Thank you Mike and good morning and welcome to all of you. It's a good reason to be here because it's raining outside. I hope I won't make you sleep here. We're gonna discuss the advances what we've seen in the past year. As you can see CT has been advancing quite rapidly and every year when we come back to this presentation, half of my slides are almost obsolete. I have to refresh all my slides to the new technology. Some of the topics which I am going to be covering within the time permitted, is the detector designs of the multiple-row detector CT, I'm going to call it as MDCT ranging from 4-32 section scanners. I also want to discuss briefly about the issue of pitch, the definition of the pitch and also what we call the geometric efficiency with multidetector CT, which is a very key factor, especially if one is scanning with very thin slices. Since CT is more diverted towards cardiac area, we're going to touch bases on the fundamentals of cardiac CT imaging and the basic of temporal and spatial resolution, we wanna touch how it has improved in the past few years and we're also gonna briefly talk about the next generation MDCT, such as 40 slice and the 64 section scanners. And also show some initial slides I got from the manufacturers about what they are working on the volumetric CT. Finally, we are gonna conclude what we have learned from today's lecture. CT has evolved considerably since it's evolution in 1972. We can consider there were two major evolutionary leaps occurred during the last decades. One was the spiral CT in the early 90’s and multiple-row detector CT in the late 90’s to present. The advances in spatial and temporal resolution helped transform CT from being just a transaxial cross-section imaging to a true, three-dimensional imaging modality. With the MDCT, the single row of detectors are being replaced with the multiple row of detectors in the Z direction. This automatically increased the scan volume per rotation. It also provided an opportunity to have multiple thin sections, enabling higher Z-axis resolution. This being one of the limitations of the single-row detector CT, so this really was an evolutionary leap in terms of the technology. Just to give you a brief description of differences between what the single-row detector CT and the multiple-row detector CT's all about, this cartoon kind of tries to explain the differences between the SDCT and the MDCT. Except for the number of rows of detectors in the Z direction, everything else is the same. In fact, in the MDCT, the x-ray beam is opened in the Z direction making the x-ray tube to be better utilized and in this example, there are a number of asymmetrial detectors I am showing here, and the number of thin sections per rotation is usually defined at the isocenter of the scanner and the dimension that you see as we talk about detectors, we’re gonna define at the isocenter, even though in reality the detector actual size is slightly larger, but for our convenience we usually define the section of the detector at the isocenter. In these type of scanners, we're able to obtain four sections per rotation compared to single section per rotation, enabling almost four times the volume coverage for the same scan time. So, this is the added advantage of obtaining a larger volume data set. However, all the detectors elements used in MDCT are very fundamental, here is an example of a detector design showing a uniform detector array having about 16 detector modules and in this case, the number of sections obtained per rotation is defined by the number of simultaneous channels of data available to be pulled out. So here in these cases, in the first generation so called MDCT, all of the manufacturers were able to do only four sections per rotation, because they had only four switching channels. So, irrespective of the number of detector modules, ie, actual physical detectors available, the numberof sections obtained per rotation are defined by the number of switching channels per rotation. Here's an example when a particular scan mode, for example, how a detector is configured and the data is pulled out for let's say 4 x 1.25 mm. In a situation of 4 x 1.25 mm, the x-ray beam is widened such that the four of the detector elements are absorbing the x-rays and because of the four switching channels, we are now able to reconstruct the data into four thin sections of 1.25 mm and so on. Similarly, the same scanner now acquired in a 4 x 2.5 mm, even though the number of detectors is still16 and the number of detector channels seen absorbing the x-rays are now increased to eight because of the limitation of the switching channels, it's still the data from two adjacent

switching channels are combined towards one switching channel, so thereby still enabling only four sections per rotation. So this is usually the scan mode for these early detectors; and they are all obtaining four simultaneous sections per rotation. Here's the description of the detector design for the four section MDCT scanner manufactured from all the major four players. We have classified them as the following: as a uniform set of detectors and also a second type called non-uniform detectors and third one called the hybrid detector design. As you can see here, each one slightly varied in the detector configuration. The uniform detector had 16 detector channels of the same dimensions. The non-uniform detectors had very thin detectors in the center and thick detectors in the adjacent row. The hybrid design accommodated both of these things together, had thin detectors in the center and adjacently thick detectors similar to that of uniform

detector designs. These as you can see, the maximum coverage per rotation is given under the total width of the detector design, and here it is about a range of 20 mm to 32 mm scan coverage. These were in the early 1998, so called first generation MDCT scanners, which were all able to limited to four sections per rotation. We have moved towards 16 section MDCT scanners enabling more thin sections. In fact, the MDCT technology is now, little bit more matured in terms of the majority of them are all 16 section scanners. And the way the 16 section scanners work is basically instead of four switching channels they enhanced the number of switching channels to switch 16 channels enabling simultaneously to pull the data from 16 of these detector modules enabling 16 scanner sections per rotation. Here's an example showing a detector design of a hybrid in nature, which has thin detectors in the center and thick detectors in the adjacent.

