

# IMAGING TECHNIQUES

## 2-3 Mammograms

Mammography is a specific type of imaging that uses low-dose to examine the human [breast](#). Mammography is widely used in both the investigation of symptomatic breast disease and is the modality used for breast screening; it has a high sensitivity in the detection of breast cancers, particularly invasive carcinomas and the ductal carcinoma in situ (DCIS), but it has a lower specificity (Reaney, 1999). The goal of mammography is the early detection of [breast cancer](#), typically through detection of characteristic masses and/or microcalcifications. Mammography is believed to reduce mortality from [breast cancer](#). In many countries, routine mammography of older women is encouraged as a screening method to diagnose early breast cancer. The [United States Preventive Services Task Force](#) recommends screening mammography, with or without clinical breast examination, every 1–2 years for women aged 50 and older (Mammography Quality Scorecard, 2010). Altogether clinical trials have found a relative reduction in breast cancer mortality of 20%, but the two highest-quality trials found no reduction in mortality (Rees, 1994). Like all x-rays, mammograms use doses of [ionizing radiation](#) to create images. Radiologists then analyze the image for any abnormal findings. It is normal to use longer [wavelength](#) X-rays than those used for [radiography](#) of [bones](#). Mammography has a false-negative (missed cancer) rate of at least 10 percent. This is partly due to dense tissues obscuring the cancer and the fact that the appearance of cancer on mammograms has a large overlap with the appearance of normal tissues. During the procedure, the breast is compressed by a dedicated mammography machine to even out the [tissue](#), to increase image quality, and to hold the breast still (preventing [motion blur](#)). Both head-to-foot (craniocaudal, CC) view and angled side-view (mediolateral oblique, MLO) images of the breast are taken. Diagnostic mammography may include these and other views, [Deodorant](#), [talcum powder](#) or lotion may show up on the X-ray as [calcium](#) spots, and women are

discouraged from applying these on the day of their examination (Mammography Frequently Asked Questions, 2007).

Until some years ago, mammography was typically performed with screen-film cassettes. Now, mammography is undergoing transition to digital detectors, known as Full Field Digital Mammography (FFDM). This progress is some years later than in general radiology, this is due to the fact that digital mammography have never been shown to be superior to film-screen mammography for the diagnosis of breast cancer (Wikipedia, 2011).

In young women, the usefulness of mammography is restricted by high prevalence of dense fibroglandular tissue, which impairs both the detection and the differentiation of the lesion (Smith, 2009). Hormone replacement therapy may decrease the sensitivity of mammography by increasing the breast density and enlarging benign masses, such as cysts and fibroadenomas (LaTrenta, 1994). After breast surgery, mass-like scars and areas of distortion may mimic a tumor or hide subtle signs of malignancy (David, 1995). Radiation after surgical treatment of breast carcinoma leads to skin thickening and increased focal or diffuse density of the breast due to edematous changes (Sinha, 2002).

There are also some conditions for which mammography have a limited role in assessment as they often have no clearly recognizable features on mammograms. These conditions are recognized as being precancerous; some have little risk of progressing to malignancy whilst others have a significantly high risk e.g. ductal carcinoma in situ has a greater risk of developing into an invasive carcinoma than atypical lobular hyperplasia (Allen, 1994). Breast implants can also impede accurate mammogram readings because both silicone and saline implants are not transparent on x-rays and can block a clear view of the tissues behind them, especially if the implant has been placed in front of, rather than beneath, the chest muscles (Allen, 1994). That is why breast services do not stop at mammography but incorporate other imaging modalities such as ultrasound, magnetic resonance imaging, cytology and clinical assessments.

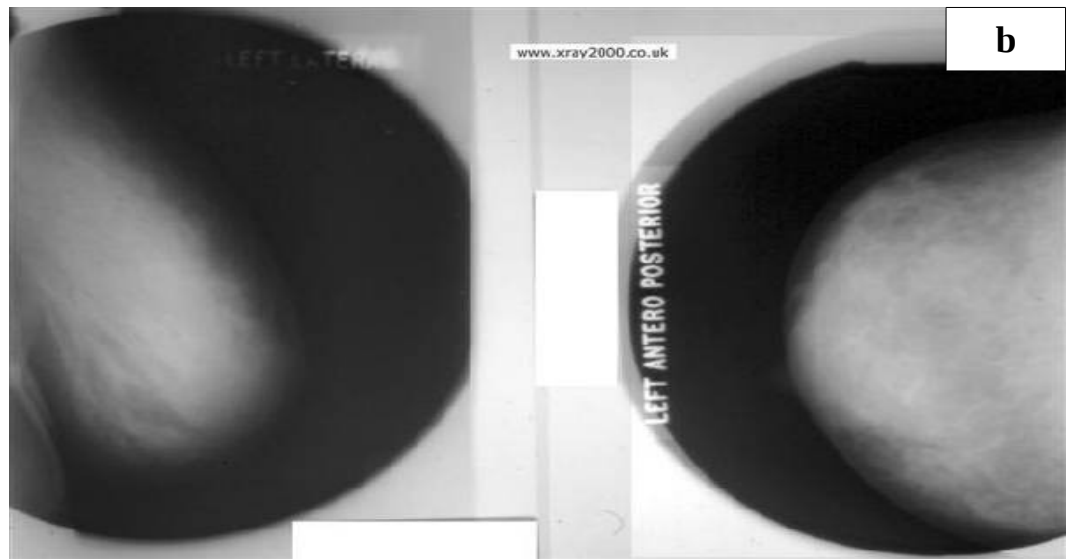
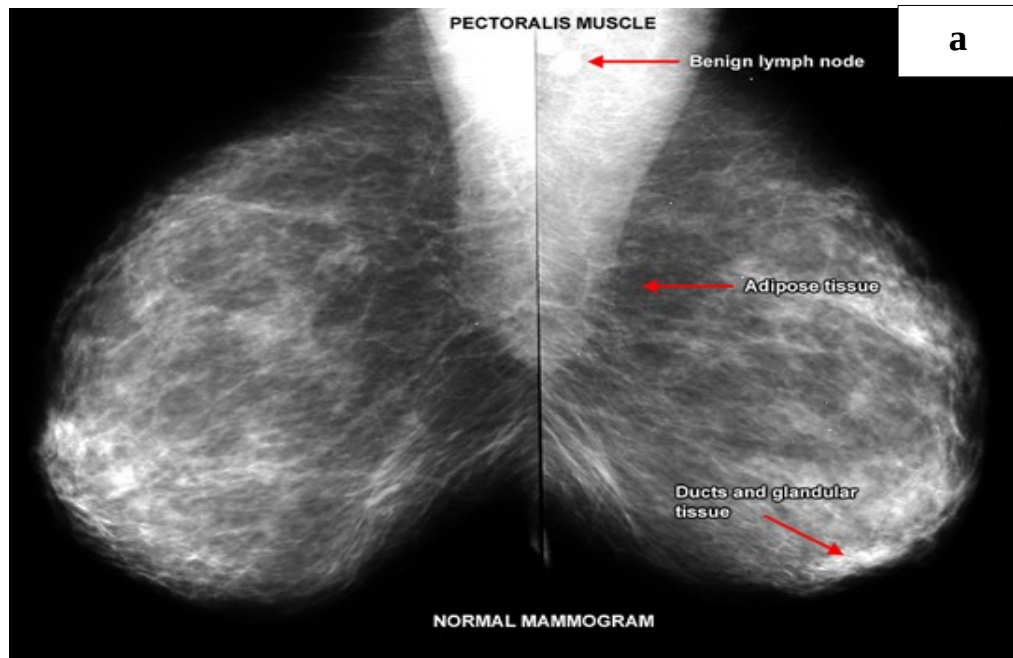


