DIGITAL SUBTRACTION ANGIOGRAPHY (DSA)
the technique and an analysis of the physical factors influencing the image quality

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1. INTRODUCTION.

1.1. Angiography, an essential examination technique for medical diagnosis.

In medical X-ray diagnostics examinations concerning cardiovascular
diseases are playing an increasingly important role. The reason for
this is that the present mortality of cardiovascular cases is about
half the total mortality (46% in the Netherlands; Arntzenius,1978).
An investigation by the World Health Organisation revealed that in
developed countries about 80% of the male population show some
atherosclerotic lesions by the age of 32, while with women this per­
centage is reached at age 42 (Arntzenius,1978). Notorious in this
respect is the occlusion of coronary arteries which may lead to
cardiac infarction. Other notorious regions in the vascular system are
the carotid, pulmonary, renal and femoral arteries.

In order to evaluate the condition of the vascular system it is
essential to be able to visualize the suspected arteries. It is
obvious to try the X-ray imaging technique for this purpose but a pro­
blem is then that both blood vessels and tissue consist for 60-70 % of
water, and therefore produce equivalent X-ray opacity. The difference
between blood vessels and tissue does not show up in an X-ray photo­
graph unless special measures are taken.

Angiography is a special X-ray imaging technique during which a
contrast medium is injected into the vascular system.
Because of the high velocity of the blood flow in arteries (up to 50
cm/s) the contrast medium must be injected very rapidly and in a high
concentration to prevent dilution. Moreover the place of injection has
to be as close as possible to the vessels under examination or else
the dilution of the contrast medium will be so high that the vessels
can not be visualized on the X-ray films. Therefore the injection is
mostly done by means of a catheter which has been introduced at a
nearby position in an artery.

Arterial angiography has been common practice for more than twenty
years, and images of excellent quality can be obtained. There are
several reasons however why research is still going on to find new
ways of visualizing the vascular system:

1. Conventional angiography is an invasive technique which is not
   altogether without risk for the patient. The catheterization pro­
   cedure or the fast injection may dislodge pieces of atheroscle­
   rotic plaque which in turn may cause embolization in more dan­
   gerous places such as the cerebral arteries.
Also local complications such as bleeding and acute thrombosis may occur in the entry vessel.
For these reasons the patient is kept hospitalized for one or two days after a catheterization procedure.

2. The complete catheterization procedure is complicated and therefore time consuming. A sterile environment and great personal skill of the examiner is required. It is not uncommon that even an experienced physician may need one to two hours for one examination.
The X-ray dose to which the patient is exposed during the examination is rather high, partly due to the long duration of the fluoroscopic period.

3. Some patients are in such a bad vascular condition that the catheterization procedure is not justified.

An alternative angiographic imaging technique that accommodates these objections uses intravenous angiography. The injection of contrast material into a peripheral vein is much less inconvenient for the patient than selective injection into an artery. With intravenous angiography the contrast dilution becomes so high that special techniques must be employed to produce useful images. Digital Subtraction Angiography (DSA) proves to be a very useful imaging method.

1.2. Scope of the thesis.

Although angiography is now a common diagnostic procedure, the reader of this work may be unacquainted with the details of this technique. Therefore a survey is given in chapter 2, in which the history and current status of conventional angiography is described. Chapter 2 also reviews some hard facts on why the non-invasive intravenous injection procedure offers so much benefit for the patient.
Image subtraction is the basic trick to improve the "conspicuity", and chapter 3 explains methods and equipment for performing the subtraction.
As low image contrast is to be expected, the ultimate imaging performance will be quantum noise limited. In chapter 4 the Rose model is introduced, which relates contrast level and detail size to the required radiation exposure dose.
Up to now the appraisal of contrast medium dilution after an intravenous injection has been just guesswork. Chapter 5 introduces a model for the calculation of the dilution factor, based on an inverse application of the dye dilution technique. The influence of important parameters such as injection site, injection rate and injection volume can be predicted by the proposed method.

In chapter 5 an analysis is presented of how the contrast in the image depends on contrast medium concentration, vessel diameter, tube voltage and scatter. There appears to be an optimum tube voltage.

As the signal-to-noise ratio in DSA is of the utmost importance, chapter 7 analyzes the TV noise and proposes unconventional methods to improve the signal-to-noise ratio of the TV camera.

Another cause of noise and image artefacts could be the inappropriate application of the sampling theory. The critical factor appears to be amount of lowpass filtering used, and chapter 8 deals with the requirements of the filtering in relation to the sampling frequency.

Digitization of video signals introduces irreversible signal distortion, which can be reduced at will, however, by using enough bit depth per sample. DSA imaging requires more bit depth than is commonly used for digital (unsubtracted) video. Chapter 9 discusses the requirements, taking the log conversion accuracy also into account.

Except for visualization of low contrasts, small detail visualization is also important when small vessels are being examined. Unfortunately the small detail capability of the current DSA equipment is less than that of the conventional technique. Chapter 10 explains which factors contribute to this resolution aspect, and some indications are given of how improvements can be made.

Chapter 11 condenses the results of the previous chapters into an overall performance assessment. Conclusions can be drawn about what is possible with intravenous injections and when selective arterial injections are needed.

Chapter 12 discusses image disturbances which are typical of the DSA technique. New requirements have to be fulfilled by some components in the system to reduce these artefacts to an acceptable level.

References:

2. MEDICAL BACKGROUND.

2.1. Historical review.

2.1.1. Introduction.

In this chapter a short general overview of angiography will be given. This information has been gathered from medical textbooks (Abrams, 1971; Weibel and Field, 1969; Loose and van Dongen, 1976) in order to inform the medically-untrained reader about this interesting examination technique.

The knowledge of the anatomy of blood vessels in man was very limited until the beginning of this century. Until then dissections of dead bodies was the only method of gaining experience in this field. When X-rays were discovered (Röntgen, 1895) the possibilities of this new technical aid were immediately recognized by the medical society. Apart from the new "Photography" of traumatic lesions of bone, the applicability to studying the blood vessels was also recognized. Within two months after Röntgen's first publication, the first "angiogram" of an amputated hand was published in Vienna (Haschek and Lindenthal, 1896). The injected contrast medium was composed of chalk, red mercury sulphide and kerosine. Other X-ray opaque substances used at the time were metallic mercury, colloidal gold-silver emulsions and suspensions of bismuth and oil. These vascular studies on cadavers led to vastly improved knowledge of the anatomy of the vascular system. In 1920 an X-ray atlas was published in England (Griff, 1920), which showed the various blood vessels in cadavers very clearly. Since that time a continuing effort has been applied towards finding non injurious methods to outline the arteries in living man. The success of the present day angiography has emanated from a combined progress in three different areas:

1. The contrast medium;
2. The way of administering the contrast medium to the patient;
3. The radiological equipment.

2.1.2. Development of contrast media.

For angiography a non-toxic fluid contrast medium is needed. Many of the media initially used were of unacceptably high general and local toxicity. The first patient of Moniz in 1927 died after an attempt to visualize the intracranial vessels (Moniz, 1927).

A breakthrough in the development of contrast media were the water soluble organic preparations of the Diodrast type. They were first reported by Gross (1939) and subsequently extensively used for nearly 20 years. The Diodrast group of contrast media were far from ideal however, because they had an irritating effect on the blood vessels leading to thrombosis and other serious complications. For many years the development of safe and satisfactory contrast media lagged behind the improvement in radiological equipment and techniques for cannulating or catheterizing specific arteries. The high rate of toxic side effects presented a serious impediment to the wide acceptance of angiography.

The introduction of two other groups of water soluble organic iodized preparations (Diatrizoates in 1956 and Meglumine iothalamates in 1961) was a significant advance in angiography. These agents were much less hazardous and at the same time gave better visualization than Diodrast.

A recent improvement pioneered by Almén is the availability of non-ionic contrast media. These contrast media, characterized by a much lower hypertonicity with respect to human blood plasma, were discovered in 1968 (Almen, 1969) and after a long period with animal experiments used for the first time in clinical practice in 1977 (Almen, 1977). The lower hypertonicity is reported to be responsible for smaller transient hemodynamic effects in the blood circulation and for less pain during injection. The general use of these agents is very limited today due to the prohibitive price (Ludwig, 1979).

2.1.3. Development of various injection methods.

The administration of the contrast medium to the patient has various degrees of invasiveness depending on whether an open method, a puncture or a catheterization is used.

Moniz in 1927 accomplished his cerebral angiographic procedure after surgical exposure of the artery. Since then this so-called open method was used for nearly 15 years.

Dos Santos (1929) proposed direct puncture of the abdominal aorta for visualization of this vessel and its branches. Adaptions of this "translumbar aortography" (TLA) are still being used to-day. Direct punctures with the intention of visualizing the aortic arch and its main branches in the thorax were found to be too risky to be employed.

In 1938 Robb and Steinberg (1939) developed a safe intravenous method for thoracic angiography but the method did not receive much acceptance. The dilution of the contrast medium during passage of a
cardiopulmonary circuit caused that a large amount of it had to be injected; due to its toxicity only one injection per examination was possible.

Direct puncture of peripheral arteries, the femoral arteries for the legs and brachial or cubital arteries for the arms, pose no special problems as these arteries can easily be found by palpation, after which the direct percutaneous puncture can be made. The injection of highly-concentrated contrast media into peripheral arteries is rather painful and therefore a general anesthetic is sometimes used during peripheral angiography (Loose, 1976).

Percutaneous puncture of the common carotid artery was first described by Loman (1936), but received general acceptance only in the mid-1940's. Although cerebral angiography by injection into the carotid is considered as very valuable, it displays only a part of the intracranial system. The vertebral arteries which also contribute to the cerebral circulation are much more difficult and dangerous to reach by percutaneous techniques.

These vessels as well as the thoracic vessels had to await the catheterization technique. Forssman was the repudiated prophet in this area. He reported (Forssman, 1929) to have inserted a catheter via an arm vein into his own atrium, after which he walked to the X-ray department to record this fact on a chest film. He was hooted after his lecture because this intervention was considered as far too dangerous by his colleagues.

The catheter method for angiography was first introduced by Parinas (1941) but the technique was seriously handicapped by the lack of suitable flexible thin-walled catheters. The introduction of polyethylene tubing allowed Pierce (1951) a percutaneous insertion of a catheter into the femoral artery without the time-consuming surgical exposure of the artery. Seldinger (1953) improved the percutaneous insertion by first feeding a flexible guide wire through the needle into the vessel. After withdrawing the needle a catheter of the same bore as the needle can be fed over the guide wire into the vessel. This Seldinger technique is the most widely used catheterization method nowadays.

Catheterization provides a relatively safe method of injecting contrast medium into the aortic arch, but intracranial vessels produce a less than optimal opacification with this method due to the dilution of the contrast medium.

Selective catheterization, first described by Radner (1951), allowed injection of the individual trunks of the aortic arch after manipulating the catheter tip into the wanted position under fluoroscopic control. Successful catheterization of coronary arteries via an open brachial artery approach has been described by Bony (1962).

Percutaneous approach to the coronaries via the femoral artery was developed by Judkins (1967). Both heart catheterization methods which require special preformed catheters, are still being used today. Depending on the skill of the examiner even superselective catheterizations which approach the second order branches are practised now.

2.1.4. Development of radiographic equipment.

The most important aspect which separates dedicated angiographic equipment from conventional radiographic equipment concerns the high imaging rate which has consequences for various parts of the total system.

In order to capture the progress of rapidly-injected contrast medium in the vessels, a rapid series of X-ray exposures has to be made; usually the filming rate has to be at least two images per second. Numerous home-made film changers were constructed by pioneers (e.g. Ruggles, 1925) because X-ray companies showed little interest in the development of such machines. After working with Moniz on a manually operated cassette changer, Sanchez-Perez (1934) constructed an improved model. This model eventually developed into the still very popular motor-driven cassette changer that handles 12 cassettes of 10 x 12" film at a rate of 2/s. Higher rates are impossible with cassette changers, but transporting only the film into and out of the exposure position makes it possible to achieve higher speeds. The most common rapid film changer to-day is the Elema-Schönander unit, also referred to as AOT, an abbreviation for AngiOTable because the apparatus was once part of the examination table. This unit can handle 30 films of 14 x 14" size at a speed of 6/s.

When examining heart diseases a higher recording speed is needed to permit the study of motion. Direct cinefluorography is the high speed motion picture recording of events as they are produced on a fluoroscopic screen that is directly exposed to X-ray radiation. The limited light available from the fluoroscopic screen required an extremely fast lens for sufficient illumination of the film. Only a few centres in the world have practised this approach in the past. During the mid-fifties the first X-ray image intensifiers became available which permitted the cine camera to utilize the intensified light output.

In the beginning period 16 mm film was also used but as the resolution capabilities of the image intensifier and lenses improved, a universal acceptance of 35 mm film occurred.
In most cine cameras the shutter has an opening of 180 degrees, which means that 50% of the total time is reserved for film transport and 50% is available for film exposure. If X-rays are generated while the shutter is closed, the patient is exposed to undue amounts of radiation which must of course be prevented. The solution is to pulse the X-rays, so that the radiation is on only when the film gate is open (<10 ms for 50 images/s). Switching the high voltage supply for the X-ray tube so rapidly requires some special measures in the X-ray generator. Top-class generators use high-power tetrodes in series with the X-ray tube for interrupting the power. Exposure times as short as 1 ms can be realized by this method. The shorter the exposure times used, the higher the delivered current must be to produce enough energy per exposure. Present day generators are able to produce current pulses of more than 1 A at a voltage range of 40 - 150 kV; the power is of the order of 200 kW.

Injection of the contrast medium could be done by hand, but the flow rate required varies considerably, depending on the blood flow velocity and the size of the vessel at the injection site. Large vessels such as the aorta require injection rates up to 30 to 50 ml/s. Due to the length of the catheters, the small size of the lumen and the end holes of the catheters and the viscosity of the contrast medium, the attainment of such a high flow rate may require a pressure up to 50 atm. This pressure can be accomplished only with mechanical injectors. Early constant-pressure models of mechanical injectors used compressed gas to activate a hydraulic system. Information about the inner diameter and length of the catheter, the viscosity of the contrast medium and the rate of delivery were needed to determine the required pressure. Modern injectors have an electromechanical drive and provide automatically, within practical limits, a constant flow rate regardless of catheter size, viscosity and other previously impeding factors.

The introduction of television systems in combination with X-ray equipment was hindered by the same sensitivity problem as direct cinefluorography. Sturm and Morgan tried to couple an orthicon TV camera to a fluoroscopic screen by means of an optical mirror system, but the performance of the system was hardly better than conventional direct screen fluoroscopy (Morgan, 1951). The introduction of the image intensifier by Teves and Tol (1952) was a major step forward and soon thereafter the first X-ray television system using a Vidicon camera tube was reported (Stauffer, 1955).

A characteristic property of the Vidicon tube is the lag which appears especially at low light levels. A much better camera tube, as far as the lag is concerned, is the Plumbicon which was first reported to be used in an X-ray television chain by Feddema and Marquerink (1964). Plumbicon camera tubes are better suited for capturing fast movements, while Vidicon tubes produce less-noisy images due to the lag. Both types of camera tubes are used today and the choice depends on the type of medical examinations to be done.

Recent developments in electronic imaging and picture processing (this thesis only deals with one special type of processing) will eventually change the role of the television chain in X-ray imaging from a convenient auxiliary to the main data acquisition part of the system.

In addition to the morphological information of the images, quantitative data may also be used as a help in the diagnosis. Once the video signals of the images have been digitized for image processing reasons, the quantitative data are so readily available that computer manipulations of these data are very appealing. The main goal of the computer manipulations is to find a figure that quantifies the functioning of an organ to be examined. Various attempts in this direction are currently being tested in clinical environments.

2.2. Complications and risk factors of invasive angiography.

In the introduction of a recent book on angiography (Johnsrude, 1979), the following statement can be found:

The diagnostic possibilities of an angiographic study must be carefully weighed against its potential complications; the potential for doing good is sometimes offset by its capacity to harm. If the danger outweighs the potential information then the angiographer should consider noninvasive techniques.

Several publications have reported statistics of complications in angiography. Remarkable is the large spread in the published figures. Some authors report quite high complication rates; 23.2% complication rate (Blain, 1966) and recently (Paugh, 1979) 12.2% cerebral complications in patients with transient ischemic attacks and even 35% with patients having critical stenoses of the carotid artery. The complication rates reported are often misleading because they do not discriminate between the clinically-important permanent complications and those that are transient and clinically less significant.
Eisenberg (1980) elaborates on the spread in reported figures by neurologists, surgeons and radiologists.

A detailed recent study (Hessel, 1981) of complications that occurred in 1975 in 514 U.S. hospitals included a total of about 120,000 examinations. In this study transfemoral, translumbar and transaxillary angiography are compared and the mostly-used transfemoral approach appeared to have the lowest complication rate (see Table 2.1).

Table 2.1.: Overall complication rates from Hessel (1981)

<table>
<thead>
<tr>
<th>Angiographic technique</th>
<th>femoral</th>
<th>axillary</th>
<th>lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td>No. of cases</td>
<td>83068</td>
<td>4590</td>
<td>4118</td>
</tr>
<tr>
<td>Total complications</td>
<td>1441 (1.7%)</td>
<td>151 (3.3%)</td>
<td>119 (2.8%)</td>
</tr>
<tr>
<td>Deaths</td>
<td>24 (0.03%)</td>
<td>4 (0.09%)</td>
<td>2 (0.05%)</td>
</tr>
</tbody>
</table>

These figures, rating both the total complications and deaths, are less than half of the figures reported in the past (Lang, 1963). Evidently a marked decrease of the risk factors has occurred in a period of 18 years experience.

The transfemoral figures given in Table 2.1. correlate very well with another recent study (Mani and Eisenberg, 1978) of transfemoral catheterization in cerebral angiography (see Table 2.2.).

Table 2.2.: Complications in 5000 angiographic procedures from Mani (1978).

| Local: | Minor transient | 2 (0.04%) | 0% |
|        | major transient | 3 (0.06%) | 0.20% |
|        | major permanent | 5 (0.10%) | 0% |
|        | death           | -         | - |
| Central nervous system: | minor transient | 5 (0.10%) | 0% |
|        | major transient | 28 (0.56%) | 0.92% |
|        | major permanent | 2 (0.04%) | 0% |
|        | death           | 1 (0.02%) | 0% |
| Systemic: | minor transient | 7 (0.14%) | 0% |
|          | major transient | 5 (0.10%) | 0% |
|          | major permanent | -         | 0.24% |
|          | death           | -         | - |
| Total    |                  | 1.36%     | 0% |

Both Hessel and Mani find that the most-frequent deficit is of a transient neurological nature (temporary blindness or paralysis). Also both studies have found that the angiographic complication rate depends significantly on the following factors:

1. catheter size
2. catheterization time
3. operator proficiency
4. presence of occlusive vascular disease in the patient

Catheterization time and operator proficiency are related factors of course. In this respect Hessel reported that the total complication rate increased to 2.9% for hospitals with a case load of less than 200 per year. Mani reported that training hospitals had a complication rate (3.9%) that is approximately 4.5 times higher than that of non-training centres (0.9%), which have operators of greater experience.

As many patients are candidates for angiographic examinations nowadays, it is of utmost importance to decrease the complication rate still further in order to make possible a relatively safe screening procedure for a check on the vascular system.

A considerable decrease in complications may be expected if the contrast medium is injected intravenously, as all of the reported central nervous complications had to do with catheter or guide wire manipulations or injections.

Detachment of trombi or atheromatous particles and air embolism are presumed to be the cause of the cerebral complications, and all of these problems are virtually impossible with peripheral intravenous injections. Local complications such as the painful extravasation by contrast medium are still possible with intravenous injections, but apart from the temporary pain they are always harmless for the patient. Systemic complications such as hypotension, cardiac problems and renal problems are presumed to be associated with allergic reactions to the contrast medium. Hypotension and cardiac problems occur within 2 hours after the procedure. Renal problems are predisposed for patients with pre-existing renal impairment and need careful postangiographic monitoring of high risk patients for 24 hours.

Conventional angiography usually requires a 24 hour postangiographic after-care for all patients, but due to the removal of the cerebral risk factors with intravenous injections, only the predisposed renal problem patients still need hospitalization.
Since its introduction in 1980 an increasing number of hospitals in the world practice DSA now. The number of DSA machines of various manufacturers currently installed is estimated to handle about 1 million examinations per year. Experience with the DVI apparatus in the Antoniushospital in Utrecht indicates that extravasations occur in about 0.25% of the examinations. This complication belongs in the category "local transient" effects and is thus higher than indicated in Table 2.2. Systemic transient effects caused by allergic reactions to the contrast medium also occur with DSA, and by lack of better data, the same 0.24% of Table 2.2 is presumed. The "central nervous" complication category can be excluded due to the non-invasive venous injection method and this leads to an estimation at the DSA complication rate of about 0.5%, compared to the 1.36% in the study of Mani.

Even more interesting to note is that the main risk factor for death of patients is removed, and also the importance of operator proficiency is reduced by excluding the cerebral complication category. In conclusion it can be stated that the risk factors in DSA are so much smaller that the examinations can be regularly carried out on an outpatient basis.

On the other side of the balance it must be admitted that for intravenous angiography the dose of contrast medium per injection is usually larger than with conventional selective catheterizations. The total dose of contrast medium per examination is limited in both types of examination by the toxicity which the human body (especially the kidneys) can handle. The larger dose for intravenous injections, especially when administered with a high injection rate, may cause problems for patients with a very bad cardiac condition. Stenocardia and/or acute cardiac decompensation may occur in those patients.

A recent new application which is being explored by experienced physicians is to combine the DSA technique with intra-arterial injections. The advantage of this procedure is that a much less than usual contrast dose can be used, which in turn makes it possible to use much thinner catheters than before. The vascular lesion at the site of catheter introduction is so small with this catheter that these examinations can also be applied on an outpatient basis.

2.3. Alternative examination procedures.

As conventional intra-arterial angiography has a certain non-negligible risk of complications, considerable effort has been devoted to developing simple and safe noninvasive examination methods.

The simplest method is definitely auscultation. This method is not reliable enough on its own, but when arterial souffles are detected in a patient with suspected vascular disease, an additional examination will usually be carried out.

In particular the diagnosis of carotid arteries may employ several non-invasive alternative examination methods (de Vries, 1984) like: (1) OPG (Oculopneumopletysmography), (2) Supra-orbital Doppler Ultrasound and (3) Duplex ultrasound scanning.

Real time Ultrasound scanners with high resolution are able to trace peripheral arteries in tissue up to 4 cm depth (van Kaick, 1979). All the noninvasive techniques are not to be considered as replacements of the angiographic procedure, but as aids to arrive at the proper indication for angiography (de Vries, 1984).

In the past, conventional angiography has been regarded as the "gold standard" for vascular diagnosis. Currently the intravenous DSA method has proved to be a valuable substitute for the gold standard.

In those examinations where DSA does not provide enough diagnostic information, catheter-angiography still has to provide the decisive answer. This happens to be required in about 15% of the DSA cases. The reason for an unsuccessful DSA examination is usually an uncooperative patient who can not remain immobile during the exposure sequence. Other drawbacks of the DSA technique are the reduced spatial resolution compared with film recording, and the simultaneous filling of vessels which leads to overprojection problems.

In spite of these apparent important drawbacks of the intravenous DSA technique, the diagnostic information obtained is so valuable and is acquired in such a safe and simple way that it will probably evolve into a main vascular imaging technique in the future.

In the Antonius hospital in Utrecht the number of DSA examinations has grown so much in the last three year period that it presently significantly surpasses the number of conventional catheter angi examinations.
References


3. THE IMAGE SUBTRACTION TECHNIQUE.

3.1. Introduction.

Differentiation of images which are similar in appearance has always been a difficult task. For instance the detection of pulmonary nodules in chest radiographs is a frequently encountered diagnostic problem. Studies have shown (Breckenridge, 1977; Guiss, 1960) that competent observers miss about 30% of the lesions at the first reading. Apart from the contrast and size of the lesion, the complexity of the normal anatomical structures that form the surround of the lesion also plays an important role.

Visual conspicuity is defined by Engel (1976) as the set of factors determining the probability that a visible object will be noticed against its background.

For chest radiographs Kundel tried to quantify conspicuity mathematically by calculating the surround complexity in various ways (Kundel, 1976; Revesz, 1977), but these attempts have not been very successful (Hallberg, 1978).

Irrespective of its quantitative definability, it is clear that qualitatively the conspicuity improves significantly by image subtraction, as the surrounding complexity is removed altogether and only image differences remain.

3.2. Film subtraction.

The basic trick of photographic image subtraction was published as early as 1900 by Galton (1900) who tried to isolate the particulars of human portraits. Successful application of image subtraction in astronomy was reported in 1904 (Pickering, 1904). Ziedses des Plantes (1934, 1935) described the application of subtraction on medical X-ray images.

The early attempts of Robb and Steinberg (1939) to carry out intravenous angiography would certainly have been more successful if they had used the subtraction technique. Either they were not aware of this possibility, or the photographic technique at that time was not yet able to provide good results.

A prerequisite for good film subtraction is the availability of a film that has a gamma of approximately 1.0, because only in that case does the sum of the densities of the original film and of the positive duplicate add up to a constant density level, irrespective of the contrast range in the original film.
As film subtraction is most relevant for angiographic studies, the interest in the subtraction technique grew with the widespread use of catheterization angiography during the late fifties. Especially neurological examinations of the cerebral arteries adopted the subtraction technique as a postprocessing procedure to enhance vessel visibility. In 1961 Ziedes des Plantes wrote an excellent treatise (Ziedes des Plantes, 1961) on the subtraction technique. In addition to the important factors such as the radiation spectrum of the source and the dynamic range of the detector, more subtle topics such as ECG triggering, logarithmic amplification, Time Interval Difference (TID) and energy subtraction are also discussed in this monograph.

During an angiographic procedure at least one radiograph should be made before any contrast medium is present in the vascular structures. From this so called basic film a positive replica, the Mask image, is prepared by contact printing on the 1.0-gamma subtraction film. When the mask is superimposed on contrast-filled radiographs, the observed density will be essentially uniform except where the mask and the later radiograph differ, specifically where the contrast medium is present. A contact copy on a third film is used to store the subtraction result. Altogether, considerable darkroom handling has to be carried out to get the ultimate subtraction result.

Strangely enough, the first international report of the use of film subtraction in combination with intravenous angiography instead of arterial angiography, was published only quite recently (de Lahitte, 1980). A much earlier publication in Dutch (Vlassenroot, 1961) has apparently not received much attention.

The film subtraction technique does indeed improve the conspicuity of the angiographic images by removing bone and tissue shadows. The objective contrast however, expressed in density differences, is hardly increased as there is practically no contrast gain mechanism working (Verhoeven, 1981). For this reason de Lahitte had to use 80 ml of contrast medium per injection in order to obtain acceptable images. This large amount of contrast medium limits the number of injections per patient to two.

Another disadvantage of the film subtraction technique is that the examiner is shooting blind with regard to making the exposures, because he does not know when the contrast medium will arrive and disappear again; timing and duration of the exposure series is just guesswork.

As the photographic subtraction requires time-consuming darkroom handling it is unfeasible to obtain the subtraction results while the patient is still on the table. The second series with the second injection can thus not be guided by the results of the first series.

3.3. Video subtraction.

The first step into the direction of video subtraction was done by Hollman (1963), who proposed to use two vidicon cameras to look at the two radiographs to be subtracted. The time consuming photographic subtraction is avoided by this method and additional amplification of the video difference signal can provide extra contrast enhancement before viewing on a TV monitor. A practical problem of this method is that it is virtually impossible to produce two video cameras which are identical enough to produce a zero difference signal.

This problem was avoided by Oosterkamp (1964), who proposed to use one TV camera and a magnetic image memory. The first image is stored in the magnetic memory and the video camera signal of the second image is subtracted from the stored video signal. In fact this Oosterkamp method is still the basis of the present-day DSA technique. The instant availability of the subtraction result made real-time techniques possible in combination with television fluoroscopy. Reliability problems with the video disc memory at that time precluded development of this video subtraction system beyond the experimental stage. About 10 years later Mistretta and his co-workers picked up the thread of the video subtraction technique by starting out with storage tubes for the video memories (Ott, 1971).

The smaller the contrast differences one wants to see after the subtraction, the higher the requirements are for the stability and reproducibility of the storage media. With the advent of extremely fast analogue-to-digital converters and digital memory chips of high density, both at affordable prices, it became obvious that the digital technique provided the best way to go.

Mistretta's group started building such a digital video processing system in 1976 (Kruger, 1978). A lot of clinical experience was gathered with studies involving dogs (Kruger, 1979) before computerized fluoroscopy, as Mistretta called his method, started to be used on human patients in Sept 1979 (Brother, 1980).

