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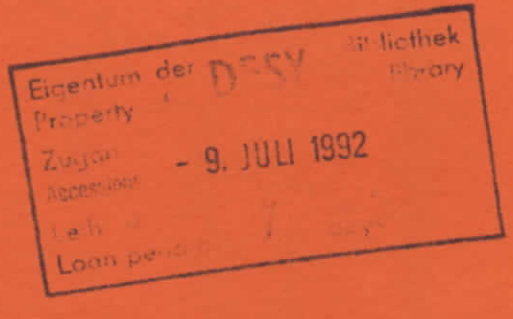

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CORONARY ANGIOGRAPHY USING SYNCHROTRON RADIATION - STUDIES IN HUMAN SUBJECTS WITH THE SYSTEM NIKOS II -

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Abstract

Worldwide at many synchrotron radiation laboratories different systems have been developed for Digital Subtraction Angiography (DSA) in energy subtraction mode (dichromography). Aim of the work is to visualize coronary arteries down to 1 mm diameter with an iodine mass density of 1 mg/cm^2, thus allowing non-invasive investigations by intravenous application of the contrast agent. The two images for subtraction are simultaneously taken with photon energies just below and above the iodine K-edge (33.17 keV) in a line scan mode.

In the Hamburger Synchrotronstrahlungslabor HASYLAB at DESY the system NIKOS II (NIKOS = Nicht-invasive Koronarangiographie mit Synchrotronstrahlung) consists of six main parts: the 20 pole wiggler HARWI, a two beam monochromator, a safety system, a fast scanning device, a fast low-noise two-line detector, and a computer system.

In 1990 the first patients were investigated with the system NIKOS II. 30 ml of contrast medium were injected within 2 s into the vena cava superior. The scan velocity was 9 cm/s corresponding to 5.6 ms per line. The resulting images have a size of 10.0 cm (horizontal) × 12.3 cm (vertical). The pixel size in the images is 0.5×0.5 mm^2. The radiation dose per scan (two energy images) was 25 mGy on the average.

There are two main problems with the intravenous investigations using DSA in energy subtraction mode for visualization of coronary arteries:
- The correct time T from the injection of the contrast material up to the beginning of the scan depends on the individual circulation time of the patient.
- In intravenous investigations the small structures of interest (coronary arteries) can be superimposed by large structures (aorta, left atrium and ventricle).

The images from the investigations are compared to angiograms, conventionally obtained. They have not yet the quality necessary for clinical routine but show the potential of the method if the system is optimized.

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Introduction

Myocardial infarction is one of the leading causes of death in the industrialised world. In most cases the reason for an acute myocardial infarction is an acute thrombotic occlusion of a coronary artery at the site of a pre-existent stenosis. Therefore, it is of considerable interest to identify stenoses in the coronary arteries before the incidence of an acute myocardial infarction in order to undertake preventions like bypass surgery or angioplasty.

At present, the only method to view stenoses in coronary arteries is invasive coronary angiography where a catheter is inserted via the arterial system up to the ostia of the coronary arteries and radiopaque iodine is injected. With modern X-ray devices this method gives excellent images[1] but has a certain risk for the patients with respect to morbidity (1.2 to 2.2%; 0.5% severe) and mortality (0.07 to 0.23%). That is why efforts are being made to develop a method without entering the arterial system by a catheter which then may be called "non-invasive". This method could be used to investigate ambulatory patients with mild symptoms of coronary heart disease, to monitor disease progression routinely, to perform follow-ups after surgery, to study effects of therapeutic interventions etc.

If the contrast medium is introduced into the brachial vein, it is diluted by a factor of at least 40 by the time it enters the coronary arteries. Assuming injection of a bolus of 10 ml of commercially available iodine solution (370 mg/ml) within 1 s, the concentration in the coronary arteries would be about 10 mg/ml, corresponding to a mass density of 1 mg/cm^2 in a coronary artery of 1 mm diameter. Imaging of mass densities of this order is not possible by classical X-ray techniques.

For non-invasive investigations as well as for invasive coronary angiography the Digital Subtraction Angiography (DSA) is used for contrast enhancement of the images. In DSA two images with a large difference in the iodine contrast and practically no difference in the contrast from all other structures, e.g. bone and soft tissue, are subtracted, thus strongly enhancing the signal from iodine. In the DSA methods used in clinical studies the two images for subtraction are produced in time subtraction mode. In this mode one image is taken before injection of contrast material and one image after injection. Because of the time difference between the two images and the fast motion of the coronary arteries - the right coronary artery with up to 6 cm/s in RAO-30^o projection - this method leads to artifacts for small structures. Therefore, this method is not suited for imaging of coronary arteries with non-invasive coronary angiography but only for suppression of large structures in images from invasive coronary angiography. Electrocardiographic triggering cannot overcome this problem for structures which measure as little as 1 mm because of the nonperiodic motion of the heart (translation and rotation) even if the respiration is stopped.