Fig. (2-8) a. and b. shows: Normal mammogram (Allen, 1994).

## 2-4 Ultrasounds

Ultrasound (U/S) imaging, also called ultrasound scanning or sonography, involves exposing part of the body to high-frequency sound waves to produce pictures of the inside of the body. Ultrasound exams do not use ionizing radiation (as used in x-rays). Because ultrasound images are captured in real-time, they can show the structure and movement of the body's internal organs, as well as blood flowing through blood vessels (Michael, 2009).

Ultrasound imaging is a noninvasive medical test that helps physicians diagnose and treat medical conditions.

Ultrasound imaging of the breast produces a picture of the internal structures of the breast. During a breast ultrasound examination the sonographer or physician, performing the test may use Doppler techniques to evaluate blood flow or lack of flow in any breast mass. In some cases, this may provide additional information as to the cause of the mass (Michael, 2009). All of the breast lesions including calcifications whatever small they are, could be detected on systematic U/S examination, and the possibility of the development of breast cancers are investigated using a strict criteria with respect of height to width ratio, mobility, vascularity and presence or absence of calcifications and degenerative changes but still the result has to be confirmed by histopathology (Bushra, 2004).

Ultrasound imaging of the breast is performed to:

1. Determining the Nature of a Breast Abnormality

The primary use of breast ultrasound today is to help diagnose breast abnormalities detected by a physician during a physical exam (such as a lump or bloody or spontaneous clear nipple discharge) and to characterize potential abnormalities seen on mammography.

Ultrasound imaging can help to determine if an abnormality is solid (which may be a non-cancerous lump of tissue or a cancerous tumor) or fluid-filled (such as a benign cyst) or both cystic and solid. Ultrasound can also help show additional features of the abnormal area. Doppler ultrasound used to assess blood supply in breast lesions (Michael, 2009).

## 2. Supplemental Breast Cancer Screening

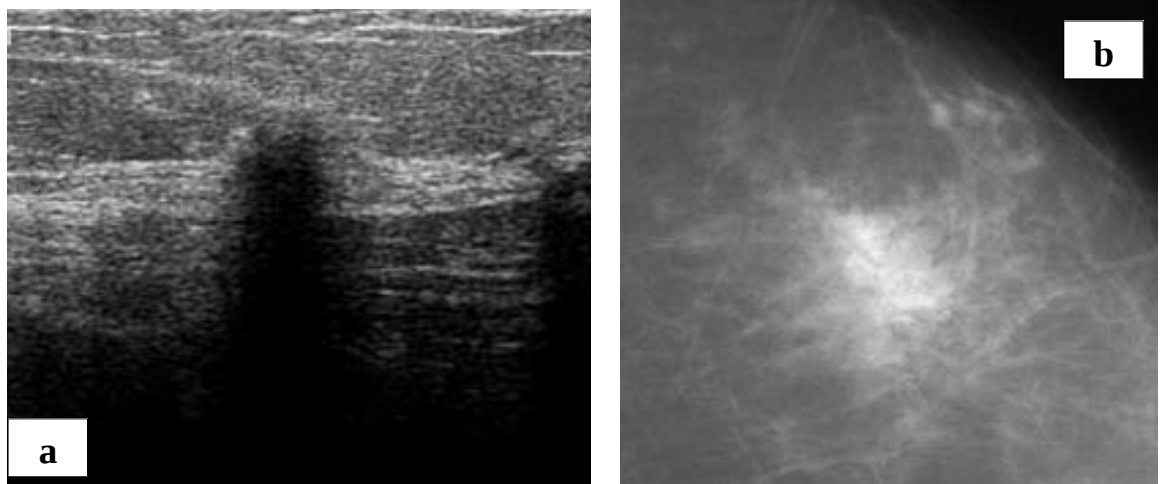
Mammography is the only screening tool for breast cancer that is known to reduce deaths due to breast cancer through early detection. Even so, mammograms do not detect all breast cancers. Some breast lesions and abnormalities are not visible or are difficult to interpret on mammograms. In breasts that are dense, meaning there is a lot of glandular tissue and less fat, many cancers can be hard to see on mammography. Many studies have shown that ultrasound and magnetic resonance imaging (MRI) can help supplement mammography by detecting small breast cancers that may not be visible with mammography. This is usually only considered when the breast tissue is dense. It is hoped that by detecting such cancers, these other screening tests might help to further prevent deaths due to breast cancer beyond what is achieved with mammography alone. When ultrasound is used for screening, many abnormalities are seen which may require biopsy but are not cancer (false positives), and this limits its cost effectiveness (Michael, 2009). Today, ultrasound is being investigated for use as a screening tool for women who have dense breasts, silicone breast implants and very little tissue can be included on the mammogram, pregnant or should not to be exposed to x-rays (which is necessary for a mammogram) also those who are at high risk for breast cancer based on family history, personal history of breast cancer, or prior atypical biopsy result.

