A new generation FFDM / tomosynthesis fusion system with selenium detector

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ABSTRACT

A new generation of digital breast tomosynthesis system has been designed and is commercially available outside the US. The system has both a 2D mode and a 3D mode to do either conventional mammography or tomosynthesis. Uniquely, it also has a fusion mode that allows both 3D and 2D images to be acquired under the same breast compression, which results in co-registered images from the two modalities. The aim of this paper is to present a technical description on the design and performance of the new system, including system details such as filter options, doses, AEC operation, 2D and 3D images co-registration and display, and the selenium detector performance. We have carried out both physical and clinical studies to evaluate the system. In this paper the focus will be mainly on technical performance results.

Keywords: breast tomosynthesis, FFDM, image reconstruction, conic grid, selenium detector, AEC, dose, DQE, CNR.

1. INTRODUCTION

The Selenia Dimensions™ is the latest breast imaging platform from Hologic, which has been developed with a focus on the screening applications. Comparing with our early tomosynthesis system [1], this system exhibits several distinctive advantages. The system has both a 2D mode and a 3D mode to do either conventional mammography or tomosynthesis. Uniquely, it also has a fusion (or combo) mode that allows both 3D and 2D images to be acquired under the same breast compression, which results in co-registered images from the two modalities. In the screening mode, the system can finish a 3D scan in 3.7 second, which can greatly reduce the potential patient motion problem in tomosynthesis. The system can be conveniently configured into other imaging modes to address different needs, e.g., wider angle scan, or higher dose imaging. The detector is a new amorphous selenium image receptor that supports high frame rate and high DQE at low dose – both are highly needed for tomosynthesis. This paper is the first scientific presentation of the system. At this time clinical trials are underway to validate its clinical performance. We will describe the system on its capability, configuration, characterization and performance.

2. METHOD

2.1 Basic system configurations

The Selenia Dimensions is a breast imaging system that can do both 2D mammography and 3D tomosynthesis. It also serves as a basic platform for other advanced applications like up-right biopsy, dual energy imaging, and contrast enhanced imaging. The system has a tungsten target x-ray tube and five filter options, with three available to users in the commercial release. The 50 micron thick Rh and Ag filters are used in the 2D mode and a 0.7 mm thick Al filter is used for the 3D mode. The fourth filter is a 0.3 mm thick copper filter for dual energy imaging and contrast-enhanced imaging, and the fifth filter option is reserved for future new applications. The x-ray generator has a maximum tube current of 200 mA, and an energy range from 20 up to 49 kVp. Similar to Selenia™ FFDM system, a High Transmission Cellular (HTC) anti-scatter grid [2] is installed and used in the 2D mode. Since tomosynthesis does not use anti-scatter grid, the HTC grid can be automatically extracted out of the x-ray field in 3D mode. The distance of x-ray focal spot to detector’s selenium layer is 70 cm, and the breast support tray is 2.5 cm above the detector surface. The tube head gantry allows a

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maximum of 30 degree scan angle around the breast in 3D tomosynthesis mode, with the rotation center located at the center of detector surface.

Both 2D and 3D modes have automatic exposure control (AEC) with kVp switching capability, with x-ray technique optimized at each thickness. When the system operates in AEC mode, a low exposure scout shot at pre-selected kVp is delivered to the breast first, and then the total mAs is calculated for the main exposure(s). If the mAs exceeds certain limit, a higher kVp will be selected to keep the mAs in range. The geometrical calibration is done by scanning a fiducial phantom, and image reconstruction is carried out with specialized filtered backprojection method into a conic grid that matches the geometry of the 2D projection. The conic grid configuration allows precise co-registration and co-display of 2D mammogram and 3D reconstructed slices acquired from the combo mode. The selenium detector has a field of view of 29 cm x 24 cm, and an intrinsic pixel size of 70 μm. The readout time is about one second in full resolution mode, and the frame rate in the 2x2 binning mode is around 4 frames per second. The system can be conveniently configured into different modes for varying applications. The basic operation mode for screening application is selected to be 1.2 mGy for the 2D mode, and 1.45 mGy for the 3D mode, with 15 projections acquired over 15 degree scan angle in 3.7 seconds, with 2x2 binning readout. The gantry with x-ray tube continuously moves during a tomosynthesis scan, which allows a fast scan time to reduce potential patient motions during image acquisition. The system is also capable of running with a wider scan angle up to 30 degrees, more projections, and higher breast dose for diagnostic tomosynthesis applications.