In January 1979 the author of this work visited Mistretta's group to get acquainted with the technique. A small development group at Philips started in August 1979 building a prototype which was installed at the St. Antonius hospital in Utrecht in July 1980. The prototype unit used a 256² matrix size and the digitally subtracted image was converted to analogue again for short-time storage
on an analogue magnetic video disc with 500 tracks.

Right from the beginning the noninvasive character of the examination procedure proved to be a clinical success. The image quality steadily improved because close cooperation with the medical staff at Utrecht gave us the possibility to learn how to improve the apparatus. Approximately 4000 patients have been examined with the prototype machine.

In July 1982 the old prototype was replaced by an improved version which in the meantime had been developed into a commercial product. This new apparatus uses a $512^2$ matrix and digital short time storage on a high capacity (330 Mbyte) Winchester disc. Many technical innovations are used in this product, several microcomputers being incorporated which control the various parts of the system. Interesting details of the hardware and software structure could be discussed, but only a general survey of its features will be presented in the next section.

3.4. Digital subtraction angiography (DSA) equipment.

A block diagram of the DSA equipment is shown in Fig.3.1. All the user interfaces are connected to a central microcomputer which in turn sends all the relevant control messages in the proper sequence to the hardware parts of the system.

In the past the image intensifier (II) and television camera (TV) combination had been used only for fluoroscopic purposes, that is real-time visualization of an X-ray image on TV with an X-ray exposure rate as low as possible, with the purpose of (1) finding the proper location for making an X-ray exposure, or (2) guiding the catheter through the vascular system in preparation for selective arterial angiography. Normally the fluoroscopic exposure rate is of the order of $50 \mu R/s = 1 \mu R/TV$ field, while X-ray exposures require about $200 - 1000 \mu R$ per shot to acquire a reasonably noise-free radiograph.

Note: According to the latest international standardization meeting (International Commission on Radiological Units; ICRU, 1980) the exposure unit shall be expressed in the SI unit $C/\text{kg}$. The older special unit of exposure, röntgen (R), may be used temporarily, and $1R = 258 \times 10^{-6} C/\text{kg}$. (Greening, 1981).

Because of the practical acquaintance that most people will have developed, the older unit röntgen will be used throughout in this text.

As the DSA equipment is aiming at visualizing small contrast differences, fluoroscopic exposure levels are too low (the images too noisy)
and normal exposure dose levels must be used. With this high exposure dose level the TV camera receives much more light from the image intensifier, and in order to prevent TV camera saturation, special measures such as a neutral density filter and/or an adjustable iris diaphragm have to be provided to reduce the light flux by 1 to 2 orders of magnitude. This previously nonexistent function in the II-TV chain is controlled by the microcomputer system as a function of various other imaging technique factors. Other important technique factors are the kV and the mA of the X-ray tube and the exposure time, which all take part in determining the exposure dose per shot. Which combination of the three variables is chosen depends on the clinical application and again the microcomputer system takes care of the control.

The control of all of the imaging technique factors mentioned should operate in such a way that the amplitude of the resulting video signal just fills the maximum input range of the AD converter. In this way the detrimental effect of the digitization on the signal, called quantization noise, is minimal.

It is well known that digitization of video signals with an 8 bit per sample accuracy (256 grey levels) is more than adequate to prevent any subjective deterioration (e.g. contouring) of the displayed image (Devereux, 1974). As soon as two digitized images of nearly equal amplitude are subtracted and subsequently contrast enhanced, a higher accuracy of the quantized raw data is needed.

The sampling frequency of the AD converter depends on the spatial resolution (pixel matrix size) that we want to achieve. As the overall spatial resolution is affected by more factors (e.g. X-ray tube, II, TV camera) than the pixel matrix size alone, the economics of a well designed system require that the choice of pixel matrix size shall be carefully balanced against the other factors.

From the AD converter the digitized video signal is fed to the RAM video memories Mem 1 and Mem 2. The first image of a series, the mask M, is stored in Mem 1; all subsequent images are stored temporarily in Mem 2.

An example of a timing diagram is shown in Fig. 3.2. After an intravenous injection at $t = 0$ it takes some 4 - 6 seconds before the injected contrast medium has passed the right heart, the lungs and the left heart. During this period no superfluous exposures are made of course. The first DSA exposure, producing the mask M, is initiated automatically at the end of a user-adjustable waiting period. Subsequent exposures $I_1$, $I_2$, $I_3$ etc. come at an imaging rate which is also user selectable. A maximum rate of 3/s is adequate for most examinations; only heart studies require a much higher rate.

As Fig. 3.2 shows, during the pauses in between the exposures the difference images $(M-I_1)$, $(M-I_2)$, $(M-I_3)$, etc. are displayed by reading...
the memories Mem1 and Mem2. The fact that the TV camera is not receiving a continuous light input, but flashes of light with dark periods in between, poses special problems which require dedicated measures in the TV camera. Before digital subtraction of the contents of Mem 1 and Mem 2, the image data are logarithmically converted by means of digital look-up tables (LUT in Fig. 3.1).

The need for a log conversion can be mathematically explained as follows.

In general the attenuation of the X-ray flux $\psi$ through an object follows an exponential law:

$$\psi_{\text{out}} = \psi_{\text{in}} \exp(-\mu t)$$  \hspace{1cm} (3.1)

where $t$ is the object thickness and $\mu$ is a material constant (for more details, see chapter 6).

The radiation profile behind an object corresponds to the varying object thicknesses; see Fig. 3.3. The presence of blood vessels of diameter $t_3$ is also indicated. The resulting video signal levels $S$, see Fig. 3.3-b, are assumed to be linearly dependent on the X-ray flux received at the detector.

$$S_1 = S_0 \exp(\mu_1 t_1)$$
$$S_2 = S_0 \exp(\mu_1 t_1 t_2 \exp(\mu_2 t_2 t_3))$$
$$S_3 = S_0 \exp(\mu_1 t_2)$$
$$S_4 = S_0 \exp(\mu_1 t_2 t_3)$$ \hspace{1cm} (3.2)

Linear subtraction would give:

$$S_1 - S_2 = S_0 \exp(\mu_1 t_1) - \exp(\mu_2 t_2 t_3)$$
$$S_3 - S_4 = S_0 \exp(\mu_1 t_2)$$ \hspace{1cm} (3.3)

So although the vessel has the same dimension in both parts of the object, a linear subtraction image would result in a location-dependent vessel contrast.

Log conversion before subtraction ensures that X-ray transmission differences of a fixed percentage will be of equal magnitude in the subtraction image. Log subtraction gives:

$$\ln S_1 - \ln S_2 = (\mu_2 t_2 t_3) \ln S_0$$
$$\ln S_3 - \ln S_4 = (\mu_2 t_2 t_3) \ln S_0$$ \hspace{1cm} (3.4)

In practice not only object thickness variations, but more so bone structures and air pockets in the field of view require the

![Fig. 3.3-a Phantom object including vessels](image)

![Fig. 3.3-b Radiation profile behind the phantom object](image)
logarithmic processing of the signal. Even deficiencies such as non-uniformity of the system (such as the heel-effect of the X-ray tube, vignetting and structural noise) are cancelled in the logarithmic subtraction process.

A demonstration of the pixel value manipulations is shown in Fig. 3.4. Digitized video signals representing a line of the image are shown both with and without the injected contrast material. Due to the high dilution of the contrast material in the vessels the subtraction result has a very small amplitude and signal amplification is required to make the subtle contrast differences visible on display. Shifting the digital words one bit position in the MSB (most significant bit) direction is a simple way of obtaining amplification factors which are a multiple of a factor of two. Special measures have to be taken to prevent possible bit overflow spoiling the pictures; a digital peak clipping circuit has to be incorporated.

In addition to signal amplification, background adjustment is also needed to bring the subtraction signal within a range suitable for the picture display monitor.

Signal amplification will also enhance the inevitable noise by the same amount, as indicated in Fig. 3.4. This noise limits the degree of contrast amplification that can be used in practice.

With respect to the noise it is extremely important that maximum use is made of the dynamic range of the TV camera. Various settings (kVp, mAs, optical aperture, camera gain, etc.) have to be optimized for each particular medical application area.

Several microprocessors are used to take care of the automation of the critical variables, using techniques such as APR (anatomically programmed radiography) and automatic test shots. These measures make the system very "user friendly" and also guarantee consistent optimal image quality.

After DA conversion the real-time difference signals are displayed on a TV monitor so that the examining physician sees the result immediately during and after the injection. From the visible passage of the contrast bolus he can decide when to stop the exposure series (e.g. after T_0 in Fig. 3.2).

During data acquisition the data stored in Mem 1 and Mem 2 are also sent to a digital Winchester disc for storage of the images.

Uninterrupted real-time video display is ensured by making the image memories dual-ported; that means within one pixel period (80 ns in our case) two independent accesses can be made, one for the real-time video and another at the address which turn it is to be sent to the disc.

Standard Winchester discs with their supplementary disc controller are
normally used as a computer peripheral and match standard computer
buses. Image data read from the image memories have to adapt to this
situation and must be packed into a format compatible with the com-
puter bus (16 bit words in our case).
The data transfer speed of the digital Winchester disc results in an
image transfer time of 0.2 s. Additional time required for X-ray
exposure and TV camera read-out lowers the effective imaging rate to
about 3/s.
After the exposure series has been terminated, the acquired images can
be recalled from the disc for diagnostic evaluation. During this
evaluation the most relevant images are selected for archiving on
8 x 10" film transparencies by means of a multiformat video hardcopy
unit.
The success of the image subtraction greatly depends on precise registra-
tion of the two images. Any small movement of the patient results
in serious artefacts which are the more disturbing the more that con-
trast enhancement is applied.
For this reason a so called "postprocessing" mode is available in the
equipment which allows the user to choose a new mask. In the example
of Fig.3.2, I₄ is optimal as regards contrast filling of the vessel
and consequently (M-I₄) will probably contain the most relevant infor-
mation.
If the patient has moved in the interval between I₁ and I₂, the sub-
traction (M-I₄) will contain movement artefacts, and (I₂-I₄) or
(I₁₀-I₄) will probably give a much better image quality.

Important physical aspects influencing the DSA image quality are
the following:
(1) What is the dilution factor of the injected contrast medium
after traversing a significant part of the vascular system.
(2) Given the diluted contrast medium concentration in the vessel,
what is the generated contrast in the image.
(3) Given a certain X-ray exposure dose per image, what is the
X-ray quantum noise contribution and how does it compare with
the weak contrasts of the blood vessels in the images.
(4) What is the influence of additional (electronic) noise sources
in the system, what is the influence of the digitization.
(5) What is the smallest detail one may expect to see, i.e. what is
the spatial resolution capability.

In the following chapters these basic aspects will be discussed in
detail. The end result of the study is a delimitation of the DSA exam-
ination technique and an indication of possible future improvements.

3.5. Further developments.

In spite of the various physical and technical limitations of the
DSA technique, it has already been proved that this non-invasive
angiographic method can replace a substantial part of the more risky
conventional angiography.

Invasive angiography with intra-arterial catheters remains needed
in many cases. The combination of arterial injections and the DSA
technique is more powerful than conventional angiographic examination
due to the improved contrast sensitivity and the instant availability
of the images.
For those examinations where object motion precludes the successful ap-
lication of the subtraction technique, unsubtracted imaging is
preferred which of course can also be performed with the D(S)A equip-
ment. A refinement of the "unsubtracted" imaging method is a sub-
traction mode which could be characterized as "digital shading compen-
sation" (Selbert, 1981; Arnold, 1982; Cowen, 1984). In this technique the
exposures are subtracted from a pre-stored mask which contains the
undesired nonuniformities of the X-ray source and detector (II + TV)
combination.

The introduction of digital imaging techniques in radiology started
about 10 years ago with the availability of CT scanners. CT technology
is now almost mature and a lot of research efforts are presently being
invested in finding digital alternatives for the more conventional rad-
diographic examinations (Digital Radiography).
Obvious advantages are the instant availability of the images and film
cost reductions, because not every acquired image needs necessarily be
kept for archival purpose. In the long-term future the archiving of
medical images may be even completely digital instead of using film
hard copies.

In addition to the replacement of conventional X-ray imaging tech-
niques by their digital counterparts, also previously unknown and more
sophisticated imaging techniques are also being pursued. Examples are:
dual-energy subtraction (Riederer, 1981), hybrid subtraction (Brody,
1981), tomographic DSA (Kruger, 1983) and tomosynthesis (Dümmling,
1967; Grant, 1972; Woelke, 1982).
These new imaging modes are beyond the scope of this thesis. In this
work the basic limitations of digital II-TV image acquisition are dis-
cussed with particular stress on the vascular angiography application.
References.


Grant T.G.: Tomosynthesis: a three dimensional radiographic imaging technique. IEEE Trans. on Biomed Eng. BME 19, 1972, 2028-.


Robb G.P. and Steinberg I.: Visualization of the chambers of the heart, the pulmonary circulation, and the great blood vessels in man. Am. J. Roentgenol. 41, 1939, 1-17.

4. QUANTUM LIMITED LOW CONTRAST DETECTABILITY OF AN X-RAY SYSTEM.

Many of research activities have been devoted in the past to the perceptibility to the human eye of low contrasts in images. Rose (1948) developed a theory which explained the sensitivity performance of the eye at various luminance levels. According to the Rose theory, the ultimate performance level of any picture pick-up device is limited by the number of quanta used and its associated fluctuation noise. At low luminance levels the number of light photons sets the performance limit. Present-day X-ray imaging systems provide so much light output that ample light photons are available; the bottleneck, or quantum sink as it is sometimes called, is now located in the X-ray beam.

When an image is generated from a uniform X-ray exposure with a fluence of \( \phi \) photons/cm\(^2\), then each square image area of size \( d \times d \) receives on the average:

\[
N = \phi d^2
\]  
(4.1)

Fluctuations around this average level occur and as the emission and attenuation of X-ray photons is a Poisson process, the r.m.s. deviation is given by:

\[
\sigma = \sqrt{N}
\]  
(4.2)

This deviation may occur within a picture element (pixel) of size \( d \times d \) when subsequent images are displayed dynamically (e.g. during fluoroscopy). The storage time of the human eye (about 0.2 s) has an effect on the appreciation of the image quality, as this time determines how many X-ray photons are gathered or "noise integrated" to build a subjective image.

When static images (e.g. radiographs) are viewed, the properties of the human eye become more-or-less irrelevant, as storage time does not improve the static images and other viewing conditions such as luminance level, colour, field of view, etc. can be chosen as desired.

In a static image the deviation given by Eq(4.2) occurs for neighbouring pixels in the image and causes a nonuniform or mottled appearance of the image. If a single pixel in the image receives an intentionally different average fluence, the resulting difference signal \( \Delta N \) can be detected only if it markedly surpasses the noise level, or:
The minimum required contrast, $C$, of the pixel is thus:

$$C = \frac{\Delta N}{N} > k_T$$  \hspace{1cm} (4.3)$$

The threshold signal-to-noise ratio $k_T$ has been determined in the past by subjective experiments and various investigators (Rose, 1948; Tol, 1955; Webster, 1962) reported $k_T$ values in the range 2-5.

In addition to experimental determinations of $k_T$, statistical detection probability theory has also been applied to estimate $k_T$ values (Schnitzler, 1973; Rose, 1974). The results are broadly the same and spread as much with circumstances as the experimental values do.

Part of the reported inconsistency in the experimentally-determined values of $k_T$ is probably caused by inaccurate knowledge of the actual number of detected photons. Both the spectral composition of the radiation and the quantum efficiency of the detector for this spectral mixture play an important role in the quantification of the number of photons involved. Both of these factors were inaccurately known in the past.

Another factor which influences the $k_T$ values is the shape of the image detail in the experiment. Very often disc-shaped test patterns of different contrast level (e.g. the Burger (1950) phantom) are used to determine subjectively which disc can be seen and which cannot. While Eq(4.4) is derived for square pixels, circular details of the same dimension have only $\pi/4 = 78\%$ of the area and thus receive correspondingly less photons.

A larger influence of the shape of the test pattern can be expected if a bar is used as the test object. A bar of width $d$ can be considered as an array of many square pixels of size $dx$. Due to this correlation the bar is easier to detect than a single pixel and a lower threshold signal-to-noise ratio $k_T$ can be expected. For circular details we will from now on use $k_T = 3$, based on the reliable experiments of Tol (1955) which show that five is definitely too high and three is the most practical value for the threshold signal-to-noise ratio.

$\Delta N > k_T$

$$C = \frac{\Delta N}{N} > k_T$$  \hspace{1cm} (4.4)$$

The product $C.d$ is thus constant for a given dose level.

Burgess (1977) reports that the contrast threshold for bars is inversely proportional to the square root of the length for lengths up to those subtending 10-20 milliradians of visual angle.

Shaw (1982) reports that slot phantoms require 5 to 10 times less contrast than discs.

These factors make it difficult to decide what the applicable $k_T$ value should be for angiography. As vessels have more resemblance to bars than to discs, one is inclined to use $k_T = 3/5 - 3/10 = 0.3 - 0.6$ as a detection criterion for DSA.

It can be argued, however, that the required criterion strongly depends on the detection task to be performed. At the lowest level the angiographer attempts to demonstrate the presence of patent vessels. This is an easy task and $k_T = 0.3 - 0.6$ could be used. Small "noise induced breaks" in the blood vessel could be tolerated here. A more relevant detection task is the search for vessel abnormalities such as stenosis (narrowing) or aneurysms (distended walls). In this case "noise induced breaks or bulges" cannot be tolerated and $k_T = 3$ for disc-shape detection must be applied.

In the course of this work the more-demanding disc detection task will be discussed and the reader may bear in mind that vessel detection is much easier.

The second important parameter in the contrast detail relationship Eq(4.4) is the particle fluence $\phi$. The conversion factor from exposure dose $D$ in Röntgen (R) to particle fluence $\phi$ depends on the photon energy as is shown for monochromatic radiation in Fig.4.1. In a practical situation the radiation is polychromatic and the spectrum depends on the tube voltage $kVp$ and the beam filtration (beam hardening) by the object. Integral results of the conversion from Röntgen to particle fluence as a function of $kVp$ and object thickness are shown in Fig.4.2. The $55 kVp - 90 kVp$ region represents the diagnostically relevant area for angiographic examinations. Many practical DSA examinations have $65 kVp$ and $13 cm H_2O$ as characteristic parameters which lead to a fluence level of $2.2 x 10^{10}$ photons/cm$^2$/R. Worst case deviations in the $55 kVp - 90 kVp$ range are smaller than $\pm 25\%$.

Substitution of $\phi = 2.2 x 10^{10}$ photons/cm$^2$/R and $k_T = 3$ in Rose's contrast-detail equation Eq(4.4) leads to theoretical predictions of low-contrast detectability performance. Plotting of the results, relating the required contrast with object size for various exposure-dose levels, gives contrast-detail curves as shown in Fig.4.3. Salient points of these curves are quoted in Table 4.1.
In the concept of contrast-detail interchangeability it is tacitly assumed that the eye integrates the noise over the entire area of the object involved, in order to get the full benefit of the available information. This requires matched spatial filtering. Human vision has the remarkable capability of applying its own matched filtering to displayed images. For instance, varying the viewing distance is a way of adapting the viewing performance to the object size, which is strongly advocated in diagnostic practice (Morgan, 1966). More-complicated filtering methods such as electronic two-dimensional lowpass filtering are not needed. Such filtering does indeed reduce the high-frequency noise level of a detail to the predicted level, but the detail detectability in the displayed image is usually not any better compared with an unfiltered display. Hence the psychovisual effect of the eye-brain combination acts similarly to matched filtering. This matched filtering theory of the eye is also supported in other publications (Schnitzler, 1973; Wagner, 1977).

Apart from quantum noise limitations, in the past low-contrast detectability was also limited by the human eye, as the human eye cannot see contrast differences smaller than about 2%. High-gamma film is usually employed in diagnostic radiography, and this enhances the incident exposure contrast by a factor of about 3. With electronic signal processing, as used in DSA equipment, any enhancement factor can be applied and the low-contrast detectability is now limited by the noise in conformity with the theory. So a 0.06% contrast difference for a 1-cm detail should now be detectable with 1 mR of X-ray exposure dose.

Table 4.1: Required contrast for 1mm details.

<table>
<thead>
<tr>
<th>Exposure level</th>
<th>Contrast</th>
</tr>
</thead>
<tbody>
<tr>
<td>1000 µR</td>
<td>0.64 %</td>
</tr>
<tr>
<td>100</td>
<td>2.0</td>
</tr>
<tr>
<td>10</td>
<td>6.4</td>
</tr>
</tbody>
</table>

As a comment on the above theoretical predictions, it must be stressed here that many other practical imaging parameters contribute to a deterioration of the low-contrast detectability, so that in practice the performance will be less than indicated in Fig. 4.3. The main deteriorating factors are:

1. Contrast reduction by scattered radiation.
2. Less than 100% quantum-detection efficiency of X-ray photons.
3. Limited resolution capabilities of the imaging system.
4. Additional noise sources (e.g. electronics noise) of the imaging system.
In subsequent chapters the influence of each of these factors will be discussed, and operational and design constraints of the imaging apparatus will be derived with the intention of approaching the theoretical performance as close as possible.

References.


5. CONTRAST FLOW AND CONTRAST DILUTION IN THE VASCULAR SYSTEM

5.1 Introduction.

As the opacification of the vessels in an angiogram is a reflection of the iodine concentration, one of the primary goals of the angiographic injection procedure is to produce a sufficiently high iodine concentration at the site of interest. Contrast agents with very high concentration (up to 90%, which means 90 g of salt in 100 ml solution) are available, but the high viscosity and the instability of the solution makes them less suitable for standard use. The most frequently used medium concentration agents (60-77%) have an iodine content in the range of 350-400 mg/ml.

With conventional angiography the injection flowrate is the most important injection parameter, as this determines the dilution of the injected contrast medium. The duration is the second important parameter as it determines how large a region of the vasculature will contain highly concentrated iodine at the moment of the X-ray exposure. Although the injected volume is also important for toxicity reasons, it is just a derivative parameter which results from the rate and duration selections.

For midstream large-vessel injections, such as in the thoracic aorta and the main pulmonary artery, a rate of 20-30 ml/s is standard. As the basal flow in these arteries is of the order of 100 ml/s (6 l/min), the dilution factor is 0.17-0.23 leading to an iodine concentration in the range of (55—80 mg/ml. The blood flow is so fast in these main vessels that 2 s of injection duration suffices to fill all the vasculature that one wants to see. The total volume of contrast medium is then 40-60 ml.

High-quality angiograms of smaller vessels, branching off from the aorta, require a higher iodine concentration, which can be achieved by selective injection techniques. Selective injections in e.g. carotid or renal arteries are usually carried out with an injection rate of about 7 ml/s. As the normal blood flow in these arteries is of the order of 8.5 ml/s (Lantz,1981), the expected dilution factor is 0.45, leading to an iodine concentration of about 160-180 mg/ml. This concentration is so high already, that very often lower concentration contrast medium (280-300 mg I/ml) is used for selective injections. With an injection duration of only 1.5 s, the total volume per injection is only 10 ml for selective examinations.

With selective injections in really small vessels (e.g. coronaries), the catheter tip frequently fills or almost fills the vessel, so that a total replacement of blood by contrast medium can occur.

Angiography of the extremities is a very special case, as due to the low blood-flow velocity in this region, an appreciably longer bolus time is needed to fill the interesting artery section. Abdominal bifurcation injections for visualizing both lower extremeties are usually carried out at a rate of 10-15 ml/s and a duration of about 4-5 s. The total volume is then 40-75 ml and with a basal flow of 6 ml/s in each leg, the expected iodine concentration in the vessels is in the range 160-220 mg/ml.

Intravenous injections, which make the DSA technique so valuable, suffer from substantial dilution of the contrast bolus in the cardiovascular system. While the contrast concentration levels for arterial injections are rather straightforward to appraise, the dilution process of intravenously administered contrast medium is of a much more complicated nature. In spite of its paramount importance for the DSA technique, very little has been published on the amount of contrast dilution that can be expected in a practical DSA situation. Kruger (1979) crudely estimated that the iodine concentration in the left heart is 20 mg/ml when 20 ml of Renografin-60 (iodine content 300 mg/ml) is injected in a dog with an injection rate of 6.7 ml/s. Niederer (1981) estimates iodine concentrations of 15-30 mg/ml when original iodine concentrations of 300-400 mg/ml have been injected in man.

In this chapter a mathematical model is developed which makes it possible to calculate the iodine concentration at various locations in the vascular system after an intravenous injection. The influence of important parameters such as: (a) injection site, (b) injection rate, and (c) injection volume can also be predicted by the model.

5.2 The intravenous injection procedure.

The least invasive and in Europe most frequently used intravenous injection procedure uses an angiocath needle (gauge 18) with teflon sheath (i.v. cannula), which is inserted into the basilic vein of the right arm (Fig.5.1). The right arm is preferably used since it gives a slightly shorter pathway to the heart compared with the left arm and also a right arm injection prevents overprojection problems with the aortic arch.

After withdrawal of the needle the teflon sheath with a length of 4.5 cm remains in the arm vein pointing in the downstream direction. A
A high pressure injector is used to deliver a preset amount of contrast medium at a preset flow rate.

For adult patients, 40 ml contrast solution of Ioxthalamate (Telebrix 38) is normally used. For children, a lesser amount can be used; in general, injection volumes of 0.5 ml/kg of body weight are suitable. The injection rate is chosen as high as possible without risking a rupture of the vein with the painful extravasation of the contrast medium.

A flow rate of 12-14 ml/s has proved to be on the safe side. The administration of the contrast solution is immediately followed by injection of 20 ml of a 5% glucose solution to provide a bolus propellant. The subsequent injections of contrast medium and glucose solution are achieved by layered filling of the injector syringe. During the injection, the arm is elevated because this prevents any postural obstruction of the subclavian vein in the armpit and also generally improves the venous flow in the arm.

The patient is instructed to hold his breath in inspiration during and after injection as the consequent negative intrathoracic pressure helps to provide a rapid venous inflow into the right heart. The so-called "Valsava maneuver", where a strained inhalation position is held, must be prevented as this condition might lead to a positive intrathoracic pressure high enough to completely block venous inflow into the heart during this period. A good coordination of the patient's breathing action is thus very important.

As stated above, the injection in the arm takes place at a rate of about 13 ml/s through an 18-gauge i.v. cannula. The inner diameter of the cannula is 1.24 mm and thus the cross-sectional area is 1.21 mm$^2$. An injection of 13 ml/s through this small a hole leads to a very high flow velocity:

$$v = \frac{13000 \text{ mm}^3/s}{1.21 \text{ mm}^2} = 10.7 \text{ m/s}$$

Probably this high flow velocity forms the limiting factor of the safely usable flow rate. Impact of the injected flow stream against the vessel wall might seriously damage or even perforate the latter. Larger bore needles such as 17 gauge and 16 gauge with diameters of 1.47 and 1.65 mm leading to flow velocity of 7.6 and 6.1 m/s are probably less harmful in this respect.

According to Lipton (1978), the jet energy remains at a high value over a distance of about 7 exit-hole diameters (8.7 mm for 18 gauge) and thereafter rapidly decays to a harmless level at a distance of about
20 exit-hole diameters (25 mm for 18 gauge). Lipton's study recommends a safe injection rate of 14 ml/s for an exit-hole area of 1.21 mm².

Under normal circumstances the blood flow in the arm vessels has been shown to be about 6% of the cardiac output (Lantz, 1981). The three contributors to this flow are the cephalic vein, the brachial vein and the basilic vein (Fig. 5.1). The relative sizes of these vessels can vary from patient to patient, but we estimate that for the average patient the contributions of the three are equal at 2% of the cardiac output.

For a cardiac output of 6 l/min = 100 ml/s the normal blood flow at the injection site is thus about 2 ml/s, which is much less than the injection flow of 13 ml/s. The question is now, how the injected vein handles this abundant supply of injected contrast solution. In practice a very distinct dilatation of the vessel can be seen and palpated. This vessel dilatation to accommodate a factor of 7.5 times more flow ((13 + 2) : 2) is by no means extraordinary because during physical exercise skeletal muscles can increase their blood flow by a factor of 20 (Rushmer, 1976).

When the injection stops, the flow conditions in the distended arm vessel gradually return to the normal situation of 2 ml/s; see Fig.5.2. The importance of injecting the 20 ml glucose solution can be now appreciated. Without the glucose injection the last part of the contrast medium would experience a prolonged clearance of the arm vessel. The glucose injection functions as a pressure head which pushes most of the contrast solution out of the arm vessel before the injection stops. The inflow of 13 ml/s of contrast from the subclavian vein via the brachiocephalic vein into the superior vena cava does not change the flow conditions in the latter vein appreciably, as the normal flow rate of about 40 ml/s (Lantz, 1981) is increased by only 32% which can be handled easily by a slight distension.

