image registration to stitch sequential 2D frames into one continuous, elongated static image (useful for measuring a long structure like an extremity muscle or a large mass) that exceeds the physical footprint/field of view of a single frame. It is a 2D stitching technique, not true volumetric 3D acquisition (contradicting A), it does not extend true penetration depth (C), and while commonly used with linear arrays, it is a processing technique layered on top of standard imaging, not a requirement that eliminates linear array use (contradicting D).

60. D. D. Coded excitation transmits a longer pulse containing a specific, known code/pattern, which delivers more total acoustic energy (improving penetration and signal-to-noise) than a short conventional pulse would; upon reception, the system applies a matched filter that decodes/compresses this longer pulse back into an effectively short pulse, preserving good axial resolution despite the longer original transmit pulse. This is the opposite of using only a short uncoded pulse (contradicting A), is unrelated to the matching layer (B), and is a deliberately engineered, non-random coding scheme, not arbitrary frequency variation (contradicting C).

61. C. C. A 2D matrix array can electronically steer and focus the beam in both azimuth and elevation, allowing it to insonate an entire volume of tissue without the sonographer needing to physically sweep or tilt the probe, unlike earlier freehand (position-tracked) or mechanically swept single-element 3D methods. Manual sweeping with a position sensor (A) describes an older freehand 3D technique, not true real-time matrix-array 3D. It acquires a full volume, not merely a single static plane (contradicting B), and works in both grayscale and Doppler applications, not Doppler exclusively (contradicting D).

62. C. C. 4D imaging is simply 3D volumetric acquisition displayed continuously and updated in real time, so that the volume changes/moves as it is being viewed — the added 'fourth dimension' being time, distinguishing it from a single static 3D volume capture. It is not merely 2D-plus-color-Doppler (A), it is specifically the continuously updating counterpart to a single static volume, not the same thing (contradicting B), and while fetal facial 4D imaging is a popular application, the mode itself is used far more broadly (e.g., cardiac), contradicting the exclusivity claim in D.

63. B. B. PACS is the digital infrastructure that stores, archives, retrieves, and distributes medical images (and often associated reports) so they can be accessed by radiologists, referring physicians, and other authorized users across a facility or health system, replacing older film-based archiving. It does not itself acquire images at the bedside (A, that is the ultrasound system's job), it is unrelated to transducer element testing (C), and it does not generate clinical interpretations (D) — that remains the interpreting physician's responsibility.

64. D. D. Accreditation and documentation standards require that archived images be properly and accurately labeled — correct patient identifiers, laterality (right/left), anatomic site/orientation, and measurement calipers where applicable — so the permanent record clearly supports the interpreting physician's report and can be reviewed or reconstructed later. Storing only one image (A) does not adequately document a full exam, patient identifiers must be present on stored images for accurate record-keeping, not omitted (contradicting B), and representative images — not measurements alone — are required (contradicting C).

65. B. B. In B-mode ('brightness mode'), each pixel's displayed brightness is derived from the amplitude of the echo returning from that specific location, after the system applies TGC and gain processing — a stronger reflector produces a brighter pixel, a weaker one a dimmer pixel. Reflector distance (A) determines pixel position/depth on the display, not its brightness. Propagation speed (C) is generally assumed constant for depth calculation and is not what brightness encodes, and operating frequency (D) is a system/transducer setting, not a per-pixel encoded value.

66. C. C. If PRF is set too high relative to the selected imaging depth, the system may send a subsequent pulse before all the echoes from the previous pulse — including from the very deepest reflectors — have returned; those late echoes then get incorrectly attributed to the next pulse cycle and displayed at a falsely shallow depth, since the system assumes each received echo belongs to the most recent pulse sent. This is a range/depth misplacement problem, not a comet-tail artifact (A) or destructive interference between scan lines (B), and it is a genuine risk the sonographer must manage by depth-appropriate PRF selection — it does not automatically self-correct (contradicting D).