2.2 X-ray characteristics and AEC dose curves

The x-ray beam quality and tube output of the system are shown in Figs. 1 and 2. The Rh and Ag filters are measured in 2D mode in which x-ray is delivered in one long exposure during imaging. But the Al filter is measured in 3D mode with pulsed x-ray exposures of 25 ms pulse width. Compared with the 0.05 mm Rh and Ag filters, the 0.7 mm thick Al only moderately filters the tungsten spectra. At 28 kVp, the half value layer (HVL) value of Al filter is less than those of both Rh and Ag filters. The exposure rate of the Al filter is about 90% and 50% higher than Rh and Ag filters, respectively. In tomosynthesis, the x-ray output is generally not filtered as heavily as that in conventional mammography in order to allow higher x-ray output. As a result, the x-ray pulse width can be kept very short in 3D mode, which helps to reduce focal spot blur and maintain adequate image sharpness in tomosynthesis.

![HVL plots of the system for three filters. The breast compression paddle is present in the beam path during measurements. The Rh and Ag filters are measured in 2D mode, the Al in 3D mode, with approximately 25 ms x-ray pulse width.](image)

Since x-ray tube and generator operate in pulsed mode in tomosynthesis, the x-ray beam quality of the system exhibits a phenomenon of so-called pulsing effect different from mammography. The origin and characterization of this phenomenon have been fully discussed previously [3]. In short, the measured exposure rate in 3D mode depends on the
ray pulse width. Even with the “effective mAs” method implemented in the system, that is, the mAs is integrated over the range of the mA time waveform from 80% to 80% of the peak magnitude of mA, the 3D mode still exhibits the pulsing effect at reduced magnitude. When the x-ray beam quality is measured with pulse width less than 25 ms, one can see the overall exposure rate curve of the Al filter will shift down slightly by a few percent. In practice, it is very important to account for the pulsing effect in studies related to x-ray dose calculation, contrast to noise ratio and image quality evaluation in tomosynthesis, whenever the x-ray pulse width is changed.

![Exposure rate plots at 63 cm away from the focal spot. The breast compression paddle is present in the beam path during measurements. The Rh and Ag filters are measured in 2D mode, and Al in 3D mode, with approximately 25 ms x-ray pulse width.](image1)

Fig. 2 Exposure rate plots at 63 cm away from the focal spot. The breast compression paddle is present in the beam path during measurements. The Rh and Ag filters are measured in 2D mode, and Al in 3D mode, with approximately 25 ms x-ray pulse width.

![AEC dose versus thickness plots of 2D and 3D in screening mode, measured with BR50/50 phantoms.](image2)

Fig. 3: AEC dose versus thickness plots of 2D and 3D in screening mode, measured with BR50/50 phantoms.

The AEC generated x-ray dose and detector pixel count plots are shown in Figs. 3 and 4, respectively. The doses of the system are calibrated to 1.2 mGy for the 2D mode and 1.45 mGy for the 3D mode with an ACR phantom, with the compression paddle display thickness set to 4.2 cm. In general, the AEC selected x-ray mAs in the 3D mode is below 100 mAs in order to keep the exposure time very short for each x-ray pulse. On the other hand, the mAs used in the 2D
mode is usually above 100 mAs in our system. In the 3D mode the Al filter is always selected for imaging; while in 2D mode the Rh filter is used for thin breasts (<7 cm) and the Ag filter is for thicker ones. Comparing the two dose curves in Fig. 3, the shapes are similar but there is an offset between them. However, there is a major difference in AEC-generated detector count plots between 2D and 3D modes shown in Fig. 4. The pixel count remains constant with each filter in 2D mode but it increases quickly as thickness goes up in the 3D mode. There are two contributing factors. 1) In 3D mode we do not use anti-scatter grid in our system. As breast thickness gets thicker, the scatters increase quickly [4]. Most of increase in detector count in 3D AEC count plot is due to scatter. 2) When we design the AEC technique for the screening mode, we tend to use higher kVp for thicker breast to lower the mAs and thus reduce the focal spot blur to the image. A higher kVp technique usually generates more detector counts.