In order to study the transient phenomena in the arm, some cinefluorographic X-ray films (50 images/s) of the arm have been taken during injections. Apart from a gradual build-up of contrast concentration in the early stage, and the vessel distension and full contrast concentration later on, two other important phenomena have been observed:

1. When the arm is not elevated during injection, a definite sedimentation or layering of the contrast medium may occur, due to the higher density of the contrast solution compared with blood (ρ = 1.49 g/ml for Telebrix and ρ = 1.0 g/ml for blood). (see also Swart, 1962).
The glucose part of the injection may then prove to be useless as a pressure head, as it slides over the contrast layer which remains at the bottom of the vessel.

2. When rapid venous flow is obstructed due to either an anatomical reason or mechanical obstruction in the armpit as a result of an unelevated arm, injection can even produce a reverse flow of contrast material into the lower arm, where fill of the other two main veins takes place via inter connecting veins.

Both observations emphasize the importance of the arm's position, since for good bolus conservation the dwell time of the contrast medium in the arm must be as short as possible.

With a correct injection procedure, the bolus-front appearance is observed after about 0.6 s at the armpit and about 1.6 s at the right heart.

Another often-used intravenous injection technique uses a catheter that has been advanced via the arm vein up to the vena cava or even into the right heart. If enough side holes are present at the catheter tip the injection rate may be increased to 30-40 ml/s, so a more compact bolus can be administered by this method.

The contrast bolus improvement with this more invasive method is indisputable, especially for the nearby pulmonary arteries. In the following sections it will be shown that it is the amount of injected iodine, and not so much the injection rate, that really matters when more additional path length has to be traversed for the imaging of more distal vessels.

5.3 The dye dilution technique.

The dye dilution technique is a classical method of determining the cardiac output. With this method a rapid intravenous injection of an amount of $m_i$ of indicator ("dye") is made and the concentration $C(t)$ in g/ml is subsequently measured by taking blood samples at a point downstream in the arterial system. Conservation of mass requires that the mean flow $F$ times the concentration $C(t)$ at any time must add up to the injected mass, i.e.:

$$m_i = \int_0^\infty F C(t) \, dt$$

(5.1)

So if $m_i$ and $C(t)$ are known, the flow $F$ in ml/s (or l/min) can be calculated.

In this chapter a different use is made of Eq.(5.1); instead of trying to find the cardiac output, this quantity is used as a known factor, together with the shape of the $C(t)$ curve, to find $C(t)$.

As an approximation we assume a $C(t)$ curve of triangular shape as in shown in Fig.5.3. Application of Eq.(5.1) gives:

$$m_i = F \int_0^\infty C_{\text{max}} \, dt$$

or:

$$C_{\text{max}} = \frac{2m_i}{F T_s}$$

(5.2)

where $T_s$ = bolus spread time.

Although the $C(t)$ shape is only approximated in this example, Eq(5.2) indicates the influence of the main relevant parameters.

Substitution of some practical values which are applicable to intravenous DSA studies such as $m_i = 40 \times 0.38 = 15.2$ g of iodine, $F = 100$ ml/s and $T_s = 7.5$ s, produces $C_{\text{max}} = 40$ mg/ml.

Of course a more precise description of $C(t)$ than in the example above is needed. Many publications (Newman,1951; Stow,1954; Bassingthwaighte,1966) give a theoretical or empirical mathematical expression for the shape of the dilution curve. Most of them just produce a curve-fitting expression without the intention of representing the physical effects leading to its formation. Especially the appearance time of the bolus and its relation to the shape of the curve has had hardly any attention so far in the mathematical expressions. As both the appearance time as well as the concentration are of importance for the DSA technique, this chapter proposes a new model that tries to include the physical factors involved in the formation of the dilution curve.

5.4 Contrast dilution curve as a convolution product.

After the intravenous injection the contrast medium has to traverse a considerable part of the circulating system before it arrives at the site of interest (see Fig.5.1 and Fig.5.4).

A useful approach for determining the contrast dilution curve at various locations downstream of the injection site is to consider the vascular circuit as a sequence of cascaded linear filter sections.

Each filter section has its own impulse response function $h_j(t)$ and the response of the circuit at a certain location can be found by a convolution of the stimulating waveform $x(t)$ (the injection) with each of the subsequent impulse response functions $h_j(t)$.
The contrast concentration then becomes:

\[ C(t) = \frac{m_i}{F} x(t) * h_1(t) * h_2(t) * h_3(t) \cdots \]  

(5.3)

The scalar constant \( \frac{m_i}{F} \) results from an application of the conservation of mass principle as expressed by Eq (5.1).

When \( C_i, F, T_i \) are the injected concentration, the injection rate and the injection duration respectively, then it follows that:

\[ m_i = C_i F T_i \]

So:

\[ C(t) = \frac{F T_i}{C_i} x(t) * h_1(t) * h_2(t) * h_3(t) \cdots \]

(5.4)

The appearance of the individual waveshapes to be convolved follows from the dispersion mechanism that prevails in the relevant section. The main factors determining the dispersion in the vascular system are: (1) the velocity profile in each vessel; (2) mixing chamber effects in the ventricles and (3) variation in pathway lengths. Each of these factors will now be discussed separately.

5.4.1 Dilution due to the velocity profile.

Due to the friction forces in the viscous blood fluid, the velocity in the middle of the vessel is higher than near the wall. It is generally accepted (Mc Donald, 1974) that laminar blood flow exists in most vessels of the vascular system with the possible exception in the aortic arch where turbulence may be expected.

Laminar flow is usually assumed to be a Poisseuille type with a parabolic velocity distribution.

\[ v(r) = \nu_0 (1 - r^2/R^2) \]

(5.5)

where \( v(r) \) is the velocity at radius \( r \) and \( R \) is the inner radius of the vessel. Gonzales-Pernandes (1962) indicated methods of calculating dispersion with this kind of velocity distribution; an impulse type of injection produces an impulse response:

\[ h(t) = 0 \quad \text{for } 0 < t < T_L \]

\[ h(t) = \frac{T_L}{t^2} \quad \text{for } t > T_L \]

(5.6)
where \( T_l \) is the arrival time of the bolus front after traversing a distance \( L \).

The problem with this impulse response is that the tail of the curve is too long to be compatible with practical observations regarding contrast dispersion in blood vessels. A judicious correction of the velocity profile has to have the result that a smaller percentage of particles will travel at the very low velocity; the velocity profile must be much blunter. This conclusion has also been drawn by Bassingthwaighte(1974). In order to obtain a better fit, an impulse response with a shorter tail is proposed:

\[
h(t) = 0 \quad \text{for} \quad 0 < t < T_l
\]

\[
h(t) = \frac{3}{T_L} \exp[-3(t/T_L - 1)] \quad \text{for} \quad t > T_l
\]

(5.7)

An exponential relationship has been chosen intuitively as these forms are usually easy to manipulate. The desired shape of the curve has been obtained by the selection of the factor 3 in the expression. Fig.5.5 shows that the new impulse response is much shorter. Because of the shorter duration, the amplitude of the impulse response must be higher as impulse responses must by definition have unit area under the curve.

The radial velocity distribution applicable to this impulse response is given by:

\[
v(r) = \frac{v(0)}{1 - \frac{1}{3} \ln(1 - r^2/R^2)}
\]

(5.8)

The interrelationship between Eq(5.7) and Eq(5.8) will now be shown by starting with Eq(5.8) as this sequence more logically follows the physical effects in the vessel.

Laminar flows, especially of highly viscous fluids like blood and contrast solution, show a disinclination to mix with each other as proved by Kjellberg(1943). For this reason a well-defined bolus front develops after a steplike start of an injection; see Fig.5.6.

Each particle in the fluid travels at its own velocity and the distance covered by a particle of the bolus front at radius \( R_l \) and at time \( t \) is:

\[
L = v(R_l) t = \frac{v(0) t}{1 - \frac{1}{3} \ln(1 - R_l^2/R^2)}
\]

(5.9)
Solving for $R_L$ gives:

$$R_L = R \sqrt{1 - \exp(3(t - v_0)/v_L)} \quad \text{for } \frac{v_0 t}{L} > 1 \quad (5.10)$$

By defining a tip-of-the-bolus front arrival time $T_L$, Eq(5.10) transforms into:

$$R_L = R \sqrt{1 - \exp(3(t - T_L)/v_0)} \quad \text{for } t > T_L \quad (5.11)$$

Due to the non-mixing behaviour of the bolus, the local average cross-sectional concentration $C(L,t)$ at distance $L$ and time $t$ is determined by the area ratio (Fig.5.6):

$$C(L,t) = \begin{cases} 0 & \text{for } t < T_L \\ C(0,0) \frac{R^2}{\pi R} & \text{for } t \geq T_L \end{cases} \quad (5.12)$$

The second factor in this expression can be considered as the step response function $H(t)$ of a steplike injection. The impulse response $h(t)$ of an infinitely short injection can be found by differentiating $H(t)$. The result is Eq(5.7).

Comparison of the parabolic velocity distribution Eq(5.5) with the newly-proposed distribution Eq(5.8) shows that the latter is much blunter indeed, as was needed for the physical explanation of the observed facts; see Fig.5.7.

A characteristic attribute of the velocity distribution which will be used later on is the relationship of axial velocity $v(0)$ to the average velocity $v_{av}$.

The total flow in the vessel follows from:

$$F = \pi \int_0^R \sqrt{\frac{R}{2r dr}}$$

$$= \pi \int_0^1 \frac{R^2 v(0)}{\ln(1 - x)}$$

The solution of this integral can be found in Gradshteyn(1965):

$$F = \pi R^2 v(0) \cdot \frac{2}{3} \cdot \frac{2}{3}$$

Fig. 5.7. Comparison of velocity profiles
where \( E_i \) denotes the exponential-integral function. A tabulation of this function is given by Jahnke (1945):

\[
F = 0.786 e^{\gamma}(w(0)
\]

Thus the average flow velocity is:

\[
v_{av} = \frac{F}{r} = 0.786 v_{w(0)}
\]

(5.14)

An interesting feature of the dispersion mechanism described is that a single easily-determinable parameter, \( T_L \), completely characterizes both the arrival time and the corresponding degree of dispersion in a single vessel route.

As mentioned in section 5.2, the observed arm to right heart transit has a bolus tip travel time \( T_L = 1.6 \) s. The travel time in other vascular sections will be discussed in section 5.4.4.

5.4.2. Dispersion in the heart chambers.

The right ventricle and left ventricle can be considered as ideal mixing chambers where any existing radial-dependent contrast concentration is dissolved in such a way that, after these mixing chambers, a new flattened bolus front is ejected. During further transit a new conically shaped bolus front will reappear due to the laminar flow conditions. As a result of the presence of the two mixing chambers the impulse responses of the venous circuit, the pulmonary circuit and the systemic circuit can be considered as independent of each other.

In addition to their function as a separating stage, the heart chambers also have an impulse response function of their own.

Suppose that a small amount of iodine \( m_i \) is instantaneously introduced into a ventricle during diastole.

If the end-diastolic volume is \( V_d \) and if complete mixing is achieved, the iodine concentration will initially be:

\[
C_0 = \frac{m_i}{V_d}
\]

(5.15)

During the first following systole a quantity of blood (stroke volume) \( V_s \) is ejected, leaving a residual volume \( V_r = V_d - V_s \).

During the next diastole, new blood supply dilutes the remaining iodine in the ventricle to a concentration.

\[
C_1 = \frac{m_i}{V_d V_g}
\]

(5.16)

The same dilution process will take place at each subsequent heart beat, so after a heartbeats the concentration will be reduced to:

\[
C_n = \frac{m_i}{V_d V_g}(1 - EF)^n
\]

with \( EF = \frac{V_s}{V_d} \)

(5.18)

In this formula \( EF \) is the ejection fraction of the ventricle.

When the heart rate is 80/min, each step in the dilution process takes \( T_h = 0.75 \) s as is shown in Fig. 5.8. The pulsing character of the blood flow is represented by the staircase shaped dilution curve. As this staircase waveform is difficult to handle analytically, the impulse response function is approximated by the dashed curve, which has an exponential character.

The impulse response of the heart chamber is then given by:

\[
h(t) = \begin{cases} 
0 & \text{for } t < \frac{T_h}{2} \\
\frac{1}{\tau} \exp(-t/T_h) & \text{for } t > \frac{T_h}{2} 
\end{cases}
\]

(5.19)

So in addition to an exponential downslope a delay of 0.5 \( T_h \) is also involved. The time constant \( \tau \) of the curve follows from:

\[
\frac{m_i}{V_d} \exp(T_s/\tau) = \frac{m_i}{V_d} (1 - EF)
\]

(5.20)

Solving for \( \tau \) gives:

\[
\tau = \frac{T_h}{\ln(1 - EF)}
\]

(5.21)

For the average healthy adult \( EF = 0.67 \) and under resting conditions \( T_h = 0.75 \) s. Substitution in Eq (5.21) produces \( \tau = 0.68 \) s.

After a heart infarction \( EF \) may be reduced substantially. Compensation mechanisms, such as a slightly increased heart rate and a gradually by developing dilation of the ventricle, attempt to restore the cardiac output at rest.

When the ejection fraction is reduced to for instance \( EF = 0.43 \) the time constant \( \tau \) of the ventricle becomes twice as high or \( \tau = 1.36 \) s.
5.4.3. Dispersion in the capillary bed.

The capillary bed of lungs and tissue consists of a branching system with numerous parallel paths. The variation in the pathway lengths contributes to the dispersion of the injected material. When a uniform pathway-length distribution is assumed, a square-shaped impulse response will result. Recent studies of Axel (1983) indicate that this assumption fits reasonably well for the capillary bed in the brain. For the pulmonary capillary bed the transition times are assumed to be uniformly distributed over the 1.5 - 3.0 s range (Fig. 5.9).

The impulse response of this section is then:

\[ h(t) = \begin{cases} \frac{1}{\tau} & \text{for } r < t < 2\tau \\ 0 & \text{for } t \leq r \\ 0 & \text{for } t \geq 2\tau \end{cases} \quad (5.22) \]

In addition to the capillary flow response, the response of the pulmonary arteries and veins (velocity profile dispersion) has also to be taken into account in order to obtain the complete pulmonary circulation impulse response.

5.4.4 Appraisal of flow velocity in the vascular system.

The travel time \( T_0 \) of the tip of the bolus front for various vascular sections, as needed in the velocity profile calculations of Eq (5.12), can be estimated if both section length and axial velocity are known.

Due to the beating of the heart, the velocity is highly pulsatory in the main vessels and consequently the progress of the blood particles has a stepwise character. For our purpose it is more convenient to use the average flow velocity over the cardiac cycle.

Average velocity estimations follow from flow rate and vessel diameter data, as presented in Table 5.1. Flow rates in this table are taken from the Lantz (1981) data and an assumed cardiac output of 100 ml/s. Vessel diameter data are taken from Ray (1974); Shah (1978); Caro (1978); Katck (1979); Stein (1979); Callum (1983) and Cronenwett (1983).

Table 5.1 shows that the flow velocity is highest in the ascending aorta at about 25 cm/s. Along its course down the aorta a substantial flow branches off, while the aorta tapers only slightly. This causes the flow velocity in the distal abdominal aorta to be reduced to about 10 cm/s.
Flow velocity in large veins, such as SVC will be about 10 cm/s, and in peripheral veins about 7 cm/s. These estimates hold for the average healthy adult at rest in the supine position.

As discussed in section 5.4.2 the momentary blood flow has a pulsatory character, so the progress of the bolus front will also show a kind of step and wait character. The calculations discussed here neglect this detail and average data are considered, based on the average flow measurements of Lantz.

Table 5.1. Estimations of average blood flow velocity

<table>
<thead>
<tr>
<th>Section</th>
<th>Basal flow</th>
<th>Lumen diam.</th>
<th>Cross sectional area</th>
<th>( \bar{V_B} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Basilic v.</td>
<td>2 ml/s</td>
<td>6 mm</td>
<td>0.28 cm²</td>
<td>7.1 cm/s</td>
</tr>
<tr>
<td>Axillary v.</td>
<td>4</td>
<td>8.4</td>
<td>0.56</td>
<td>7.6</td>
</tr>
<tr>
<td>Subclavian v.</td>
<td>6</td>
<td>10</td>
<td>0.79</td>
<td>7.6</td>
</tr>
<tr>
<td>Brachiocephalic v.</td>
<td>20</td>
<td>16</td>
<td>2.0</td>
<td>10</td>
</tr>
<tr>
<td>Superior Vena Cava</td>
<td>40</td>
<td>23</td>
<td>4.0</td>
<td>10</td>
</tr>
<tr>
<td>Main pulm. a.</td>
<td>100</td>
<td>23</td>
<td>4.0</td>
<td>25</td>
</tr>
<tr>
<td>Ascending aorta</td>
<td>100</td>
<td>7.6</td>
<td>0.45</td>
<td>19</td>
</tr>
<tr>
<td>Carotid a.</td>
<td>8.5</td>
<td>12-21</td>
<td>1.2-3.5</td>
<td>10-18</td>
</tr>
<tr>
<td>Abdominal a.</td>
<td>12-62</td>
<td>8.0</td>
<td>0.50</td>
<td>12</td>
</tr>
<tr>
<td>Iliac a.</td>
<td>6</td>
<td>6.0</td>
<td>0.28</td>
<td>12</td>
</tr>
<tr>
<td>Superficial femoral a.</td>
<td>3.5</td>
<td>4.5</td>
<td>0.16</td>
<td>11</td>
</tr>
<tr>
<td>Popliteal a.</td>
<td>1.7</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 5.2. Bolus front travel times.

<table>
<thead>
<tr>
<th>Circulation section</th>
<th>( T_L )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Basilic v. - SVC</td>
<td>1.6 s</td>
</tr>
<tr>
<td>Pulm. a. + Pulm v.</td>
<td>1.5</td>
</tr>
<tr>
<td>LV - carotid a.</td>
<td>0.7</td>
</tr>
<tr>
<td>LV - cerebrum</td>
<td>1.5</td>
</tr>
<tr>
<td>LV - renal a.</td>
<td>1.4</td>
</tr>
<tr>
<td>LV - bifurcation</td>
<td>2.4</td>
</tr>
<tr>
<td>LV - popliteal a.</td>
<td>5.6</td>
</tr>
<tr>
<td>LV - foot</td>
<td>10.6</td>
</tr>
</tbody>
</table>

5.4.5. Summary of waveshapes to be convolved.

A summary of the impulse response functions \( h_j(t) \) of the subsequent stages is shown, more-or-less to scale, in Fig.5.10. The time constants which will be used in the convolution calculations are:

\[
T_1 = 1.6 \text{ s} \\
T_2 = T_5 = 0.70 \text{ s} \\
T_3 = 0.75 \text{ s} \\
T_4 = 1.5 \text{ s} \\
T_6 = \text{ dependent on site of interest, see Table 5.2}
\]

5.5 Practical application of the contrast dilution model.

After the extensive discussions of the contribution of the subsequent parts of the circulation to contrast dilution, the overall results can now be obtained by carrying out the convolution product Eq(5.4).

The results of convolutions calculated by a computer are shown in Fig.5.11 and Fig.5.12. The circumstances used are typical, that means the patient is the average healthy human at rest.

The flow rate \( F \) to be substituted in Eq(5.4) requires a remark. Due to the injection the venous supply to the right heart temporarily increases and for \( C(t) \) calculations of the pulmonary arteries this higher flow rate has to be substituted.

Although on the average the outputs of the right heart and left heart must be equal, it is reasonable to expect that the output of the left
Fig. 5.10 Summary of waveshapes to be convolved

\[ x(t) = \frac{1}{T_1} \quad \text{for } 0 < t < T_1 \]

Arm - right heart

\[ h_1(t) = \frac{3}{T_1} e^{-\frac{t}{T_1}} \quad \text{for } t > T_1 \]

Right ventricle

\[ h_2(t) = \frac{1}{T_2} e^{-\frac{t - 0.5 T_1}{T_2}} \quad \text{for } t > 0.5 T_1 \]

Pulm a. + pulm v.

\[ h_3(t) = \frac{3}{T_3} e^{-\frac{t}{T_3} - 1} \quad \text{for } t > T_3 \]

Pulm cap bed

\[ h_4(t) = \frac{1}{T_4} \quad \text{for } t < T_4 \]

Left ventricle

\[ h_5(t) = \frac{1}{T_5} e^{-\frac{t - 0.5 T_1}{T_5}} \quad \text{for } t > 0.5 T_5 \]

Arterial route

\[ h_6(t) = \frac{3}{T_6} e^{-3\frac{t - 1}{T_6}} \quad \text{for } t > T_6 \]

Fig. 5.11 Calculated \( C(t) \) for arm injection

Conditions: 40 ml contrast volume + 20 ml flushing volume at a rate of 13 ml/s, \( T_i = 3 \) s

The sequence of the different curves is: (1) pulm a., (2) aortic arch, (3) cerebral a., (4) popliteal a.

Fig. 5.12 Calculated \( C(t) \) for SVC injections (\( T_i = 0 \))

Conditions: 40 ml of contrast volume at a rate of 30 ml/s, \( T_i = 13 \) s

Same sequence as in Fig. 5.11
The heart does not immediately respond to the injection. The pulmonary blood will act as a buffer and the short-period increased right heart output is accommodated by the left heart during a much longer transient period. For simplicity of calculation, the output of the left heart is assumed to be unaffected by the injection.

The important practical bolus data such as appearance time $t_a$, peak concentration level $C_{\text{max}}/C_1$, bolus peak time $t_p$ and bolus spread time $t_s$ are summarized in Table 5.3 and Table 5.4. Bolus spread time $t_s$ is defined as the time difference at a height of one-tenth of the peak concentration.

Table 5.3. Bolus data for arm injections.

<table>
<thead>
<tr>
<th>Location</th>
<th>$t_a$</th>
<th>$t_p$</th>
<th>$t_s$</th>
<th>$C_{\text{max}}/C_1$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pulmonary a.</td>
<td>2.0 s</td>
<td>4.8 s</td>
<td>5.2 s</td>
<td>0.113</td>
</tr>
<tr>
<td>Aortic arch</td>
<td>6.0 s</td>
<td>9.2 s</td>
<td>6.7 s</td>
<td>0.103</td>
</tr>
<tr>
<td>Cerebral a., Renal a.</td>
<td>7.6</td>
<td>10.9</td>
<td>6.9</td>
<td>0.099</td>
</tr>
<tr>
<td>Popliteal a.</td>
<td>12.1</td>
<td>15.9</td>
<td>9.4</td>
<td>0.075</td>
</tr>
</tbody>
</table>

Table 5.4: Bolus data for SVC injections.

<table>
<thead>
<tr>
<th>Location</th>
<th>$t_a$</th>
<th>$t_p$</th>
<th>$t_s$</th>
<th>$C_{\text{max}}/C_1$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pulmonary a.</td>
<td>0.4 s</td>
<td>1.5 s</td>
<td>2.7 s</td>
<td>0.210</td>
</tr>
<tr>
<td>Aortic arch</td>
<td>4.1 s</td>
<td>6.2 s</td>
<td>5.0</td>
<td>0.140</td>
</tr>
<tr>
<td>Cerebral a., Renal a.</td>
<td>5.6</td>
<td>7.9</td>
<td>5.5</td>
<td>0.129</td>
</tr>
<tr>
<td>Popliteal a.</td>
<td>10.0</td>
<td>12.8</td>
<td>8.3</td>
<td>0.087</td>
</tr>
</tbody>
</table>

The parameter $t_a$, $t_p$ and $t_s$ are in very close agreement with observations during DSA examinations. Of course there is a patient-dependent spread in the timings, but the data presented in Tables 5.3 and 5.4 are a good representation of what can be expected in the case of a reasonably-healthy adult patient. This was to be expected of course, as in the proposed model, which is based on an understanding of the physical factors involved, practical numerical factors are substituted to match reality.

A real validation of the predicted contrast dilution factor can be done by taking blood samples at various places during an examination. This type of experiments were not under control of the investigator and were not carried out. Experimental data of Burbank (1983), who measured contrast dilution in dogs, confirm the results of the dilution model proposed in of this chapter.

The practical benefits of the contrast dilution model presented are that predictions can be given of:

1. the contrast dilution factor for various anatomical regions,
2. the influence of injection parameters such as: injection rate, injection volume and injection site,
3. the shape of the time-concentration curve, which can be used for determining the proper exposure moments.

As an example, the influence of injection rate on the concentration curve is calculated and presented in Fig. 5.13. In an additional example the influence of injection volume on the concentration curve is calculated; see Fig. 5.14.

5.6 Discussion.

The results of the contrast dilution calculations indicate that the most frequently examined anatomical areas such as extracranial, intracranial and abdominal arteries have about equal dilution factors. As a practical example: arm injection of 40 ml of contrast agent with iodine concentration $C_i = 380$ mg/ml will give a peak iodine concentration at the carotids of about $0.10 \times 380 = 38$ mg/ml.

For the lower peripheral arteries it takes longer for the bolus to arrive, and therefore also more contrast dilution occurs. Partial occlusions may even aggravate this.

SVC injections with a venous catheter instead of needle injections in an arm vein are somewhat better as regards the peak bolus concentration. The difference between the two injection techniques becomes less, the more remote the site of observation is from the injection site.

The explanation for this is that injection durations, which are short compared with the overall impulse response of the circulation trajectory considered, can be considered almost as Dirac-type excitations. The response shape is then determined only by the impulse response of the system and not by the excitation waveform itself.
Bolus peak-time differences of SVC injections and arm injections are about 3.0 s, partly due to the shorter pathlength and partly due to the shorter injection period (the contrast dose is earlier in the circulation).

The predicted improvement in peak concentration, obtained by the more invasive SVC injection, is about 35% for carotid arteries. In the literature both Hetzel (1954) and Modic (1983) came to the same conclusion. The model presented here also concludes that for arteries in the legs the difference is only 16%.

An attempt to improve the concentration level by increasing the injection rate above the usually applied levels (13 ml/s for arm injections and 30 ml/s for SVC injections) is predicted to be marginally effective; see Fig. 5.11.

Injection of more contrast dose results in a nearly linear increase in peak concentration as long as the injection time $T_i$ is short compared with the system impulse response; see Fig. 5.14. The 80 ml dose curve in this figure (with $T_i = 6$ s) shows already some leveling; the peak concentration is only 27% higher than the peak of the 40 ml curve.

From the shape of the contrast concentration curve for various locations in the body we are able to decide at what times the exposures should be made to acquire a useful series of images. Predicted appearance times $t_a$ can be used to set the "waiting time" as defined in section 3.3 and Fig. 3.2.

Patient influence on appearance time has been investigated by Schad (1981) who advocated the use of the heart rate as a measure of the flow velocity in the vascular system. He reported that after an arm injection of contrast medium the concentration peak of the bolus arrives in the left ventricle after 11 heart beats + 4% for a heart rate range of 50-100/min.

As the injection period reported by Schad is very short compared with ours, we have the concentration peak at the arch a little later at about 12 heartbeats.

Actually, the cardiac output is a better measure of flow velocity in the blood vessels. Heart rate is related to cardiac output by:

$$\text{cardiac output} = \text{stroke volume} \times \text{heart rate}$$

Thus the heart rate may serve as an indicator for cardiac output, provided that the stroke volume is constant. This latter condition is hardly realistic, especially for the elder and unhealthy patient. This category of patients was therefore excluded by Schad in his study.
In spite of the limited patient category for which Schad's heart beat timing holds, this method may be used to make a first estimate of the "waiting time" after injection to make the first "mask" image of a DSA series.

In addition to the determination of the useful "waiting time", the C(t) curves may also be used to deduce the appropriate imaging rate. A rate as low as possible should be used because it reduces the radiation dose to the patient. Moreover, if the available heat load capacity of the X-ray tube is spread over a smaller number of exposures, the higher heat load per exposure allows the use of a lower kV setting for a given dose level, which in turn improves the radiographic contrast. (see chapter 6).

On the other hand, if too low an imaging rate is used, the sampling points on the concentration curve become so widely spaced that the peak concentration which gives the best image can be missed easily. A useful imaging rate is obtained if 8 sampling points (7 intervals) are distributed over the bolus spread time. With this rate we are sure that there is always a sample with less than 5% deviation from the peak concentration; usually this is the 3rd or 4th after arrival of the contrast medium.

The above criterion leads to a maximally needed rate for arm injections of about 1/s; for SVC injections about 1.4/s is maximally needed.

Pulmonary artery examinations form a special case as the pulmonary arteries are so close to the beating heart, that in order to obtain good subtraction images, the exposure rate has to be synchronized with the heart rate (ECG triggered exposures) (Ludwig, Verhoeven,1983).

According to Shillingford, an increased bolus spread can be expected with increased central blood volume. This situation of pulmonary congestion is typical for patients with cardiac failure.