67. A. A. Grating lobes are a specific consequence of array transducer construction: because the array consists of discrete, periodically spaced elements (rather than one continuous aperture), interference patterns create secondary, weaker beams (grating lobes) directed at other angles; if these grating lobes strike a strong reflector, the resulting echo may be displayed erroneously as though it originated from the main beam's actual direction. This is a structural array-geometry phenomenon, not caused by excessive gain (B), a single failed element (C, which instead produces dropout, not grating lobes), or a TGC setting error (D).

68. C. C. Side lobes are off-axis, generally lower-intensity secondary beams of energy that exist even for a single, continuous-aperture transducer element (unlike grating lobes, which specifically require the periodic spacing of a multi-element array), and can produce spurious weak echoes displayed as though they came from the main beam. This makes option A's restriction to array transducers backward (side lobes are actually the more general, single-element-applicable phenomenon, while grating lobes are array-specific). Side lobes are a B-mode/general acoustic phenomenon, not Doppler-spectral-exclusive (contradicting B), and are unrelated to skin-transducer reverberation (D).

69. A. A. The diaphragm is a strong, curved specular reflector, and mirror-image artifact classically occurs when the beam reflects off the diaphragm, continues to a real structure (e.g., a liver lesion), reflects back off the diaphragm again, and returns to the transducer with a longer total travel time — causing the system to falsely display a duplicate 'mirrored' copy of that structure symmetrically positioned on the far side of the diaphragm, typically appearing within the lung field where no such structure truly exists. This classic teaching case is specific to strong curved reflectors like the diaphragm, not routinely described at a distended bladder (B), the endometrial stripe (C), or a superficial lymph node (D).

70. B. B. Because simple fluid attenuates ultrasound far less than the surrounding soft tissue through which the beam also travels, more acoustic energy survives to reach and be reflected from tissue deep to the fluid-filled structure — making that posterior region appear brighter (enhanced) relative to adjacent tissue at the same depth. This is essentially the opposite phenomenon from what occurs deep to bone (A) or a calcified stone (C), both of which produce shadowing, not enhancement, from their high attenuation, and bowel gas (D) similarly tends to produce dirty shadowing/reverberation rather than enhancement.

71. C. C. Gas has a very large acoustic impedance mismatch with soft tissue and reflects/scatters the beam chaotically at the gas-tissue interface, producing a characteristic 'dirty' shadow — a zone of indistinct, low-level reverberation echoes deep to the gas — which looks visually different from the clean, sharply marginated acoustic void seen deep to a smooth, strongly attenuating reflector like bone (contradicting A, which describes clean rather than dirty shadowing). Gas produces shadowing, not enhancement (contradicting B), and it very much affects the beam, contrary to D.

72. C. C. When the ultrasound beam strikes a curved interface (such as the wall of a cyst or fluid collection) at an oblique angle near its lateral margins, refraction bends the beam's path, effectively diverting energy away from a straight return path and producing a thin acoustic shadow extending from those lateral edges — distinct from the enhancement seen directly posterior to the fluid (contradicting A), from comet-tail artifact's mechanism (small strong reflector reverberation, contradicting B), and from grating lobe artifact's array-spacing mechanism (contradicting D).

73. D. D. Speckle arises because tissue parenchyma contains many scatterers smaller than the system's resolution capability; the echoes from these numerous scatterers interfere constructively and destructively with one another, producing a granular, mottled pattern that does not represent individual true anatomic structures but is nonetheless a real, characteristic, and diagnostically useful (texture-based) feature of the image. It is a fundamentally different phenomenon from reverberation (A), grating lobes (B), or range ambiguity (C), each of which has a distinct separate mechanism described elsewhere in this exam.