For intravenous coronary angiography a different mode of DSA, the energy subtraction mode (dichromography)[2] is suitable. Using this method the two images for subtraction are taken at the same time when the contrast material (iodine) is present. The method uses the discontinuity of absorption at the K-edge of the contrast material (Fig.1). Iodine differs by a factor of 6 in the absorption coefficient below and above

the K-edge at 33.17 keV. If the two images for subtraction are simultaneously taken with two different monochromatic energies - one energy below and the other energy above the K-edge - the contrast from iodine is strongly enhanced after subtraction. For an energy separation of the two monochromatic energies of 300 eV the sensitivity to iodine is 10000 times higher than that to soft tissue.

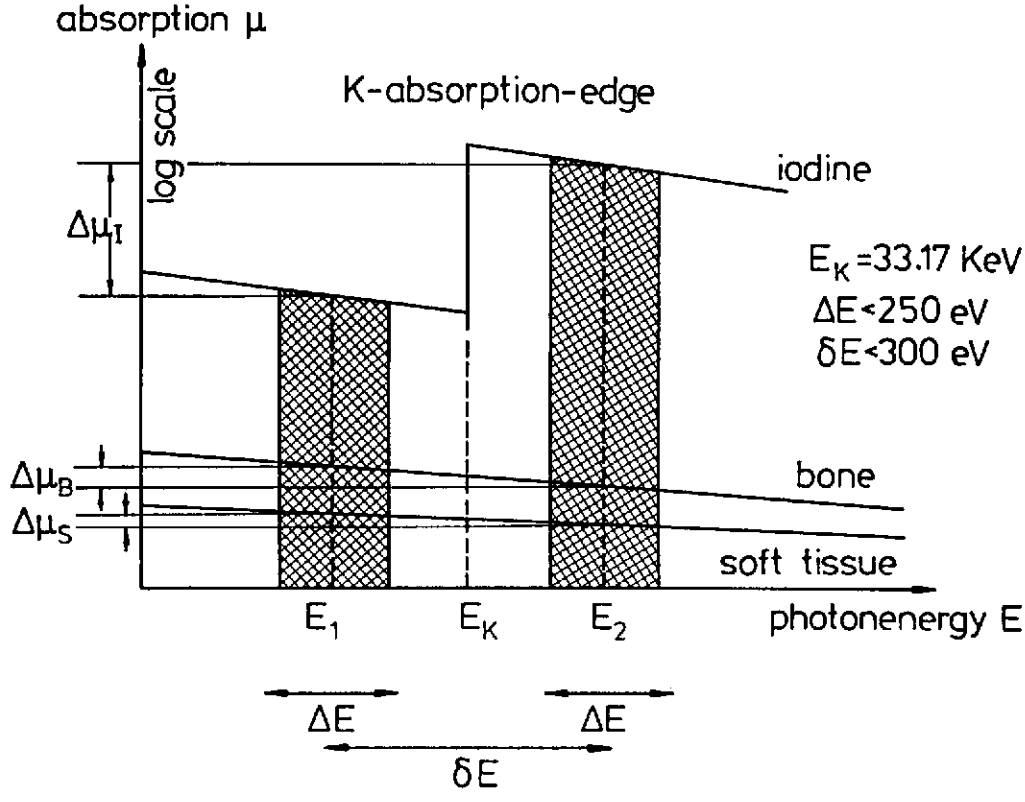

Figure 1: K-edge of iodine at E_K = 33.17 keV. E_1 and E_2 denote the two energies of the quasi-monochromatic beams for dichromography with bandwidth ΔE.

This method requires monochromatic X-rays with very high intensity. Therefore, the development of systems for dichromography only started when these X-rays were available at synchrotron radiation facilities. In 1979 the first systems were installed in Stanford, USA[3] and Novosibirsk, USSR[4]. Since 1981 the system NIKOS is developed in Hamburg, Germany[5]. In Tsukuba, Japan[6] work started in 1983. In 1989 the Stanford system was transferred to Brookhaven, USA[7]. Other synchrotron radiation laboratories like ESRF in Grenoble, France or Argonne, USA plan to start intravenous coronary angiography as well. All existing systems have been used for in-vivo investigations of animals. Up to now human studies were only performed with the systems in Stanford[3]/Brookhaven[7] and Hamburg.

Parameters for dichromography systems

A system for intravenous coronary angiography using dichromography must fulfil the following medical and physical requirements (for more details see refs (8) and (9)):

In order to suppress the background caused by scattered radiation most systems include line scan detectors. These detectors must have a large dynamic range (at least 8400:1) and a high detecting quantum efficiency (DQE) of 60% or more. The dynamic range is necessary because small structures with a mass density of 1 mg/cm^2 (see above) only show a 3% difference between the two energy images. This small contrast must be visible behind lung tissue as well as behind bone tissue. The ratio for the transmission through lung tissue and through bone tissue in a normal patient is about 60:1.