The result of valvular incompetence can be calculated by assuming a decreased ejection fraction. Substitution of EF = 0.40 for the left ventricle results in only a slightly lower $C_{\text{max}}$ and a slightly longer tail of the curve, which is in accordance with Shillingford's findings.
5.7. Conclusions.

The proposed mathematical model for calculating the contrast dilution after intravenous injections provides a detailed insight into the dilution process. Reasonably accurate predictions of the iodine concentrations at various locations in the vascular system can be obtained, and the influence of non-standard conditions can also be evaluated.

In general it can be concluded that SVC injections do result in a higher bolus concentration than arm injections, but the difference is not great and becomes smaller the more remote the location of interest is from the site of injection. The need for the more invasive SVC injections is thus questionable in this respect.

Increasing the injection rate above the normally used levels is predicted to be marginally effective for increasing the peak concentration level. This is quite contrary to common experience with intra-arterial injections where contrast usually does increase with injection rate.

With intravenous injections, the main factor determining the peak concentration is the total amount of injected iodine. Increasing the injection volume is effective only as long as the injection period remains short compared with the system impulse response.

As a recapitulation, a comparison of expected contrast dilution factors for intravenous and conventional arterial injections is shown in Table 5.5.

Table 5.5: Iodine concentration in carotid artery for different injection procedures (contrast agent with 380 mg I/cm³).

<table>
<thead>
<tr>
<th>Injection procedure</th>
<th>Injection rate</th>
<th>Basal flow</th>
<th>Dilution factor</th>
<th>Iodine content</th>
</tr>
</thead>
<tbody>
<tr>
<td>i.v. arm injection (40 cm³)</td>
<td>13ml/s</td>
<td>2ml/s</td>
<td>0.10</td>
<td>38mg/ml</td>
</tr>
<tr>
<td>i.v. SVC injection (40 cm³)</td>
<td>30</td>
<td>100</td>
<td>0.137</td>
<td>52</td>
</tr>
<tr>
<td>Conventional aorta injection</td>
<td>30</td>
<td>100</td>
<td>0.23</td>
<td>88</td>
</tr>
<tr>
<td>Conventional selective carotid injection</td>
<td>7</td>
<td>8.5</td>
<td>0.45</td>
<td>170</td>
</tr>
</tbody>
</table>

The iodine content in the vessels is thus appreciably lower for intravenous injections. In a quantitative study of conventional angiography it has been established (Fisher, 1976) that for intracranial vessels at least 60 mgI/ml is needed to obtain clinically useful images, and a good image contrast is reached in the 90-150 mgI/ml range. Above this level further radiographic advantage was not perceived.

The iodine levels of the intravenous injection methods are definitely too low for conventional imaging and the contrast enhancement of the DSA method is needed to provide clinically useful images.
References.

Amiel M., Barbe R and Duc C.: Effect of methylglucamine salts of io­


Barbe R., Kirkorian G. and Amiel M.: Effects of contrast media on circ­


Cronenwett J.L. and Garret H.E.: Arteriographic measurement of the ab­
dominal aorta, iliac and femoral arteries in woman with atherosclero­


Gradshteyn I.S. and Ryzhik I.M.: Tables of integrals, series and pro­


von Kaick G. und Naser V.: Echographische Echtzeitdarstellung der Ar­

Kjellberg S.: Die Mischungs- und Strömungsverhältnisse von wasserlös­


Lantz B., Forderer J., Link D.P. and Holcroft J.W.: Regional distribu­
tion of cardiac output: Normal values in man determined by video dilu­

Lipton M., Abbot J., Kosek J., Hayashi T., Lee F. and Bishop E.: Car­

Ludwig J., Verhoeven L., Kersbergen J. and Overtoom T.: Digital sub­
traction angiography of the pulmonary arteries for the diagnosis of pul­


alysis of factors shaping the time - concentration curves. Circ. Res. 4, 1951, 735.

Modic M.T., Weinstein M.A., Pavlicek W., Gallagher J., Duchesneau P.M., Buonocore H. and Meany T.P.: Intravenous digital subtraction an­

6. RADIOGRAPHIC CONTRAST GENERATION.

6.1. Introduction.

Differences in X-ray attenuation in an object lead to an X-ray contrast in the beam. The contrast in the beam arriving at the detector plane depends on many variables such as: kVp value, kV waveform, tube filtration, off-focal radiation, geometry, patient thickness, field size, antiscatter grid performance, etc. The abundance of parameters and their complicated relationships form a continuously studied subject in diagnostic radiology. The availability of computers more-or-less invited several investigators to search for an all-enfolding computer model (Pillai, 1981; Zamenhof, 1982; den Boer, 1983).

In this chapter only the main parameters which contribute to the X-ray contrast will be discussed without diverting into too much detail.

The question of the X-ray contrast produced is the next logical step after the determination of the dilution factor of the injected contrast medium as described in the preceding chapter. Moreover, contributions to the discussion of whether and how DSA imaging deviates from conventional angiography require an understanding of the X-ray image generation process.

Subject contrast is defined as the ratio of the X-ray intensities transmitted by two selected portions of the subject. The X-ray intensities in this definition could be based on the number of photons arriving at the detector plane. However, photons of different energies have different impacts during the detection process and therefore a better way of defining the subject contrast is in terms of X-ray energy differences (den Boer, 1983).

Subject contrast expressed in this way has limited practical usefulness however, as it presumes a detection process which is independent of the photon energy. When the energy dependence of the detector is also taken into account, the contrast is best defined as the ratio of the light intensities produced by the detector. We designate this by "fluoro contrast" as fluorescence is involved in the detection process of X-ray image intensifiers.
6.2. Contrast with monochromatic radiation.

It has been suggested that monochromatic X-rays might produce better images than those obtained by conventional radiation, which has a continuous spectral distribution. This subject is considered extensively in the famous paper of Oosterkamp (1961), which presented also a method of calculating the contrast performance for various beam quality conditions.

The Oosterkamp method will be used in this chapter to calculate the contrast in the beam. However, an important difference is that, instead of working with a constant incident dose on the object, we will consider the situation with a constant energy level at the detector plane behind the object, which is more in accordance with exposure controlled practical systems.

As a first step in the contrast calculations we start with monochromatic radiation, not because of any expected practical usefulness, but as an easy stepping stone to the more complicated calculations for polychromatic radiation.

The X-ray attenuation of an X-ray pencil beam (no scatter) in a homogeneous object is given by the exponential relationship:

\[ \phi_{\text{out}} = \phi_{\text{in}} \exp(-\mu t) \]  

(6.1)

where \( \phi_{\text{in}} \) is the incident particle fluence on the object.

\( \phi_{\text{out}} \) is the particle fluence leaving the object.

\( \mu \) is the linear attenuation coefficient (cm\(^{-1}\)).

\( t \) is the object thickness (cm).

Objects of inhomogeneous composition require repeated application of Eq (6.1) for the different parts of the object traversed by the pencil beam.

A simple but mathematically very useful object is shown in Fig. 6.1. The fluence \( \phi_0 \) of the source is attenuated to \( \phi_1 \) and \( \phi_2 \) for the different parts of the object.

\[ \phi_1 = \phi_0 \exp(-\mu_1 t_1) \]

\[ \phi_2 = \phi_0 \exp(-\mu_1 t_1 - \mu_2 t_2) \]  

(6.2)

Fig. 6.1. Simple object used in contrast calculations.

Fig. 6.2. X-ray mass-attenuation coefficient for iodine, bone and tissue (from Mika, 1960)
The contrast in the beam is:

\[ C = \frac{\mu_2}{\mu_1} \]

\[ = 1 \cdot \exp(-\mu_2 t_2) \tag{6.3} \]

We choose the model to represent a clinical situation, where \( t_1 \) is the thickness of a patient and \( t_2 \) is the diameter of a blood vessel in the body. The composition of both regions is only slightly different; \( t_1 \) contains only tissue (= equivalent to water) and \( t_2 \) contains blood which is for the largest part equivalent with water too, but additionally contains the diluted iodized salt particles. Due to the dilution, the salt particles fill only a fraction of the blood volume and therefore:

\[ \begin{align*}
\mu_1 &= \mu_W \\
\mu_2 &= \mu_W + \rho_1 \
\end{align*} \tag{6.4} \]

where \( \mu_W \) and \( \mu_i \) represent the linear attenuation of water and iodine.

Substitution in Eq(6.3) gives:

\[ C = 1 \cdot \exp(-\mu_2 t_2) \]

\[ = \mu_2 / (\mu_W t_2) \tag{6.5} \]

where \( \rho_1 \) is the mass density of iodine (g/cm\(^3\)) and \( (\mu/\rho)_1 \) is the mass attenuation coefficient of iodine (cm\(^2\)/g).

The approximation in Eq(6.5) holds for low contrast values such as \( C \) 10%.

Notice that the contrast in the beam is independent of the composition \( M_1 \) and thickness \( t_1 \) of the object in this simplified monochromatic case.

The mass attenuation coefficient \( (\mu/\rho) \) of a material strongly depends on the photon energy as is illustrated in Fig.6.2.

Calculations of the contrast generated by a thin layer of diluted iodine solution have been carried out for different photon energies. It has been assumed that the injected iodine solution, of density \( \rho_j = 380 \, \text{mg/cm}^3 \), is diluted by a factor of 12.6 producing \( \rho_j = 30 \, \text{mg/cm}^3 \). The substituted thickness of the layer is 2 mm, which is of the same order as the diameter for a relatively small blood vessel.

The product \( P_{j2} \), which has been proposed as a figure of system iodine sensitivity (Riederer, 1981), has the value 6 mg/cm\(^2\) in this case.

The calculated contrast values are shown in Fig.6.3-a. A photon energy just above the K-absorption edge at 31.2 keV gives the highest...
contrast for the diagnostically relevant range ($>20$ keV).

The photon fluence $\psi_1 = \psi_2$ at the detector is related to the energy fluence $\psi_1$ at the detector by:

$$\psi_1 = \psi_2 \frac{1}{h \nu}$$

(6.6)

The relative noise level $R$ at the detector is thus given by:

$$R = \left\{ \frac{\psi_1}{\psi_1^2} \right\}^{-1}$$

(6.7)

For a given energy fluence level $\psi_1$, the relative noise $R$ thus increases with photon energy $h \nu$, (more energy per photon accompanies fewer photons); see Fig.6.3-b.

Due to the increasing noise at higher energies, the contrast-to-noise ratio $C/R$ thus even degrades somewhat faster than the contrast degradation alone; see Fig.6.3-c.

6.3. Contrast with polychromatic radiation.

The results of the preceding section indicate that both the iodine contrast and the signal-to-noise ratio sharply decrease with photon energy. The same general conclusions will hold for practical situations, e.g. radiation generated at a tube voltage of 100 kVp will give inferior results compared with 50 kVp radiation. In addition to this general statement, more detailed knowledge is wanted, which requires a closer inspection of the whole radiation spectrum.

If the radiation source has a spectral density of the energy fluence $d\psi/d\nu$, and various subsequent absorbing materials are in the beam, the filtered spectral density at the detector is:

$$\frac{d\psi_1(\nu)}{d\nu} = \frac{d\psi_2(\nu)}{d\nu} \exp\left[ -\sum_{j} \mu_j(\nu) \right]$$

(6.8)

The total fluence at the detector can be found by integration over the energy range:

$$\psi_1 = \int_{0}^{\infty} \frac{d\psi_2(\nu)}{d\nu} \frac{d\nu}{h \nu}$$

(6.9)

With an ideal detector the output is proportional to the input radiation energy, so the contrast ratio is then (similar to Eq(6.3)):

$$C = \frac{\psi_1^2}{\psi_1}$$

(6.10)

The r.m.s. noise depends on the total number of photons involved and similar to the monochromatic case, we find:

$$\phi_1 = \int_{0}^{\infty} \frac{1}{h \nu} \frac{d\psi_1}{d\nu} d\nu$$

(6.11)

$$R = \left\{ \frac{\phi_1}{\phi_1^2} \right\}^{-1}$$

(6.12)

Using Eq(6.9) we can rearrange to give:

$$R = \left\{ \frac{1}{\psi_1} \frac{1}{\psi_1^2} \frac{d\psi_1}{d\nu} \frac{d\psi_1}{d\nu} \right\}^{-1}$$

(6.13)

The radiation spectrum emitted by the anode of an X-ray tube is assumed to obey the theoretical Kramers-Kulenkampf law (Kramers,1923):

$$\frac{d\psi_0(\nu)}{d\nu} = K(\nu_{\text{max}} - \nu)$$

(6.14)

Deviations from this expression occur at the low energy side of the spectrum due to the total inherent filtration $T_w$ of the radiation source (glass envelope, etc.); see Fig.6.4. This inherent filtration is usually equivalent to 2.5 mm of aluminium and this can be introduced as one of the absorbing materials in Eq(6.8).
At the high energy side, deviations from Eq.(6.14) occur due to the characteristic radiation of the tungsten anode, but this effect can be neglected since the energy contribution is very small; 1.4% at 90 kVp and 4.1% at 98 kVp have been measured (Epp, 1966).

The contrast $C$, the relative noise $R$ and $C/R$ are worked out as a function of X-ray tube voltage $kVp$ (which sets $\gamma$), with object thickness as a parameter. The iodine concentration $\rho_I$ is the same as was used in the preceding section. The results of the calculations are shown in Fig. 6.5. A detection efficiency of 100% is assumed here until further notice.

In comparison with Fig. 6.3, it appears that $C$ and $C/R$ do not attain the value generated by the optimal monochromatic radiation of 34 keV. The dependence of $C$, $R$ and $C/R$ on tube voltage $kVp$ is less than the dependence on photon energy in the monochromatic case. $R$ hardly depends on object thickness, but $C$ and $C/R$ have steeper slopes for greater object thickness. The reason for this is that a greater thickness produces more beam hardening and hence a higher-energy part of the spectrum is effectively used (higher $keV_{eq}$), which leads to reduced contrast generation. At the low end of the tube voltage range, an optimal $kVp$ value appears to exist in the neighbourhood of 45 kVp. The reason for the decline below a tube voltage of 45 kVp is that the radiation then has its maximum spectral intensity at a photon energy below the $K$-edge of iodine, so an increasing part of the total radiation energy hardly contributes to contrast generation.

In practice the optimal $kVp$ value can seldom be used, as the transmittance of the object here is so low that, in order to reach a usable signal level behind the object, the incident exposure dose must be so high as to lead to an unacceptably high skin exposure for the patient, and also to an inadmissible $kW$-loading of the X-ray tube. For thin objects such as the lower arm and hand, lower leg and foot, this transmittance problem is not so severe and the lower $kVp$ values can be used beneficially, with the object of visualizing smaller vessels than in the other parts of the body.

6.4. Influence of X-ray detection efficiency.

In the preceding section an ideal X-ray detection mechanism was assumed, with which all of the X-ray energy contributes to fluorographic image formation. In practice only a part of the X-ray energy leaving the object is detected, because:
1. The X-ray-sensitive screen does not absorb all the incident photons.
2. The material of the image intensifier (II) input window as well as the usually used antiscatter grid in front of the II cause additional X-ray attenuation.

The X-ray detection capability of the screen QDE (= quantum detection efficiency) depends upon the thickness and the composition of the detection screen. Practical QDE data for contemporary II screens with a 0.2 mm thickness of CsI-Na are shown in Fig.6.5-a. Fortunately the efficiency is optimal in that part of the X-ray spectrum where the maximum iodine contrast signal is generated.

As X-ray detection via an image intensifier is a complicated process involving several energy-conversion steps in cascade, the noise turns out to be somewhat higher than predicted by the number of absorbed photons alone. The parameter DQE (= detective quantum efficiency) has been defined to quantify this. For reference, see e.g. Dick(1981) and Rowlands(1983). Detailed analysis of the DQE concept is beyond the scope of this thesis. Instead the DQE can, roughly speaking, be considered as a noise-equivalent QDE which is always somewhat lower (0.75 - 0.95) than the actual QDE. This difference factor is comparable to the inverse square root of the "noise factor", used in low-noise electronic amplifier concepts.

Transmission loss through the II input window is about equivalent to that of 2-mm Al; see Fig.6.6-b.

Loss of the primary radiation by transmission through an antiscatter grid is composed of an overall loss of about 15% due to the lead content, and an energy-dependent loss, equivalent to 2-mm Al, caused by the interspace material; see Fig.6.6-b.

The overall effect of the transmission losses T[(I) of grid and input window, the QDE of the II-screen, and the noise factor is shown as the DQE of this detector combination in Fig.6.6-c.

Now that we know how detector performance deviates from the ideal situation, the contrast C, the relative noise R, and C/R can be recalculated using the following formulas:

$$C = \frac{\int T(\nu)QDE(\nu) d\nu}{\int T(\nu)QDE(\nu) d\nu}$$

$$R = \frac{\int N(\nu)QDE(\nu) d\nu}{\int T(\nu)QDE(\nu) d\nu}$$

$$C/R = \frac{\int T(\nu)QDE(\nu) d\nu}{\int N(\nu)QDE(\nu) d\nu}$$

(6.15)
Fig. 6.6-a
QDE of CsI Na screen

Fig. 6.6-b
Transmission $T$ of grid (33/8) and input window of image intensifier

Fig. 6.6-c
DQE of grid-III detector combination

The results of the calculation are shown in Fig. 6.7. Comparison of Fig. 6.7-a and Fig. 6.5-a reveals that in general the contrasts are slightly higher with the II as a detector, which can be explained by the fact that the spectral sensitivity of the II benefits the lower energy range where the maximum iodine signal is generated. As it is now clear that, due to the properties of the detector, the detected contrast is not necessarily equal to the incident radiation contrast, a clear distinction between these two has to be made. For this reason the detected contrast has been named "fluorocontrast".

Comparison of Fig. 6.7-b and Fig. 6.5-b shows that, due to the non-ideal detection efficiency, a higher noise level is present. As roughly speaking, $\text{DQE} \approx 0.30$ in the range of interest (see Fig. 6.6-c), a factor of about $\sqrt{\frac{1}{0.3}} = 1.82$ more noise can be expected which is in agreement with the more-detailed results of the calculations. The non-ideal detection characteristic has serious consequences for the contrast-detail performance as discussed in chapter 4; a dose higher by a factor of 3 is needed to attain a given performance level. The contrast-to-noise ratio $C/R$ shows similar slopes in Fig. 6.7-c and Fig. 6.5-c, so in spite of the spectral dependence of the practical detector situation, an equal kVp dependence occurs in the ideal and non-ideal cases. In both cases the lower kVp values are optimal.

The contrast curves shown in Fig. 6.7-a are the result of carefully summing the contributions of each part of the generated X-ray spectrum, where the spectral sensitivity of the image intensifier is also taken into account. Instead of working out Eq (6.15) for each new situation, it is much easier in practice to use a simple formula like Eq (6.5). This is possible by defining an "effective mass attenuation coefficient" which depends on the tube voltage kVp. This $\omega/p_{\text{eff}}$ for iodine can be derived from Fig. 6.7-a by using Eq (6.5). The results are shown in Fig. 6.8.
Fig. 6.7-a  
Fluoro contrast for 6 mg/cm³ of iodine

Fig. 6.7-b  
Relative quantum noise

Fig. 6.7-c  
Contrast-to-noise ratio

Fig. 6.8  
Effective mass-attenuation coefficient for iodine
6.5. Influence of exposure time and ECG triggering.

The X-ray exposure dose for an image is determined by the tube voltage kVp, tube current mA and exposure time. The product of the latter two is commonly referred to as the "mAs" value of the exposure. In the previous section it was established that low kVp exposures are favoured for high iodine contrast generation. For various reasons such as: 1) Skin-dose protection, 2) X-ray generator limitations and 3) X-ray tube loadability, one is often forced to deviate from the optimal kVp towards higher values. The results of such a deviation are drastic as the output of the X-ray tube increases and the X-ray attenuation in the object decreases with kVp.

If the higher kVp value is exchanged for lower mAs, but with the X-ray energy fluence at the input of the II constant, a considerable decrease of C and C/R (up to a factor of 2.5) may occur in the diagnostically relevant range of 50-90 kVp; see Fig.6.7.

For this reason the exposure control system should take this effect into consideration and aim at the lowest kVp and the highest mAs per exposure within the given boundary conditions. A straightforward way of getting high mAs per exposure is to use a long exposure time. Long exposures however, may cause motion blurring of moving vessels. Especially the vessels in the thorax, such as the heart and aortic arch, move so rapidly (up to 20 cm/s) that exposures shorter than 10 ms are usually applied in this area. As the movement of the vessels is correlated with the pulsations of the heart, a definite periodicity in the movement is detectable and even a relatively long period of rest occurs during end-diastole. Beneficial use can be made of this end-diastole rest period by triggering the DSA exposure with ECG pulses (Ludwig, Verhoeven, 1983). With triggering at end-diastole, an exposure time of 200 msec proved to be feasible without disturbance by motion blurring. The prolonged exposure time (factor 20x) is an appreciable help in precluding the need for high kVp settings.

6.6. Influence of scatter and the antiscatter grid.

The discussions of the preceding paragraphs have dealt only with the primary radiation which carries the X-ray image. In most X-ray examinations, scattered radiation plays an important role as it frequently makes up 50 to 90% of the total number of photons emerging from the patient. This scattered radiation contributes no useful information; it is just an additional uniform exposure which spoils the contrast of the primary radiation image.

The scatter contribution in an image depends very much on the irradiated field of view on the object, and on the object thickness. The influence of tube voltage on scatter production has been rather vague for a long time, but a recent publication (Burgess, 1981) revealed theoretically and by measurements, that the generated scatter fraction $I_s/(I_p + I_s)$ is nearly independent of photon energy and tube voltage. Concise mathematical representation of scatter behaviour is beyond the scope of this thesis, but the condensed results of $(I_p + I_s)/I_p$ measurements in various practical situations are presented in Fig.6.9. The "without grid" measurements proved to be nearly independent of the tube voltage and are presented as a single curve. For a large field diameter (14" = 35 cm) of a thick object (H = 35 cm), which could occur in an abdomen examination, the scatter generated could be 10 times higher than the primary radiation. For very small fields of view all the curves converge to the origin as an infinitesimally small pencil beam has negligible scatter radiation. The beneficial effect of an antiscatter grid is also shown in Fig.6.8. With the aid of Fig.6.9 it is possible to estimate roughly the primary fraction $I_p/(I_p + I_s)$ for some practical examination conditions; see Table 6.1.

Table 6.1: Estimates of scatter conditions expressed as primary fraction for various examination conditions with grid.

<table>
<thead>
<tr>
<th>Examination type</th>
<th>Examination field size</th>
<th>Object Thickness</th>
<th>Tube voltage</th>
<th>$I_p/(I_p + I_s)$</th>
</tr>
</thead>
<tbody>
<tr>
<td>carotid</td>
<td>6&quot;</td>
<td>10 cm</td>
<td>60 kVp</td>
<td>0.92</td>
</tr>
<tr>
<td>cerebral</td>
<td>6&quot;</td>
<td>17</td>
<td>70</td>
<td>0.81</td>
</tr>
<tr>
<td>cerebral</td>
<td>9&quot;</td>
<td>17</td>
<td>70</td>
<td>0.75</td>
</tr>
<tr>
<td>abdomen</td>
<td>9&quot;</td>
<td>22</td>
<td>80</td>
<td>0.65</td>
</tr>
<tr>
<td>abdomen</td>
<td>14&quot;</td>
<td>22</td>
<td>80</td>
<td>0.58</td>
</tr>
</tbody>
</table>

The presence of scattered radiation not only reduces the contrast of the radiation image, but also significantly influences the quantum noise level as more photons arrive at the detector plane. In order to determine the influence of scatter on the contrast-to-noise ratio, the phantom of Fig.6.1 is considered for four different cases.
A. Imaging without scatter.

\[ \psi_1 = \psi_p \]
\[ \psi_2 = \psi_p \exp(\mu_2 t_2) \]

\[ C = \frac{\psi_2}{\psi_1} = \mu_2 t_2 \]
\[ R = \left( \frac{hr}{\psi_p} \right) \]
\[ C/R = \mu_2 t_2 \left( \frac{\psi_2}{hr} \psi_p - \psi_1 \right) \]

B. Imaging with scatter; equal exposure dose on object as in A.

\[ \psi_1 = \psi_p + \psi_s \]
\[ \psi_2 = \psi_p \exp(\mu_2 t_2) + \psi_s \]

\[ C = \mu_2 t_2 \frac{\psi_p}{\psi_p + \psi_s} \]
\[ R = \left( \frac{hr}{\psi_p + \psi_s} \right) \]
\[ C/R = \mu_2 t_2 \left( \frac{\psi_2}{hr} \psi_p - \psi_1 \right) \]

In this case the exposure dose supplied to the detector is greater by a factor \((Ip + Is)/Ip\). This is not a practical situation however, as in a practical system the dose level at the detector plane will be stabilized by adjusting the mAs value. The presence of the scatter on the detector means that less radiation from the source will suffice. Therefore the following case C represents what really happens if there is scatter.
C. Imaging with scatter; equal radiation dose on detector as in A.

\[ \phi_1 = \frac{\phi_p}{\phi_p + \phi_s} (\phi_p + \phi_s) \]

\[ \phi_2 = \frac{\phi_p}{\phi_p + \phi_s} \left\{ \phi_p \exp(\mu t_2) + \phi_s \right\} \]

\[ C = \mu t_2 \frac{\phi_p}{\phi_p + \phi_s} \]

\[ R = \left( \frac{\mu t_1}{\phi_p} \right)^{1/2} \]

\[ C/R = \mu t_2 \left( \frac{\phi_p}{\phi_p + \phi_s} \right)^{1/2} \]

In this case the relative noise \( R \) is equal to that of case A. Both \( C \) and \( C/R \) are reduced by the factor \( Ip/(Ip + Is) \).

If we want to restore \( C/R \) to the same level as in case A, then an increased dose level is required.

D. Imaging with scatter; increased dose level.

\[ \phi_1 = \frac{\phi_p + \phi_s}{\phi_p} (\phi_p + \phi_s) \]

\[ \phi_2 = \frac{\phi_p + \phi_s}{\phi_p + \phi_s} \left\{ \phi_p \exp(\mu t_2) + \phi_s \right\} \]

\[ C = \mu t_2 \frac{\phi_p + \phi_s}{\phi_p + \phi_s} \]

\[ R = \left( \frac{\mu t_1}{\phi_p} \right)^{1/2} \]

\[ C/R = \mu t_2 \left( \frac{\phi_p + \phi_s}{\phi_p + \phi_s} \right)^{1/2} \]

From this last case D, which gives contrast-to-noise ratio \( C/R \) equal to that of the theoretical scatterless case, it can be concluded that the increase in exposure level at the detector is equal to \( (Ip + Is)/(Ip)^2 \) compared with the scatterless case, but the required increase in dose incident on the object is only \( (Ip + Is)/Ip \).

### 6.7. Dynamic range of the fluoro image and log processing accuracy

The phantom considered so far has been of an exaggerated simplicity. In practice the human body has a much more complicated structure than the homogeneous object of Fig. 6.1. One of the troublesome deviations is the presence of bone in certain parts of the object, which causes a locally much more attenuated X-ray profile.

A more comprehensive phantom that, for instance, could simulate an a.p. abdominal examination is shown in Fig. 6.10. The main part of the phantom \((t_1)\) consists of tissue-equivalent material \((= \text{water})\). Two small blood vessels with diameter \(t_2\) and also a small compartment \(t_3\) with bone equivalent material is present.

In order to approximate a real situation, the phantom dimensions will be chosen to simulate as closely as possible the abdominal examination; e.g. \(t_1 = 20 \text{ cm}, t_2 = 0.2 \text{ cm}\) and \(t_3 = 1.7 \text{ cm}\). The dimensions of the spine depend on the location, being largest in the abdominal region. The largest part of a vertebra is composed of spongy bone and it is estimated that this part is about equivalent to 1.7-cm thickness of compact bone with \(\rho = 1.70 \text{ g/cm}^3\) \((\rho = 2.89 \text{ g/cm}^3)\).

Results of fluorocontrast calculations using Eq (6.15) are presented in Table 6.2. Several kvp cases are considered and in all cases the maximum signal level is normalized at 100 because for a given kvp, the optimum exposure conditions are set by adjusting the mAs of the X-ray tube.