## 3. Ultrasound-guided Breast Biopsy

When an ultrasound examination cannot characterize the nature of a breast abnormality, a physician may choose to perform an ultrasound-guided biopsy. Because ultrasound provides real-time images, it is often used to guide biopsy procedures<sup>[32]</sup>. In conventional ultrasonic imaging, contrast between diseased and healthy tissues is often low because acoustic impedance differences between various tissues are small. Quantitative ultrasound (QUS) techniques have been used to increase the contrast between healthy and diseased tissues in ultrasonic imaging. Specifically, statistically significant differences between healthy and diseased tissues have been observed in preliminary studies from rodent models of breast cancer using QUS imaging, whereas, these differences were not observed in conventional ultrasonic imaging. In these

studies, QUS made use of estimates of average scatterer diameter (ASD) and average acoustic concentration (AAC) from the ultrasonic backscattered power spectra and the ratio of periodic to randomly spaced scatterers (k parameter) from the envelope statistics. In each of these cases, the contrast in QUS images was due to both microstructural differences and impedance differences between tissues. Improvements in modeling, identification of scattering sources responsible for scattering, multiparameter analysis, and signal processing techniques have led to further improvements in QUS techniques.

The procedure is performed with the patient lie on her back and her arms raised above her head on the examination table. A clear gel is applied to the breast being studied to help the transducer make secure contact with the body and eliminate air pockets between the transducer and the skin. The sonographer (ultrasound technologist) or radiologist then presses the transducer firmly against the skin and sweeps it back and forth over the area of interest. For this exam no preparation is needed (Michael, 2009).



**Fig. (2-9)** a. and b. shows: Normal Ultrasound of Breast tissue (Michael, 2009).

## 2.5 MR Imaging

Magnetic resonance imaging (MRI) is a diagnostic procedure to view areas of the body without using x-rays. Magnetic field and radiowaves are used to detect the size and location of tumors using a large, donut-shaped magnet (Elizabeth, 2005). MRI of the breast offers valuable information about many breast conditions that cannot be obtained by other imaging modalities, such as mammography or ultrasound. MRI of the breast is not a replacement for mammography or ultrasound imaging but rather a supplemental tool for detecting and staging breast cancer and other breast abnormalities (Uematsu, 2001).

**MR imaging of the breast is performed to:** Assess multiple tumor locations, especially prior to breast conservation surgery, identify early breast cancer not detected through other means especially in women with dense breast tissue and those at high risk for the disease also evaluate abnormalities detected by mammography or ultrasound, distinguish between scar tissue and recurrent tumors, determine whether cancer detected by mammography, ultrasound, or after surgical biopsy has spread further in the breast or into the chest wall, assess the effect of chemotherapy, provide additional information on a diseased breast to make treatment decisions and determine the integrity of breast implants (Uematsu, 2001).

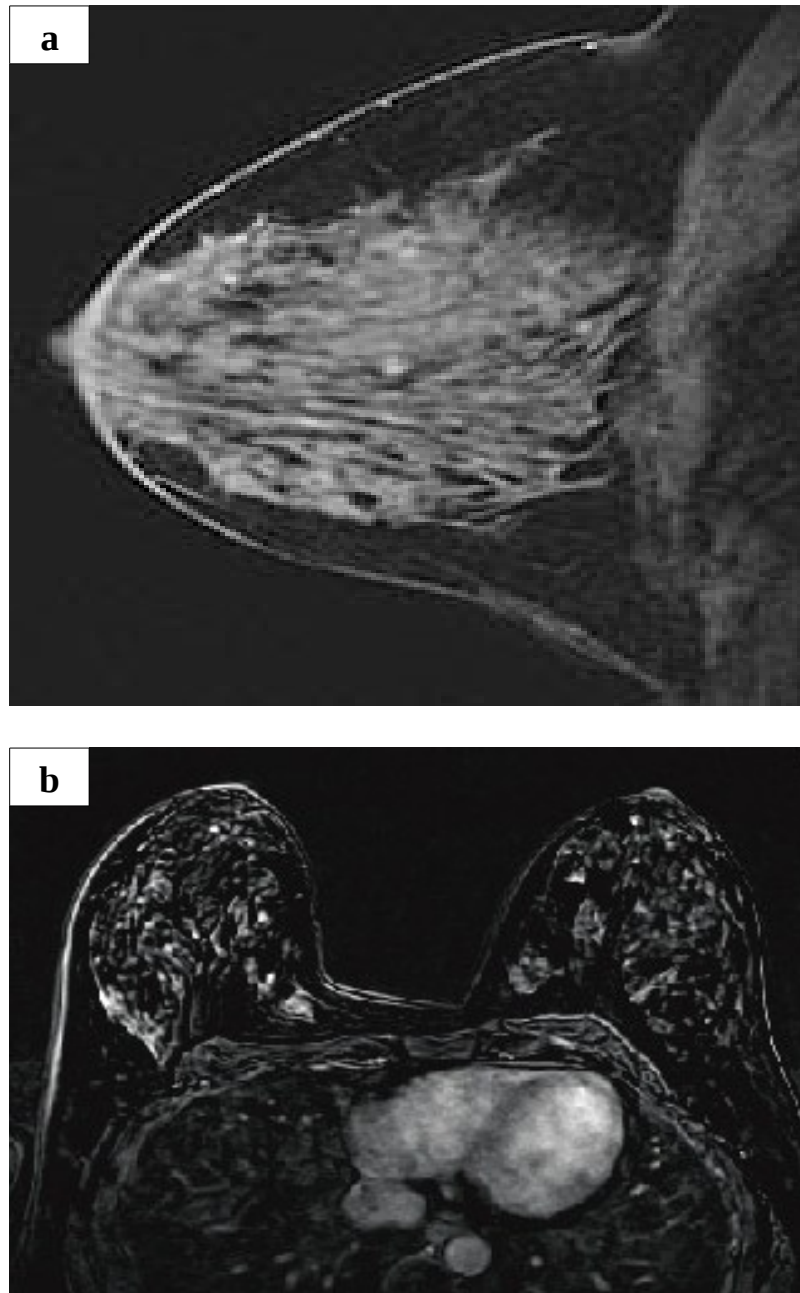
**Without contrast material, an MRI of the breast can show:** Breast tissue density, cysts, enlarged ducts, hematomas, leaking or ruptured breast implants and the presence of enlarged lymph nodes.