74. D. D. Because spatial compounding requires acquiring several separate frames from different steering angles before combining them into one displayed composite frame, the effective frame rate for the composite image is lower than for a conventional single-angle acquisition — a real-world temporal resolution trade-off for the artifact-reduction benefit compounding provides. This directly contradicts an increase (A) or no effect (B), and the effect is a general consequence of the compounding acquisition process, not limited to Doppler applications (contradicting C).

75. C. C. Refraction-based edge shadowing and many reverberation artifacts depend heavily on the specific angle of insonation relative to the reflecting interface; because spatial compounding averages frames acquired from multiple distinct angles, an artifact prominent from one angle is often absent or reduced from another angle, so averaging tends to suppress these angle-dependent artifacts while reinforcing consistent true anatomy. It does not specifically address patient breathing motion (A) or electronic interference (B), which are unrelated to beam-angle artifact physics, and spectral Doppler aliasing (D) is a PRF/Nyquist phenomenon unrelated to spatial compounding's angle-averaging mechanism.

76. B. B. Because the described problem is specifically localized to the far field while the near field is acceptable, the appropriate first step is a depth-selective adjustment — increasing far-field TGC (and/or using persistence to smooth residual noise) — rather than a uniform whole-image change like overall gain, which would also alter the already-acceptable near field. Decreasing overall gain uniformly (A) would worsen, not fix, a far-field visibility problem, switching to a higher frequency (C) would actually reduce far-field penetration further (higher frequency attenuates faster), and eliminating focal zones entirely (D) would degrade lateral resolution broadly rather than targeting the specific depth-related noise issue described.

77. C. C. The Doppler equation solves for flow velocity using the measured Doppler shift, the known transmitted frequency, the assumed propagation speed of sound in blood/soft tissue, and the cosine of the angle between the ultrasound beam and the direction of flow — vessel diameter, viscosity, and cardiac output (A) are not variables in the Doppler equation itself, transducer footprint and BMI (B) are irrelevant, and the angle term is a required, not optional, part of the equation (contradicting D).

78. A. A. Because the Doppler equation includes the cosine of the angle between the beam and flow, and the cosine of 90 degrees is zero, a perpendicular insonation angle mathematically drives the calculated Doppler shift toward zero regardless of the true flow velocity — making velocity measurements unreliable or invalid near 90 degrees. This is the opposite of a maximal shift (B), directly contradicts the claim that angle has no effect (C), and there is no automatic correction that overcomes this fundamental cosine relationship (contradicting D).

79. C. C. Because the error in calculated velocity grows disproportionately large as the angle approaches 90 degrees (due to the nonlinear behavior of the cosine function near 90 degrees), the accepted clinical standard is to keep the Doppler angle at 60 degrees or less, with angles closer to 0 (beam nearly parallel to flow) being even more accurate. This is essentially the opposite of aiming for 90 degrees (A), angle-dependence is a real and unavoidable part of the Doppler equation, not eliminated by modern systems (contradicting B), and there is no fixed universal requirement of exactly 45 degrees (D).

80. D. D. The Nyquist limit is the maximum Doppler shift frequency that a pulsed system can accurately and unambiguously sample, and it is mathematically defined as one-half the pulse repetition frequency (PRF/2); any true Doppler shift exceeding this limit will be misrepresented (aliased) rather than measured correctly. This concept is unrelated to reflection coefficient (A), attenuation (B), or acoustic impedance (C), which are separate physical properties governing reflection and beam intensity loss, not sampling-rate limits.

81. D. D. Aliasing is fundamentally a sampling-rate problem: when the actual Doppler shift frequency is higher than the Nyquist limit (half the PRF), the pulsed system cannot correctly sample that rapidly changing signal and instead displays it as an ambiguous or wrapped value — appearing as spectral wraparound in spectral Doppler or as an abrupt color reversal in color Doppler. An excessive Doppler angle (A) causes underestimation of velocity, not aliasing; wall filter setting (B) affects low-velocity signal rejection, not the Nyquist limit; and overall gain (C) is a receiver amplification setting unrelated to the sampling-rate cause of aliasing.