<table>
<thead>
<tr>
<th>kvp</th>
<th>50</th>
<th>60</th>
<th>80</th>
<th>100</th>
</tr>
</thead>
<tbody>
<tr>
<td>20 cm water</td>
<td>100</td>
<td>100</td>
<td>100</td>
<td>100</td>
</tr>
<tr>
<td>19.8 cm water + 0.2 cm vessel</td>
<td>88.0</td>
<td>90.5</td>
<td>93.2</td>
<td>94.6</td>
</tr>
<tr>
<td>absolute difference</td>
<td>12.0</td>
<td>9.5</td>
<td>6.8</td>
<td>5.4</td>
</tr>
<tr>
<td>18.3 cm water + 1.7 cm bone</td>
<td>38.4</td>
<td>46.7</td>
<td>57.1</td>
<td>62.5</td>
</tr>
<tr>
<td>18.1 cm water + 1.7 cm bone + 0.2 cm vessel</td>
<td>34.0</td>
<td>42.5</td>
<td>53.6</td>
<td>59.6</td>
</tr>
<tr>
<td>absolute difference</td>
<td>4.4</td>
<td>4.2</td>
<td>3.5</td>
<td>2.9</td>
</tr>
</tbody>
</table>

**Dynamic range**

2.9:1 2.3:1 1.9:1 1.7:1
This table shows that the dynamic range of the primacy signal, defined as the ratio of maximum signal to minimum signal, decreases for higher tube voltages. The absolute signal differences produced by the contrast-filled vessel are considerably lower in the bone-shadow area, and the difference factor with the no-bone situation increases for lower tube voltages. As explained in section 3.4, the logarithmic amplifier in the electronic part of the DSA system should equalize this difference by applying a larger small-signal differential-gain factor for the lower signal levels. An ideal logarithmic amplifier will raise the contrast of the bone-shadowed vessel to the same level as that of the unshadowed vessel. Table 6.3 shows the results of applying ideal logarithmic processing to the data of Table 6.2.

Table 6.3: Results of logarithmic processing applied to Table 6.2.

<table>
<thead>
<tr>
<th>kvp</th>
<th>50</th>
<th>60</th>
<th>80</th>
<th>100</th>
</tr>
</thead>
<tbody>
<tr>
<td>log difference in tissue only = ( \Delta_1 )</td>
<td>0.0555</td>
<td>0.0433</td>
<td>0.0306</td>
<td>0.0241</td>
</tr>
<tr>
<td>log difference behind bone = ( \Delta_2 )</td>
<td>0.0529</td>
<td>0.0409</td>
<td>0.0275</td>
<td>0.0206</td>
</tr>
<tr>
<td>( \Delta_2/\Delta_1 )</td>
<td>0.95</td>
<td>0.94</td>
<td>0.90</td>
<td>0.86</td>
</tr>
</tbody>
</table>

It turns out that even with this ideal logarithmic processing, the vessel contrast \( \Delta_2 \) in the bone-shadow area is slightly smaller than \( \Delta_1 \) in the tissue-only area \( \Delta_2/\Delta_1 = 0.86 - 0.95 \). This effect, which is caused by beam hardening, is not of main concern for imaging quality as it is only a second-order effect.

The inevitable presence of scattered radiation has a much more detrimental effect on the accuracy of the logarithmic processing. As an example of the influence of scatter, the abdomen case of Table 6.2 is reconsidered by adding scatter level values of respectively 35, 45, 65 and 85 to the primary level values in the 50, 60, 80 and 100 kvp column, and subsequently renormalizing to a peak value of 100. The chosen scatter values correspond to primary fractions of 1/1.35, 1/1.45, 1/1.65 and 1/1.85 (scatter fractions 26%, 31%, 39% and 46%) which can be read from Fig.6.9 for an abdominal examination with a patient thickness of 20 cm and a 14" field of view. The resulting signal levels are given in Table 6.4.
Table 6.4: Fluoro profile including scatter for abdomen phantom with diluted contrast material (30 mg/ml iodine) in the vessels.

<table>
<thead>
<tr>
<th>kVp</th>
<th>50</th>
<th>50</th>
<th>80</th>
<th>100</th>
</tr>
</thead>
<tbody>
<tr>
<td>20 cm water</td>
<td>100</td>
<td>100</td>
<td>100</td>
<td>100</td>
</tr>
<tr>
<td>19.8 cm water + 0.2 cm vessel</td>
<td>91.1</td>
<td>93.4</td>
<td>95.9</td>
<td>97.1</td>
</tr>
<tr>
<td>Absolute difference</td>
<td>8.9</td>
<td>6.6</td>
<td>4.1</td>
<td>2.9</td>
</tr>
<tr>
<td>18.3 cm water + 1.7 cm bone</td>
<td>54.4</td>
<td>63.2</td>
<td>74.0</td>
<td>79.7</td>
</tr>
<tr>
<td>18.1 cm water + 1.7 cm bone + 0.2 cm vessel</td>
<td>51.1</td>
<td>60.3</td>
<td>71.9</td>
<td>78.2</td>
</tr>
<tr>
<td>Absolute difference</td>
<td>3.3</td>
<td>2.9</td>
<td>2.1</td>
<td>1.5</td>
</tr>
<tr>
<td>Dynamic range</td>
<td>2:0:1</td>
<td>1:7:1</td>
<td>1:4:1</td>
<td>1:3:1</td>
</tr>
<tr>
<td>Log difference in tissue only = $\Delta_1$</td>
<td>0.0404</td>
<td>0.0294</td>
<td>0.0183</td>
<td>0.0129</td>
</tr>
<tr>
<td>Log difference behind bone = $\Delta_2$</td>
<td>0.0268</td>
<td>0.0204</td>
<td>0.0126</td>
<td>0.0068</td>
</tr>
<tr>
<td>$\Delta_2/\Delta_1$</td>
<td>0.66</td>
<td>0.69</td>
<td>0.69</td>
<td>0.67</td>
</tr>
</tbody>
</table>

From this table it can be concluded that the dynamic range is much lower in the presence of scatter.

Comparison of Table 6.4 and Table 6.3 shows that the difference signals are substantially reduced by the scatter. The reduction factor for high-level signals is equal to the previously-mentioned primary fractions 1/1.35, 1/1.45, 1/1.65 and 1/1.85 as expected. For the signals at a lower level (e.g. behind bone), an additional reduction occurs. Due to the uniform bias of the scatter, the behind-bone vessel signal arrives at the wrong level in the logarithmic curve, which results in a differential gain factor which is too low.

Problems similar to those just discussed for the abdominal case will be encountered in other areas of the body. Imaging of intracranial vessels will be suboptimal behind the petrous bone. The same holds for the region behind the pericard and sternum when thoracic examinations are carried out.

In addition to local regions of higher opacity due to bone structures, some local regions of lower opacity may also occur in an object due to e.g. air pockets or varying object thickness.

The concurrent presence of both low and high transmittance extremes may cause the primary-beam dynamic range to reach much higher levels than the figure of 3:1 put forward in Table 6.2. In fact, practical DSA experience has shown that abdominal examinations pose the least dynamic range problems.

The most difficult region of the body as regards the dynamic range is in the chest area. Equivalent tissue thicknesses of different areas of the chest vary from 10 cm in the lungs to about 25 cm in the mediastinum (Niklason, 1981). Primary beam attenuation differences due to this varying tissue thickness alone account for a primary beam dynamic range of 25:1 to 50:1 depending on the kVp chosen. In a practical situation (with grid), the mediastinum signal level is determined mainly by the scattered radiation from the lung fields into this area (cross scatter; Niklason, 1981). Due to this scatter the dynamic range within the object is, even with an antiscatter grid, very unlikely to exceed 6:1.

In principle compensation of the scatter bias can be devised by adding a certain amount of negative offset to the electrical video signal before inputting this signal to the log amplifier. The problem, however, is that the amount of correction required is unpredictable in practice, as it depends on so many variables (e.g. object dimensions, field of view, kVp, etc). No sensible compensation voltage can be chosen for the real-time display of the images during the examination (during "live mode"), so preferably no compensation at all should be used here, as too much compensation will lead to destructive clipping of the lower video range.

During the later evaluation and diagnosis of the stored images ("view mode") one of the postprocessing functions could offer a user interactive method for experimentally determining an optimal scatter bias compensation offset.

6.8 Glare, bright spots and bolusising.

In the previous section it has been explained that, due to the presence of scattered radiation, the dynamic range of the X-ray profile of an anatomical object, even in a most-demanding situation such as chest examinations, will rarely exceed a 6:1 ratio. Non-uniform irradiation from the X-ray source, caused by the heel effect, may have an influence on this dynamic range figure, but for the difficult chest region it probably tends towards a smaller dynamic range.
One is inclined to presume therefore that the detection of such a limited X-ray profile is not particularly difficult for an II, as lead-pil contrast performance of present-day IIs is in the (15 - 25):1 range. Lead-pil contrast ratios (as defined by NEMA) are measured with a lead disc covering 10% of the input area. Other disc diameters will produce different contrast ratios and therefore the lead-pil figure is of limited practical usefulness.

A more comprehensive method of describing the contrast loss caused by the II is to define a low frequency drop figure, LFD, in the modulation transfer function MTF (Kühl, 1979; Beekmans, 1982). Fourier transformation of the MTF gives the point spread function PSF, which can be used to explain the spatial domain performance aspects. Usually the PSF is made up of a small Gaussian-shaped "nucleus" spot surrounded by a much-broader glare spot which may contribute a non-neglectable portion of the total light flux of the PSF, see Fig. 6.11. A detailed discussion of CRT spots including glare is given by Schade (1973). With the glare spot concept the performance limitations in a number of difficult imaging situations can readily be predicted.

If the II is uniformly lit by X-rays, except for a small region behind a lead disc, the "feet" of all the PSFs of the irradiated area add up to an unwanted light contribution at the output of the II in the pil area. When the lead disc covers 10% of the input area (in accordance with NEMA pil ratios 15-25) a 4-7% glare level may be found in practice (note 1/25 = 4%; 1/15 = 6.7%).

As a rough approximation it may be assumed that the glare level in the covered area is proportional to the area that is uncovered and irradiated. Extrapolation by this rule to an irradiated area of 10% instead of the 90% considered above, would thus give a 9x lower glare level (0.44-0.74%) in the covered area.

The glare of an II produces a similar effect on the image as the X-ray scatter discussed earlier. In most imaging situations with not too much differential attenuation in the object (small dynamic range), the glare levels are uncritical compared with the ever-present scatter. In some difficult imaging situations, however, it may happen that some unattenuated X-rays can pass alongside the object and can directly hit the II. This problem is particularly relevant during examinations of peripherals and of the neck area. The resulting so-called "bright spots" can, depending on the object thickness, exceed the object radiation level 10-50 fold, which results in at least so high a dynamic range of the primary beam.

Depending on the intensity and the relative area of the bright spot, a significant glare level will be added to the interesting object area of the image.

Conventional screen-film systems will hardly suffer from bright spots in the image, as such a detector system has virtually no glare and the high intensity radiation just causes the film image to saturate in high density without detrimental effects on other parts of the image.

When imaging with IIs, glare is inevitable, and the difficult bright spot situation has to be prevented as much as possible by judicious use of X-ray shutters, or by applying "bolussing" material such as copper plates or mouldable sandbags in the beam path of the problem area.

A second problem, related to the presence of bright spots, is the behaviour of the X-ray exposure-control system. In order to prevent the TV camera and video AD converter from saturating, the X-ray exposure factors kVp and mAs are usually adjusted (manually or automatically) by the exposure-control system in such a way that the detected peak level just fits into the signal range of the TV system. If this peak level happens to be a bright spot then the other more-interesting parts of the image will correspond to so low a radiation level at the detector that they will be flooded with quantum and electronic noise.

To circumvent this problem, a user-adjustable "measuring field" has been incorporated in the DSA equipment. With this feature the user can adjust the area in the image where the signal amplitude will be measured. Difficult bright spots can thus purposely be excluded from the measuring field. The signal amplitude within the measuring field will then have the appropriate level. Image areas outside the measuring field will have a definite risk of signal saturation, but the saturation can sometimes be accepted if it occurs in an uninteresting part of the image.

Instead of using the adjustable measuring field, the bolussing method mentioned earlier can also be used to solve the exposure control problem. The ultimate responsibility for how to use the system is that of the examining physician.

6.9. Conclusions.

In this chapter the image contrast, C, generated by small iodine concentrations in the beam is analysed.

Due to the particular iodine mass attenuation characteristic of iodine with its K-edge at 33 keV, and due to the beam filtration which occurs in a practical situation, there appears to be an optimal kVp condition at about 45 kVp. For practical reasons the voltage used is...
usually above 50–60 kVp.

For an object thickness of 15 cm H2O, the relative level of generated contrast is approximately inversely proportional to the relative kVp variation (e.g., if kVp changes from 60 to 75 kVp then the contrast decreases by a factor 60/75 = 0.8). For greater object thicknesses the contrast variation changes more steeply with the kVp variations.

The presence of scatter is detrimental to the image quality as it reduces the image contrast. For an equal dose at the detector as in the scatterless case the quantum noise level is equal, but both the contrast C and the signal-to-noise ratio S/N are decreased by the primary fraction Ip/(Ip+Is). The application of an antiscatter grid helps to reduce the influence of the scatter, but nevertheless an appreciable degradation still remains. In a practical situation with a grid, primary ratios in the range of 0.55–0.90 will be encountered. As the scatter-reduction capability of grids is better for low kVp radiation (see Fig.6.8), this is an additional reason for aiming at low kVp techniques.

The presence of scatter also influences the log processing accuracy; overprojection of vessels and bone results in reduced vessel contrast in the bone region. Electronic postprocessing, such as shifting the electronic signal in the logamp range, may result in a proper enhancement factor in the bone region. Together with the increased differential gain for vessels in the bone-shadow region, the noise of the scattered radiation will also be more enhanced by the postprocessing method proposed. For this reason the postprocessing scatter compensation method is not as good as the well-known scatter prevention measures, which can be taken before the exposures are made.

For low iodine concentrations, as in intravenous DSA, the contrast of a vessel is proportional to its diameter. For high concentrations (as in conventional arterial angiography) the vessels become nearly opaque to X-rays, and a contrast saturation effect occurs. For this reason the difference between small and large vessels will be more apparent in DSA than in conventional angiography.

Exposure control systems, especially those which are based on video peak measurements as in DSA, are particularly sensitive to "bright spots" in the image. Proper bolussing and/or an adjustable measuring field are needed to get the appropriate dose level.

Low-frequency drop (LFD) or glare of the image intensifier produces effects similar to X-ray scatter, although in most circumstances the glare can be neglected in comparison to the scatter. Only when a very bright spot occurs, may glare become troublesome to an extent which is proportional to its area.

The limited X-ray detection efficiency of the grid + II combination (DQE=0.3) produces about \( \sqrt{3.3} = 1.82x \) more noise than ideal detection. This has important consequences as regards the low-contrast detectability as discussed in chapter 4. DQEs of rare-earth screen-film systems + grid are of the same order, so in this respect there are no serious differences from angio "gold standard".

A recent publication (Zamenhof, 1982) reported a kVp optimization analysis for small amounts of iodine, where the patient entrance dose was kept constant. For a given patient thickness of 20 cm, a broad optimum in the range 60–70 kVp was calculated; for 10-cm thickness the optimum was 50–60 kVp. In this chapter a different approach (constant energy at the detector) has led to an optimum at about 45 kVp. In practice, the limitations of the X-ray tube and generator lead to a kVp range which just about covers the range advised by Zamenhof.

H. den Boer (1983) proposed that instead of selecting an X-ray exposure dose for a particular imaging task, an exposure dose should be selected which fully loads the X-ray tube. For a thin object a higher exposure dose can then be applied with a correspondingly better image quality (less X-ray noise). With this philosophy the optimum tube voltage depends on the object thickness, and he found that for 10-20-30 cm H2O the optimal kVp is 55–77–87 kVp.
7. NOISE CONTRIBUTION OF THE TV CAMERA.

7.1. Introduction.

Television systems have been integrated in modern diagnostic X-ray systems for many years, but their use has been limited to fluoroscopy. As the intention of fluoroscopy is to produce X-ray images using a minimum possible exposure rate, the images are usually rather noisy. Due to the high amount of X-ray quantum noise no special requirements are needed for the signal-to-noise ratio of the TV in this application.

For radiographic applications such as DSA, an exposure level much higher than during fluoroscopy will be used in order to acquire images with less X-ray noise. As the inherent noise of the TV system must not be allowed to dominate, a higher dose level and a better signal-to-noise ratio of the TV camera are required, compared with the conventional fluoroscopic application.

The mode of operation of the TV camera is unconventional as in radiographic applications no continuous light flux is supplied to the camera tube, but isolated flashes appear instead. Each exposure (limited to a speed of a few images per second) gives a flash with dark periods in between. Instead of the conventional interlaced scanning, a non-interlaced (sequential) scanning of the TV camera tube is needed, as will be explained further on in this chapter. Analysis of the signal-to-noise ratio of the TV camera will show that the scanning method has an important influence upon this figure.

The noise of the camera preamplifier will be analyzed in detail as intuitively one expects this noise to be the limiting factor for the low contrast detection capability of the system.

7.2. Maximum camera output current.

It is self-evident that with a certain fixed preamplifier noise, the S/N ratio of the camera improves when more signal current is delivered. Practical preamplifier noise levels are of the order of 1 nA r.m.s. for a standard bandwidth of 5 MHz. Usual signal currents in standard TV applications are 100-200 nA, so the S/N ratio is 100-200 or 40-46 dB. This S/N level is excellent for normal direct viewing, so no higher signal levels are usually pursued because they would require a much higher (and costly) illumination level in broadcast studios and/or a much faster lens system (with reduced depth of focus).
Because of the abundant amplification of the X-ray image intensifier, a much higher signal current may be used in radiographic applications. The maximum signal handling capability of a camera tube is limited by two factors: (1) beam current and (2) target voltage excursion.

7.2.1. Beam current limitation.

The maximum beam current available to discharge the target potential of the camera tube is determined by the electron-gun capability. In order to obtain a small spot diameter, a small aperture with a diameter of about 50 μm is present in the gun structure. About 99% of the electrons produced by the cathode are stopped here (Hasker, 1977) and only 1% passes through this aperture. After deflection and focusing of the beam, just before it reaches the target, a part of the beam (about 50%) is again obstructed by the mesh, so that finally only a very small part of the cathode current is able to land on the target. Cathode and gun are designed in such a way that normal broadcast operation allows at least 600 nA of beam current to land on the target. For a signal current of 200 nA, a beam current reserve of a factor 3 is then available to discharge inadvertent peak highlights.

Most standard plumbicons with triode guns will be able to supply still higher beam currents with another Vg setting. The newest diode gun plumbicons (Franken, 1980) are specified as able to deliver up to 2500 nA of beam current. Using a higher beam current will reduce the life of the tube and slightly decrease the MTF (Franken, 1978). For DSA application the S/N performance is so highly important that the slightly-decreased MTF that accompanies the higher beam currents is purposely accepted. The life of the tube is preserved by switching on the higher beam current only when it is needed. This temporary "beam boost" during the 10-20 sec period which is typical for a DSA run, is automatically controlled by the microcomputer system in the apparatus.

7.2.2. Limitation of target voltage excursions.

A very important parameter of a TV camera tube is the storage capacitance of the photosensitive layer. The layer capacitance resulted from a trade-off between two undesirable effects (de Haan, 1963). If the layer capacitance is too high, more than one read scan is needed to discharge the target, which results in an unwanted "discharge lag". If on the other hand the layer capacitance is too low, the voltage excursions on the target may become too large, which results in "beam pulling" effects. Beam pulling causes the path of the electron beam to be bent by the potential pattern on the target, so that a strong white area of the image will appear larger than it actually is. In order to keep this effect negligible, both 5 V (Theile, 1957) and 10 V (de Haan, 1963) have been indicated as the maximum target voltage excursion that can be tolerated. As the relative influence of a small amount of beam pulling depends on the dimensions of the image on the target one may expect that a tube with a larger target area may tolerate more absolute beam pulling, and thus a larger target voltage excursion. Moreover the beam pulling in DSA application does not necessarily need to be negligible because the subtraction process will remove its visible effects on the displayed difference image.

A second limiting factor for the voltage excursion on the target is due to the fact that the excursion should be small with respect to the target supply voltage. If this is not so, the actual drift-field for the generated charge carriers in the photosensitive layer, made up of the difference between the target supply voltage and the target voltage excursion, will become so low that part of the charge carriers recombine and do not contribute to the charge accumulation. In this way a deviation from the tube's linear characteristic arises, which eventually leads to complete saturation of the transfer characteristic. As the advised target supply voltage for plumbicons is 45 V, a target voltage excursion limit \( \Delta V = 20 \text{ V} \) seems tolerable in this respect.

The target voltage excursion \( \Delta V \) depends on the amount of light energy supplied. Absorbed light photons create charge carriers which are accumulated. The charge \( Q \) and the layer capacitance \( C \) are related to \( \Delta V \) by:

\[
Q = C \Delta V
\]

The layer capacitance \( C \) depends on the physical properties of the layer and the scanned area. Standard 1" Plumbicons have a layer capacitance of about 500 pF/cm\(^2\) (Schut, 1972), so the maximum charge density that is allowed to be accumulated is therefore:
This figure is a property of the plumbicon layer and, depending on the target size used, a corresponding maximum signal current can be calculated.

The area of one pixel $a_{\text{pix}}$ and the scan time of one pixel $t_{\text{pix}}$ (they will be discussed in more detail in the next section) are needed to determine the signal current $I_{\text{g}}$. As a preliminary example:

If $a_{\text{pix}} = 17 \, \mu\text{m} \times 17 \, \mu\text{m}$ and $t_{\text{pix}} = 58 \, \text{ns}$,

$$Q_{\text{max}} = \frac{e_{\text{g}} \max}{t_{\text{pix}}} = \frac{(17 \times 10^{-6})^2 \times 10^{-14}}{68 \times 10^{-9}} = 410 \, \text{nA}$$

(7.3)

7.2.3. Influence of scanning parameters.

In relation to the maximum available signal current from the camera tube, two parameters are important: (1) scanning area and (2) scanning rate.

Scanning area depends on the target dimensions of the camera tube used but also on the relative utilization of the target area by the scanning format. In this respect the scanning format used in X-ray TV systems deviates substantially from that of the conventional broadcast scanning. The influence is illustrated in Fig. 7.1.

According to the manufacturer's specifications the quality area of a 1" camera tube has a diameter of 16 mm. Broadcast images with a 4:3 aspect ratio which just fit into this area must have scan dimensions of $H \times V = 12.8 \times 9.6$ mm.

As the number of active scan lines in a 625 line system is $N_y = 625 \sim 50 = 575$, the vertical dimension of one pixel on the target is:

$$d_{\text{pix},V} = \frac{V}{N_y} = \frac{9.6 \, \text{mm}}{575} = 17 \, \mu\text{m}$$

(7.4)

Fig. 7.1-a
Standard broadcast scanning

Fig. 7.1-b
Scanning for medical X-ray television

Fig. 7.2. Simplified diagram of a camera preamplifier
For simplicity square pixels are assumed, so:

\[ d_{\text{pix}} = d_{\text{pix}}, H = d_{\text{pix}}, V \]

\[ s_{\text{pix}} = (d_{\text{pix}})^2 = \left(\frac{V}{H}\right)^2 \]

The maximum charge of one pixel element is therefore:

\[ Q_{\text{pix}}, \text{max} = d_{\text{pix}}, \text{Qs}, \text{max} = 10 \times 10^{-4} = 28 \times 10^{-15} \, \text{C} \]

With a 4:3 scanning aspect ratio and square pixels the number of horizontal pixels on a line must be

\[ N_h = \frac{4}{3} N_v = \frac{4}{3} \times 575 = 767. \]

The value of \( t_{\text{pix}} \) is obtained by dividing the active line period \( T_h \) by the number of horizontal pixels \( N_h \).

\[ t_{\text{pix}} = \frac{T_h}{N_h} = \frac{52 \, \mu\text{sec}}{767} = 68 \, \mu\text{sec} \]

The maximum signal current in the broadcast situation described above is thus:

\[ I_s, \text{max} = \frac{Q_{\text{pix}}, \text{max}}{t_{\text{pix}}} = \frac{28 \times 10^{-15}}{68 \times 10^{-6}} = 410 \, \text{nA} \]

Note:
The assumption of square pixels is made for simplicity. Usually a so-called "Kell factor" is taken into account, which allows the use of a bandwidth lower than that corresponding to square pixels. The assumption about the horizontal pixel dimensions does not influence the calculated signal current as both \( d_{\text{pix}} \) and \( t_{\text{pix}} \) change to the same degree.

As the images from X-ray image intensifiers are always circular, a larger scanning area on the same 1" tube can now be used while still staying within the 16 mm quality area with the image part of the scanning, see Fig.7.1-b.

For this situation the calculations proceed as:

\[ d_{\text{pix}}, V = V = \frac{16 \, \text{mm}}{16 \, \mu\text{m}} = 28 \mu\text{m} \]

\[ Q_{\text{pix}}, \text{max} = 77 \times 10^{-15} \, \text{C} \]

\[ t_{\text{pix}} = 58 \, \mu\text{sec} \]

\[ I_s = 1140 \, \text{nA} \]

Thus due to the more-effective use of the target area, nearly 3 times more signal current can be accommodated by the same 1" camera tube. It is evident of course that the higher signal current also needs a larger light energy input, but as said before, this poses no problem in combination with image intensifiers.

Usually the actual optical demagnification of the image intensifier output screen is designed to give a 14.5 mm diagonal image on the target to provide some allowance for mechanical shift tolerances. This does not influence the above calculations however; the only difference is that the visible circular image on the TV monitor covers 90\% of the monitor height instead of 100\%.

Specifications of different Plumbicon tubes are compared in the following table.

<table>
<thead>
<tr>
<th>Table 7.1: Signal handling capabilities of Plumbicon tubes with XTV scanning format.</th>
<th>V (mm)</th>
<th>( d_{\text{pix}} ) (( \mu\text{m} ))</th>
<th>( C_s ) (pF/cm)</th>
<th>( Q_{\text{pix}}, \text{max} ) (C)</th>
<th>( t_s, \text{max} ) (( \mu\text{sec} ))</th>
<th>( I_s, \text{max} ) (nA)</th>
</tr>
</thead>
<tbody>
<tr>
<td>XQ 1072</td>
<td>16</td>
<td>28 ( \mu\text{m} )</td>
<td>570</td>
<td>88 \times 10^{-15} , \text{C}</td>
<td>1300</td>
<td>1300</td>
</tr>
<tr>
<td>XQ 1073</td>
<td>16</td>
<td>28</td>
<td>570</td>
<td>88</td>
<td>1300</td>
<td></td>
</tr>
<tr>
<td>XQ 2070</td>
<td>16</td>
<td>28</td>
<td>730</td>
<td>113</td>
<td>1660</td>
<td></td>
</tr>
<tr>
<td>XQ 1022</td>
<td>21.4</td>
<td>37</td>
<td>410</td>
<td>114</td>
<td>1680</td>
<td></td>
</tr>
<tr>
<td>XQ 1023</td>
<td>21.4</td>
<td>37</td>
<td>500</td>
<td>139</td>
<td>2040</td>
<td></td>
</tr>
<tr>
<td>45 XQ</td>
<td>26</td>
<td>45</td>
<td>500</td>
<td>204</td>
<td>3000</td>
<td></td>
</tr>
</tbody>
</table>

* detailed information via private communication with Drs A. Franken, Philips Elcoma div.
The first two tubes in this table (1072 and 1073) are currently the Plumbicons most-frequently used in XTV cameras. XQ 2070 has a diode gun with a high beam current capability. XQ 1022 and 1023 are of the 1 1/4" family and 45 XQ is the newest so-called "Frogs head" Plumbicon.

As the images from X-ray image intensifiers are circular it is sometimes advocated to use a non-standard TV system with 1:1 aspect ratio. The advantage is that for imaging the same small details, a smaller electrical bandwidth is sufficient. The consequence of the slower scanning is, however, that \( t_{\text{pix}} \) becomes longer and therefore \( I_{\text{signal, max}} \) becomes lower by a factor of \( \frac{3}{4} \). Weighing the disadvantage of a lower signal current against the advantage of a lower bandwidth requires a more-detailed knowledge of the noise spectrum and we will deal with this in a later section.

7.2. Sequential versus interlaced scanning.

The signal current calculations in the preceding section will hold only when each scan line reads only its "own" charge pattern on the target. Theoretically this needs a scanning spot size which is exactly equal to \( d_{\text{pix}} \) in the vertical direction. It is known (Franken, 1978; Kurashiye, 1982) that the effective spot size adapts to the instantaneous discharge current needed, provided that the scan lines are not too far apart in relation to the spot size.

In most broadcast applications the scanning spot is broader than the line pitch, so most of the charge of the target is read out in one field of an interlaced frame. If no new light flux is supplied after this first scan, only about 25% of the signal amplitude would be present in the second field of the frame. This means that instead of gathering the charge of one TV line, there is a partial overlap and the charge of about 1.6 TV lines is collected during interlaced scanning.