The way it works is, here's an example of all the manufacturer's detector elements in the 16 section MDCT scanners. The common feature, which we see here is migrating from four sections to 16 sections, all of the manufacturers have gone towards what we call it as hybrid detector design, which have thin detectors in the center of the scan region and thick detectors in the adjacent. So, it becomes a little bit more easier andless confusion in terms of the scan modes for these scanners, because the acquisition modes are now very simply either you acquire as thin section or as thick section. All of these are again limited to only 16 channels per rotation. Since most of these scanners can also scan half a second rotation per gantry, we can simultaneously get almost like 32 sections per second scan time. How does it work? Here's an example of different detector combinations resulting in possible section widths for reconstructed images. For this

particular detector module, let's say it has the 16 thin detectors of .75mm and four 1.5 mm adjacent. If you are acquiring this scan of .75mm collimation, means that acquiring the collimation of 16 x .75 collimation, the acquired data can be reconstructed into whatever slice section width and the section width has to be created and the acquired slice collimation cannot be less than 0.75 mm. So, it can range usually by greater than .75 anywhere from 1-10 mm reconstructed images. On some acquisition modes, you can also acquire a thicker slice thickness when you don't need the fine resolution, making it more faster and larger area to cover. In those situations, the data from two adjacent channels in the thinner section is combined towards as one channel. Thereby enabling again 16 of these 1.5 mm coverage resulting in reconstructed images of two, three, all the way to ten. These scanners are also able to acquire datah in the helical or spiral mode, but they are also capable to perform a sequential or axial mode, similar to the conventional scanner like “step and shoot”. So a lot of the head protocols are still done in the axial mode and also a lot of the cardiac protocols are done in terms of the sequential of axial mode. Right now, as we speak, the technology is rapidly moving so fast, every time we turn around and some emails we get, there's a new technology coming back. So, in RSNA there was announcement about the 32 section scanners and this is one of the early scanners we have been fortunate to have in our site. We had some experience working on this scanner and basically, there are minor differences between 16 and 32slice MDCT, in terms of general body imaging. The radiologists are not finding major improvement or something. But the major emphasis has been in the area of cardiac imaging and that's one of the reasons why you might see most of the manufacturers are now putting the label out on most of the scanners as cardiac CT and sending them to the cardiologists directly. Because a lot of the new applications are opening up in the area of cardiac area. The highest number of slices and higher thinnest number section scanners do provide a much nicer ability to scan the heart at a fast rate. This is the detector design of a 32 section scanner, from one of the manufacturers which we have currently installed in our site. This particular scanner has given us an opportunity to comparebetween the same manufacturer of 16 section scanners and 32 section scanner. As you can see here, in the 16 section scanner we had a hybrid detector design. We had 16 of the thin detector of 0.5 mm and thick detectors to the adjacent. They have changed the detector module now from this type to a more uniform thin detectors. In fact, the detectors physically have 64 small detectors of 0.5 mm. However, currently they can only acquire 32 thin sections. Right now the limitation has been how fast it can pull the data out of the scanners and the number of detector channels they have. The way it works here is like they double the switching channels, so that now they have like 32 switching channels, even though the detector module is 64, basically allowing it to upgrade to a 64 section scanner pretty soon. So they’re already claiming that by fall of this year, they're going to be marketing the 64 channel scanner of this particular manufacturer. The only difference here is they're going to double switching channels, as they already have physical detectors spliced into very thin detectors of .5 mm. So, if you acquiring at a 32 x .5 mm acquisition, you are basically covering only the center part of the detector region about 32 of these detectors are absorbed in the main x-ray acquisition and getting the scans, the x-ray which has reconstructed to 32 thin sections. The same scanner now can be acquired on 32 x 1 mm scans basically in the following way: as I showed in the 16 section scanner, the same thing applies to the 32 section scanner also. Although now you can acquire either on 32 of a very thin section or 32 of a thick section basically combining the data from the adjacent channels to combine as 1 mm sections and of course other scan modes are also currently available for 16, four sections on these scanners.