82. B. B. Because aliasing occurs when true velocity exceeds the Nyquist limit (PRF/2), several corrective options exist: increasing PRF/scale directly raises the Nyquist limit, shifting the baseline effectively extends the usable display range in the direction of flow, and decreasing transmitted frequency lowers the Doppler shift produced by the same true velocity (since Doppler shift is directly proportional to transmitted frequency) — any of which can resolve or reduce aliasing. Simply decreasing the angle without changing scale (A) does not raise the Nyquist limit and would not reliably resolve aliasing at very high velocities. Decreasing gain (C) affects signal amplitude/noise, not the sampling-rate limit. Aliasing is a well-recognized, correctable phenomenon, not something requiring abandonment of spectral Doppler altogether (contradicting D).

83. B. B. Because CW Doppler uses separate continuously transmitting and continuously receiving crystals rather than pulsing, it has no PRF and therefore no Nyquist limit — it cannot alias, even at very high velocities such as severe stenotic jets. The trade-off is that, because it is receiving continuously along the entire beam path, it cannot distinguish where along that path a given signal originated (no range/depth resolution), unlike PW Doppler, which gates reception to a specific depth (sample volume) at the cost of being limited by PRF and therefore subject to aliasing. This directly contradicts A (CW does not alias like PW), C (CW lacks range resolution, not the reverse), and D (they are functionally quite different).

84. D. D. PW Doppler transmits a pulse and then listens only during a specific, timed window after transmission — since echo return time corresponds directly to depth (given the assumed propagation speed), this range gating allows the system to selectively sample the Doppler signal from a chosen depth (the sample volume) rather than the entire beam path at once. This is different from CW's dual continuously active crystal design (contradicting A, which describes CW rather than PW), it is not random sampling (B), and it depends on echo timing, not amplitude alone (contradicting C).

85. C. C. Vessel walls and surrounding soft tissue move slowly but produce very high-amplitude Doppler signals compared with blood; the wall filter is a high-pass filter that removes signals below a chosen frequency (velocity) threshold, filtering out this low-velocity, high-amplitude wall-motion clutter while preserving the genuine blood-flow signal for display. It is not a uniform amplifier (A), has no effect on the Nyquist limit, which is set by PRF (contradicting B), and does not correct for Doppler angle, an entirely separate measurement parameter (contradicting D).

86. C. C. Because the wall filter removes signal below its set frequency/velocity threshold, setting it too high risks removing not just unwanted wall-motion clutter but also genuine slow-velocity flow signal — such as venous flow, which is inherently low-velocity — potentially producing a falsely absent or underrepresented flow signal on a vessel that is actually patent. This is the opposite of improved visualization (A), the wall filter affects any low-velocity signal regardless of arterial or venous origin (contradicting B), and an over-filtered venous signal is more likely to appear diminished or absent than aliased, which is a distinct PRF/Nyquist-related phenomenon (contradicting D).

87. D. D. The internal carotid artery supplies the brain, a continuously metabolically active, low-resistance vascular bed, and normally demonstrates a low-resistance waveform with continuous forward (antegrade) flow throughout diastole. The external carotid artery (A) supplies high-resistance structures (face/scalp) and normally shows a more high-resistance pattern with minimal diastolic flow. A resting skeletal muscle artery (B) is also normally high-resistance at rest (low diastolic flow), becoming low-resistance only with exercise. Acute renal transplant rejection (C) is a pathologic state that characteristically increases, not decreases, resistance — the opposite of a low-resistance pattern.

88. D. D. Peripheral limb arteries at rest supply a high-resistance vascular bed (resting skeletal muscle), and normally show a triphasic waveform: a sharp forward systolic peak, a brief reversal of flow in early diastole (reflecting high downstream resistance), and a small forward component in late diastole. This triphasic high-resistance pattern is not the expected normal appearance of the internal carotid artery (A, which is low-resistance/continuous diastolic forward flow), the renal artery (B, a low-resistance organ-perfusion pattern), or the hepatic artery (C, also typically a lower-resistance, continuous-forward pattern feeding the liver).