Today the spatial resolution of the detectors used for human studies is 0.5 mm which seems to be adequate for the visualization of stenoses in arteries of 1 mm diameter. Nevertheless the detectors of the next generation will have a spatial resolution of 0.25 or 0.30 mm, respectively.

In order to avoid artifacts due to the line scan mode in image formation the images should be taken in the slow motion phase of the heart. Assuming the heart is 12.5 cm high, the spatial resolution is 0.5 mm, and the duration of the slow motion phase of the heart is 250 ms, the exposure time per line must not exceed 1 ms.

The energy separation δE (see Fig.1) must be smaller than 300 eV, otherwise after subtraction bone and soft tissue give higher contrast than the smallest artery. This means that the bandwidth ΔE must be less than 300 eV, too.

The K-edge of iodine is at 33.17 keV. Photons of this energy in soft tissue have an absorption length of only 2.1 cm. Given a pixel size of 0.5×0.5 mm^2, a signal/noise ratio (SNR) of at least 3 for a 1 mm thick vessel, 20 cm of soft tissue and an assumed DQE of the detector of 60%, this leads to a required photon flux of $\Phi_0 = 10^8$ photons/mm^2 in front of the patient. This flux corresponds to a skin dose of K = 11 mGy per scan (two energy images). For an exposure time per line of 1 ms an intensity of 10^{11} photons/(mm^2·s) must be available. Today only synchrotron radiation facilities can deliver this intensity.

System NIKOS II

During the development of the system NIKOS for intravenous coronary angiography starting in 1981 there existed different versions of the system. In the following the version NIKOS II (Fig.2) is described which was used for the human study runs in May 1990. This system has six main components: The hard X-ray wiggler HARWI as a synchrotron radiation source, a two beam monochromator, a safety system, a fast scanning device, a fast low-noise two-line detector and a computer system. A short description of these components is given in this chapter. For more details see ref.(5).

The 20-pole **wiggler** HARWI[10] is installed in the storage ring DORIS at DESY in Hamburg. Its total length is 2.4 m plus 0.12 m for the endpoles. It is constructed with

permanent magnets ($Co_{17}Sm_2$) and soft iron (hybride design). The magnetic gap is 42 mm high and the maximum field strength is 0.95 T. The horizontal opening angle of the radiation is 6.4 mrad or 4.4 mrad, respectively, depending on the energy of DORIS (3.7 GeV, 5.3 GeV). The wiggler fulfils the special demands of dichromography, i.e. a broad beam and a high flux at 33.17 keV.

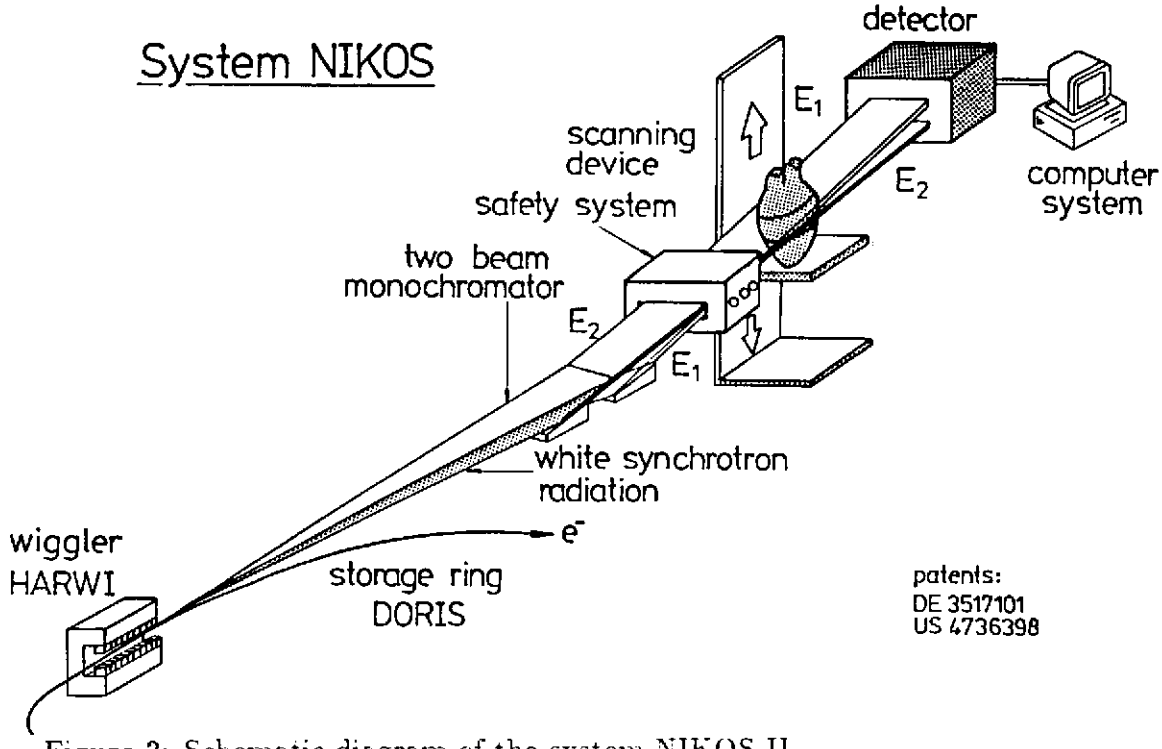

Figure 2: Schematic diagram of the system NIKOS II.