For the broadcast application the removed charge of a field read-out is continuously replenished, so a maximum continuous signal current \( I_{\text{signal}} \) higher than calculated in the preceding section may be allowed before the target voltage excursion limit is reached. For XTV scanning formats and/or larger target dimensions, the line pitch becomes larger in relation to the spot size which may decrease the 1.6x figure to a somewhat lower value.

The very same effect as discussed above forces us to deviate to a noninterlaced (sequential) scanning mode if video is to be acquired after a single light flash. Interlaced scanning would result in very different video amplitudes of the two acquired fields, producing an intolerable flicker in the stored and displayed interlaced picture. Sequential scanning of 625 lines in a vertical sweep period of 40 ms instead of interlacing two video fields of 20 ms and 312.5 lines each, provides a solution to this problem. Each line of the sequentially-read frame is then subjected to the same read-out procedure. Because the nonstandard sequential scanned video signal is unsuitable for direct display, the digital video memories in the system have to perform a scan conversion: sequential read-in and interlaced read-out.

The sequential scanning procedure implies that the effective scanning spot dimension in the vertical direction is equal to \( d_{\text{pix}} \), so that the signal level calculations of the preceding section will hold provided again that the line pitch is not so great that unstabilized regions remain in between the scanned lines (Franken, 1980).

7.3. TV camera noise.

The noise in the video output signal of the TV camera comes partly from the preamplifier and partly from the camera tube. In a simplified diagram of a video preamplifier, Fig. 7.2, three main sources of noise are indicated: (1) shot noise \( I_{\text{n1}} \) of the camera tube, (2) thermal noise \( I_{\text{n2}} \) of the load resistor \( R_L \), and (3) equivalent noise voltage \( e_{\text{n3}} \) of the amplifier itself. A detailed analysis of preamplifier circuit noise is given by Sadashige (1964) and Theile (1973).

7.3.1. Shot noise.

Even with an ideal noiseless preamplifier \( I_{\text{n2}} = 0, e_{\text{n3}} = 0 \) we still have to accept a limited signal-to-noise ratio because of the shot noise which is produced by the camera tube itself. The actual origin of the shot noise is the corpuscular character of the signal current; each pixel has a charge composed of a high but finite number of electrons. Although shot noise is fundamental and inevitable, it is of interest to compare its importance relative to the other electronic noise sources produced by the nonideal preamplifier. With a signal current \( I_s \) and a bandwidth \( B \) we have a shot noise contribution.

\[
I_{\text{n1}} = \sqrt{2eI_sB}
\]
where $q$ is the charge of one electron ($q = 1.6 \times 10^{-19}$ C).

Substitution of $B = 5$ MHz and some practical signal levels gives:

<table>
<thead>
<tr>
<th>$1_n$</th>
<th>$i_n$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1000 nA</td>
<td>1.26 nA</td>
</tr>
<tr>
<td>100</td>
<td>0.40</td>
</tr>
<tr>
<td>10</td>
<td>0.126</td>
</tr>
</tbody>
</table>

7.3.2. Thermal noise of the load resistor.

The thermal noise of the load resistor $R_L$ within a video passband $B$ is given by:

$$i_n^2 = \frac{4kT_B}{R_L}$$  \hspace{1cm} (7.11)

The maximum possible $R_L$ must be used for low-noise operation, but the higher $R_L$, the more will be the bandwidth limiting effect of the parasitic input impedance $C_L$. The required video bandwidth can be achieved by either correcting the amplifier frequency response with a slope of +5 dB per octave, or a current feedback circuit which reduces the effective input impedance of the preamplifier (virtual earth concept). Both methods are equivalent as regards the noise behaviour and also both methods are limited by the finite gain-bandwidth product of the preamplifier. In standard TV cameras therefore the practical choice of $R_L$ is confined to the 0.5 - 1 Mohm range.

Substitution of practical data such as $R_L = 0.5 \text{ Mohm}$ and $B = 5$ MHz will give $i_n^2 = 0.4$ nA.

7.3.3. Preamplifier noise.

The amplifier noise is usually considered to be caused by a (fictitious) equivalent noise resistance $R_{eq}$ of the first amplifier stage. The amplitude in an incremental frequency bandwidth $df$ is given by:

$$d_i^3 = \sqrt{4kT_{eq} df}$$  \hspace{1cm} (7.12)

Due to the frequency characteristic of the correction amplifier we will find at the output of the video preamplifier a noise amplitude:

$$d_i^3 = \frac{A_i}{R_{eq}} \sqrt{4kT_{eq} \frac{df}{R_L}}$$  \hspace{1cm} (7.13)

If we transform this noise contribution to an equivalent noise current $i_n^3$ at the input we find:

$$i_n^3 = \frac{d_i^3}{A_i R_L} = \frac{4kT_{eq}}{R_L} \sqrt{\frac{df}{R_L}}$$  \hspace{1cm} (7.14)

The spectral density of the noise thus increases with frequency. The integral of the noise spectral density over the video passband follows from:

$$i_n^3 = \frac{4kT_{eq}}{R_L} \int_0^B \left[ 1 + \frac{4kT_{eq}}{R_L} \frac{df}{R_L} \right] df$$

$$\approx \frac{4kT_{eq}}{R_L} \left[ B + \frac{4kT_{eq}}{3} \frac{df}{R_L} \right]$$

$$\approx 4kT_{eq} \frac{4kT_{eq}}{3} \frac{df}{R_L}$$  \hspace{1cm} (7.15)

An important design parameter for low-noise performance is thus $C_L \sqrt{R_{eq}}$, and all possible measures must be taken to keep this product as low as possible.

The input capacitance $C_L$ is composed of two components: (1) $C_{target}$ mainly determined by the mechanical construction and (2) the input capacitance $C_{amp1}$ of the first amplifier stage.

Typical data for standard 1" plumbicons in XTV cameras are:

- Internal capacitance of camera tube: 2.5 pF
- Capacitance to deflection coils: 4.0
- Capacitance to TV lens: 2.0
- Wiring capacitance to preamp: 2.5

$C_{target} = 11$ pF.
New developments in plumbicon design (LOC tube, Franken, 1981) as well as minimizing wiring capacitance by mounting the first preamplifier stage close to the target connection inside the yoke, may allow a reduction of the target capacitance by some 4 pF in future camera designs.

A junction FET transistor (JFET) is the preferred choice for high impedance signal sources. The choice of a JFET with the lowest input capacitance is not always the best, as input capacitance and amplifying power (transconductance $g_m$) appear to be interdependent. Theoretical derivation (van der Ziel, 1962) shows that the following holds for JFETs:

$$R_{eq} = \frac{2}{3} \cdot \frac{1}{g_m}$$  \hspace{1cm} (7.16)

In practice however somewhat more noise is found (Motchenbacher, 1973 and see also product data of BFW 11), so that a better approximation is:

$$R_{eq} = \frac{1}{g_m}$$  \hspace{1cm} (7.17)

For low noise performance we have to look for a FET which combines high transconductance with low input capacitance. A good example of current technology is J310 with $g_m = 12$ mA/V and $C_{FET} = 6.1$ pF. Older designs often use BFW 11 with $g_m = 4.5$ mA/V and $C_{FET} = 4.9$ pF. Sometimes it is advocated to connect several FETs in parallel. The transconductance of such a combination is higher, but the input capacitance also. Analysis of such a combination will show that there appears to be an optimum number of FETs to use:

$$n_{opt} = \frac{C_{target}}{C_{FET}}$$  \hspace{1cm} (7.18)

For our typical conditions $n_{opt} = 2$, but the difference in $C_1 \sqrt{R_{eq}}$ with $n = 1$ is so small (about 14%) that for S/N reasons it is hardly worthwhile to make the effort of using parallel FETs. Another reason why two parallel FETs are used sometimes is that the gain-bandwidth product of the first (cascode) amplifier stage is doubled, which makes it easier (1) to achieve the required bandwidth, and (2) to diminish noise contributions from subsequent amplifying stages. Additional noise from subsequent amplifying stages is

---

![Fig. 7.3](image-url)  
Camera noise currents as a function of signal level

![Fig. 7.4](image-url)  
Signal-to-noise ratio of the TV camera
negligible if the first stage has sufficient voltage gain; for
instance more than 10.

While parallel connection of FETs is not worthwhile for S/N impro­
vement, a measure which does make sense is the selection of FETs with a
high transconductance. Manufacturer's product data usually give typi­
cal figures and a relatively large spread, especially in transconduc­
tance $g_m$, may be expected (63% - 160% of typical is common). A dif­
ference in noise performance by a factor 1.6 (4 dB) between the worst
case and the best case of a particular type can thus be expected.

The practical amplifier noise level that can be expected for the usual
single FET input stage follows from Eq(7.15) by substituting $B = 5 \text{MHz}$
and $C_i = C_{\text{FET}} + C_{\text{target}} = C_{\text{FET}} + 11 \text{pF}$.

The results are:

<table>
<thead>
<tr>
<th>$C_i$ (pF)</th>
<th>$F_{\text{eq}}$</th>
<th>$i_{n3}$ (nA)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Older designs</td>
<td>15.9</td>
<td>222</td>
</tr>
<tr>
<td>New designs</td>
<td>17.1</td>
<td>83</td>
</tr>
</tbody>
</table>

7.3.4. Overall TV noise.

The result of the preceding noise calculations is illustrated in
Fig.7.3. The shotnoise $i_{n1}$ is dependent on the video signal level $I_s$, while Johnson noise $i_{n2}$ and preamplifier noise $i_{n3}$ are independent of
the signal level. For simplicity $i_{n3} = 1.0 \text{nA}$ is chosen, which is a
performance level halfway between the old and new amplifier designs.

The amplifier noise $i_{n3}$ appears to dominate over a large part of the
signal range.

As the three noise sources are independent, the total noise current
can be found by powerwise addition of the three:

$$i_{n_{\text{tot}}} = \sqrt{i_{n1}^2 + i_{n2}^2 + i_{n3}^2}$$

The signal-to-noise ratio of the TV camera, which is defined as
$(S/N)_{\text{tot}} = I_s/i_{n_{\text{tot}}}$, is shown in Fig.7.4. Assymptotes in this figure
are formed by $S/N_1 = I_s/i_{n1}$ and $S/N_2 = I_s/i_{n2}$.

7.4. Camera noise for larger details.

So far the camera noise has been considered on a per pixel basis.
All three noise sources $i_{n1}$, $i_{n2}$ and $i_{n3}$, depend on the video band­
width $B$. Especially the preamplifier noise, $i_{n3}$, with its rising speci­
cral density is heavily dependent on bandwidth.

If only details larger than a pixel are of interest, the system
could as well use a lower bandwidth $B$, yielding a considerable gain in
noise performance. If, for instance, the region of interest (e.g. the
vessel) in the image has a width of 4 pixels, then a bandwidth of
$1/4$th would suffice. The differences in noise level for $B = 5\text{MHz}$
applied to the preamplifier discussed earlier are presented in the
following table.

<table>
<thead>
<tr>
<th>$B$ (MHz)</th>
<th>$i_{n1}$ (nA)</th>
<th>$i_{n2}$ (nA)</th>
<th>$i_{n3}$ (nA)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$5\text{MHz}$</td>
<td>1.26</td>
<td>0.40</td>
<td>1.00</td>
</tr>
<tr>
<td>$1.25\text{MHz}$</td>
<td>0.63</td>
<td>0.20</td>
<td>0.63</td>
</tr>
</tbody>
</table>

An important conclusion which can be drawn now is that for the lower
bandwidth, $i_{n3}$ no longer dominates; see also Fig.7.5 and Fig.7.6.

If the region of interest in the image is increased in both the
horizontal and vertical dimensions, a two-dimensional lowpass filter
could be used to improve the signal-to-noise ratio even further.

The lowpass filtering and its consequences on signal-to-noise per­
formance have been discussed simply to illustrate what the effect
would be on the various noise sources. This does not mean that such
filtering also has to be executed in order to obtain the predicted S/N
improvements. In practice the human psychovisual interpretation will
give a similar result while judging the noisy images. The human eye
synthesizes its own adapted filtering while concentrating on a detail
of a certain size; the human observer functions as a signal-matched
noise integration system (Chester,1981). The same philosophy is the
basis of the generally-accepted Rose theory (chapter 4).

In standard broadcast television the response of the human visual sys­
tem to noise is usually taken into account by a so-called "noise
weighting function", which measures the noise in a bandwidth lower
than the signal bandwidth. Recently, two-dimensional noise weighting
functions have also been proposed. (Fuijo,1980).

7.5. Alternative scanning methods.

The more detailed knowledge of the influence of bandwidth $B$ on
noise allows a return to an earlier unfinished discussion about 4:3
aspect ratio versus 1:1.
As concluded earlier, a change from a 4:3 to a 1:1 aspect ratio decreases the signal current $I_g$ by a factor $3/4$ due to the longer read-out period $t_{pix}$, if the charge $Q_{pix}$ per pixel is kept constant. Due to the slower read-out the bandwidth $B$ can be reduced by a factor of $3/4$. Substitution of the change factors in the noise equations Eq(7.10), Eq(7.11) and Eq(7.15) will lead to:

\[
\begin{align*}
S/N_1 &= \frac{I_g}{h_1} \\
S/N_2 &= \frac{I_g}{h_2} \\
S/N_3 &= \frac{I_g}{h_3}
\end{align*}
\]

\[
\begin{align*}
S/N_1' &= \frac{I_g \times 0.75}{h_1 \times 0.75} = S/N_1 \\
S/N_2' &= \frac{I_g \times 0.75}{h_2 \times (0.75)^{0.5}} = 0.875 S/N_2 \\
S/N_3' &= \frac{I_g \times 0.75}{h_3 \times (0.75)^{1.5}} = 1.15 S/N_3
\end{align*}
\]

From this overview it can be concluded that the signal-to-shot noise ratio $S/N_1$ remains constant. This is understandable because the shot noise is actually caused by the limited number of electrons in the charge packet $Q_{pix}$ of one pixel, and this charge packet is not dependent on the scanning method.

The signal-to-thermal-noise ratio $S/N_2$ decreases by about 13% while the signal-to-amplifier-noise ratio $S/N_3$ improves by about 15%. The overall effect of the changes depends on the signal current level. For noise on a per pixel basis, and a low signal current, $S/N_3$ dominates and the full 15% improvement will be effective. For high signal currents, where shot noise is no longer negligible, the advantage becomes much less than 15%. If the camera noise is considered for larger details as is done in the previous section, then thermal noise becomes dominant for low signal currents, and the fact that this $S/N_2$ becomes worse for 1:1 scanning is a disadvantage. In conclusion it can be stated that there is no marked advantage for the 1:1 scanning method.

A similar discussion can be held for the sometimes advocated slow scan method, which purposely makes a bigger step in the slow scan direction than the 1:1 aspect ratio just discussed. When, as an example, a 4 times slower scantime is considered with again an unchanged charge level on the target, we have:
The deterioration of $S/N_2$ is now even more explicit and in practice may cause $S/N_2$ to dominate over $S/N_3$ for low signal levels.

Schreibet (1964) noticed that the relative thermal noise contribution does not need to increase with slower scan speeds if the load resistor is made inversely proportional to the bandwidth. In that case not only the shot-noise contribution but also the thermal-noise contribution become independent of the scan speed. In practice this solution can not be used because an X-ray TV camera always has to be capable of normal fluoroscopy at the standard scanning rate, which limits the usable upper level of load resistance $R_L$ in this mode. Switch over of the TV camera to a slow scan mode is feasible for the scanning circuitry, but switching $R_L$ to a higher value during slow scan must be considered as unfeasible as $R_L$ is at the most-sensitive point of the whole TV chain.

A third scan method is the "high resolution" standard which has twice the number of TV lines and twice the number of pixels on a line. One pixel in such a system has thus 1/4th of the standard pixel area, and thus $Q_{pix}$ is 4x smaller also. The read-out time per pixel is 4x shorter and the bandwidth needs to be 4x larger. In this situation we have:

**Standard scanning**

<table>
<thead>
<tr>
<th>$S/N_1$</th>
<th>$S/N_2$</th>
<th>$S/N_3$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\frac{l_1}{n_1}$</td>
<td>$\frac{l_2}{n_2}$</td>
<td>$\frac{l_3}{n_3}$</td>
</tr>
</tbody>
</table>

**High resolution scanning**

<table>
<thead>
<tr>
<th>$S/N_1$</th>
<th>$S/N_2$</th>
<th>$S/N_3$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\frac{l_1 \times 0.25}{n_1 \times 0.25} = 0.50 S/N_1$</td>
<td>$\frac{l_2 \times 0.25}{n_2 \times 0.25} = 0.50 S/N_2$</td>
<td>$\frac{l_3 \times 0.25}{n_3 \times 0.25} = 0.125 S/N_3$</td>
</tr>
</tbody>
</table>

In particular the contribution of the amplifier noise appears to be much larger (8x). The much higher amplitude of the very fine grained noise in this case is subjectively not as bad as the 8x amplitude suggests. Lowpass filtering by the human eye (noise weighting as mentioned earlier) causes the visual effects of the high amplitude noise to be less dramatic.

7.6. Noise caused by deflection instability.

A peculiar noise phenomenon which appeared unnoticeable in normal operation of high-quality TV camera, became evident with the DSA application.

Uniformly-illuminated images showed the problem most strikingly. The nature of the visible disturbance is a random horizontal streaking in the image. Each horizontal TV line seemed to have a slightly different amplitude; video experts would be inclined to suspect clamp noise as the cause of this interference. The problem proved to be related to the scanning mechanism as slight defocusing of the scanning beam made the interference disappear. Similar problems are reported by Schade et al (1970) in high-definition television systems.

A plausible explanation of the disturbance is the following.

If the scanning beam size is smaller than the raster pitch, some target charge remains unscanned between the raster lines. If subsequent raster scans do not retrace the same path exactly, a certain amount of the left-over "in between" charge will be stabilized, resulting in an increased video signal amplitude.

The cause of the scan instability is the noise produced by the vertical deflection amplifier. The allowed vertical instability $\Delta d$ is only a fraction of the raster pitch $D$, e.g.:

$$\frac{\Delta d}{D} < \frac{1}{10}$$

$$\frac{\Delta d}{V} = \frac{\Delta d}{576 D} < \frac{1}{5750}$$

Any noise voltage superimposed on the deflection waveform must thus have a peak amplitude of 75 dB (20 log 5750) below the sawtooth peak-to-peak voltage. The design of the vertical deflection amplifier needs special attention to meet this "hifi" requirement. Judicious choice of low noise opamp ICs and restricting the bandwidth to the real basic need (~ 5kHz) has resulted in a satisfactory performance.

The reason that the effect described is unnoticeable in standard TV images is that the subtraction and subsequent contrast gain of the DSA...
system bring these small disturbances over the threshold of the just visible.

7.7. Practical implications of intermittent TV operation.

For most of the vascular DSA applications, the TV camera is used in a snapshot mode, in contradistinction to the continuous mode of operation during conventional television applications. The TV camera receives its input as repetitive flashes of light. In between the flashes no light is supplied to the camera. If the camera keeps scanning and reading off the target in these dark periods, the target becomes negatively charged due to the velocity distribution of the electrons in the beam. Not all the electrons of the beam arrive with zero velocity at the target; some have a slightly higher velocity and will be able to land, which charges the target negatively. It has been observed that the amount of negative charge buildup depends on the duration of the dark period. When an exposure is subsequently made, the positive charge resulting from the light has first to surpass the negative charge in order to be readable. This effect is the origin of the so-called "build-up lag" of the camera tube. The origin of this effect can be reduced appreciably if the camera target is not scanned in the dark periods, which has been realized by suppressing the beam current during these dark periods.

Another method of reducing the build-up lag of the camera tube is the application of a small amount of light biasing, enough to compensate or even overcompensate the negative build-up. The nonzero video signal corresponding to this artificial dark current may be offset by a constant voltage in the TV chain, provided that the dark current is uniform over the whole image area. Good uniformity of bias light is very difficult to achieve and in order not to be disturbed by the shading, a very low bias light level is usually applied.

In addition to the build-up lag, the decay lag of the camera tube also requires special attention during the intermittent mode of operation. Even if the beam current is superfluous, some charge still remains after a single scan of the target. This is caused by the inherent lag of the target layer itself and by the lower beam acceptance for the lower target-voltage excursions (v.d. Polder, 1967). If subsequent snapshot images are not to be disturbed by leftovers from the preceding image, a so-called "target erase" scan cycle has to follow the normal target read cycle. When beam current suppression (as discussed above) is used, the beam switching has to take this erase cycle into account. Although a video signal is produced during the erase cycle, the amplitude is much smaller than the original video signal from the read cycle. Due to this reduced amplitude the signal-to-noise ratio of the video during the erase cycle is much lower and not worthwhile for use for data-acquisition; summation with the original video would worsen the signal-to-noise ratio rather than improve it.

The implementation of the above-mentioned special measures during intermittent operation requires careful synchronization of events. During the light flash the beam current must be suppressed. Depending on the expected movement of the object, an exposure time in the range of 5-200 ms may be used. After completion of the exposure, the beam must be switched on at the first coming V-drive pulse, initiating the (sequentially-scanned) read-out cycle. Only the first video read-out is used for data acquisition. Depending on the situation, one or more target erase cycles can be applied before the beam is switched off again. With light bias, the beam switch off may wait until just before the following exposure.

In the DSA system all these events are switched V-synchronously under microprocessor control, as both the exposure time and the imaging rate have an influence on the required switching waveforms. Conversely, the sequence of the three events, (1) exposure, (2) read and (3) erase, sets a maximum imaging rate. In a 625 line TV system, the frame period of 40 ms is the logical choice for the event clock, so a minimum repetition time of $3 \times 40 = 120$ ms holds, which corresponds to a maximum imaging rate of $25/3 = 8.33$ exposures/s. In practice the imaging rate is also limited by the speed of the image data transfer to the digital disc. With present-day technology this takes about 200 ms for 512 images, resulting in an effective maximum imaging rate of about 3/s.

7.8. Frame integrations.

When the object to be visualized does not contain fast-moving details, a relatively long exposure time may be tolerated. Instead of splitting up the acquisition cycle into separate expose and read-out periods, the two can be combined in a read-while-expose period, provided that an integer number of frame periods is involved. Several TV frames can be electronically added or "integrated" with the intention of reducing the noise. Frame integration can readily be performed with digital video memories, and in this digital world an obvious choice for the number of frame integrations is 1, 2, 4 or 8, resulting in effective exposure times of 40, 80, 160 or 320 ms. Longer exposures are considered unsuitable, and even 320 ms is marginally usable.
As the signal level adds up linearly with summation and the uncorrela-
ted noise adds up powerwise, the S/N improvement factor is equal to
the square root of the number of integrations, thus 1, 1.4, 2 and 2.8
for 1, 2, 4 and 8 integrations respectively. The improvement factors
hold for each individual pixel regardless of the noise in neighbouring
pixels. This means that the spectral dependence of the noise remains
unaltered; e.g. pink noise (rising with frequency) of the preamplifier
stays pink, but at a reduced level.

Interesting is that both the quantum noise and the electronic noise
are reduced by the video integration. When, however, the same amount
of X-ray dose per image is used irrespective of the exposure time,
the X-ray quantum noise is constant. Longer exposure times have less
dose per TV frame and thus more X-ray quantum noise per TV frame, but
after the summation the X-ray quantum noise has settled to the value
which theoretically belongs to the applied total dose. In practice
this means that frame integration is especially effective for elec-
tronic noise.

A special problem related to the frame integration mode is the
build up of the charge on the target of the TV camera. Even when the
start of the X-ray exposure and the start of the TV scanning are per-
fectly synchronized, the read-out of the first video frame is still
usable, as during this first scan the charge build-up process in the
camera tube is still busy. The first (upper) lines of the frame have
hardly experienced any light exposure when the beam passes these lines
to discharge, while the last lines have had a nearly complete expo-
sure. The video signal of the first frame is thus useless and must be
discarded. Subsequent frames will have the full amplitude and can be
used for data acquisition.

This means that always one extra frame time of exposure is needed
before the effective expose and read-out period. As the X-ray dose of
this first frame is not used effectively, the frame integration mode
requires more dose than a single sequential scan mode operation. For
this reason the choice of 2 integrations is useless as a factor of
(1+2):2 = 1.5x more dose will be used for an improvement of (only) the
electronic noise by a factor of 2 = 1.4. For 4 integrations the situ-
atation is not as bad because here 4/5 = 80% of the X-ray dose is effec-
tively used.

7.9. Conclusions.

A high camera read-out signal current is advantageous for an
optimal S/N ratio.

Many Plumbicon camera tubes (especially the diode gun types) have
the capability of realizing a much higher current than is usually
applied.

Limitations of applicable signal current level are formed by beam
bending effects and target saturation. A large target layer
capacitance is advantageous in this respect (large area and/or large
specific layer capacitance) although image lag in the tube will then
also be larger which compromises the imaging of fast dynamic effects.

A further important parameter is the size of the scan area on the
camera tube. Scanning a 4:3 rectangle just within the circular target
area (as done in broadcasting) or scanning a 4:3 rectangle just
comprising the circular area (as done in medical TV cameras) makes a
difference of about a factor 2.8 in area and thus in available signal
current.

Interlaced scanning of the TV camera tube is less impeded by target
saturation effects then the sequential scanning method. Depending on
the spot size, a factor of between 1 and 2 more signal current is
allowed in the interlaced mode as the charge of more than one TV line
is acquired during each field scan.

The overall S/N ratio of the TV camera is dependent on the signal
level. For high signal currents ( > 500nA) the shot noise of the TV
camera tube is dominant.

A S/N ratio of 60 dB (= 1000:1), which is often mentioned in the DSA
literature, is hardly achievable when the shot noise contribution is
taken into account.

For lower signal currents ( <500 nA, as is usual practice in
broadcast TV) the preamplifier noise is dominant.

Application of "noise weighting" reveals that for details larger than
the pixel size, the dominance of the preamplifier noise wears off and
the thermal noise of the load resistor becomes more important at low
signal levels.

Of the alternative scanning methods, a 1:1 aspect ratio has no
worthwhile advantages.

Slow scan TV is good for reducing the preamplifier's noise contribu-
tion, but at the same time the thermal noise contribution is in-
creased. A dedicated preamplifier design with an extra high load
resistor may reduce the thermal noise problem, but this makes the
camera less suitable for standard scan rates.

The high resolution scan standard needs so high a bandwidth
that 3 times more amplifier noise will be present. As this higher noise amplitude is concentrated mainly in the high frequency part of the spectrum, the visual disturbance is not as dramatic as the amplitude may suggest (noise weighting).

Vertical deflection instability has been discovered to be much more critical in DSA applications than in standard unsubtracted TV. S/N degradations of the video signal can be prevented by proper design of the deflection circuitry.

Intermittent TV operation requires special measures to counteract the build-up and discharge lag of the camera tube. This together with the (present-day) limited digital transfer speed to disc, limits the exposure rate to 1/s maximum.

Note: Digital discs with higher transfer speed are becoming available so future developments will certainly provide an imaging rate of 25/s.

Frame integrations improve the S/N of the TV camera, but at the expense of one extra frame time of X-ray exposure. For this reason 2-frame integration is useless. In view of the exposure duration, 8-frame integration is too long (320 ms) for most examinations, so 4-frame integration remains as the most suitable option.

References


Schut Th.G.: Recente ontwikkelingen bij opneembuizen van het type "Plumbicon". De Ingenieur, 84, 17 nov. 1972, ET136-142.
3. **SAMPLING THEORY APPLIED TO VIDEO WAVEFORMS.**

If an image is displayed as a matrix of picture elements (pixels), the most obvious conclusion which can be drawn immediately is that the finite pixel size limits the display of very fine spatial details. Judicious choice of a suitable matrix size will be discussed in chapter 10. Apart from this most obvious effect some video waveform distortions may also occur when the sampling theory is not applied properly.

8.1. **The ideal sampling system.**

The sampling theory requirements are best understood via the frequency domain. Suppose the input signal of a sampling system has a bandwidth $B$ as schematically shown in Fig.8.1-a. Sampling this input signal at a sampling frequency $f_s$ results in a frequency spectrum which still contains the original input spectrum, but also contains higher order spectral components centered at multiples of $f_s$; see Fig.8.1-b. (Reference textbook Carlson,1968). Fig.8.1-b shows that the original input can be separated from the spectral mixture, provided that the sidebands do not overlap. Obviously two conditions are necessary to prevent overlapping sidebands:

1. The input signal must be band limited, say in $B$.
2. The sampling frequency $f_s$ must be sufficiently high; $f_s > 2B$.

The theoretical minimum sampling frequency $f_s = 2B$ is called the Nyquist rate.

For ideal waveshape matching of the input and output signals of a sampled data system, it appears that two lowpass filters are required: (1) an input filter which restricts the bandwidth and ensures that the baseband and sidebands are separable, and (2) an output filter which does the separation.

8.2. **Practical sampling with nonideal filters: aliasing.**

In practice we always have to live with nonideal lowpass filters which have a finite cut-off slope.