By comparing breast images taken before and after contrast material injection, an MRI exam can determine: If there are breast abnormalities and whether an abnormality looks benign (non-cancerous) or malignant (cancerous) also the size and location of any abnormality that looks malignant. For MR exam patient may be asked to wear a gown during the exam or she may be allowed to wear her own clothing if it is loose-fitting and has no metal fasteners, also all opacities and metals should be removed (Uematsu, 2001). For MRI of the breast, the patient lies on her stomach with both breasts hanging freely into a cushioned recess containing the signal receiver (also known as the breast coil). The entire bed on which

she is lying is advanced into the opening of the magnet (a tube-like machine that looks like a giant donut--open at both ends). The subject will be asked to lie still for up to 15 minutes at a time while the computer acquires the images; the total examination is made up of several scans, usually 5 to 15 minutes in length and the patient is usually in the magnet for 40-60 minutes (Uematsu, 2001). If a contrast material will be used in the MRI exam, a nurse or technologist will insert an intravenous (IV) line into a vein in patient's hand or arm. A saline solution will drip through the IV to prevent blockage of the IV line until the contrast material is injected. Patient will be moved into the magnet of the MRI unit and the radiologist and technologist will leave the room while the MRI examination is performed. If a contrast material is used during the examination, it will be injected into the intravenous line (IV) after an initial series of scans. Additional series of images will be taken following the injection. MRI exams generally include multiple runs (sequences), some of which may last several minutes. The imaging session lasts between 30 minutes and one hour and the total examination is usually completed within an hour and a half. MR spectroscopy, which provides additional information on the chemicals present in the body's cells, may also be performed during the MRI exam and may add approximately 15 minutes to the exam time (Uematsu, 2001).



## Normal MR Breast Images



**Fig. (2-10)** a. and b. shows: Normal MRI breast tissue (Uematsu, 2001).

## **2-5-1 Dynamic Contrast Enhanced (DCE) MRI**

The capabilities of MR imaging in breast imaging have been investigated since the 1970s. With the introduction of contrast agents and the first encouraging results of contrast-enhanced MR imaging in the 1980s it emerged as a promising modality for breast diagnostics (Riham, 2009).

Breast cancer imaging represents an important application of Magnetic Resonance Imaging. In addition to providing anatomical images without the need for ionizing radiation, MR is especially well suited for breast cancer imaging for another reason: contrast Magnetic resonance is capable of providing a wealth of information through the use of various contrast mechanisms. One method for controlling the contrast in MRI is to change various aspects of the imaging pulse sequences. However, in breast cancer imaging, very good contrast is desirable to not only detect, but diagnose the cancer (Bahri, 2008).

In breast cancer imaging, a method known as dynamic contrast enhanced magnetic resonance imaging is often employed. In DCE-MRI, a contrast agent, most often Gadolinium (Gd) is used to differentiate cancerous tissue and normal tissue. This method has several key features which make it well suited for breast cancer imaging. First, contrast agents can be designed to preferentially go to cancerous tissue. Therefore, an increase in image brightness post-contrast injection should correspond to a tumor. Secondly, recent advances in DCE-MRI allow for the simultaneous acquisition of high spatial resolution and high temporal resolution data sets. However, this is still a challenge (Bahri, 2008).

As with other modalities of perfusion imaging, DCE-MRI uses repeated imaging to track the entrance of diffusible contrast agents into tissue over time. A paramagnetic contrast agent, gadolinium-DTPA, is injected intravenously circulates through the body and diffuses over time into the extravascular extracellular space. As the mean contrast agent concentration within a voxel increases, the signal intensity from that voxel increases. From the known properties of the imaging sequences it is possible to convert the relative signal increase into a quantitative measure of contrast agent over

time in tissue. From these curves we can obtain semiquantitative analogs of blood flow (Uematsu, 2001).

The most widely used form of DCE-MRI analysis is the assessment of the type of time-signal intensity curve (i.e. kinetic curve) by categorizing the washout pattern of a gadolinium contrast agent. These patterns are classified as type I, persistently enhancing (progressive), which is suggestive of benignity; type II, plateau type, which has an intermediate probability for malignancy; and type III, washout type, which is indicative of malignancy (Kinkel, 2000). The washout patterns are typically assessed qualitatively, with intra- and interobserver variability reported to vary widely from 0.27 to 0.80 (Stoutjecdijk, 2005).

Based on changes in signal brightness, doctors can track changes in contrast agent concentration, which in turn provides information on the presence of tumors. As an example of pre and post contrast images, and how they might be used to determine the presence of cancer (Bahri, 2008).

To see the relationship between the increase in signal intensity and the presence of particular tumor types, the figure below shows how several tumor types respond to contrast.

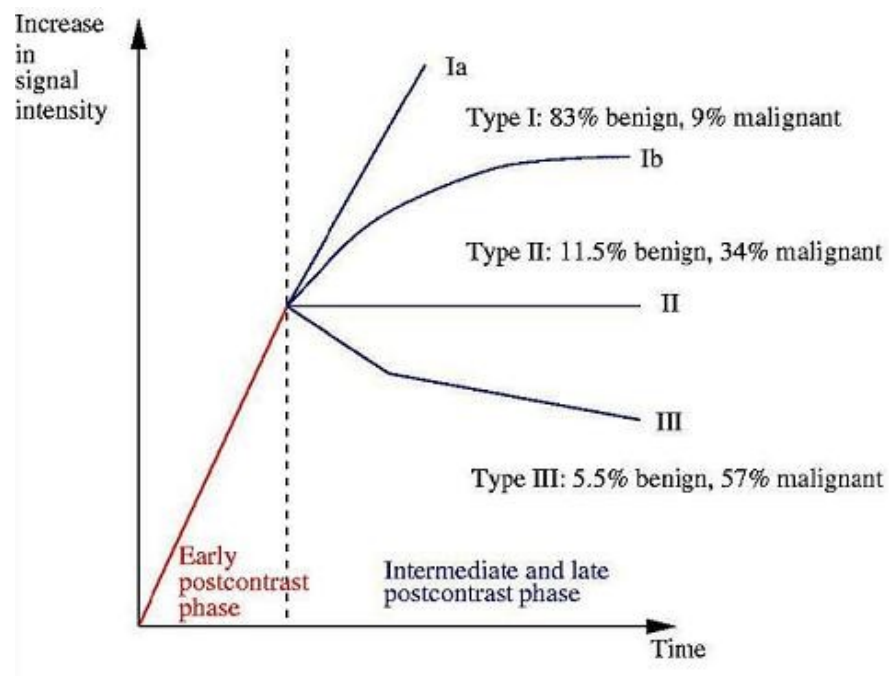


Fig. (2-11) shows how several tumor types respond to contrast (Bahri, 2008). normal