In the **two beam monochromator** two perfect Ge(111) crystals are used for filtering the quasi-monochromatic beams out of the white synchrotron radiation beam. The bandwidth of the beams is 55 eV with a separation of $\delta E = 350$ eV. The crystals vertically split the incident synchrotron radiation beam into two parts. Each crystal is delivering one of two monochromatic beams, 10 cm × 0.7 mm. They are installed in a He-filled tube and cooled by water. The Bragg angle is $\Theta = 3.28°$ for X-rays of $\lambda = 0.373$ Å corresponding to 33.17 keV, leading to a tilt of the beams of 6.56° behind the monochromator. The two beams cross in front of the patient due to the slight difference in the reflection angle of 0.03°.

The crystals are cut asymmetrically in order to increase the flux in front of the patient. The asymmetry angle is 2.55° corresponding to a factor 2.7 for the flux. A flux of maximal $0.13 \cdot 10^{11}$ photons/(mm^2·s) was measured in the monochromatic beams for a current of 44 mA in the storage ring DORIS (5.2 GeV, 2 e$^-$-bunches).

This type of monochromator is sensitive to small vertical motions of the incident beam and of the crystals. These motions result in large changes in the intensity at the detector. In the human study runs intensity fluctuations with a frequency of 50 Hz of up to 10% were registered in the images. These fluctuations are enhanced in the

subtraction image. At least part of them came from motions in the ground which were coupled to the crystals.

Furthermore in the monochromator X-rays of 99.51 keV are reflected as well which lead to beam hardening effects. ((222)-reflections (66.34 keV) are "not allowed" in Ge). The portion of these X-rays in front of the patients was estimated to be 2.4% (field strength of 0.95 T in the wiggler HARWI and 5.3 GeV in the storage ring DORIS). In a future mode the wiggler is expected to be run at 1.26 T and 4.5 GeV in DORIS resulting in a higher harmonic content of 1.9%. A decrease of image quality due to beam hardening effects could not be observed. An estimate of the beam hardening in the patient lead to a factor of 2.5.

The central part of the **safety system** consists of three independent beam shutters[11]. Because in front of the patient the dose rate in each beam amounts up to 5.5 Gy/s, these shutters must close very fast within less than 10 ms (measured 8.8 ms or less). The greatest accident would be a sudden stop of the scanning device or a change of the scanning speed during the exposure of the patient. Therefore, the velocity of the chair which determines the patient exposure is monitored permanently by two independent angular encoders. Malfunction of the scanning device or power failure or breakdown of electronic components in the safety system switch automatically off the beam using two of the three shutters. The beam is switched off also if the patient leaves the scanning device, if the physician steps forward to the beam, if the shielding is not fixed etc. The third shutter is controlled by the computer and is only opened during the exposure of 256 lines of the image.

For the line scan mode used in the NIKOS system a **scanning device is** needed. The maximum scan velocity is 50 cm/s and is kept constant over a distance of 20 cm with a precision of 1%. During the human studies the device was run with 9 cm/s. The total lift of 40 cm in the device includes an additional 20 cm for acceleration and deceleration. The actual vertical position of the chair is determined by a precise optical scale which triggers a new readout of the detector every 0.5 mm (precision 5 μm). The moving force of the scanning device is generated in a hydraulic system with proportional valves under computer control. It can move loads of up to 300 kg. The chair allows for a patient rotation of $\pm$ 40° about a vertical axis and $\pm$ 20° about a lateral axis in order to set the appropriate projection angles. The patient is scanned with raised arms which accelerates the flow of contrast material, and avoids overlap problems with the arms in extreme projection angles. The arms rest on pads where the cables for ECG and tubes for the injector can be fixed. For a precise positioning of the patient a light mask is projected on the patient's body which indicates the irradiated area. For the positioning vertical and horizontal adjustment of the scanning device is possible. The device is started ECG-triggered by the computer.

In NIKOS II a fast low-noise **two-line detector**[12] is installed which simultaneously records the two lines with the energy above and below the K-edge, respectively. This detector has the advantage that it makes better use of the photon flux than a one-line detector combined with a beam switcher and the afterglow problems are reduced.

Also no fast moving mechanics must be installed in the system.