To prevent overlapping sidebands, the sampling frequency is very often made somewhat higher than the Nyquist rate; for instance, 3.4 kHz speech signals are usually sampled at 8 kHz ($f_s/B = 2.35$). By this means a so-called guard band is created, accommodating the finite slopes of the filters.
Even when the sampling frequency is higher than the Nyquist rate, overlap of the tails of the sidebands is usually unavoidable; see Fig.8.1-c.

When this occurs then some frequencies originally outside the nominal message band (e.g. \( f_1 \)), will appear at a much lower frequency inside the message band (\( f_2 = f_S - f_1 \)). This phenomenon of downward frequency translation is called "aliasing", and an important design aspect of a sampled system is how much of this aliasing can be tolerated.

One can imagine that in audio systems the aliased frequencies would sound like high-frequency garble or hissing. For sampled television signals there is no simple interpretation of the effect of aliasing on the video waveform.

### 8.3. Edge busyness effect.

As a first step in trying to understand the bad effects of aliasing on TV signals, a situation is considered where a unit-step signal is subjected to sampling without applying any input bandlimitation at all.

The samples taken from this input signal are designated \( S_1, S_2, S_3, \ldots \). Depending on the relative phase of the sampling pulses, different situations may occur as shown in Fig.8.2. The reconstruction of the original signal from these samples can be done with a first approximation by connecting \( S_1, S_2, S_3, \ldots \) with straight lines. The results show that the 50% points of the reconstructed signal depend on the phase of the sampling pulses; a timing instability \( \Delta t = T_S \) is possible. On a television display this timing instability causes a jitter or "edge busyness" of the contours in the image.

As the ideal sampling process with the appropriate filtering is able to restore the original input signal ideally, one can imagine that an intermediate situation with practicable (nonideal) filters and a small amount of aliasing will result in a minute amount of edge busyness.

### 8.4. Transient analysis of a sampled system.

More detailed information about the aliasing distortion requires transient analysis.

A single unit-step waveform has a spectrum content:

\[
V_f(\omega) = \frac{1}{1 + j\omega} \quad (8.1)
\]
After applying this signal to an input lowpass filter with frequency characteristic \( F_1(\omega) \) we obtain the filtered input spectrum.

\[
V_2(\omega) = \frac{1}{j\omega} F_1(\omega)
\]  
(8.2)

The sampling process which results in a repetition of the input spectrum at multiples of the sampling frequency \( \omega_s \), can be represented by:

\[
V_3(\omega) = \sum_{n=-\infty}^{\infty} V_2(\cdot n\omega_s) e^{j\omega \cdot n\omega_s}
\]

\[
= \frac{F_1(\omega)}{j\omega} + \sum_{n=0}^{\infty} \frac{F_3(\omega_n\omega_s)}{j(\omega_n\omega_s)} e^{j\omega_n\phi}
\]  
(8.3)

Parameter \( \phi \) in this equation represents the phase of the sampling waveform with respect to the input transient.

After application of an output lowpass filter with frequency characteristic \( F_2(\omega) \) the reconstructed waveform spectrum is:

\[
V_4(\omega) = \frac{F_1(\omega) F_2(\omega)}{j\omega} + \sum_{n=0}^{\infty} \frac{F_3(\omega-n\omega_s) F_2(\omega)}{j(\omega-n\omega_s)} e^{j\omega \cdot n\phi}
\]  
(8.4)

The first term in this equation is equivalent to the output of an ideal sampling system; the waveform is just the input step applied to a cascade of two lowpass filters. The second term in Eq(8.4) is the distortion generated by the nonideal sampling system. A time-domain expression of the output signal is obtained by Fourier transformation:

\[
v_4(t) = \frac{1}{2\pi} \int_{-\infty}^{\infty} V_4(\omega) e^{j\omega t} \, d\omega
\]  
(8.5)

As we are not particularly interested in the fixed delay caused by the two lowpass filters, we might as well simplify the calculations by assuming \( F(\omega) = F(\omega) \) which means that the filters have no phase shift at all. We further assume that the lowpass filters attenuate sufficiently at \( \pm \omega_s \) which means that only the first-order
sidebands \((n = \pm 1)\) have to be taken into account.

Then it will follow that:

\[
v_4(t) = \frac{1}{\pi} \int_{0}^{\pi} F_1(\omega) F_2(\omega) \frac{\sin \omega t'}{\omega} d\omega + \frac{1}{\pi} \int_{0}^{\pi} F_1(\omega - \omega_s) F_2(\omega - \omega_s) \frac{\sin(\omega t' - \omega_s t)}{\omega_s} d\omega
\]

\(= v_{4,0}(t) + e(t)\)

Again the first term \(v_{4,0}(t)\) is the wanted result and \(e(t)\) is the distortion.

In order to get an idea about the nature of the disturbance \(e(t)\), some computer calculations have been carried out evaluating Eq (8.6) for a certain set of practicable filters. For simplicity, identical input and output filters of the Butterworth type of order \(m\), phase-corrected for a symmetrical stepresponse, are used.

\[
F_1(\omega) = (1 + \frac{\omega}{\omega_c})^{-2m} \frac{1}{2m} \frac{\sin 2 \frac{\omega}{\omega_c}}{2 \pi \frac{\omega}{\omega_c}}
\]

\[
F_2(\omega) = (1 + \frac{\omega - \omega_s}{\omega_c})^{-2m} \frac{1}{2m} \frac{\sin 2 \frac{\omega - \omega_s}{\omega_c}}{2 \pi \frac{\omega - \omega_s}{\omega_c}}
\]

The second factor in \(F_2(\omega)\) represents the inevitable additional low-pass filtering caused by the sample-and-hold circuitry.

For a given cutoff frequency \(f_c\) of the Butterworth filters (i.e. a given video bandwidth), the order of the filter \(m\) and the sampling frequency \(f_s\) are parameters available to reduce the disturbance. Fig.8.3 shows the results of one particular case: \(m = 4\) and \(f_s/f_c = 2\).

The presence of \(e(t)\) means that, dependent on the sampling phase, both edge displacement and slope modulation of the transient may occur. If the sampling frequency is asynchronous to the TV deflection all phases may occur at a particular spatial point in the image, resulting in the previously-predicted edge busyess of contours in the image. The nature of \(e(t)\) appears to be similar for all kinds of other filters; generally it has a bell-shaped envelope which is filled in with a frequency of about \(1/2 f_s\), the phase of which is determined by the sampling phase.

The peak amplitude \(e_{max}\), which may be considered as a measure for the annoyance in the displayed image, appears to be about proportional to the area under the aliased part of the spectrum (shaded area in Fig.8.1-c).

Summarized results of \(e_{max}\) calculations for different Butterworth
filters and sampling frequencies are shown in Fig. 8.4. It is no surprise that the closer one wants to approach the Nyquist rate, the more complicated the filtering must be to preserve a low distortion level.

8.5. Line-locked sampling.

When subtracted images are being displayed, as is the case with DSA, the aliasing distortion, which will most probably be different in the two original images due to the asynchronousness of the sampling clock, may show up as a rather disturbing zipper-like patterning. Subtraction results from two images taken from the same stationary object, should show a flat unmodulated uniform field, but the aliasing distortion shows the zipper-like structures at those positions where high contrast contours are present in the unsubtracted original images.

A very useful method of reducing the visibility of aliasing distortion effects is to use a synchronous sampling clock. The sampling frequency must then be a multiple of the horizontal scan frequency. This gives an integer number of sampling points per line and a sampling pattern which is stationary with respect to the TV frame. In that case the aliasing distortion is still present in both unsubtracted images, but will cancel out after subtraction because both images contain the same distortion.

In this way the filtering problem can appreciably be alleviated. Even a display of unsubtracted images, which still contain the distortion, is subjectively much more acceptable because the edge busyness has now turned into a stationary distortion of high contrast contours, visible as jagged sloping contours. This kind of disturbance appears to be much more tolerable.


The best stability of a sampling clock is obtained when starting out with a crystal oscillator at the sampling frequency and subsequently dividing the horizontal and vertical scan frequencies for the TV system from this stable oscillator. Unfortunately with X-ray TV systems we are usually forced to deviate from this approach. The reason for this is that the high voltage of the X-ray tube always contains a rather-high mains frequency-related ripple, which is reflected as an even-stronger ripple component in the X-ray output. In order to prevent this ripple becoming visible in the TV images the vertical scan rate is usually phase-locked to the mains
frequency. Generating the sampling frequency requires an additional phase-locked-loop circuit that locks on the TV line rate. The cascade of two PLL circuits is less optimal in the sense that a small phase-jitter may exist, resulting in a spatially-unstable sampling pattern; the alignment of the frame-memory matrix with the object is unstable in the horizontal direction.

The effect of this instability is identical to the edge busyness and zipper effect in the difference images as discussed in the previous section. An important difference, however, is that improved filtering will not help to reduce the effect; the sampling clock has to be stable regardless of the amount of applied video filtering. The error voltage, \( e \), generated in a subtracted image at the position of a high contrast contour is proportional to the steepness of the contour waveform and to the shift amplitude. The steepness of a properly filtered step-waveform of amplitude \( E \) is \( E/T_s \), see Fig. 8.5.

When there is a clock jitter, \( \Delta t \), between the two images to be subtracted, the error voltage \( e \) is then:

\[
e = \frac{\Delta t}{T_s} E
\]

(8.9)

If the error voltage is required to be a fraction of the contrast step amplitude \( E \), the required clock stability must be within the same fraction of the sampling interval. In practice this requirement is very hard to fulfill, but fortunately the steepness of contours in X-ray images is limited, which alleviates the problem. For electronically generated transients in the image, such as the electronically-gated circle blanking, the worst-case situation given by Eq(8.9) holds, and noticeable nonideal subtraction for these waveforms can therefore be expected.

8.7. Aliasing of noise.

In section 8.2 it has been demonstrated that when unsufficient low-pass filtering is applied before sampling, downward frequency translation of the highest part of the input spectrum takes place. In addition to disturbing the transient waveform, this effect also has serious consequences for the noise characteristics of a sampled system.

Suppose that the object to be imaged is only a uniform unmodulated flat field. The corresponding video signal is just a d.c. voltage, but some inevitable noise is always superimposed due to TV preamplifier noise and the X-ray quantum noise. Very high frequency noise (\( > 1/2 f_s \)), which may be hardly visible in the unsampled video signal, will show up as visible noise of lower frequency (\( < 1/2 f_s \)) in a sampled video system.

If instantaneous sampling (practically zero aperture time) is supposed, it is easy to grasp that the (quasi) peak-to-peak value of the noise remains the same after the sampling process; only the spectral composition of the noise will have been changed. In particular if we recall that the TV preamplifier noise has a triangular spectrum (see section 7.3.3) it is clear that for this type of noise especially a proper lowpass filter before sampling is required.

Again we may ask here what proper lowpass filtering means as regards the complexity of the filter. For this purpose a calculation has been made of the effect of a Butterworth lowpass filter upon a triangular noise spectrum. When the order of the filter is \( m \), the filtered r.m.s. noise voltage \( e_n \) follows from:

\[
e_n^2 = \int_0^{\omega} \frac{\pi^2}{1 + \left( \frac{f}{f_c} \right)^{2m}} df
\]

(8.10)

\[
e_n^2 = \frac{1}{3} K f_c^3 \frac{3\pi}{2m} \frac{3\pi}{m} \frac{2m}{\sin 3\pi}
\]

where \( K \) is a constant (Volt\(^2\)Hz\(^{-3}\)).

The results for various filter orders \( m \) is given in the following table.

<table>
<thead>
<tr>
<th>( m )</th>
<th>( e_n )</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>1.83</td>
</tr>
<tr>
<td>3</td>
<td>1.25</td>
</tr>
<tr>
<td>4</td>
<td>1.13</td>
</tr>
<tr>
<td>5</td>
<td>1.08</td>
</tr>
<tr>
<td>6</td>
<td>1.05</td>
</tr>
<tr>
<td>( \infty )</td>
<td>1.00</td>
</tr>
</tbody>
</table>
From this table it can be concluded that a fifth-order filter appears to be close to the theoretical performance of the ideal filter ($m = \infty$).

8.8. Visibility of pixelization.

In this chapter it has been demonstrated that both an input filter and an output filter are needed in a sampled system in order to restore as close as possible the waveform of the input signal. If this is done with the appropriate filtering it will be impossible to recognize individual pixels in the image; the pixel boundaries are smoothed by the output filter.

In practice it has turned out that this is not always a wanted situation. As a matter of fact some viewers even prefer to see the pixelization which shows up when the output of the sampled system (= output of the digital-to-analogue converter) is fed to the TV monitor without using a lowpass output filter at all. This waveform has a staircase-like approximation identical to the waveform from a sample-and-hold circuit. The possible presence of steep edges at pixel boundaries indicates that the higher-order sidebands are still present and the output waveform does not look as similar as the input waveform in comparison with the waveform which uses an output filter.

A possible psychovisual explanation for this preference is that when the viewer knows that a pixel matrix has been used in the system, he wants to judge the actual smallness of the pixels instead of being in doubt about the pixel size used.

Similar behaviour has occurred in the beginning of broadcast television. Some television set manufacturers tried to improve the picture quality by reducing the visibility of the raster lines. Vertical spot wobble, elliptical spot shapes or optical means such as lenticular screens have been tried for this purpose. The appreciation of these measures was negative. For most viewers the visibility of the raster lines at close viewing distance was proof that the resolution of the monitor tube was at least good enough to present the individual raster lines.

The controversy about the application of output filtration of digitized images becomes even stronger when a picture processing algorithm such as zoom is applied. Some viewers prefer to see the pixel boundaries even though the pixels may be very coarse at high zoom factors. Other viewers prefer the "bilinear interpolated zoom" where digital two-dimensional filtering is applied to remove the "blocky-ness" of the zoomed image.

8.9. Conclusions.

If a band-limited signal is sampled at a frequency higher than the Nyquist rate ($f_s > 2B$), it can be completely reconstructed from the sampled wave.

Input filtering before sampling is needed to assure a band-limited signal. Output filtering, or reconstruction filtering, removes the higher order spectral components of the sampled signal.

Practical filters are not ideal and this causes partially overlapping sidebands or "aliasing".

The most obvious effect of aliasing on video waveforms is a certain amount of "edge busyness" in the image if an asynchronous sampling clock is used.

If the sampling clock is locked to the horizontal scan frequency of the TV, the aliasing distortion becomes stationary and is subjectively much less disturbing.

Ideal image subtraction would remove all stationary (= reproducible in each image) aliasing distortion components. Slight instabilities of the sampling clock, however, reintroduce disturbances in the subtracted images at those positions where high contrast gradients are present in the unsubtracted image. Sampling clock stability is of crucial importance.

Although the image information in the video signal is band-limited by various causes (see chapter 10), the TV noise can have a much wider frequency spectrum. Allasing of this TV noise must be prevented by the application of a suitable video lowpass filter.

Omission of the reconstruction filter at the output of the sampled system causes the pixelization in the image to become visible.

Although the signal waveform is then less matched to the input signal, some viewers prefer this type of image display.

References.

9. THE INFLUENCE OF DIGITAL QUANTIZATION.

9.1. Introduction.

Digital signal processing and storage systems have the outstanding advantage that once a signal is presented to such a system, no unintentional signal degradation will occur, while in analogue systems signal distortion and S/N degradation will always be present to a certain degree.

But in order to code a continuous signal, we must start by quantizing it into discrete steps of amplitude. This quantization is an irreversible process as the instantaneous values of the continuous signal can never be restored exactly. We have an artificially-introduced quantization noise which can be reduced however to any desired degree by choosing fine enough quantum steps, i.e. generating enough bits per sample.

The requirements for the quantization accuracy of a video signal can be derived from two different points of view: (1) the quantization noise must be negligibly small with respect to the fluctuation noise already present in the input signal or, (2) the quantization noise must be subjectively invisible in the displayed image. Depending on the application area, criterion (1) or (2) will be the most severe. Both aspects will be considered in this chapter with the purpose of determining the required digital hardware configuration for a DSA system.

9.2. S/N degradation due to quantization.

When a continuous input is quantized with equal spacing, \( a \), between the representative levels, the quantization error \( e(t) \) is bounded by \(-0.5a \leq e(t) \leq 0.5a\). It is usually assumed that \( e(t) \) has a uniform probability distribution and the mean squared value of \( e(t) \) will then be (Carlson, 1968):

\[
\sigma_e^2 = \frac{1}{2} \int_{-a}^{a} e^2 \, de = \frac{a^2}{12} \tag{9.1}
\]

If the fluctuation noise contained in the input signal has an r.m.s. amplitude \( v_{n, in} \), the average increase in noise due to the quantization follows from:

\[
\frac{v_{n, out}}{v_{n, in}} = \left[ 1 + \frac{\frac{1}{12} \frac{a}{2v_{n, in}}}{1 + \frac{\frac{1}{12} \frac{a}{2v_{n, in}}}} \right] \tag{9.2}
\]

Results shown in Fig.9.1 show that up to \( a = v_{n, in} \) no appreciable noise performance degradation occurs, and even \( a = 2v_{n, in} \) might be usable as the increase in noise voltage is only 15% in that case. These preliminary conclusions must be used carefully as the above calculations do not take into account the typical characteristics of video waveforms. Especially the assumption of a uniform probability distribution of \( e(t) \) might be true on average for a whole image, but can be false in a certain region within an image. Some images may contain regions of nearly-uniform video amplitude. With coarse quantization \( (a \gg v_{n, in}) \) such a region may show either a fluctuation between \( e = 0.5a \) and \( e = -0.5a \) within that region if the uniform video level is at a decision threshold; or it may show a uniform \( e = 0 \) if the video level is just in the middle between two decision levels. Therefore we have to consider the subjective effect of the quantization errors in a video signal.

9.3. Subjective effects of quantization in unsubtracted video images; contouring.

If the criterion of the preceding section were the only one that determines the required bit-depth of the digitization, an electronically-generated test pattern with virtually infinite S/N would theoretically require an infinite bit-depth. This is not the case as the ultimate decision instrument as regards image quality is the human eye, which has only a limited contrast detection capability; beyond a certain digitization accuracy the eye no longer appreciates the improvement.

Several investigators have carried out subjective measurements in order to determine a sensible performance target. As is usual with subjective measurements, the outcome of these measurements is rather vague; bit depths of 6, 7 and 8 are reported as being acceptable. Of course the conditions in which the subjective tests are carried out have a big influence on the results. An important viewing parameter is for instance the amount of ambient illumination, as small contrast differences may be wiped out by the ambient light. Another important parameter is the type of image which is used for testing the visibility of the digitization in an image. In this respect, a busy scene with...
a lot of detail will mask the visibility of quantization errors. Not surprisingly it has been found that the simplest image, like a uniformly illuminated area, represents the worst-case situation for visibility of quantization errors.

When such a scene is presented to a TV camera, the corresponding video signal that is read out from the camera is never perfectly flat, but always shows some degree of shading. The video signal will be at its maximum in the middle of the image and falls off somewhat towards the edges. The causes of this effect are (1) vignetting of the optics and (2) the spatially dependent read-out efficiency of the camera tube. If a video signal containing such “vignetting” is too-coarsely digitized, it will be displayed on the TV monitor as a recognizable pattern of stepwise-increased luminance towards the middle of the image. Concentric isoluminance regions are more-or-less circular in the image and appear like the annual rings in the cross-section of a tree. This effect is called "contouring" and although most easily recognized in flat illuminated images, more complex images will usually also contain flat areas with visible contouring if the applied bit-depth is not sufficient.

In addition to subjective tests with digitized video signals and video equipment, it is interesting to make a theoretical forecast of the bit-depth requirement based on the sensitivity of the human eye to just perceptible luminance differences.

For a large range of luminance levels the Weber-Fechner law holds, which states that for large area contrast (not limited by MTF deficiency), the smallest visible difference is a certain percentage of the luminance itself; $\Delta L/L = \text{constant}$. It is generally accepted that this sensitivity threshold is at about 2% luminance. In radiological terminology this is equivalent to the statement that the minimum noticeable density difference is $\Delta D_{\text{min}} = \log 1.02 = 0.0086$. Relating this sensitivity figure to bit-depth requirements involves the TV monitor transfer characteristic and the ambient light level. As a TV monitor tube is a nonlinear device, a fixed quantization error voltage in the video waveform supplied to the monitor has a different impact at different video voltage levels. The light output of the tube is given by:

$$L = k \cdot v^\gamma$$

(9.3)

where $k$ is a contrast, $\gamma$ is the "gamma" of the tube and $v$ is the video voltage.

If an ambient light level $L_0$ is present, then relative contrast of a quantization error $\Delta v$ is:
For simplicity we take $k = 1$ and $0 < v < 1$ in the evaluation of Eq(9.4). Common gamma values of TV monitor tubes are reported to be in the range $\gamma = 2.2 - 2.8$; for the calculations we take $\gamma = 2.5$. Fig.9.2 shows the results of the evaluation with $L_0$ as a parameter. It appears that maximum visibility $\frac{\Delta L}{L}$ of a quantizing step $\Delta v$ occurs at low video levels, thus in the darker grey regions. This is in accordance with practical observations (Note: For the same reason other "noises" then quantizing noise are also most visible in this same region).

A very common viewing condition is with $L_0 = 3\%$ of the video peak brightness. From Fig.9.2 we can deduce that for this ambient light level the maximum sensitivity occurs at $v = 0.25$. The perceived contrast is then:

$$
\frac{\Delta L}{L} = 5 \frac{\Delta v}{V_{\text{max}}} \tag{9.5}
$$

By substituting the sensitivity figure of the human eye $\frac{\Delta L}{L} = 0.02$, it follows that the quantizing step $\Delta v$ has to meet:

$$
\frac{\Delta v}{V_{\text{max}}} = \frac{1}{8} \times 0.02 = \frac{1}{256} \tag{9.6}
$$

This means that according to these calculations an 8-bit accuracy (256 levels) is needed for digitized video representations.

Extensive measurements by Devereux(1974) have proved indeed that critical pictures required 8 bits per sample in order to reduce the impairment below the just-perceptible level. During Devereux's experiments synchronization pulses and headroom for colour subcarrier overswing were included in the range of the AD converter input; thus the actual black-to-peak white range contained only about 150 levels instead of 256. As mentioned by Devereux, with this digitization accuracy most picture sources will provide sufficient inherent random noise ($v_{\text{in}} > v_{\text{max}}/150$) to eliminate coherent patterning effects such as contouring; the noise acts like a "dither" signal that masks the contouring.
So far in this section only the required accuracy of unsubtracted digitized images has been discussed. For the DSA technique the above considerations hold with regard to the image finally displayed. The digital subtraction and subsequent contrast gain, however, require a higher accuracy prior to subtraction. The presence of a logarithmic conversion in the chain complicates the discussion of the ADC requirements, so the implications of the log conversion will be analyzed separately.

9.4. Log conversion by analogue methods.

As explained earlier (see section 3.4) logarithmic signal processing is needed to obtain DSA images in which vessel detail visibility is not hampered by overlying bone and tissue. The realization of the required logarithmic transfer characteristic can be achieved by: (1) an analogue logarithmic amplifier or by (2) a Look-Up-Table (LUT), with a logarithmic transfer characteristic, in the digital domain. In our early experiments we used the first approach. An inexpensive integrated circuit exist which can do the job, but in the course of these experiments serious doubts arose about the d.c. stability of the circuit. The digital solution with LUTs is ideal as regards reproducibility, and moreover it provides a flexible means of changing the shape of the transfer characteristic. Depending on the application use of the DSA processor, various transfer characteristics may be needed, which can easily be loaded into the RAM-LUT by software control. Because of these advantages newer DSA processors will contain LUTs for log conversion.

Before discussing the digital log conversion method in detail, it is very instructive to discuss first a configuration where an analogue logarithmic amplifier is followed by an analogue-to-digital converter (ADC).

The first fact one has to face is that a logarithmic transfer characteristic is possible only for a limited input signal range. The ideal log curve approaches infinite slope (or differential gain $dv/dv$) when the input signal approaches zero. This is impossible to meet in practice. For this reason one has to accept a log transfer characteristic which is valid only in a limited Dynamic Range ($DR = V_{i,\text{max}}/V_{i,\text{min}}$). This is no serious limitation as $DR$ seldom exceeds 20 in DSA practice; $DR = 100$ is already enough. As soon as the concept of a limited dynamic range is accepted, the actual meaning of a quantization step after the log conversion can be discussed. Each step now represents a certain relative increase or decrease of signal level.

For small differences one can express this as a percentage per quantum step, but when large differences must also be dealt with, an expression in decibels (dB) per quantum step is more appropriate. So as an example, when we want to cover a dynamic range of 40 dB ($DR = 100$) with 8-bit digital words, each quantum step represents $40 \times 2^{56} = 0.16$ dB or 1.8%. Several other combinations of ADC accuracy and dynamic range $DR$ are listed in the following table.

<table>
<thead>
<tr>
<th>ADC accuracy</th>
<th>$DR = 100$ (40 dB)</th>
<th>$DR = 50$ (34 dB)</th>
<th>$DR = 25$ (28 dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>12 bit</td>
<td>0.0098 dB/step</td>
<td>0.0083 dB/step</td>
<td>0.0068 dB/step</td>
</tr>
<tr>
<td>11</td>
<td>0.019 dB/step</td>
<td>0.017 dB/step</td>
<td>0.014 dB/step</td>
</tr>
<tr>
<td>10</td>
<td>0.039 dB/step</td>
<td>0.033 dB/step</td>
<td>0.028 dB/step</td>
</tr>
<tr>
<td>9</td>
<td>0.078 dB/step</td>
<td>0.066 dB/step</td>
<td>0.055 dB/step</td>
</tr>
<tr>
<td>8</td>
<td>0.16 dB/step</td>
<td>0.130 dB/step</td>
<td>0.11 dB/step</td>
</tr>
<tr>
<td>7</td>
<td>0.31 dB/step</td>
<td>0.27 dB/step</td>
<td>0.22 dB/step</td>
</tr>
<tr>
<td>6</td>
<td>0.62 dB/step</td>
<td>0.53 dB/step</td>
<td>0.44 dB/step</td>
</tr>
</tbody>
</table>

From this table it can be concluded that the influence of the dynamic range selection is rather small. If the goal of our imaging system is to detect small contrast differences of the order of 1% of the input signal, it appears from the table that a 9-bit ADC accuracy is marginal for this purpose; at least 10-bit should be used. This performance seems inferior to that of a linear system. Without log conversion a 9-bit ADC can discriminate 0.28% contrast in the upper most part of the signal range (discriminate between 511 and 512). At lower signal levels the discrimination capability becomes increasingly worse with linear digitization; for instance at 1/20 of the maximum signal level the discrimination capability is only 4% (discriminate between 5 en 26). The application of a logarithmic amplifier preceding the ADC has the effect that the accuracy performance of the ADC is more uniformly spread out over the signal range, but together with this trade-off, the performance in the high signal range is compromised.

Another way of looking at this effect is to consider the actual differential gain of the logamp circuitry. If we have an ADC with a 1V input signal range, a logamp producing this output voltage range with a 1V input signal range must have the transfer characteristic (see...
also Fig. 9.3):

\[ V_u = \frac{1}{\log DR} \log v_i + 1 \]  

with \( \frac{1}{DR} < v_i < 1 \)  \hspace{1cm} (9.7)

The differential gain of this circuit is:

\[ \frac{dv_u}{dv_i} = \frac{1}{\log DR \log 10} \frac{1}{v_i} \]  

For 34 dB range (DR = 50):

\[ \frac{dv_u}{dv_i} = 0.256 \frac{1}{v_i} \]  

As Eq(9.9) shows, and can also be concluded from Fig.9.3, the slope or differential gain of the configuration is less than one for the largest part of the input signal range \( (0.256 < v_i < 1) \).

This means that most of the small contrast differences which we want to detect are being compressed before the digitization takes place. This seems not to be the optimal approach and together with the earlier mentioned problem of d.c. drift stability, this is sufficient reason to reject such a solution.

9.5. Required ADC accuracy.

After the decision to put the log conversion in the digital domain after the ADC, the question of the requirement of the ADC's accuracy can be answered.

An accuracy as high as possible is needed because this opens the possibility of using high contrast-gain after subtraction. On the other hand, a quantization step size which is smaller than 1-2 times the lowest r.m.s. noise amplitude in the video signal is overkill as has been shown in section 9.2.

The lowest r.m.s. noise in the video signal occurs when both high exposure dose conditions (say infinitely high for the sake of simplicity) and low video levels are present. In that case the only noise sources are the thermal noise \( n_{th} \) and the preamp noise \( n_{prec} \) of the TV camera; see Fig.7.3.