89. C. C. Resistive Index = (peak systolic velocity − end-diastolic velocity) ÷ peak systolic velocity, producing a unitless value that increases as diastolic flow decreases relative to systolic flow (i.e., as downstream resistance rises) and decreases as diastolic flow becomes more prominent (lower resistance). This is not simply diastolic-over-systolic velocity (A, which is a related but different ratio), not mean-over-peak-systolic (B), and does not involve vessel diameter at all (D).

90. B. B. Color Doppler estimates the mean Doppler frequency shift (and therefore an estimated mean velocity) at each sampled location within the color box, along with the direction of flow relative to the transducer, and encodes this information using a color map (commonly red/blue with brightness indicating relative velocity magnitude). It does not encode anatomic tissue type (A), acoustic impedance (C), or attenuation coefficient (D) — those are separate acoustic properties unrelated to the color-flow calculation.

91. D. D. The BART mnemonic (Blue Away, Red Toward) describes the standard default color-map convention: flow moving away from the transducer is shown blue and flow moving toward the transducer is shown red, based on direction relative to the transducer — not based on whether the vessel is an artery or vein, which is a common misconception addressed by ruling out option A (color does not intrinsically denote artery versus vein; a vein flowing toward the probe would show red just like an artery flowing toward the probe). This default map is a convention, not an unchangeable law of physics — the color map can be manually reversed on most systems, so it is not universally fixed without exception (contradicting C), and it is a defined, non-random standard (contradicting B).

92. B. B. A larger color box encompasses more scan lines, and because color Doppler requires multiple pulses along each individual scan line (a 'packet') to estimate the mean velocity/direction at each point (unlike single-pulse B-mode line acquisition), increasing box size directly increases the total number of pulses needed to complete one color frame — lengthening frame time and lowering the color frame rate. This is the opposite of an increase (A), the effect is real and predictable, not absent (contradicting C), and frame rate obviously cannot become infinite (D).

93. D. D. Packet size (ensemble length) is the number of pulses sent down each individual scan line specifically for the color Doppler calculation; more pulses per line allow a more accurate estimate of the mean Doppler shift (improving color sensitivity and reducing noise) but require more total transmit time per frame, which further reduces frame rate — a distinct concept from the physical box size itself (contradicting A), unrelated to element activation count (B), and not a measure of box depth (C).

94. A. A. Power Doppler encodes the total integrated amplitude (power) of the Doppler signal returning from a region, rather than the mean velocity/frequency shift and direction that conventional color Doppler displays — because it is not trying to encode velocity or direction, it can use signal from a broader range of angles and slower flows, making it substantially more sensitive to low-volume or low-velocity flow (at the cost of losing directional and precise velocity information). This is fundamentally different from displaying mean velocity (B) or direction alone (C) as conventional color Doppler does, and it has no relationship to tissue stiffness/elastography (D).

95. D. D. Because power Doppler encodes signal amplitude rather than a directional mean-frequency-shift calculation, it does not alias (there being no directional/velocity value near the Nyquist boundary to wrap around) and is much less angle-dependent than conventional color Doppler, along with offering greater sensitivity to slow or low-volume flow — making it valuable for detecting subtle perfusion, such as in a small structure or at a tumor's periphery. This is the opposite of showing direction (A, which power Doppler specifically does not display), it is not simply faster in every case (B), and it is more, not less, sensitive to slow flow (contradicting C).

96. C. C. Flash artifact appears as a sudden, diffuse burst of color across some or all of the color box, generally caused by physical motion — patient movement, breathing, transmitted pulsation from an adjacent structure, or transducer pressure changes — being misinterpreted by the color processor as a genuine large-scale Doppler shift, rather than reflecting true blood flow. It is not a persistent true vascular finding (A), not primarily a hardware calibration issue (B), and it is specifically a color/power Doppler phenomenon, not a spectral-Doppler-exclusive one (contradicting D).