### Dynamic Enhancement Patterns

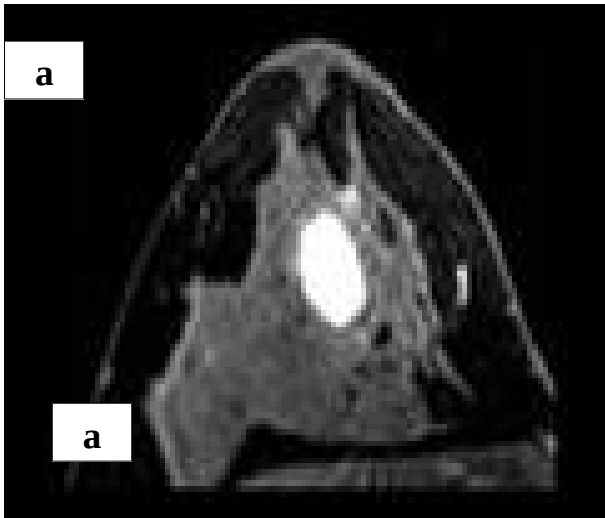
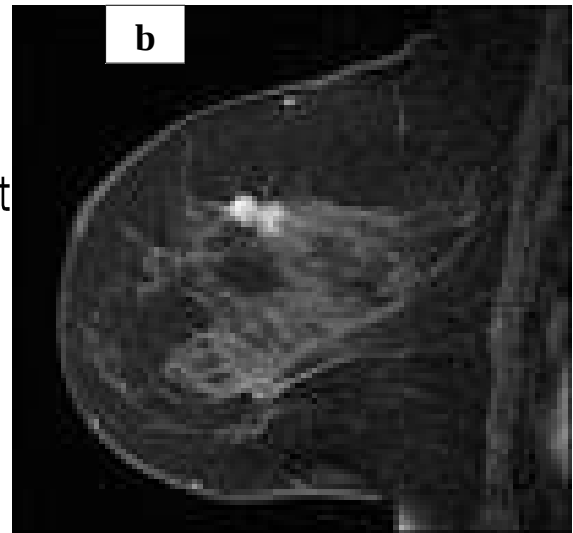


Fig. (2-12) Shows: **a.** (DCE MR) Benign Breast Lesion and **b.** Benign progressive Enhancement Pattern (Elizabeth, 2005).



washout  
normal

Fig. (2-13) shows: **a.** (DCE MR) Malignant Breast Lesion and **b.** Malignant Enhancement Pattern (Elizabeth, 2005).

### 2-5-2 MRI Diffusion Technique

Diffusion MRI is a magnetic resonance imaging (MRI) method that produces in vivo images of biological tissues weighted with the local microstructural characteristics of water diffusion. The field of diffusion MRI can be understood in terms of two distinct classes of application—diffusion weighted MRI and diffusion tensor MRI. Diffusion tensor imaging (DTI) is a magnetic resonance imaging (MRI) technique that

enables the measurement of the restricted diffusion of water in tissue in order to produce neural tract images instead of using this data solely for the purpose of assigning contrast or colors to pixels in a cross sectional image. It also provides useful structural information about muscle—including heart muscle, as well as other tissues such as the prostate (Evelyn, 2007). In diffusion weighted imaging (DWI), each image voxel (three dimensional pixel) has an image intensity that reflects a single best measurement of the rate of water diffusion at that location. The mean or average diffusivity in tissue is quantified by an index called the Apparent Diffusion Coefficient (ADC). DWI is a modification of regular MRI techniques, and is an approach which utilizes the measurement of Brownian motion of molecules. Regular MRI acquisition utilizes the behavior of protons in water to generate contrast between clinically relevant features of a particular subject. The versatile nature of MRI is due to this capability of producing contrast, called weighting. In a typical  $T_1$ -weighted image, water molecules in a sample are excited with the imposition of a strong magnetic field. This causes many of the protons in water molecules to precess simultaneously, producing signals in MRI. In  $T_2$ -weighted images, contrast is produced by measuring the loss of coherence or synchrony between the water protons. When water is in an environment where it can freely tumble, relaxation tends to take longer. In certain clinical situations, this can generate contrast between an area of pathology and the surrounding healthy tissue (Merajver, 2004).

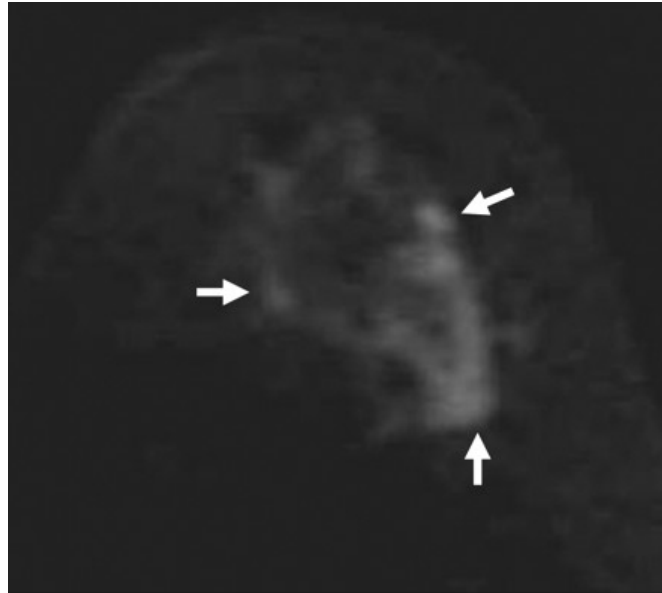
In biological tissues, microscopic water molecular motion is induced by both intravascular water movement (flow) and an extravascular component (diffusion) (Philadelpho, 2011).

With respect to the extravascular component, the state of the extra cellular space is the most important factor that regulates diffusion. If a tissue is made up of tightly packed cells, as occurs in a malignant tumor, the extracellular space is reduced, and diffusion of water is decreased. This phenomenon results in a higher DWI signal intensity, restricted signal intensity on the ADC map, and a lower ADC value. In contrast, in benign lesions in which the cells are more separated, the extracellular space is larger, diffusion of water is less restricted, and the ADC value is higher (Philadelpho, 2011).