The detector is based on commercially available photodiode arrays (Reticon RL 1024 SF) which integrate the light generated in a phosphor and which is amplified through image intensifiers (Proxitronic Proxifier BV 2543 KX25). At the entrance face two lines of scintillators (Gd_2O_2S:Tb powder) convert the incoming X-rays into visible light. The lines are 0.5 mm thick, the portion of the powder is about 80%. Each line has 250 pixels (0.5×0.5 mm^2 each). The lines have a vertical spacing of 2 mm and the light is guided by special glass fiber optics through the two two-stage proximity focussing image intensifiers onto the two photodiode arrays (1024 photodiodes each, 3 active + 1 dummy for each pixel). One photodiode array per line is needed and the readout of the two lines is controlled by a common digital control board triggered by the linear optical encoder of the scanning device. The data are digitized with a 12 bit ADC (DATEL 505 BMC) and the values of 3 diodes are summed up. Hence the full scale of the output signal of one pixel is represented by theoretically 12288 grey levels but the electronics saturates at 10700 grey levels. Furthermore an offset of 200 or 800 grey levels exists, respectively, depending on the channel of the detector.

The data are transmitted to the computer via a glass fibre link (maximal transfer rate 4.4 Mbaud). The maximum readout frequency for the lines of the detector is 0.5 kHz per line which corresponds to a scan velocity of 25 cm/s. For this frequency the DQE was determined. For technical reasons it was different in the two channels of the detector: 4% in the E_1-channel and 14% in the E_2-channel. The dynamic range was measured to D = 3000:1. For a frequency of 0.1 kHz the dynamic range is D = 8000:1.

The **computer system** in NIKOS II consists of a PDP 11/73 for the control of the complete system and a MicroVAX-WSII/GPX for data acquisition and image processing and presentation. These two machines are connected via a serial line and are interfaced with the DESY computer center.

The raw data of a flat phantom without any corrections show a standard deviation of up to 35%. These variations are not statistical but can be corrected by the following procedures:

- The dark current is measured and subtracted.
- The overall fixed pattern of the detector is determined by the scan of a flat phantom. This reference image is taken for correction of the raw data. Although the amplitude of the fixed pattern is very large, the pattern itself is very stable and the correction reduces the standard deviation of the raw data drastically.
- Any irregularities of the scanning device are measured and corrected for.
- As between the moments when the reference image is taken and the patient is scanned, the beam shape changes slightly, the correction by the reference image does not remove the fixed pattern completely. The residual pattern (vertical fringes in the image) is determined by software and eliminated[8].
- Fast intensity oscillations as mentioned above cause a complex horizontal pattern. A special software for recognition of this pattern was developed, but a substantial loss of quality in the images was observed.

All these procedures are applied to each of the two energy images. After all the

standard deviation which was measured with the raw data to be 35% is reduced to 6%. For a proper application of the method the latter should be 0.1%. The main part of the remaining variations in the images is due to the fast intensity oscillations. The mentioned procedures introduce further information into the images and hence increase their quality.

Then the two images are subtracted logarithmically. The resulting image is corrected for residual fixed pattern and grey level adjustment is made. Afterwards an unsharp masking algorithm is applied to this image. That means the image is filtered with a large median filter (29 pixels $\times$ 29 pixels) and the filtered image is subtracted from the original subtraction image in order to enhance the small structures and suppress the large structures like ventricle, atrium or aorta. Finally edge preserving smoothing[13,14] of the image is performed. Although the information content of the image is decreased by these procedures the structures of interest are enhanced.

Studies on human subjects

After investigation of 16 anesthetized dogs[5] with the system NIKOS in May 1990 the first three human subjects were studied. All patients were under treatment for coronary artery disease in the Universitäts-Krankenhaus Hamburg-Eppendorf. They had conventional coronary angiograms within the last 2 years.

During the investigations in this study a 6 f catheter was introduced into the brachial vein. The tip of the catheter was positioned into the vena cava superior. 30 ml of contrast material (Ultravist-370) were injected within 2 s. The contrast material was transported by the blood into the coronary arteries after passing the right heart, the lungs, the left heart and the aorta.

In our opinion the dose for a simple scan is prohibitive of taking a series of scans per injection. Therefore, only one scan per injection was performed. This caused at that time the problem to find the correct time T from the injection up to the moment when the coronary arteries are filled with contrast material. This time strongly depends on the individual circulation time of the patient as well as on his condition at the moment of the investigation and the site of injection. With bolus injection into the vena cava superior T can range from 8 to 13 s in normal subjects. On the other hand the optimal time window for the scan is only 1 to 3 s. Therefore, not all scans taken during these first studies were started at the optimal time T thus not showing the highest possible contrast of the coronary arteries.

A known problem with non-invasive investigations is the superposition of the small iodinated structures of interest by large structures which are filled with iodine, too. If the contrast material is injected into the vena cava superior within 2 s then left heart, aorta and part of the lung veins are opacified in the moment when the contrast material shows up in the coronary arteries. To overcome this problem and to visualize all segments of the coronary arteries a careful selection of the projection angles for the investigations of the patients is of great importance. A pre-study[15] has demonstrated that the best projection for the right coronary artery is left anterior oblique (LAO)

40°, caudo cranial (CC) 20° and for the left anterior descending artery (LAD) is right anterior oblique (RAO) 40°, CC-20°. Part of the left circumflex coronary artery (Cx) and in most cases the left main stem cannot be projected free from other iodinated structures. In order to visualize these structures image processing algorithms like unsharp masking (see above) must be used.