Again, these scanners can do both the helical mode and also sequential axial mode. What does this 32 section MDCT scanner provide? It's pretty much the same as 16, but double the number of sections. Basically it provides faster scan acquisition, because the gantry speed has been reduced, indeed it has been coming down, they're trying to come down as fast as possible. Right now the gantry speed ie., scan time per rotation is getting less than half a second. Scanners are

ranging from 370 msec to 400 msec per gantry rotation. And most of these scanners can also do what is called a partial scan, which is an attractive feature for the cardiac imaging and with the partial scan, scan time can go down to almost 0.25 sec or 250 msec. These scanners also enable to provide an isotropic volume data set, the holy grail of imaging to provide isotropic resolution for 3D imaging These scanners have very thin section ranging from .5 onwards, enabling this large volume data sets enabling reconstruction in a very fine resolution. It’s also providing a large scan volume in a very short time, which is critical in imaging trauma patients or in cardiac patients where the time is critical, so that you can cover a larger area in a very short period of time. And the main emphasis has been in the area of cardiac imaging, because a lot of new applications are opening with the cardiac imaging, such as the CTA competing head to head with

the cath angiography. Just to give an example, on some of the volume coverage in the scanners, here's an example to demonstrate showing single section scanner, , as much as, at the same with the 5mm scan slice thickness with the rotation time of 1 sec per gantry rotation. We were able to cover this particular region of volume data on a single-row detector CT, about 150 mm, which would take approximately 30 seconds. Now, with the arrival of the four section multidetector scanner, the section thickness coming down to a 3 mm let's say, with a pitch of 1.5 and rotation time cut down to half a second, now you are basically covering for the same scan time and scan region of 1000 mm or 100 cm. With the arrival of the 32 section scanner, basically with every other parameters keeping the same, you basically cut down the scan time to 10 sec and this reduction in the scan time makes a lot of improvement in terms of breadth hold technique for

imaging in the heart area or for like pediatric cases or in the trauma cases covering a larger area. Just to give you a flavor of what's the technology advance and this technology, the multidetector technology, especially with the 16 and 32 has also enabled a lot of the new application and some of the controversial application such as CT screening because now it's much faster for the sites to do the whole body screening at a reasonable rate. Pitch. When I'm talking about pitch I also put this word as the definitions and also the confusion arriving with the pitch. As we all know the concept of pitch came into use with the helical CT scanner. With the helical or spiral CT, we define pitch in order to understand how much of the table feed per gantry rotation, we transfer the table feed and the beam collimation and we also understand the concept of pitch being greater than one, less than one and all those things. However, with the arrival of the MDCT,

initially there was some confusion with the definition of the pitch even though some of it still now translates into this confusion, because there are two different definitions of pitch who have floated in this field. One was called the beam pitch, and the other one is called the detector pitch. This resulted in a lot of confusion of understanding the concept of pitch and it's relationship with image quality and also with respiratory radiation dose impact. However, fortunately in the past last year, all the manufacturers have commonly come to an agreement along with the IEC standard to define a common definition of pitch, which can be applicable to both the single-row detector CT and also multiple-row detector CT. That can be simply defined as the following: here is a situation where T is called the thickness, a single dash channel width, and I is defined as the table feed per rotation. Beam pitch is defined as the table feed divided by the width, means

the beam with as such. This definition is pretty much as the one which we all knew in terms of the SDCT. The detector pitch basically defined the table feed per rotation divided by the single dash channel width. That resulted in a lot of the confusion now the definition which has been uniformly accepted across the board is equal to the beam pitch and is defined as the table feed divided by and the product of the number of dash channels times the width of the single dash channel. Basically, this product is the same as how much of the x-ray has while opened up, the width of the x-ray beam. How does it translate? When it comes to here, what we have is like when we have a pitch greater than one, it implies extended imaging basically the coil has sprung out, there is some gap in anatomy and there is an impact on the reduction of the radiation dose and with some reduction to axial resolution depending on the pitch value. At the same time, a