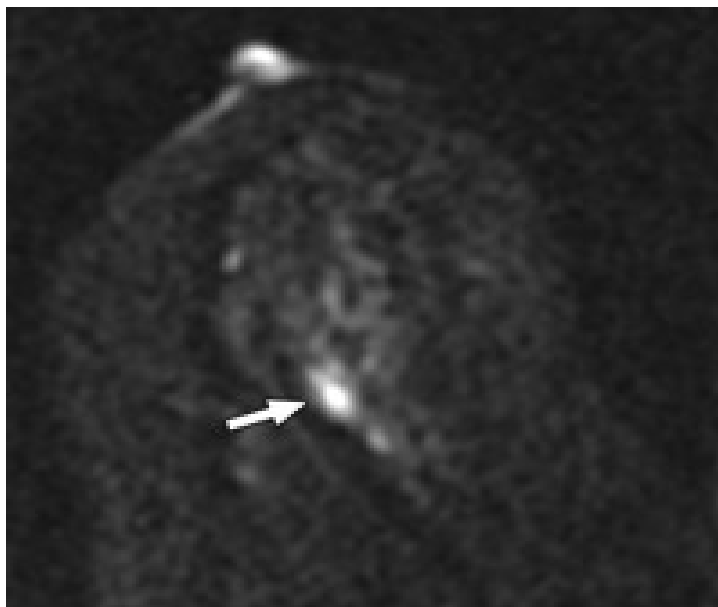
Tumor cellularity is inversely correlated with the ADC value, and malignant breast tumors exhibit higher cellularity and lower ADC values than benign breast tumors.

For clinical MR imaging scanners, the diffusion sensitivity can easily be altered by changing the parameter known as the b-value. Diffusion images are produced using at least 2 different b-values, and the loss of signal between these images is proportional to the amount of diffusion. Images acquired with low b-values are less diffusion-weighted because they use less of a gradient. The diffusion sensitivity is also more affected by microperfusion when low b-values are used, which leads to higher ADC values. On the other hand, high b-value images are strongly-weighted. Highlighting signals from malignant tumors and eliminating signals from normal tissues, but have a lower SNR and, consequently, more image distortion (Philadelpho, 2011). In DWI the normal breast gland has a high signal in images acquired with low b-values and low signal in images acquired with high b-values. Ideally, the background signal for the breast gland should be suppressed to emphasize the tumor signal (Philadelpho, 2011). A trend towards a decreased ADC has been observed during the second week of the menstrual cycle, and an increased ADC during the final week before menstruation (Philadelpho, 2011).

Variations in the ADC occur in the breast as a result of normal or normal hormonal fluctuations associated with the menstrual cycle. The reduced ADC in the second week is correlated with reduced water content of the breast, and the increased ADC during the week before menstruation has been attributed to increases in secretion activity, stromal edema, and water volume in the extracellular matrix. In women with less dense breasts, the ADC values for breast tissue may be artificially reduced as a result of partial volume effects of fat tissue. The mean ADC values in normal breast tissue vary from  $1.51 \times 10^{-3}$  to  $2.37 \times 10^{-3} \text{mm}^2/\text{s}$  for sequences acquired with b-values ranging from 0 to  $1074 \text{s}/\text{mm}^2$  (Philadelpho, 2011).



**Fig. (2-14)** shows: False-negative case at DW imaging in 64-year-old woman with a diagnosis of invasive ductal carcinoma (Reiko, 2010).



**Fig. (2-15):** MRI-detected lesion (*arrow*) shows reduced diffusivity with hyperintensity on diffusion-weighted image. MRI-guided biopsy determined lesion to be benign fibrocystic change with hyperplasia, with no evidence of atypia or malignancy (Savannah, 2009).

### 2-5-3 MR Spectroscopy

(MRS / MRSI - Magnetic Resonance Spectroscopic Imaging) A method using the NMR phenomenon to identify the chemical state of various elements without destroying the sample. MRS therefore provides information about the chemical composition of the tissues and the changes in chemical composition, which may occur with disease processes.

Although MRS is primarily employed as a research tool and has yet to achieve widespread acceptance in routine clinical practice, there is a growing realization that a noninvasive technique, which monitors disease biochemistry can provide important new information for the clinician (Magnetic Resonance-Technology Information Portal, 2003).

The underlying principle of MRS is that atomic nuclei are surrounded by a cloud of electrons, which very slightly shield the nucleus from any external magnetic field. As the structure of the electron cloud is specific to an individual molecule or compound, then the magnitude of this screening effect is also a characteristic of the chemical environment of individual nuclei.

In view of the fact that the resonant frequency is proportional to the magnetic field that it experiences, it follows that the resonant frequency will be determined not only by the external applied field, but also by the small field shift generated by the electron cloud. This shift in frequency is called the chemical shift. It should be noted that chemical shift is a very small effect, usually expressed in ppm of the main frequency. In order to resolve the different chemical species, it is therefore necessary to achieve very high levels of homogeneity of the main magnetic field  $B_0$ . Spectra from humans usually require shimming the magnet to approximately one part in 100. High resolution spectra of liquid samples demand homogeneity of about one part in 1000 (Magnetic Resonance-Technology Information Portal, 2003).

In addition to the effects of factors such as relaxation times that can affect the NMR signal, as seen in magnetic resonance imaging, effects such as J-modulation



or the transfer of magnetization after selective excitation of particular spectral lines can affect the relative strengths of spectral lines.

In the context of human MRS, two nuclei are of particular interest - H-1 and P-31. (PMRS - Proton Magnetic Resonance Spectroscopy) PMRS is mainly employed in studies of the brain where prominent peaks arise from NAA, choline containing compounds, creatine and creatine phosphate, myo-inositol and, if present, lactate; phosphorus 31 MR spectroscopy detects compounds involved in energy metabolism (creatine phosphate, adenosine triphosphate and inorganic phosphate) and certain compounds related to membrane synthesis and degradation. The frequencies of certain lines may also be affected by factors such as the local pH. It is also possible to determine intracellular pH because the inorganic phosphate peak position is pH sensitive (Magnetic Resonance-Technology Information Portal, 2003).