In the following we give two examples of the studies: The first patient was a 48-year-old man who had an aorto coronary vein bypass (ACVB) surgery in 1989. In 1990 angina pectoris during exercise showed up again. In March 1990 a repeated selective coronary angiography revealed that the LAD bypass is patent while the RCA bypass is closed. The native LAD is occluded proximally and filled retrograde via the bypass.

The height of the patient was 173 cm and his body weight 62 kg. During the investigation the blood pressure was 135/70 mm Hg, the pulse rate was 70 beats per min. Two intravenous angiograms were taken plus an additional positioning scan with decreased dose before. Fig.3 shows the angiogram in RAO-30°, CC-0° projection. This scan was started 9.3 s after injection of the contrast material. The storage ring DORIS ran with 40 mA at 5.2 GeV (two e^--bunches). The dose to the patient was measured with thermo luminiscence dosimeters (TLD) made of LiF which were put on his breast. For one scan the dose amounted to 30 mGy. The resultant image in Fig.3 is shown together with an image of the same patient from March 1990 taken with normal selective angiography and an image taken with a normal X-ray device after intravenous injection of the contrast material.

The patient received a total of 60 ml Ultravist-370 and 66 mGy during the investigation. The images from this study show that the LAD bypass is patent though the patient has angina pectoris again.

The second example is an image of a 62-year-old man who had a first ACVB surgery in 1982. In 1987 the closure of all venous grafts was documented by selective coronary angiography. In 1988 a repeated ACVB surgery was performed.

The height of the patient was 163 cm and his body weight 69 kg. During the investigation the blood pressure was 140/85 mm Hg, the pulse rate was 80 beats per min. Again two intravenous angiograms were taken. Fig.4 shows the angiogram in LAO-30°, CC-1° projection. This scan was started 7.6 s after injection of the contrast material. The storage ring DORIS ran with 33 mA at 5.2 GeV. The dose to the patient was measured to be 19 mGy. The resultant image shows no RCA bypass though the aorta is filled with contrast material. No decision can be made from this image whether the bypass is occluded or the scan was performed too early. The patient received a total of 50 ml Ultravist-370 and 32 mGy during the investigation.