pitch less than one implies there's an overlapping of the tissues enabling higher radiation dose to the patient with also providing some higher axial resolution. In general, the radiation dose to the patient has this definition. It has a direct relationship to the mAs per rotation and in most, related to the pitch factor. So the greater the pitch value, the lesser the radiation dose and vice versa. How does it translate into some of the now common scan modes which we see in the MDCT? Here I give some examples. Here's a table speed per rotation, here's a detector combination in scan mode, the detector combination, which we're usually selecting, and here's the detector of the helical pitch, which is even now some manufacturers still uses, and here is the actual pitch value pertaining to the common definition of the pitch. One of the scanners has this mode called HQ and HS, is basically is a pitch of three and a pitch of six in terms of the detector pitch, in

reality is a pitch of .75 and 1.5. So as you can see here, with most of the cardiac scans done these days are done at a very low pitch so directly to have a major impact on the radiation dose delivered to the patient, because there are some inherent limitations of the cardiac scans makes the protocol to be scanned at a very low pitch. Usually we see ranges from .25 to .4 pitch values in most of the cardiac protocols. With respect to the pitch, there is also one thing I would like to arrange, this one particular manufacturer, even though varying pitch results in increase or decrease of radiation dose to patient, however, in some scanners, especially of one particularly manufacturer's in the 16 section scanner, the image noise and the patient exposure is maintained independent of the pitch. They basically vary the mAs in proportion to pitch and they use the definition called as effective mAs. And the effective mAs is basically a ratio of the actual mAs

divided by the pitch factor. So one needs to understand or get in the back of the mind that when using the effective pitch values in some of these scanner. Because that can really throw off the understanding of the pitch because they're trying to settle the scan mode independent of the pitch value so they have folded the pitch into their setting and they use a concept called effective mAs. So depending on the scan region and depending on the defined scan, they vary the mAs to accommodate for the same image noise that's why. And the effective mAs is basically the relationship of the mAs pitch value. The concept of dramatic efficiency becomes very critical in MDCT because we are worried about all the statements that radiation dose is higher in MDCT as compared to other scanners and other things. How we can understand is one aspect is the geometric efficiency. When we say geometric efficiency this is defined as the amount of radiation excluded, such as the penumbra part excluded relative to the radiation collected by the detectors in forming an image. In the single-row detector CT, the penumbra was pretty much falling on the detector and didn't influence anything else. So this contributes both to the image and also patient dose in SDCT. However, in the MDCT, in order to have the same amount of x-ray flux in all the detectors, the penumbra has to be excluded, because otherwise there would've been much more noisy image of the edges of the scan, edge detectors compared to central detector. So basically what we use is a term called over beaming where this contributes to the patient dose, but not to the image and we call it this as overbeaming to ensure image quality. And penumbra as you all know is caused by the finite focus spot size. How does this geometric efficiency to a large extent be reduced when we are acquiring 16 and 32 section scanner; can be answered in the following way. The common feature is the geometric efficiency decreases with thinner sections and fewer detector elements. When you have fewer detector elements for the scan acquisition and you acquiring at a very thin section, the contribution from the penumbra also becomes a dominant factor when compared to the primary part. So, in this particular situation when you are acquiring a very thin section, you are having a stronger effect with the thin collimation, however, when we are acquiring a larger number of detectors, the penumbra effort becomes less dominant and the effect decreases with the thicker collimation. And with the 16 section scanner or with the 32 section scanner, which are acquiring large number of thin sections, we see this penumbra of geometric efficiency becoming less influencing on the radiation dose and this improves with thicker section or with more actual thin section. Right now the mantra in the collimation acquisition is to acquire thinner section and more of it at the same time thereby providing to a large extent this geometric efficiency problem However, that can be defined here. The dose in a four slice scanner grows markedly with dense collimation, but less so with 16 and 32 scanners. This is one of the penalties which we saw in the early multidetector CTs. On all of the manufacturers which were acquiring four section scanners, when they went to the scan mode of a very thin section as you can see here, say a 4 x .5, the radiation dose penalty was significantly higher compared to the same scanner acquiring at a thicker section because of the geometric efficiency factor. Here, the data are normalized to the 16 section for the 16 x .5, as you can see here there is not much of a significant difference between the 16 and 32 section scanners when you are acquiring of the .5 mm. Comparatively, if we acquire on a very thin section 4 x .5 you see a dose penalty almost doubles or triples in the thin collimation. The same thing applies even to the model scanner when you are acquiring in thin sections, that's why. The dose gross markedly with collimation and fewer activel detector elements, but less so for thicker collimation and four detectors. What I've shown here is like a comparison between a 32 section scanner and a four section scanner, normalized to the 4 x 4 acquisition as you can see here. Irrespective of the detector's design, if you are acquiring in very thin sections and since you are using only very few of the detectors, you are gonna have dose penalty . So, the word is, to use more detectors when you are using a very thin detectors such as here as you can see here, acquiring a 32.5 in this acquisition you pretty much see that the dose value, not much of a difference between the thicker collimation acquisition. So, the bottom line here is like if you're going to acquire a scan model, if you wanna acquire the thinnest collimation, better thing is use a lot more of actual detectors instead of having very few detectors. The clinical impact in terms of the CT as you all know has expanded in all directions. Just to give a list of applications and the list of applications keeps growing every day as we talk; cardiac imaging is one of the areas of which I am mainly interested. I wanna talk about some of the fundamentals of cardiac imaging. With the MDCT, the key is using cardiac imaging is fast imaging resulting in a high temporal resolution to stop cardiac motion. Fine imaging, the higher spatial resolution to resolve small lesions in any plane, radiation dose, which is a consequence of the CT imaging, which this as a cardiologist requires our medical physicist help to understand some of these concepts to optimize the protocols. When I say temporal resolution, the approach