Fig. 4: AEC detector pixel count versus thickness plots of 2D and 3D in screening mode, measured with BR50/50 phantoms

2.3. Combo mode and conic grid reconstruction

As mentioned early, the new system has a very unique combo mode that allows both 2D and 3D breast images to be acquired quickly under single breast compression, thus results in co-registered 2D and 3D breast images. Based on our studies at Hologic, we find that 2D conventional image has the advantage of better image sharpness while 3D tomosynthesis images can remove structure noises and improve mass lesion detection. The combo images of 2D and 3D together are expected to combine good calcification detection of 2D and good mass detection of 3D, to improve the detection sensitivity and to reduce the recall rate. We have developed novel reconstruction and display methods to help radiologists to maximize the advantage and benefit in reviewing breast images acquired in combo mode.

Mammography as a projection imaging method is imbedded with the magnification effect. Breast feature located at the upper part of breast appears larger and further away from the chest wall than that at the bottom, due to different magnification ratio at each height. This is an intrinsic phenomenon in mammography that radiologist are accustomed to. The magnification effect in mammography is illustrated in Fig. 5.1. There are two objects “a” and “b” of the same size being placed on the upper and the lower surface of a breast phantom, and they are also positioned physically right on top of each other. Due to the cone beam geometry and the magnification effect, the projection image of object “a” separates from object “b”, and it also appears larger in the mammogram in Fig. 5.1.

In tomosynthesis, the most common way of doing image reconstruction is to use a rectilinear grid [5, 6]. As Fig. 5.2 shows, the rectilinear grid is able to eliminate the magnification effect in mammography and the object “a” and “b” appear to have same size and location in slices “A” and “B”, respectively. However, when slices “A” and “B” of Fig. 5.2 are compared with the mammogram in Fig. 5.1, we find that features like “a” in slice “A” appear both smaller and shifted toward chest wall. This is a phenomenon that can be easily noticed when the combo-mode generated images are co-
displayed together, and the root cause is that 2D image has the magnification effect and 3D slices with rectilinear grid do not. Therefore there exist discrepancies in both shape and location of breast features between 2D mammogram and 3D slices with rectilinear grid method.

Fig. 5.1: Mammography has magnification effect intrinsically so that object “a” appears larger and further away from chest wall. Fig. 5.2: Rectilinear grid eliminates the magnification effect so that objects “a” and “b” have same size and locations in reconstruction slices. But the object in slice and in mammogram does not match in location and size anymore. Fig. 5.3: Conic recon grid retains the mag. effect so that objects “a” and “b” in slice “A” and “B” have the same sizes and locations to itself in mammogram, respectively. Skin line in slice “A” looks similar to slice “B” without any shrinkage and shifting.

In Fig. 5.2, though the object “a” in slice “A” and the object “b” in slice “B” appears to have the same size and location, but the apparent breast outline in slice “A” looks much smaller than that in “B”. It should be noted that, the apparent breast outline shown in each slice does not necessary define the physical location of skin line of breast inside the slice. The outline actually comes from the projection of the out-most skin lines of middle slices between “A” and “B” in the compressed breast. The actual boundary of skin line in slices “A” and “B” is smaller and inside the projection shadows, which can not be detected and visualized due to the poor depth resolution in tomosynthesis. This is why the breast appears much larger in slice “B” than that in “A”. Another related effect in tomosynthesis is that, when 3D slices in Fig.
5.2 are displayed in a cine mode slice-by-slice, one will find that the apparent shape of breast appears to become smaller, and shifted toward the chest wall direction as the reconstruction slice height increases.