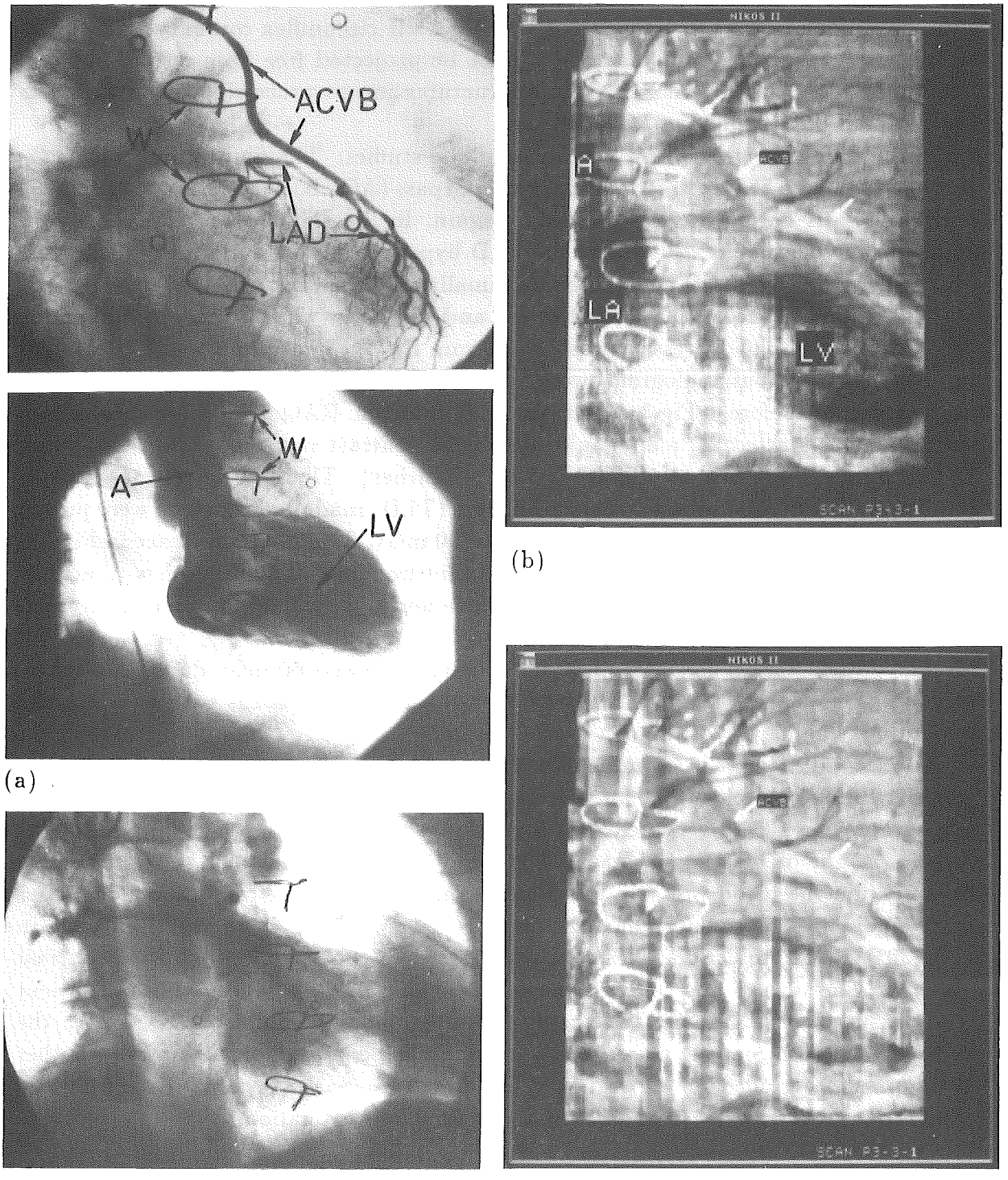

Figure 3: (a) Selective angiograms of a 48-year-old man for comparison with the intravenous angiograms. On the top the native LAD occluded proximally and filled retrograde via the bypass, the aorto coronary vein bypass (ACVB) and wire structures (W) are visible. At the bottom the contrast material is injected into the left ventricle. Left ventricle (LV) and part of ascending aorta (A) are visible. The anatomical relation

between left ventricle, aorta and LAD bypass are demonstrated.
(b) Intravenous angiogram of the same patient (same projection angles). Left ventricle (LV), part of ascending aorta (A), part of the left atrium (LA), aorto coronary vein bypass (ACVB), LAD inclusive bifurcation, pulmonary veins, wire structures, a surgical silver ring near origin of the LAD bypass and silver clips in the LAD are visible. The iodine concentration is about 8 mg/ml (contrast material diluted by a factor of 45). The signal/noise ratio of the bypass (3.5 mm in diameter) is SNR = 2.
(c) Same intravenous angiogram as in Fig.3b after image processing (unsharp masking and edge preserving smoothing). The large structures are suppressed but the fixed pattern noise is enhanced in these areas.
(d) Intravenous angiogram with the same amount of contrast material like in Fig.3(b) but imaged with a conventional X-ray device. No iodinated structures are visible.

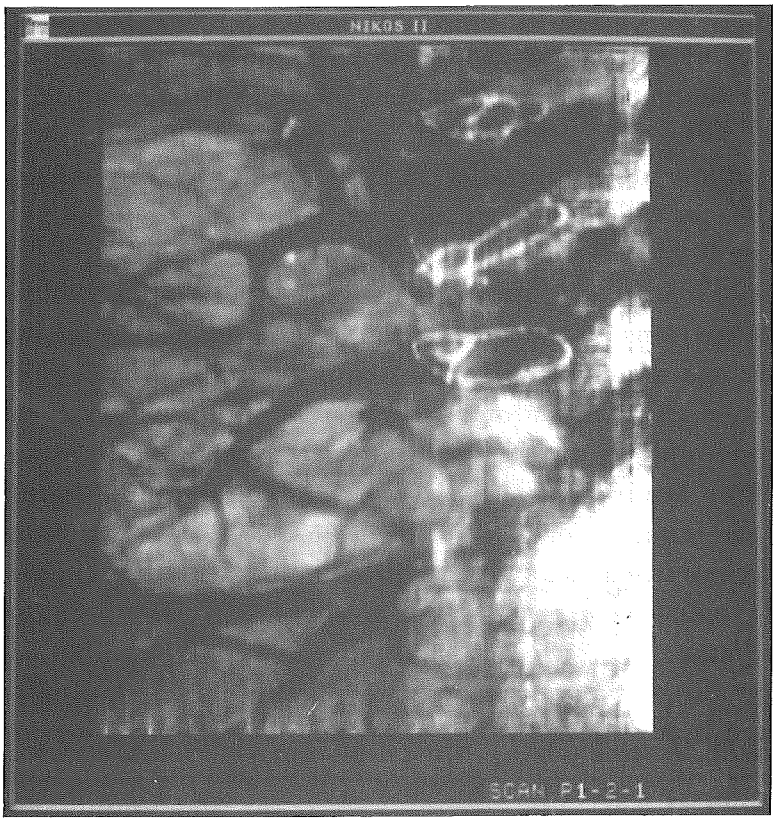

Figure 4: Intravenous angiogram of a 62-year-old man. Pulmonary veins down to a diameter of 1.5 mm, parts of the aorta and the left atrium, wire structures and surgical silver rings are visible. The RCA-bypass could not be imaged. The iodine concentration is about 22 mg/ml (contrast material diluted by a factor of 17).