taken in CT to stop the heartbeat is done in the following way. It's achieved either by prospective ECG triggering or odd retrospective ECG gating. What is prospective ECG triggering? This method of scanning is pretty much like the conventional axial scan, it does a partial scan or like a step and shoot scan, the patient is set up and the ECG's monitor and the scanner is set to turn on at the preset delay time from the RR peak when the ECG arrives at that point, the preset turns on the scanner and it acquires a date of what a partial time of the beam and the couch moves to the next position and the scanner waits for the next heartbeat and again at the preset delay time it acquires the scan again. This partial scan is usually about 180 degree past the flat angle, the time. This results in images resulting in a temporal resolution close to 200 to 200 msec depending on the gantry rotation speed. Here in this case, the radiation dose is

minimized because you are not acquiring data throughout the heartbeat, but partially. The drawback of this procedure is, of this is, only available to a very limited data set. If this limited data set is embedded with image noise because of the motion artifact, you pretty much end up doing the patient scan again. So that's where the trade off between the radiation dose and the limited data set. There's a more common approach which the cardiac protocols uses, ie the retrospective reconstruction. Among the retrospective reconstruction, there are two methods. One is called the partial reconstruction scan reconstruction, again similar to the prospective ECG triggering. You are having the x-rays turned on through the ECG cycle, you basically acquire continuously the data and then you reconstruct the data only part of the segment, depending on the heartbeat, you can define where to pick up the data set. And there is a second type of

reconstruction what we call the segmented reconstruction in which case, even though you are acquiring the data set for the whole period, you reconstruct different segments of the projection data from the same phase of the cardiac cycle at successive heartbeats, so basically to provide the same data set you choose the same data set from different cardiac cycles. As you can see, here is the difference between the two type of retrospective reconstruction, where in the x-ray is turned on throughout the scan period so you are basically acquiring the data through the cardiac cycle. Retrospectively, you come back and match with the ECG cycle of the patient and choose the area, preferably the diastolic region where is the minimal heart motion artifact, pick up the data there and reconstruct the images. This allows similar to the prospect of triggering, you have temporal resolution reaching to 200 to 250 msec with a radiation dose higher because you are

acquiring the data set throughout, but you have a good data set, in which case if you have some part of the cycle embedded with the motion artifact you can go back and reconstruct it. But the drawback here is like, now the data set is discarded, except for the part which we selected. In the segmented reconstruction data, again this is a prospective reconstruction, you are acquiring the data through the heart cycle. However, depending on the heartbeat and other things, we are basically choosing fraction of the data from different cardiac cycles and combining together to provide one image. This enables to provide, to cut down the temporal resolution even further to almost less than 100 msec comparatively to the best temporal resolution one can get is about half of the gantry rotation. So, right now they're trying to match with the EBCT, because of the compareability with all the cardiac CT's matching image quality with respect to the temporal