If the field is uniform over the volume of the sample, "similar" nuclei will contribute a particular frequency component to the detected response signal irrespective of their individual positions in the sample. Since nuclei of different elements resonate at different frequencies, each element in the sample contributes a different frequency component. A chemical analysis can then be conducted by analyzing the MR response signal into its frequency components (Magnetic Resonance-Technology Information Portal, 2003).

During frequency encoding, fat protons precess slower than water protons in the same slice because of their magnetic shielding. Through the difference in resonance frequency between water and fat, protons at the same location are misregistered (dislocated) by the Fourier transformation, when converting MRI signals from frequency to spatial domain. This chemical shift misregistration cause accentuation of any fat-water interfaces along the frequency axis and may be mistaken for pathology. Where fat and water are in the same location, this artifact can be seen as a bright or dark band at the edge of the anatomy. Protons in fat and water molecules are separated by a chemical shift of about 3.5

ppm. The actual shift in Hertz (Hz) depends on the magnetic field strength of the magnet being used. Higher field strength increases the misregistration, while in contrast a higher gradient strength has a positive effect. For a 0.3 T system operating at 12.8 MHz the shift will be 44.8 Hz compared with a 223.6 Hz shift for a 1.5 T system operating at 63.9 MHz (Magnetic Resonance-Technology Information Portal, 2003).

Chemical shift imaging (CSI) is an extension of MR spectroscopy, allowing metabolite information to be measured in an extended region and to add the chemical analysis of body tissues to the potential clinical utility of Magnetic Resonance. The spatial location is phase encoded and a spectrum is recorded at each phase encoding step to allow the spectra acquisition in a number of volumes covering the whole sample. CSI provides mapping of chemical shifts, analog to individual spectral lines or groups of lines.

Spatial resolution can be in one, two or three dimensions, but with long acquisition times of full 3D CSI. Commonly a slice-selected 2D acquisition is used. The chemical composition of each voxel is represented by spectra, or as an image in which the signal intensity depends on the concentration of an individual metabolite. Alternatively frequency-selective pulses excite only a single spectral component (Magnetic Resonance-Technology Information Portal, 2003).

There are several methods of performing chemical shift imaging, e.g. the inversion recovery method, chemical shift selective imaging sequence, chemical shift insensitive slice selective RF pulse, the saturation method, spatial and chemical shift encoded excitation and quantitative chemical shift imaging.

Chemical shift depends on the nucleus and its environment and is defined as nuclear shielding / applied magnetic field. Nuclei are shielded by a small magnetic field caused by circulating electrons, termed nuclear shielding. The strength of the shield depends on the different molecular environment in that the nucleus is embedded. Nuclear shielding is the difference between the magnetic field at the nucleus and the applied magnetic field (Magnetic Resonance-Technology

Information Portal, 2003).

Chemical shift is measured in parts per million (ppm) of the resonance frequency relative to another or a standard resonance frequency

The major part of the MR signal comes from hydrogen protons; lipid protons contribute a minor part. The chemical shift between water and fat nuclei is about 3.5 ppm (~220 Hz; 1.5T). Through this difference in resonance frequency between water and fat protons at the same location, a misregistration (dislocation) by the Fourier Transformation take place, when converting MR signals from frequency to spatial domain. This effect is called chemical shift artifact or chemical shift misregistration artifact (Magnetic Resonance-Technology Information Portal, 2003).

There are two types of MRS: either SVS: Single Voxel Spectroscopy which receives the spectrum from a single voxel only, or spectroscopic imaging (CSI: Chemical Shift Imaging) which measures spectra in projection (1D), on a slice (2D) or a volume (3D).

-Adapted data processing software.

-An adapted radiofrequency system (in the resonance frequency of the studied nucleus) (Pohmann, 1997).

-Analysis of the differences in metabolite resonance frequency can only be performed in the presence of a highly homogenous magnetic field. A heterogeneous magnetic field leads to resonance frequency dispersion, spreading out the peaks or even causing them to disappear into the background noise.

Prior to any MRS acquisition, the magnetic field is homogenized (shimming) in the region of interest. The bigger the region, the harder it is to homogenize the magnetic field throughout. Close to the bone, calcifications or hemorrhagic zones, spectroscopic quality will be poorer due to perturbations in the field generated by the differences in magnetic susceptibility compared to soft tissue. The precession frequency of the water must be optimized to adequately suppress the water peak, using selective frequency pulses and dephasing gradients (Pohmann, 1997).

The other problem with MRS concerns the weak signal-to-noise ratio. This entails multiplying the number of measurements (NSA) and limits spatial resolution (voxel of a minimum volume of roughly 3.5 cm<sup>3</sup> i.e. of dimensions 1.5 x 1.5 x 1.5 cm). Spectrum quality is evaluated according to two main criteria:

- Signal-to-noise ratio (height of metabolite peaks in relation to background noise)
- Spectral resolution (peak width, which determines whether the different metabolites can be separated).

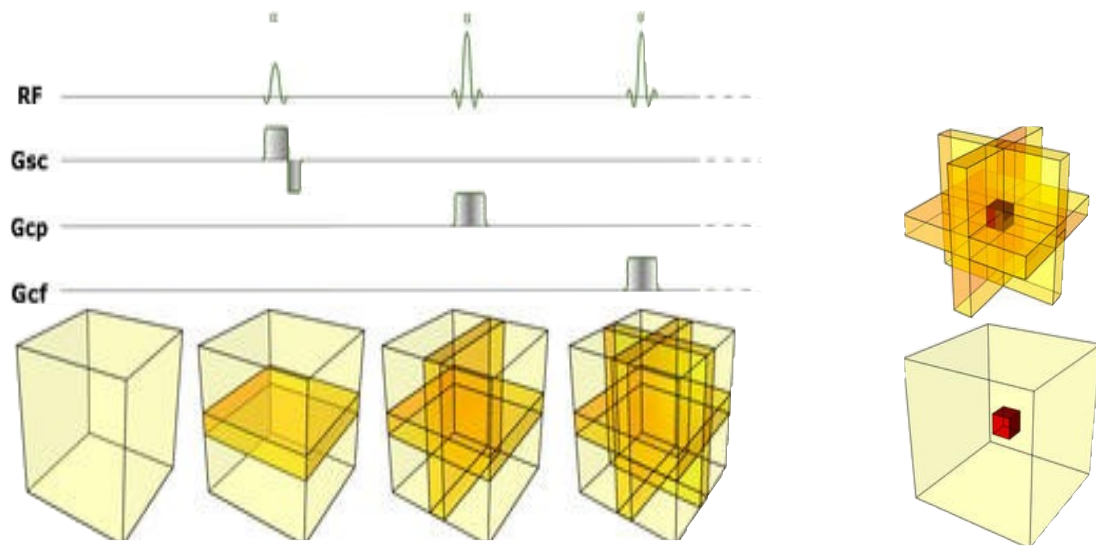
Spectral resolution will depend on the homogeneity of the magnetic field B<sub>0</sub> and on digital resolution, i.e. the precision with which the signal is sampled, determined according to sampling time ( $T_e = 1/F_e$ ) and the total number of points measured. In SVS, the signal is received of a volume limited to a single voxel. This acquisition is fairly fast (1 to 3 minutes) and a spectrum is easily obtained. It is performed in three steps:

-Suppression of the water signal: the quantity of hydrogen nuclei in the water molecules in the human body is such that the water peak at 4.7 ppm “drowns” and masks the spectroscopic signal from the other metabolites. It is therefore vital to suppress the water peak to observe the metabolites of interest.

-Selection of the voxel of interest

-Acquisition of the spectrum, for which two types of sequence are available (PRESS: Point-RESolved Spectroscopy, STEAM: STimulated Echo Acquisition Mode). The most commonly used method to suppress the water peak is CHESS (CHEMical Shift Selective). CHESS consists in applying three couples (90° RF pulses + dephasing gradients) in each spatial direction. The bandwidth of these RF pulses is narrow and centered on the resonance frequency of the water peak in order to saturate the water signal and preserve the signal from the other metabolites (Pohmann,1997). Techniques applying a 180° inversion pulse with adapted TI, like those used in FLAIR and STIR sequences, can also be used to eliminate the water signal (WEFT: Water Elimination Fourier Transform) or suppress the fat signal in breast spectroscopy, for

example. In practice, CHESS is more commonly used than WEFT. The analyzed volume is selected by a succession of three selective radiofrequency pulses (accompanied by gradients) in the three directions in space. These pulses determine three orthogonal planes whose intersection corresponds to the volume studied. Only the signal of this voxel will be recorded, by selecting only the echo resulting from the series of three radiofrequency pulses (Pohmann,1997).



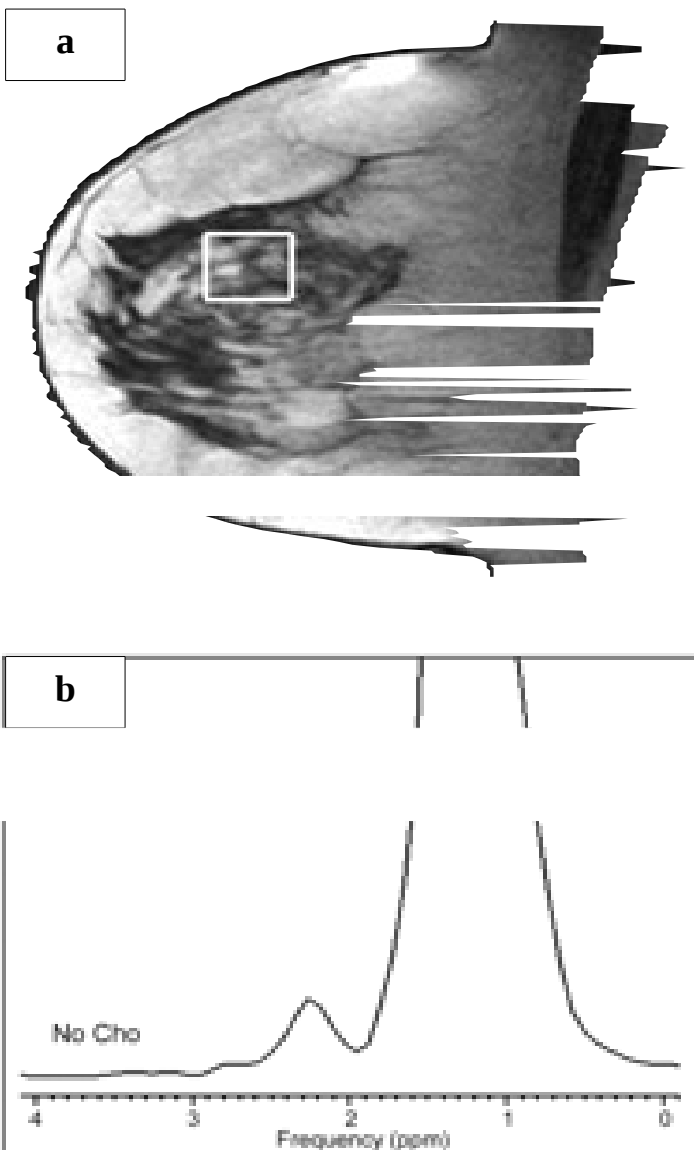
**Fig. (2-16)** shows: MRS Volume Selection

Metabolic imaging (CSI) consists in recording the spectroscopic data for a group of voxels, in slice(s) (2D) or by volume (3D). It is based on a repetition of STEAM or PRESS type sequences to which is added spatial phase encoding. The number and direction of phase encodings depend on the number of dimensions explored (1D, 2D or 3D), adding on to acquisition time. The duration of the sequence is equal to  $TR \cdot N_{ph1D} \cdot N_{ph2D} \cdot N_{ph3D} \cdot NSA$  ( $N_{phxD}$  number of phase encoding steps in direction  $x$ )<sup>[45]</sup>.

Extracting a quality spectrum from the signal involves several processing steps:

- Zero-filling or apodization filtering to complete the digitized FID signal
- Phase correction to obtain the real part of the spectrum (absorption spectrum)
- Baseline correction

Relative quantification: The results of the MRS are generally expressed as concentration ratios. Creatine peak or comparison with the healthy controlateral zone often serve as the reference values. Absolute quantification: Measuring the true concentration of metabolites by MRS comes up against several technical difficulties: the peak area has to be determined accurately then converted into concentration after calibration <sup>[45]</sup>.



**Fig. (2-17)** shows: (a) Sagittal non-fat-suppressed T1-weighted MR image  
(b) Magnified spectrum illustrates a high lipid (*Lip*) peak, but no choline (*Cho*) resonance peak is observed at a frequency of 3.2 ppm. *Lac* = lactate (Bartella, 2007).