We studied the above phenomenon and developed a new reconstruction method with a conic grid \(^\text{(7)}\) to address the problem, which is illustrated in Fig. 5.3. The pixel size of each slice in conic grid is scaled with its separation distance from the focal spot, which matches the cone beam geometry. As a result, the magnification effect in mammogram is retained in each tomosynthesis slice. The breast outline in the slice “A” and “B” in Fig. 5.3 are similar to the skin line in the mammogram in Fig. 5.1 without any size changes. Besides, the object “a” in slice “A” has identical size and location (in term of pixel numbers) as its correspondence in the mammogram. When the reconstruction slices from the conic grid method are displayed in a cine mode slice-by-slice, one will find that there is little shifting of breast outline toward and away from the chest wall.

It is important to note that, the combo 2D and 3D images with conic grid reconstruction method not only look similar when co-displayed, they are also mathematically co-registered perfectly. Any in-focused feature in a reconstruction slice of 3D images will have an exact pixel-to-pixel mapping of itself in the mammogram. The mathematically accurate co-registration of the same lesion in both 2D and 3D images is a powerful new tool in image analysis. For example, it may allow advanced CAD algorithm to be developed to utilize 2D and 3D images together to look for real lesions and reject false artifacts, therefore combine the advantages of 2D and 3D images of the same breast.

### 2.4 Detector performance

The selenium detector in the system has been redesigned to obtain significant performance improvements \(^\text{(8)}\). We are able to achieve a fast frame rate (up to 4 frames/s), a high detector gain with low electronic noise, and very small detector ghost and lag. The detector level DQE results are shown in Fig. 6 and 7, measured without the detector cover in the x-ray beam path. Both 2D mode (readout in full resolution) and 3D mode (readout in 2x2 binning) can reach a saturation DQE(0) of about 75%. In particular, the detector can reach the saturation DQE in 3D mode at an exposure level as low as only 1 mR. Good low dose DQE performance is important for tomosynthesis as it not only allows to reduce the overall breast dose in a tomosynthesis scan, but also allow to divide the total dose into more projections to reduce shadow artifacts and to improve overall image quality.

![Detector level DQE at different exposure levels, W/Rh, 28 kV, 2D mode, 0.07 mm pixel size.](image)

**Fig. 6:** Detector level DQE at different exposure levels, W/Rh, 28 kV, 2D mode, 0.07 mm pixel size.
Fig. 7: Detector DQE at different exposure levels, W/Al, 29 kV, 3D mode, 0.14 mm 2x2 binned pixel size.

Fig. 8: The DQE(0) versus exposure plots in 2D and 3D modes.

To better show the dependence of DQE on the exposure level, the DQE(0) is plotted against the exposure level for both 2D and 3D modes together in Fig. 8. Both 2D and 3D modes have similar saturation DQE(0) at higher doses as expected. Since the 3D mode operated with 2x2 pixel binning and with a higher amplifier feedback gain internally, the electronics noise in the 3D mode is relatively smaller. Fig. 8 shows that the detector can reach the saturation DQE at a smaller exposure level in 3D mode than that in the 2D mode.
2.5 Design considerations and dose mode developments

Tomosynthesis is still a developing field that demands complicated design and optimization efforts. Our R&D activities at Hologic have gone through different stages over years. Many of our studies are actually on similar technical topics as those published by others [5, 9, 10], e.g., spectra optimization, dose optimization, scan angle, view number, reconstruction method and so on. Besides we have also carried out extensive studies to access the clinical performance of the systems. In this section, we will discuss a few selected subjects among many others that have been found to be very important.