position you obtain with the electron beam CT. So we are getting around 100 msec scan. Here are some clinical images showing the same set of data with the half scan reconstruction, which has a temporal resolution of 250 msec. The reason why we want a higher temporal resolution is because in this one you see a much nicer image with much lesser motion artifact here when you have the segmented reconstruction with the higher temporal resolution of 105 msec, however, there will be some type of artifact problem in this type of segmented reconstruction sometime, because you are collecting data from different cardiac cycles. What are the factors influencing the temporal resolution? The following are the main factors, there are other factors, which influences temporal resolution, still gantry rotation speed defines the basic temporal resolution. Right now, we are able to get about .4 sec or even less than that for per gantry rotation and it

depends on the image reconstruction such as the prospective triggering or the retrospective gated imaging, and also the field of view and also post processing are algorithns to some extent influence the temporal resolution. Temporal resolution as of today depends on the gantry rotation, can range from .5 to .4 or less than that up to .37 sec with the 16 and 32 section scanner. We are able to obtain up to 105 to 250 msec, achieved through partial scans of subsequent data reconstruction and this improves with subsegment data reconstruction, but the spatial resolution decreases in subsequent data reconstruction. This is the reason why we do cardiac protocol with a very low pitch value is because the higher pitch produces a lot of the data gaps. Here is a situation showing acquisition, this is a table moving in a Z position. Here's a retrospective or a prospective, doesn't matter. If the table is moving much faster and you are acquiring partially, as

you can see for a higher pitch value there will be a data set gap in the data set. That's one of the reasons why most of these scanner cardiac protocols are all done at a very high overlaps. And also, the typical pitch values ranges from .2 to .4. In terms of spatial resolution, right now the gold standard is the coronary cath angiography procedures. Here the images obtained on the MDCT, they're trying to master the same level. Here is one published image showing one of the coronary arteries, pretty much the same, equal in size, so right now there are a number of studies going on to equate, to compare the image quality and the acquisition time and other things with respect to the coronary angiography. In terms of the spatial resolution, there is number of factors which influence the spatial resolution and is measured as the equivalent at half maximum of the slice sensitivity profile. The section width has to be larger than or equal to the section

collimation, it cannot be less than that. And this is affected by the collimation, the pitch value, the reconstruction algorithm and also the Z filter. Here's an example to show the effect of section width and also the reconstruction interval. As to the section width, if you are acquiring and reconstructing an image on a section with the 1 mm or 5 mm, the image quality improves if you have the entire section width with more overlaps. This is just to give you a comparison of the image quality improvement with a thin section, a dramatic implement in the spatial resolution and three-dimension imaging. With the section being thinner, that's one of the reasons why all the scanners are going to a very thin section acquisition. The effect of reconstruction kernels also may influence the spatial resolution and also the image noise. Here's just an example of one particular manufacturer's data, which we have here. These particular kernels are

considered as the smooth reconstruction kernels, medium kernels and the sharp kernels. As you go across these kernels in terms of the reconstruction, the image noise dramatically increases at the same time the spatial resolution increases. So there is a trade off between what you're going to see in the images with respect to the spatial resolution. When the spatial resolution is the endpoint, you reconstruct with a very sharp kernels, when the spatial resolution is not the endpoint, you want to compromise for the image quality, since the same raw data is reconstructed with smooth filter. With the computer reconstruction being so fast, the acquisition is done at the most thinnest section and then the reconstruction are done in all different corners. Here's an example of acquiring the effect of thin section versus noise. The bottom line is to acquire with very thin collimation, so that the raw data available will be the raw data at the very

thin collimation for high spatial resolution, but one can reconstruct into thick sections to improve the image noise. The raw data still exists so we can reconstruct either in the thin section or the thick section to improve image noise. Here's an example of showing a high contrast spatial resolution phantom, which we reconstructed which is enabled to show up to .5 mm spacing very nicely in the Z direction. A few more slides, I am just skipping to the new part. The axial resolution and spatial resolution right now ranges from 10 to 20 line pair per cm. The Z axis resolution is still around 7 to 10 line pair per cm. The goal is to match these two to get the complete true isotropic data sets. The advantage of the thin section again, in general, the thinner section has higher Z axis resolution and also improves partial volume. However, it requires higher tube current in order to reduce image noise, which leads to a higher radiation dose,

especially when you use very few detectors. Here is just to show a number of different vendors, three different vendors each of the respective vendor's data normalized to the 2.5 x 4 mm acquisition, which you see here. Each of these vendors respectively, when they have acquired a very thin collimation and with very few detectors, you have this dose penalty because of the poor geometric efficiency. So here are some of the images of the 16 section scanners, I'm gonna skip here to the next one and the same thing here. The one I want to show you in the next five minutes are the next generation scanners, which are coming out. All these data have been provided by the manufacturer to me so I have not done any scans on these scanners yet, but these are the detector modules which you are going to be seeing more of around fall of this year. All of these detectors like the major players, this particular player already has the 64 detector