97. C. C. Blooming occurs when color gain or transmit power is set too high, causing the displayed color signal to extend beyond the true anatomic boundaries of the vessel or jet — visually overestimating its actual size — which can be corrected by reducing color gain/power to an appropriate level. It is not an absence of signal (A, the opposite problem), not a spectral-Doppler-specific artifact (contradicting B, this is fundamentally a color-Doppler phenomenon), and unrelated to CW Doppler or aliasing specifically (contradicting D).

98. A. A. The standard optimization technique is to raise color gain until random background color noise/speckle begins to appear in tissue where no true flow should exist, then reduce gain just slightly until that background noise disappears — leaving the display set at the highest gain that still shows only genuine flow signal, which maximizes sensitivity without introducing false color noise or blooming. Simply maxing out gain (B) risks noise and blooming, minimizing it (C) risks missing genuine slow flow, and color gain is a distinct, independently adjustable control from B-mode gain, not something without its own setting (contradicting D).

99. B. B. Slow venous flow is best detected by lowering the velocity scale/PRF (which increases sensitivity to low-velocity signal, since a lower PRF corresponds to a lower Nyquist limit better matched to slow flow) and lowering the wall filter threshold (so the genuinely slow venous signal is not inadvertently filtered out along with wall-motion clutter). A high scale and high wall filter (A) would do the opposite, likely missing the venous signal entirely. Maximizing transmit power without regard to ALARA (C) is not an appropriate general practice. PW Doppler, with appropriately adjusted settings, is entirely capable of detecting venous flow — CW is not a required substitute (contradicting D).

100. B. B. Because CW Doppler continuously transmits and receives along the entire beam path without any depth (range) gating, if two vessels with different flow patterns happen to lie along the same beam line at different depths, their signals can overlap in the resulting spectral display with no way to determine which vessel a specific portion of the signal came from — a direct consequence of CW's lack of range resolution. This is a distinct concept from the Nyquist limit (A, which specifically concerns pulsed systems and does not even apply to CW), from aliasing (C, also a pulsed-Doppler-specific phenomenon that does not occur with CW), and from spectral broadening (D, a turbulence-related widening of the velocity spectrum, not a depth-ambiguity issue).

101. A. A. Normal laminar flow has red blood cells moving at fairly uniform velocity, producing a clean spectral waveform with a clear window beneath the envelope; turbulent flow — commonly seen distal to a stenosis — contains a much wider range of simultaneous velocities and directions within the sample volume, which fills in that spectral window, a finding called spectral broadening and a recognized indirect sign of disturbed/turbulent flow. This is essentially the opposite of the appearance expected with normal laminar flow (contradicting B), and it is a genuine hemodynamic finding, not simply a wall filter setting artifact (contradicting C) or a consequence of Doppler angle alone (contradicting D).

102. C. C. The sample volume (gate) length determines the axial extent of the beam being sampled for spectral analysis; sizing it appropriately to the target vessel's lumen (rather than leaving it maximally large) improves specificity by reducing the chance of also capturing signal from an adjacent vessel, the vessel wall, or surrounding tissue — while an oversized sample volume risks contaminating the waveform with unwanted extraneous signal. This directly contradicts always maximizing sample volume length (A), sample volume size clearly does affect specificity, contrary to B, and there is no requirement that it match the color box length (D).

103. A. A. With transmitted frequency, propagation speed, and angle held fixed, the Doppler equation reduces to a direct, linear proportionality between true flow velocity and the resulting Doppler frequency shift — as one increases, the other increases proportionally, which is the basis for calculating velocity from the measured shift in the first place. This directly contradicts an inverse relationship (B), the claim of no relationship (C), or an exponential rather than linear relationship (D).