2.5.1 Scan time:

In general longer scan time is the enemy of tomosynthesis, as patient motion during imaging will destroy image sharpness and degrade image quality. The scan time of our systems was 20 seconds in the beginning, and had been reduced to 10 second, 5 second, and to the current 3.7 second over time. We have observed a steady reduction in the occurrence of patient motion in our clinical cases as the scan time gets shorter. We realize that for a clinical tomosynthesis system, the scan time should be kept very fast to reduce impact from potential patient motions during image acquisition. We choose to operate the gantry with continuous c-arm motion that allows faster scan time than the step-and-shoot scan method in tomosynthesis. Besides we also choose a relatively smaller scan angle of 15 degree in the screening mode to keep the scan time fast.

2.5.2 Focal spot blur:

For a tomosynthesis scan with a continuous c-arm motion, the x-ray focal spot will travel a distance of 1 to 2 mm during each pulsed x-ray exposure. The elongation of effective focal spot length along the scan direction will smear breast image on detector and reduce the image sharpness. To reduce the impact of focal spot blur, our system employs a powerful x-ray generator with a maximum of 200 mA tube current to shorten the total exposure time. Besides, we also use a relative thin Al filter (only 0.7 mm thick) to keep the tube output rate high in 3D mode. Because of the excellent low-dose DQE performance of the new selenium detector, we could divide the total dose into more projections, which help to reduce the x-ray dose and focal spot blur in each projection image. All these approaches together help to keep the focal spot blur small in our system for good image quality.

2.5.3 Image artifacts:

In tomosynthesis, each reconstructed slice is characterized with sharp and focused real breast features, together with blurry and out-of-focus shadow artifacts. The role of reconstruction filter is to enhance the real sharp features and simultaneously to suppress the shadow artifacts in each slice. The shadow artifacts can be best removed through filtering when they appear to be uniformly spaced and having similar intensities. Therefore in tomosynthesis the total dose is always divided equally among all projection views and the angular separation between projections are kept the same in a scan. We also find that the reconstruction filter works better with smaller angular step per view in a scan. So for the screening mode it is selected as 15 projections over 15 degree scan angle (1 degree per projection), which gives good reconstructed images with high image quality.

2.5.4 High dose mode:

For tomosynthesis scan with higher dose, in general it is not a good idea to simply increase the x-ray dose per view, as it would increase focal spot blur proportionally. A better approach is to increase the total view number as well as to increase the dose per view of the new selenium detector, we could divide the total dose into more projections, which help to reduce the x-ray dose and focal spot blur in each projection image. All these approaches together help to keep the focal spot blur small in our system for good image quality.

2.5.5 3D operation modes

Our system can support the following imaging modes in 3D tomosynthesis:

<table>
<thead>
<tr>
<th>Mode</th>
<th>Screening mode:</th>
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<tbody>
<tr>
<td></td>
<td>15 degree, 15 views, 1.0x dose (1.45 mGy), 3.7 second scan</td>
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</tbody>
</table>
Mode 2 ➔ Diagnostic mode - A: 15 degree, 21 views, 1.5x – 2x dose (2.2 – 2.9 mGy), 5.2 second scan
Mode 3 ➔ Diagnostic mode - B: 30 degree, 30 views, 1.5x – 2x dose (2.2 – 2.9 mGy), 7.4 second scan

3. EXPERIMENTAL RESULTS

System level evaluation studies have been carried out on a variety of topics. In this paper, we will only report results on CNR performance. We used BR50/50 slab phantoms of five thicknesses (3, 4.5, 6, 7.5 and 9 cm), and 0.1 mm thick Al foil as a contrast object. The phantoms were imaged with 1x dose for the mode 1 and 1.5x dose for the modes 2 and 3. The x-ray techniques for each thickness were selected by AEC. The kVp of the tests are kept the same at each thickness among three modes in the study. The CNR are calculated from 3D slices from the back-projection-only reconstruction method, with the filtering disabled [10].