modules in place even though they are requiring only 32, the point of this going to upgrade to number of switching channels so they will be able to do 64 of very thin slices. This particular manufacturer is going to introduce the 64 thin detectors so they will be able to acquire 64 thin sections of .625 mm size. And this particular manufacturer is going to be using 32 thin section detectors with four adjacent and they are going to be cleverly doing a switching of, I'm gonna show you how they do, they're gonna be also obtaining a 64 section per rotation. Here's an example of what the 16 section slice, these are data slice provided by the manufacturer. A comparison here is like in a light speed of a 16 section scanner to obtain a coronary CT angiography for a 20 sec scan, you are able to obtain this amount of scan volume of high resolution, the claim is none of us with the 64, they can basically quadruple the scan volume and

reduce the contrast medium to be used four times lesser and if you wanna cover a larger area, which is a major impact on breadth hold technique for some of this acquisition scans. So that is a major implement we're going to be seeing with the 64 section scanner. All of them are geared towards the cardiac imaging. One other manufacturer, Siemen's, is emphasizing what is called straton tube, which is a new thing which you might have seen in the advertisement. The straton tube is basically, it's kind of like a combining the electron beam technology into the x-ray tube so that in order to announce the x-ray tube capacity, they don't have any cooling inside, so they don't have moving part outside. Basically, the bottom line here is that the electron beam is magnetically deflected to the different spots on the rotating anode, so they're able to switch the focus spot very precisely almost like the electron beam CT, enabling very fine focus spot

switching. This will enable them to use still the 32 section scanner, but they will not have 64 DAS channels for the same detector module, so for every detector channel they gonna take two sections every time enabling them to have it overlap data and providing almost like .4 mm section. So even though the detector module will have only 32 detector modules, they will be able to obtain 64 DAS channel. We are looking forward to getting one of these scanners pretty soon in the next couple of months. The same thing, the other important applications in these cases with one short imaging, you can do a triple rule or they call this triple ruler acquisition. Not only the cardiac part, you're also able to do a number of aspects, such as looking for the pulmonary embolization in the lung field, because the raw data set is covering the whole chest area and you can also do this aortic dissection for much better utilization. Similarly here, right

now the big thing is the holy grail in imaging, what we call it as the Plaque characterization, we are now able to see plaques in some of these thin imaging. Now the next big thing is like how do you characterize the plaque, which are more susceptible to dislodging from the location and causing a heart attack, how to characterize it is a big thing. There's a lot of quantification being done in the MDCT and the same thing here with the multiple views enabling the data's for the physician to see in multiple aspects. Irrespective of the advances, there are some dose traps in the MDCT. I just want to bring this to your attention. One is the scanner geometry. The shorter distance between the x-ray tube and the isocenter usually it's a higher dose for identical mA setting compared to the single-row detector or the MDCT. So if a clinic is switching the protocol from a single-row detector to a multiple-row detector CT, they need to adjust the mA

accordingly, because if they use the same mAs, they're going to end up having a lot more radiation dose to the patient because of the fact of the scanner geometry because they are brought more closer. The second thing is the narrow collimation as I said. The decreasing geometric efficiency leads to increasing radiation dose, so the take home message is to acquire thin sections using more detectors and then decreasing the effect of geometric efficiency. The third dose trap is the ‘effective mAs’ concept. But probably now everybody knows about this concept of effective mAs, is basically is the ratio of the mAs per pitch should not look primarily at the mAs setting, but rather choose the proper mA, which reflects the patient dose as described by the CTDI volume, which is defined as CTDI divided by pitch. In fact, pretty much all the manufacturers are required to display this value on their console. So, as a physicist when we are

doing some testing, if we just arbitrarily set some value and see how the CTDI or the display number changes, we get a better understanding of all these parameters changes when any one of the factors changes. Because this CTDI values will automatically change if you change the filter or the beam filters, if you change the mAs, if you change the pitch value, it should reflect those things. The future. I thought like far, far away, it's actually not, it's only around the corner. Towards 256 rows and the flat panel detector; right now in theJune issue of Medical Physicis there is a paper from this one particular manufacturer describing the initial result of this 256 row detector. In the prototype, the 256 rows detector and also 1024 x 1024 flat panel; right now the cone beam is currently used in imaging, especially in the rotational angiography, especially with the arrival of the flat panel in the cardiovascular room, and also some of these image intensifiers,