104. B. B. Because Doppler shift is directly proportional to transmitted frequency (for a fixed true velocity and angle), using a higher-frequency transducer produces a larger measured Doppler shift for the same actual blood velocity; since the Nyquist limit is set by PRF, not by transducer frequency, this larger shift is more likely to exceed the Nyquist limit and alias. This is the opposite of reducing aliasing risk (contradicting A), transmitted frequency very much affects the resulting shift (contradicting C), and higher frequency increases rather than eliminates aliasing risk (contradicting D) — which is part of why lower-frequency transducers are sometimes chosen specifically to reduce aliasing on very high-velocity jets.

105. D. D. Because CW Doppler transmits and receives continuously with no pulse-repetition-based sampling limit, it has no Nyquist limit and cannot alias at any velocity — making it the appropriate tool for accurately measuring very high-velocity jets (such as severe valvular stenosis) that would alias badly on pulsed techniques. Color Doppler (A) is itself a pulsed technique subject to the same Nyquist/aliasing limitation. Standard PW Doppler, even with a small sample volume (B), remains fundamentally pulsed and therefore still subject to aliasing at sufficiently high velocity — sample volume size does not change the PRF-based Nyquist limit. Power Doppler (C) does not provide a quantitative velocity measurement at all, so it cannot be used to determine peak velocity in the first place.

106. D. D. When true velocity exceeds the Nyquist limit, the peak of the spectral waveform is cut off at the display's scale limit and reappears wrapped around on the opposite side of the baseline (e.g., a high forward-flow peak appearing to spill over into the reverse-flow/below-baseline region), producing a visually distorted, discontinuous-looking waveform rather than a smooth single peak. This is a very different appearance from a complete absence of signal (A), a smooth undistorted waveform (B, the opposite of the aliased appearance), or simple baseline noise increase without waveform disruption (C).

107. C. C. The reject control sets an amplitude-based threshold, discarding weak, low-amplitude signals presumed to represent noise rather than genuine flow — this is a distinct mechanism from the wall filter, which instead excludes signal based on its frequency (velocity), regardless of amplitude, to remove high-amplitude, low-velocity wall-motion clutter. Reject does not directly change the Nyquist limit, which is set by PRF (contradicting A), it does not alter transmitted frequency (B), and it has no relationship to angle correction (D).

108. C. C. To capture the true peak velocity of a jet or area of turbulence — which is typically fastest in the central stream, not at the wall — the sample volume should generally be positioned centrally within the lumen and systematically moved/interrogated across the suspected area, since wall-adjacent placement risks underestimating true peak velocity or picking up low-velocity wall-related signal instead. This contradicts routinely favoring wall placement over central placement (A), placing the gate outside the vessel would not sample flow at all (B), and an oversized sample volume covering the vessel and surrounding tissue simultaneously (D) would reduce, not improve, specificity for localizing the true peak-velocity jet.

109. B. B. A variance color map adds a separate encoded color (frequently green) to indicate pixels where there is substantial variation in velocity and/or direction within the sample — a hallmark of turbulent, disorganized flow — overlaid on or in place of the standard direction/mean-velocity color scheme, helping highlight areas of turbulence such as distal to a stenotic jet. It is not a measure of vessel wall thickness (A), tissue attenuation (C), or a display of absolute peak systolic velocity value alone (D), which is a spectral Doppler, not color variance, measurement.

110. C. C. A regular, repeating wraparound pattern in a vessel with known genuinely unidirectional flow is the classic appearance of aliasing — the true peak velocity exceeds the Nyquist limit, so the waveform peak is displayed as if it crossed to the opposite side of the baseline, repeating each time velocity exceeds the limit during the cardiac cycle — rather than representing true alternating bidirectional flow (contradicting A). This is not an expected/normal venous appearance requiring no action (contradicting B, especially given the stated unidirectional true flow), and it is a scale/PRF setting issue correctable by adjusting the Doppler controls, not evidence of hardware failure (contradicting D).