Table 1: CNR results of three image modes, with imaging techniques and x-ray pulse width, at five thicknesses

<table>
<thead>
<tr>
<th>cm</th>
<th>mode 1</th>
<th>15 / 15</th>
<th>1x dose</th>
<th>mode 2</th>
<th>15 / 21</th>
<th>1.5x dose</th>
<th>mode 3</th>
<th>30 / 30</th>
<th>1.5x dose</th>
</tr>
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<tbody>
<tr>
<td></td>
<td>mAs</td>
<td>ms*</td>
<td>CNR</td>
<td>mAs</td>
<td>ms*</td>
<td>CNR</td>
<td>mAs</td>
<td>ms*</td>
<td>CNR</td>
</tr>
<tr>
<td>3</td>
<td>46</td>
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<td>38</td>
<td>8.29</td>
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<td>26</td>
<td>8.13</td>
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</table>

The x-ray techniques and CNR results of the study are summarized in Table 1. In particular, the x-ray pulse width of each test is given in the table for a discussion of pulsing effect in the study. The mAs of the mode 2 is approximately 1.5x that of the mode 1. The mAs of mode 2 and 3 are kept the same at each thickness. Looking at the CNR curves in Fig. 9, the plot of mode 2 and 3 are higher than the mode 1, since higher doses were used for modes 2 and 3. Though same mAs had been used at each thickness between mode 2 and 3, the CNR plot of mode 3 does not overlap with the mode 2 and is actually lower.

![Fig. 9: CNR results measured with the three dose modes, and at 5 thicknesses.](image-url)
To better illustrate the dependence of CNR on x-ray dose, the square of CNR ratio between modes are calculated and shown in Fig. 10. We find that the square of CNR ratio between mode 2 and 1 is approximately 1.5, which follows the dose difference (or the mAs difference) between the two modes. However, the ratio plot of the mode 3 over the mode 2 is around 0.9 only, suggesting that there are CNR losses in the mode 3 although the same mAs is used.

![Fig. 10: The square of CNR-ratio plots.](image)

When we compare the x-ray techniques between modes 2 and 3 in Table 1, we find that although the mAs are same, mode 3 has more projections and therefore smaller x-ray pulse width at four thicker thicknesses except the 3 cm. From discussions on the x-ray pulsing effect in section 2.2, we know that x-ray exposure of a short pulse width will deliver less x-ray than a long pulse under the “same” mAs. Therefore measurements with mode 3 might have used less x-ray than mode 2 at four thicknesses, which were responsible for the reduced CNR results. On the other hand, the detector counts in mode 3 are consistently lower than mode 2, because x-ray doses were divided into more projections. A lower detector count, in particular for the 3 cm measurement with mode 3, will be penalized more by the presence of the detector electronic noise. In conclusion, the CNR loss in mode 3 in this study is a mixed effect of both x-ray pulsing effect and the impact of detector noise under low exposure. The results of this study suggest that in tomosynthesis it is important to take the x-ray pulsing effect into account in dose and optimization studies, otherwise the experimental results could be misinterpreted. It is also a good practice to check the x-ray output with a dosimeter when pulse width varies in a tomosynthesis study.

4. DISCUSSIONS AND CONCLUSIONS

A new generation breast imaging system has been successfully developed and is currently commercially available outside the US. The new system carries several unique features compared to our early prototypes, which include: 1) The 2D and 3D fusion (combo) mode that allows conventional FFDM and tomosynthesis images are to be acquired under the one single breast compression. 2) A novel conic grid reconstruction technique that allows 2D and 3D breast images to be co-registered and displayed with matched geometry. The method can not only make the image review process convenient, but also makes further CAD development possible to combine 2D and 3D combo images to achieve better sensitivity and less false positive. 3) The new system has very short 3D scan time, which is less than four seconds, and will reduce patient motion during tomosynthesis scan. 4) The new system employs an improved selenium detector that allows high frame rate readout and good DQE performance under very low dose exposures. 5) Both 2D and 3D modes
are also equipped with AEC control with automatic kV switching capability. We have also carried out extensive evaluation studies to characterize the performance of the system at both the system level and the sub-systems levels. Clinical studies are also underway in order to validate its clinical performance.

REFERENCES