we do rotate continuously and acquire the data and reconstruct it. And again in the radiation therapy, now we see all this cone beam CT using a flat panel, but it can depict only high contrast structure and the acquisition takes a few seconds. So that's the limitation with this rotational angiography. With this volumetric CT, you will be able to acquire the entire heart area in one rotation. That's the goal towards this imaging. There are certain limitations right now which are hindering this technology and everybody is working towards solving this. Some of the major limitations right now is the scatter radiation. The scatter radiation was not an issue in CT and that's one of the reasons why we had such good contrast resolution, because we were acquiring in a very thin collimation. Now with the beam widened quite much, we are gonna have what is called as the wider scatter acceptance angle, resulting in the scatter radiation, which is going

diminish low contrast resolution automatically, they require some type of grids into the field. Also right now, the other limitation is the wide bandwidth required for fast signal transmission. To pull the data very fast, you need a much wider band width and also to sort the data for all sort of reconstruction. The other thing, which is also kind of a rough guess, is the image noise increases with decreasing collimation. So, it is roughly related as $1/x^4$ with x is the voxel dimension. Just back up to the element of calculation, if the voxel dimension decreases from 1 mm to .5 mm, the image noise increases by 16 times. So that has to be compromised by increasing the radiation dose to the patient or to the detector module. So, the next goal is to decrease the voxel dimension from 1 mm to .2mm to almost bring it to the same level as the radiograph, the image noise is going to dramatically increase, so that right now one of the

limitations we are seeing in the flat panels and also in the 256 rows detector is the amount of radiation dose increase per rotation, will be one of the limitations and manufacturers are working towards this, it's around the corner. I'm gonna skip this high power x-ray tubes before I come back to the other part. I thought I would have more time to show these slides, I apologize. Let me come down to the end of this lecture. Briefly, we can list the number of advantages and there are also some disadvantages also in the MDCT. One of the advantages, like shorter scan duration, radius motion artifact, radius contrast medium and also enabling newer applications, such as perfusion imaging and so on. There is also enabling advantage the longer scan ranges enabling doing CT angiography and doing trauma patient in a much faster scan time and you're able to obtain a thinner section. The disadvantages on the other hand, is the increasing data load.

These days the chest CT's, we're getting anywhere from 400 to 800 images. So the new rethinking of how to analyze these images are coming into picture for radiologists that cannot see all these 800 images to pull a diagnosis, so a new method of fly through of reading the images or reconstructing the three-dimensional range, then seeing the images are becoming an issue. And there is also the image processing has been a disadvantage because it takes longer time, but with increasing capability of the computer reconstruction, that part has been handled. There is also the aspect of patient dose, when you go to a thin section there will be a concept of that, because radiation dose is gonna increase because you don't have to maintain the same image noise and of course, with the technology spreading to other areas where people are using it left and right, there will be a lot of inappropriate protocols being done and there’s gonna be definitely an increase in the radiation dose. So the bottom line is the increased coverage per rotation of the MDCT is approaching isotropic resolution and three-dimension imaging is becoming a reality. So, CT is most widely used, oh I have this slide to show you why the CT is also revolutionary in radiation therapy, also because of the dual modality imaging. I know with a lot of the majority of the people in the area of radiation therapy there will be a lot more implement in terms of the treatment planning, how the data will be combined with this function as a part of the imaging to have a better understanding of the follow up of the tumor and so on. In conclusion, the advances in MDCT technology has resulted in improved spatial resolution and temporal resolution, transforming CT from transaxial cross section imaging to a true three-dimensional imaging modality. For the past five years, no other imaging modality has matched the rapid advances of MDCT. I think this is a bold statement, but I can stand behind it because CT has advanced so fast in the past year. The bottom line is it's no longer your parents old CT scanner, so we have expanded quite a bit and I just want to end with these slides where we have gone from early conventional CT's to the modern CT images and this is a slide I just wanna show you with what the future we are looking forward as part of the CT we can record the data with the fax and the CR and the DR, and the fax system we can send the images across the globe immediately and different parts can be reviewed and so on. It's just wishful thinking for the next five years.Thank you.