Discussion of the results and outlook

In this first study it was shown that it is possible to get intravenous coronary angiograms with the system NIKOS what is not possible with normal X-ray devices. On the other hand up to now the images have bad quality and are not suited for routine

investigations in a clinical environment. The main technical reasons are:
- The intensity in the monochromatic beams is insufficient; a factor 8 is missing.
- The fast beam fluctuations decrease the quality dramatically.
- The DQE and the dynamic range of the detector are too low.
These problems must be overcome before routine medical work can start.

It is planned to increase the intensity of the monochromatic beams by replacing the vacuum chamber in the wiggler and by installing a new monochromator with a different design. The design of the monochromator follows an idea of Suortti and Thomlinson[16]. The crystals for reflection are used in Laue geometry and are slightly bent. That increases the bandwidth ΔE of the monochromatic beams thus increasing the intensity by a factor of about 3 compared to the existing monochromator. A prototype of the new monochromator was tested but was not yet installed because the cooling problems could not be solved up to now. A new vacuum chamber for the wiggler was built and will be installed as soon as possible. It is a variable chamber which allows a smaller gap of the wiggler thus increasing the field to 1.26 T. Therefore, the flux at 33.17 keV will be increased by a factor of 2. During the time before these two new components are installed in the system NIKOS the low intensity of the monochromatic beams will be compensated be decreasing the velocity of the scanning device correspondently.

The new monochromator is not as sensitive to vertical motions as the old one. But for the investigations in the next period the old monochromator must be used in the system. In order to avoid the fast beam fluctuations rubber sandwiches were installed to decouple the monochromator from the ground. Furthermore parts of the monochromator are replaced by others with different weight and geometry[1].

After the human study runs the detector was completely dismantled. It turned out that one of the image intensifiers showed only 58% of the original amplification. Therefore, new image intensifiers are installed in the detector. If the fast beam fluctuations can be avoided we expect a DQE of about 20%. Furthermore new phosphor lines are installed and the electronics is rebuilt. Now it allows a readout frequency of 0.25 kHz corresponding to a scan velocity of maximal 12.5 cm/s. This change of the electronics resulted in a dynamic range of 8000:1. For the final version of the system NIKOS a different detector is under test. This two-line ionization chamber[17] is filled with Xe (80%) and CO_2 (20%) under 10 bar pressure. The prototype has a resolution of 0.4 mm, the final version of 0.3 mm. For this detector a DQE = 80% and a dynamic range of at least 10000:1 is expected.

The images from the human study runs show coronary arteries down to 1.5 mm diameter (Fig.4). It is not possible to decide from these images whether a detector with higher DQE and dynamic range than the used one and a spatial resolution of 0.5 mm can image coronary arteries of 1 mm diameter with sufficient quality. Therefore, the spatial resolution in the new detectors is doubled in order to have the possibility to use this feature. In that case the dose would be increased fourfold.

Some additional minor changes to the system will take place before the next human

[1] Meanwhile an instrumentation run showed that these changes lead to a remarkable suppression of the fluctuations.

study runs: The horizontal width of the white synchrotron radiation beam is increased thus allowing images of 12.5 cm × 12.8 cm. New monitors in the beamline are installed in order to improve the long term stability of the beam. The scanning device is redesigned. Now it is driven from below. This gives higher stability to the patient's chair. Furthermore a larger range of projection angles is accessible.

The main medical problem is the determination of the scan time T after injection. A new densitometer has been installed which allows the measurement of the circulation time when the patient is already sitting on the scanning device just before the scan[2]. With the measured circulation time T two scans per injection can be performed - one with contrast material still in the pulmonary veins and the other at the optimal time. This method would double the dose to the patient but has the advantage of better possibilities for identification of coronary arteries and pulmonary veins.

The selected projection angles for the angiograms are optimal for the visualization of RCA and LAD. An additional projection (left lateral) will be tested during the next investigations. For the visualization of Cx and main left artery more sophisticated image processing algorithms will be developed.

In the next future central venous injections like in this first study will be performed. Lateron injection into the brachial veins is planned. No decrease of the iodine concentration in the coronary arteries is expected if the brachial veins of the two arms are used simultaneously.

In the images of the human study runs a bypass of 3.5 mm diameter gives a SNR of about 2. In order to get SNR = 3 for 1 mm thick arteries the SNR must be increased by a factor of about 6. This will be possible by increasing the DQE of the detector from now 10% on the average to 80% in the new detector. Furthermore the correct determination of the optimal time T should give a higher concentration of iodine in the coronary arteries during the scan than in the existing scans.

After performing all the changes and completions to the system NIKOS described in this chapter the system is optimized for dichromography. Good quality intravenous angiograms can be expected. After optimizing the system components, several patients must be studied to ascertain the validity of this method when compared to invasive coronary angiography.

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[2]Meanwhile a study has been performed which showed that the circulation time of the individual patient can be measured with sufficient precision.

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