When the peak video current of the camera is set at \( I_g = 20000 \text{nA} \), the lowest r.m.s. noise level is at \( \text{inA/1000nA} = 1/1000 \) of the maximum range. An ADC of 9 or 10-bit accuracy is required to quantize this signal appropriately. If the peak video current of the camera is set at \( I_g = 2000 \text{nA} \), an ADC of 10 or 11-bit accuracy would be needed to comply with the lower relative noise level.

In practice the applied X-ray exposure dose is not infinitely high and additional X-ray quantum noise will reduce the accuracy requirements for the ADC, except for the very lowest video levels where the preamp noise still dominates (see later in section 11.5). Due to the limited dynamic range of radiographic images containing scatter, the lowest region of the video range will not be penetrated in practice, provided that good exposure control is used.

In conclusion it can be stated that 9 or 10-bit of ADC accuracy is sufficient with typical state-of-the-art TV cameras which run at standard TV rates. High-resolution TV cameras have a much higher preamp noise contribution (see section 7.5) and reduced ADC accuracy seems admissible in that case.

9.6. Digital log conversion; LUT size.

Digital log conversion at the speed dictated by the video sampling frequency can be executed by very fast RAM circuits in a look-up table configuration. The video input words are taken as the address input of the RAMs; the content of the RAM memory at each address location forms the output of the LUT. Input word size and output word size can be independently chosen in a design.

With a given limited word length as input to the LUT, any arbitrary shaped transfer characteristic will theoretically require an infinite word length as output to be able to code any output level. In practice of course a limited output word length must be used and this constitutes an additional quantizing effect on top of the quantizing effect caused by the AD converter. This additional quantizing effect may have a serious influence on the system performance and therefore the LUT size requirements have to be carefully analyzed.

If the optimum contrast detection capability of the ADC is to be preserved the dB/step or %/step figure at the output of the LUT must at least comply with the optimum contrast detection capability of the ADC in the uppermost part of the signal range. In other words, the differential gain figure in the uppermost part of the signal range must be equal to one. This means that a change of 1 LSB (least significant bit) of the input word (length \( n_1 \) bits) must result in a 1 LSB change of the output word (length \( n_2 \) bits) in the uppermost part of
the signal range. Further down in the signal range the change of 1 LSB of the input word results in more levels of change at the output of the LUT; there the differential gain is more than one.

When the dB/step figure and the slope of the log curve have been established in the way just described, the dynamic range obtained depends on the output word size \( n_2 \) of the LUT. As an example: a 10-bit ADC is capable of 0.098V/step or 0.0085 dB/step. If the output word size of the LUT is also 10 bit, the total range covered is \( 1024 \times 0.0085 \text{ dB} = 8.7 \text{ dB} = 2.7:1 \). This dynamic range is too small, of course, and a larger LUT is required. Various alternative combinations are presented in the following table.

Table 9.2: Dynamic Range (DR) as a function of ADC accuracy \((n_1)\) and LUT output \((n_2)\).

<table>
<thead>
<tr>
<th>ADC accuracy ( n_1 )</th>
<th>ADC contrast performance in upper part of the range</th>
<th>( n_2 = n_1 )</th>
<th>( n_2 = n_1 + 1 )</th>
<th>( n_2 = n_1 + 2 )</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>dB/step</td>
<td>8/step</td>
<td>DB</td>
<td>DR</td>
</tr>
<tr>
<td>11</td>
<td>0.0042</td>
<td>0.049</td>
<td>8.7</td>
<td>2.7</td>
</tr>
<tr>
<td>10</td>
<td>0.0095</td>
<td>0.098</td>
<td>8.7</td>
<td>2.7</td>
</tr>
<tr>
<td>9</td>
<td>0.017</td>
<td>0.19</td>
<td>8.7</td>
<td>2.7</td>
</tr>
<tr>
<td>8</td>
<td>0.034</td>
<td>0.39</td>
<td>8.7</td>
<td>2.7</td>
</tr>
</tbody>
</table>

Table 9.2 shows that a dynamic range of 54:1 is obtained when the output word size \( n_2 \) of the LUT has two more bits than the input word size \( n_1 \). This dynamic range DR = 54:1 is more than sufficient for radiographic images. One bit less \((n_2 = n_1 + 1)\) gives a hardly acceptable dynamic range DR = 7.4:1, so an appropriate design requirement is \( n_2 \geq n_1 + 2 \).

If the dynamic range of the image exceeds 54:1, or if a too-low exposure level is used, the lowest video levels will be clipped at zero output by the log conversion; see Fig.9.3. With DR = 54 this clip range is negligibly small, and moreover will never be entered in practice due to suitable exposure control and the limited dynamic range (usually < 10:1) in radiographic images.

9.7. Display range selection; enhancement.

After subtraction of two digitized images, an extra bit is generated that indicates the polarity of the difference signal (accuracy is in no way improved by this extra bit). So if we start with 12-bit LUT outputs, a 13-bit subtraction result is produced.

During display of the difference images it makes no sense to produce more than 256 levels in the video range between black and white, as the human eye is not able to discern smaller quantization steps (as discussed in section 9.3). Therefore we can use an 8-bit Digital-to-Analogue converter (DAC) and an appropriate choice has to be made as to which of the available bits to send to the DAC.

During DSA, the images to be subtracted are more or less alike, so in practice only small amplitude differences will arise. In a sign and magnitude representation this means that the most significant bits (MSBs) of the magnitude word will remain constant; the information is confined to the least significant bits (LSBs) of the word. This is also the case with the 2's complement representation usually employed. Contrast enhancement, which is the key factor of DSA imaging, is now as simple as bitshifting the real information containing bits towards the MSB inputs of the DA converter; see Fig.9.4.

The most appropriate number of bit shifts to apply depends on the magnitude of the difference signal to be expected. In DSA practice with intravenous injections, 3 to 5 bitshifts usually produce the most useful diagnostic images. Higher numbers are inappropriate because the latitude of the display range would become so small that: (1) large vessel contrasts will not fit within the range, (2) displayed noise becomes excessively disturbing in practice, and (3) the instability of most X-ray generators poses serious problems.

Some discussion of these points is in order for a better understanding of the basic limitations of the DSA technique. If we have chosen the 54:1 (or 34.7 db) dynamic range as advised in the previous section, then after subtraction the difference signal covers a range of \( \pm 34.7 \text{ dB} \). If we now apply a display range selection with 5 bitshifts, the display range covers \( \pm 34.7 \times 32 = \pm 1 \text{ dB} \).

When large vessels such as the cardiac chambers and the aortic root are opacified by the arrival of contrast medium, the iodine projected thickness is so large (a few cm) that the concomitant image contrast is usually larger than a 1 dB difference or 12%. As signal clipping is undesirable, an enhancement of 5 bitshifts is usually too much for visualization of these vessels.
The problem with X-ray generation has to do with unintentional drifts of the X-ray source output during a DSA run. This drift may consume a substantial part of the display range. Even if the kV supply and filament current of the X-ray tube are kept perfectly stable during a run, the internally-generated heat in the X-ray tube may still cause a slow drift in X-ray output of up to 10% (= 0.83 dB), which is indeed nearly equal to the available 1 dB range between mid-grey and the clip levels at peak white or black. Additional drift effects may also be caused by the build-up lag of the TV camera tube.

Contrast enhancement inevitably also enhances the inherent noise in the image. So far we have considered only the noise contribution of the TV camera quantitatively. Signal-to-noise ratios of a TV camera as given by Fig. 7.4 are translated into a noise fraction of the video level and presented in column 2 of Table 9.3. X-ray quantum noise is not considered yet.

Columns 3 to 6 of Table 9.3 show how the TV camera noise is displayed when display ranges of ±1 dB and ±2 dB are selected for viewing. Due to the powerwise addition of the noise when two images are subtracted, \( \sqrt{2} \) factor is involved in the calculations. As an example: a ±1 dB display range has \( 2 : 256 = 0.0078 \) dB/step or 0.09%/step. For 25% video level the combined rms noise of two images is \( \sqrt{2} \times 0.28\% = 0.39\% \) (see column 2 of Table 9.3) or 0.39\% : 0.09 = 4.40 steps in the display.

As the total display range has 256 steps (8-bit DAC), the 4.40 steps make 4.40 : 256 = 1.72\% of the total display range.

(Peak-to-peak noise is of course much more, related by the so-called "crest factor" which might be taken as 4-6 for Gaussian noise).

Table 9.3: TV camera noise in the display of enhanced subtractions.

<table>
<thead>
<tr>
<th>Video level</th>
<th>rms noise in ±2 dB steps out of 256</th>
<th>Fraction of display range</th>
<th>rms noise in ±1 dB steps out of 256</th>
<th>Fraction of display range</th>
</tr>
</thead>
<tbody>
<tr>
<td>100%</td>
<td>0.92</td>
<td>0.32%</td>
<td>1.64</td>
<td>0.64%</td>
</tr>
<tr>
<td>50%</td>
<td>1.30</td>
<td>0.51%</td>
<td>2.60</td>
<td>1.02%</td>
</tr>
<tr>
<td>25%</td>
<td>2.20</td>
<td>0.86%</td>
<td>4.40</td>
<td>1.72%</td>
</tr>
<tr>
<td>10%</td>
<td>4.79</td>
<td>1.87%</td>
<td>9.58</td>
<td>3.64%</td>
</tr>
<tr>
<td>5%</td>
<td>9.03</td>
<td>3.53%</td>
<td>18.06</td>
<td>7.06%</td>
</tr>
</tbody>
</table>
Table 9.3 shows that the rms noise in the display increases with the enhancement chosen and also increases for lower video levels, but nowhere becomes excessive in the situations presented. This statement has been validated by experiments where the input to the TV camera was formed by uniform illumination from a lamp instead from the X-ray image intensifier.

The practical observation that the enhancement must be confined to 3 or 5 bitshifts (display range 1-4 dB) for S/N reasons leads to the conclusion that X-ray quantum noise must be the limiting factor in practice. This subject will be dealt with in more detail in chapter 11.

When trying to find the best enhancement factor, 3, 4 or 5 bitshifts, it turns out that this choice is too limited because the sudden increase or decrease by a factor of two in displayed contrast is too coarse to allow the optimum image quality to be found. For this reason, in addition to the bitshift preselection, the user is provided with an analogue gain control. This gives him the possibility to select any enhancement factor (window width) that will suit the purpose of optimizing the quality of the displayed image. In addition to the analogue gain control an analogue level (brightness) control is also provided for optimizing the displayed image quality. Any level drifts caused by X-ray output fluctuations, as discussed earlier, are readily compensated during the process of setting the optimum display level by the user.

The capability of simultaneous operation of both controls (gain and level) by the user, using slider-type potentiometers, has proved to be a very powerful method of obtaining the optimum image quality very quickly. Despite the fame of digital methods for image enhancement procedures, implementation of these two controls (gain and level) by digital methods turned out to be less simple and responsive than the analogue method.

One last point that deserves attention in combination with bitshift selection is the possibility of overflow. When unexpectedly large image differences show up due to massive vessel contrast (large vessels) or patient movement, the amplitude of those difference signals might be larger than the selected display range. If no additional measures are taken, wrap-around of those high amplitude signals will occur, showing up as black over-range areas in the peak white part of the image, and white over-range areas in the black part.

Additional circuitry is therefore needed to clip those over-range signals at the peak white and black borders of the selected display range.


Quantization noise must be small with respect to the fluctuation noise already present in the input signal. If an unlimited X-ray dose could be used, only the TV noise would remain. Quantization noise is negligible in such a case if the quantization step is smaller than 1.5 times the r.m.s. noise voltage. For a S/N = 1000 TV system 9 or 10 bits ADC accuracy is then needed. For a S/N = 200 TV system a 7 or 8-bit accuracy is sufficient to achieve this goal.

A second criterion for digitized images has to do with the limited contrast detection capability of the human eye. The gamma of the TV monitor tube and the ambient light level during viewing also play a role in this respect. Under typical circumstances no more than 256 grey levels (8-bit) need to be displayed.

For unsubtracted images a 10-bit ADC is thus an excessive. In the DSA application, where two images are subtracted and the difference signal is enhanced for display, a higher ADC accuracy is justified; here the maximum useful ADC accuracy is determined by the S/N ratio as mentioned above.

Log conversion can in practice be accomplished only over a limited dynamic range DR. For DSA applications DR = 50 (34 dB) is amply sufficient.

Log conversion with an analogue circuit necessarily compresses the upper part of the signal range and therefore puts higher demands upon the ADC accuracy in this part of the signal range. Application of a Look-Up Table (LUT) for log conversion does not have this disadvantage. The full benefit of the ADC accuracy used is available provided that the proper curve is programmed. The signal range need not be compressed anywhere; only range expansion must be used (differential gain > 1).

Analysis of the LUT size requirement shows that in general the number of output bits must be two more than the input bits applied. So a 10-bit ADC needs a 12 bit LUT output in order to maintain the accuracy of the ADC.

The gain factor after subtraction may range from 8x to 32x in practice. The actual contrast enhancement factor depends on this gain, but also upon the slope of the log conversion used. Therefore a more general figure for enhancement setting (window width) is the display range selection expressed in dB.
Enhancement greater than $+1 \text{ dB}$ display range becomes unpractical as unintentional image differences such as noise, movement artefacts (see chapter 12) and uniform level shifts caused by X-ray instability will become intolerable.

Due to the log conversion the TV noise is more visible in the lower part of the signal range, but thanks to the restricted enhancement factor mentioned above, this noise is not the limiting factor of the DSA technique (See photograph P.14).

References.


10. SPATIAL RESOLUTION CAPABILITY.

10.1. Introduction.

The discussions of the spatial resolution performance of a DSA system are often focused on the matrix size of the digital video memories. The possibilities usually considered are $256^2$, $512^2$ and $1024^2$ matrices. The square format of the matrix more-or-less complies with the circular images detected by the image intensifier. The actual numbers 256, 512 and 1024 are powers of two which suit the nature of the digital hardware. The square pixel matrix is fed only with useful image information from within a circular area which has $\pi/4 = 78\%$ of the area of the square. The hardware of the video memories could be designed with correspondingly fewer memory places, but addressing such a memory would be so much more complicated that this effort to save memory chips is not worthwhile. Moreover such a video memory design would preclude easy access to vertically-neighbouring pixels, which is needed for some image processing functions.

High-resolution images naturally call for a large matrix. On the other hand the choice of the matrix size has important economic consequences as: (1) a large matrix requires video frame stores of high capacity, and (2) the pixel rate increases with matrix size. Both factors will increase the price of a digital imaging system considerably. Overspecification of the matrix size has thus to be avoided and the judicious choice is the one which just suits the need.

In addition to the pixel matrix size, several other system parameters such as (1) focal spot size of the X-ray tube, (2) geometric enlargement factor, (3) image-intensifier resolution, (4) resolution of the optics, (5) resolution of the TV camera tube, (6) video bandwidth, and (7) resolution of the TV display monitor, each have their influence on the overall spatial resolution capability.

A well-proven technique for combining the various resolution degrading factors is the multiplication of the respective modulation transfer functions (MTFs) of each of the factors involved. This chapter summarizes some basically well-known radiological MTF factors (focus and geometry) and combines them with lesser-known MTF factors in order to appreciate the influence of the matrix size upon the overall performance capability.

In practice the X-ray dose applied per image also has a decisive role in the spatial resolution performance of a complete system. As explained in chapter 4, the Rose model predicts a quantum noise
In this chapter the quantum noise is neglected for the time being, which means that the spatial resolution capability is discussed for the condition that an unlimited dose would be available. To what degree this spatial resolution capability will result in actual spatial resolution performance depends on the dose level, and will be discussed in chapter 11.

10.2. Focal spot blurring.

The ideal size of the focal spot of an X-ray tube is infinitesimally small. This is an impossible situation in practice as about 99% of the energy supplied to the tube is converted into heat at the focal spot location. As the tube-loading figure can be as high as 100 kW, a certain area must be allowed for dissipating the heat without the risk of melting the anode material. Table 10.1 shows a range of currently available focus sizes with their corresponding loadability.

**Table 10.1: Focal spot loadability data of available X-ray tubes.**

<table>
<thead>
<tr>
<th>Focus size $\text{mm}^2$</th>
<th>Power rating</th>
</tr>
</thead>
<tbody>
<tr>
<td>$1.2 \times 1.2$</td>
<td>100 kW</td>
</tr>
<tr>
<td>$0.6 \times 0.6$</td>
<td>30</td>
</tr>
<tr>
<td>$0.3 \times 0.3$</td>
<td>12</td>
</tr>
<tr>
<td>$0.15 \times 0.15$</td>
<td>3</td>
</tr>
</tbody>
</table>

If we assume a square focal spot with uniform intensity distribution, a pinhole object in the object plane between focus and receptor transforms into a square image of the focus on the detector. The size of this image on the detector is $(M-1)a$, where $a$ is the focus size and $M$ is the geometrical enlargement factor; see Fig. 10.1-a and Fig. 10.1-b.

An equally large image on the detector would have resulted if the focus had been infinitesimally small, and the object would have had a square of linear dimension $(1-1/M)a$; see Fig. 10.1-c. So it can be concluded that the influence of the focal spot dimension is identical to the transformation of a delta function input image into a square shaped output image of dimension $(1-1/M)a$. The MTF of such a transfer function is:

$$\text{MTF}_{\text{FOC}} = \frac{\sin \pi f(1-1/M)a}{\pi f(1-1/M)a}$$

**Fig. 10.1-a** Geometric enlargement 1.2
**Fig. 10.1-b** Pinhole focus image on receptor
**Fig. 10.1-c** Equivalent focus image in object
This MTF function (see Fig.10.2) shows zero crossings and polarity inversions (spurious resolution) which can also be found in practice by imaging the so-called "star test pattern". The zero crossover frequency of the star pattern usually corresponds to an "equivalent" spot size somewhat larger than the rated nominal value. The reason for this is that the radiation distribution of the spot is not uniform as assumed, but has a double-humped so-called "camelback" distribution. The influence of this on the MTF curve has been explained by Pfeiler (1973); the crossover frequency shifts to a lower frequency but otherwise the shape of the MTF curve up to the first crossover remains very similar to that described by Eq(10.1). For this reason Eq(10.1) is generally used in resolution calculations, but a spot size 50% higher than the nominal is substituted. This practice is within the NEMA tolerance limit for focal spot standards. Pinhole exposures of the focal spot will give an indication of spot shape and its intensity distribution, but straightforward conclusions about either the nominal spot size or the equivalent spot size can not easily be made.

In clinical angiographic practice where \( d_1 + d_2 = 100-120 \text{ cm} \) and \( d_1 = 8-35 \text{ cm} \), the geometric enlargement factor is in the range \( M = 1.1 - 1.5 \). Fig.10.3 shows how dramatically the MTF deteriorates as the magnification increases; for a 1.2 mm focus and \( M = 1.25 \), which very often occur in practice, the MTF is at 50% for 2 lp/mm. A larger magnification such as \( M = 2 \) is sometimes advocated because magnification reduces the spatial frequency which the detector has to resolve; the detector is able to do a better job in exchange for some loss in effective image field size. A second advantage of magnification is that an air gap between patient and detector is provided, which helps in reducing the scatter radiation contribution. An optimum magnification factor can be found by balancing focal spot blurring and detector blurring, but it usually turns out that \( M = 2 \) is beneficial only for 0.3 x 0.3 mm\(^2\) spot size or smaller.

More detailed analysis of X-ray tube designs will reveal that the square equivalent spot shape is obtained in the center of the image only; the spot dimension varies for offcentre radiation; it becomes smaller when moving to the anode, and larger when moving to the cathode side (Fenner, 19xx). Rao (1973) and Chaney (1974) have found that the equivalent spot size also depends on the exposure factors mA and kVp.

Strictly speaking, both effects are important enough to be taken into account, but for simplicity the calculations in this chapter will assume a stable uniform spot size.
10.3. MTF of the image intensifier.

Very often the spatial resolution performance of an Image Intensi­
er (II) is described by a single parameter; the so-called "limiting re­
solution". This single figure has a very limited useful meaning in a
systems approach for several reasons.

(1) Different opinions exist about the definition of the limiting re­
solution point; 1%, 2%, 4% and 5% MTF levels are used as a criter­
ion.

(2) Different opinions exist about the normalizing point of MTF =
100%. The internal scatter and glare degradations which cause the
so called "Low Frequency Drop" (LFD) in the MTF are sometimes com­
pletely ignored by starting the MTF curve at a certain (non-zero)
low frequency. The latest improvements in II technology, which
make use of titanium input windows and fiber-coupled output win­
dows, have achieved a drastic improvement in image quality. This
improvement shows up in the MTF curve only when the low frequency
drop region is included.

(3) Finally, even if there were agreement about the two preceding
points concerning MTF definitions, the limiting resolution figure
still represents only one point of the whole MTF curve. Other
parameters such as the steepness of descent and the LFD are equal­
ly important in contributing to the spatial resolution perfor­
ance. (see also Scheid,1980).

Physical aspects, important for the design of high quality X-ray
image intensifiers are discussed by: Teves(1952), Niklas(1964), Franz

State of the art performance is represented by the Philips 14/10/6.5"
image intensifier. The MTF of this tube is shown in Fig.10.4. Data are
provided by Rijpert of Philips (personal communication), but also Lin
(1982) has published his measurements of this tube.

The measured MTF curve shown in Fig.10.4 includes the effect of the
optics which relay the output image of the II to the input of the TV
camera tube. The optics contribute to the LFD, but additionally hardly
degrade the high frequency part of the MTF.

Johnson(1972) has pointed out that many MTFs fit remarkably with
the expression:

\[ MTF = \exp\left(\frac{-f}{k}\right)^n \]  

(10.2)
where $f_\text{c}$ is a frequency constant and $n$ is called the "device index". He also indicated a graphical method with which the parameters $f_\text{c}$ and $n$ can be read immediately.

Introduction of LFD at zero frequency gives the simplified expression (see also Arakami, 1982):

$$\text{MTF} = (1 - \text{LFD}) \exp \left( -\frac{f_\text{c}}{f_\text{c}} \right)$$

(10.3)

Curve fitting Eq(10.3) to Fig.10.4 by Johnson's graphical method produces the data shown in Table 10.2.

Table 10.2: MTF parameters of 14" II + optics.

<table>
<thead>
<tr>
<th>Field size</th>
<th>LFD</th>
<th>$f_\text{c}$</th>
<th>$n$</th>
</tr>
</thead>
<tbody>
<tr>
<td>14&quot;</td>
<td>12%</td>
<td>1.2 lp/mm</td>
<td>1.55</td>
</tr>
<tr>
<td>10&quot;</td>
<td>12%</td>
<td>1.4</td>
<td>1.48</td>
</tr>
<tr>
<td>6.5&quot;</td>
<td>12%</td>
<td>1.6</td>
<td>1.48</td>
</tr>
</tbody>
</table>

Similar procedures applied to 9" image intensifiers, produce data as shown in Table 10.3.

Table 10.3: MTF parameters of 9" II + optics.

<table>
<thead>
<tr>
<th>Field size</th>
<th>LFD</th>
<th>$f_\text{c}$</th>
<th>$n$</th>
</tr>
</thead>
<tbody>
<tr>
<td>9&quot;</td>
<td>19%</td>
<td>1.5 lp/mm</td>
<td>1.50</td>
</tr>
<tr>
<td>6&quot;</td>
<td>19%</td>
<td>1.7</td>
<td>1.50</td>
</tr>
</tbody>
</table>

10.4. MTF of the TV camera tube.

Camera tube MTF data as supplied by the manufacturers are usually incomplete for direct application because: (1) the LFD is usually ignored and (2) square-wave response curves are given instead of the sine-wave response curves required.

Transformation of square-wave response $R(f)$ to sine-wave response $\text{MTF}(f)$ is possible by (Coltman, 1954):

$$\text{MTF}(f) = \frac{\pi}{4} \left\{ R(f) + \frac{1}{3} R(3f) - \frac{1}{5} R(5f) + \frac{1}{7} R(7f) \right\}$$

(10.4)

This simplification is also included in Eq(10.3).

Usually the low-frequency glare spot, which is responsible for the LFD, is at least 20x larger in diameter than the nucleus of the point spread function. Except for the very lowest frequencies, the MTF expression can thus be simplified to:

$$\text{MTF} = (1 - \text{LFD}) \text{MTF}_{\text{high}}$$

(10.5)

Table 10.4: MTF parameters of TV camera tubes.

<table>
<thead>
<tr>
<th>$XQ$</th>
<th>LFD</th>
<th>$f_\text{c}$</th>
<th>$n$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1072</td>
<td>14%</td>
<td>20 lp/mm</td>
<td>1.50</td>
</tr>
<tr>
<td>1073</td>
<td>13%</td>
<td>22</td>
<td>1.30</td>
</tr>
<tr>
<td>45 XQ</td>
<td>8</td>
<td>22</td>
<td>1.53</td>
</tr>
</tbody>
</table>
The parameters of the 45 XQ tube in the table indicate that the target layer has a somewhat better intrinsic resolution than the standard 1" tubes (Franken, 1981). Most of the improved MTF characteristic of this tube is due to the larger target area; 26 mm of image diagonal instead of 16 mm are possible. An MTF curve which shows lp/diagonal instead of lp/mm would demonstrate this feature more convincingly.

The MTF parameters of Table 10.4 are derived from published MTF curves which are valid for the relatively low signal currents (<400 nA) which are usually applied. DSA applications require the ultimate signal-to-noise ratio and for this reason a much higher than standard signal current is used (I_s,max = 2000 nA; see section 7.2). This higher signal current will result in an increased effective read-spot diameter and thus a decreased MTF performance of the camera tube. This subject is discussed extensively by Franken (1978). His measurements for a particular situation and also our own exploratory measurements indicate that the MTF degradation can be transformed into a decrease of the frequency constant parameter, f_c, by about 25%.

The MTF data of TV camera tubes discussed use spatial frequencies defined in the target plane of the camera tube. Useful application of these data requires a transformation to the imaging plane of interest, which is usually the entrance plane of the II. The magnification factor involved is equal to the II field size diameter (340 mm, 240 mm and 163 mm for 14", 10" and 6.5" II mode) divided by the image diameter on the Plumbicon target (14.1 mm for 1" tubes and 24.3 mm for 45XQ).

10.5. MTF of the pixel matrix.

The sampling process in both the horizontal and the vertical direction sub-divides the image into basic picture elements or pixels. The impulse response function or point spread function of the sampling process is a square area of uniform contribution with the size of one pixel. The MTF which belongs to such a point spread function has the sinx/x shape identical to the focal spot blurring shown in Fig.10.2. In addition to this, a lowpass filter is required to prevent aliasing distortion as explained in chapter 8. In the horizontal direction this lowpass filtering is feasible by means of an electrical lowpass filter that limits the bandwidth of the video signals. In the vertical direction no means are available for applying the required lowpass filtering, and we have to live with the degrading effects resulting from it. Easily-recognisable effects are the jagged appearance of sloping lines and edges in every TV display.

In any case, with a lowpass filter or without, the information in the frequency range above half the sampling frequency must be considered...
as worthless because any frequency higher than half the sampling frequency (alternating black and white pixels) cannot possibly be represented. As regards image resolution capabilities we therefore have to employ:

\[
\text{MTF}_{\text{pix}} = \frac{\sin \pi f a}{\pi f a} \quad \text{for } f < \frac{1}{2a} \\
= 0 \quad \text{for } f \geq \frac{1}{2a} 
\]  

(10.7)

where \(a\) is the pixel size.

The pixel size to be substituted depends on the selected II field and the applicable matrix size. For instance in a 6.5" mode of the II and a 512\(^2\) matrix, the pixel size, \(a\), is 163 mm : 512 = 0.32 mm at the entrance plane of the II.

In the overall MTF picture of a complete system, we now have to examine the impact of the sharp cut-off of Eq(10.7) upon the overall performance.

10.6 Overall MTF in the detector plane.

The overall MTF in the entrance plane includes all previously discussed MTF curves, except for the focal spot blurring, as in the entrance plane, the geometrical enlargement factor is \(M = 1\). Thus:

\[
\text{MTF}_{\text{EP}} = \text{MTF}_{\text{II}} \cdot \text{MTF}_{\text{TV}} \cdot \text{MTF}_{\text{pix}}
\]  

(10.8)

A typical configuration with a 512\(^2\) matrix size, a 14" II and a XQ 1072 camera tube at high current level will result in overall curves as shown in Fig.10.6.

Low-frequency drop is not included in the curves, as it was felt that this parameter has nothing to do with the overall high frequency behaviour and its consequences for the matrix size. Instead of including LFD in the curves, this important parameter is indicated as an additional mark to the curves.

From Fig.10.6 it can be concluded that for both the 6.5" and 10" field size the MTF of the pixel matrix is better than those of the II and TV. For these field sizes the pixel matrix is thus certainly not the limiting factor.

For 14" field size the MTF of the pixel matrix is comparable to the MTF of the II, but again it is not the limiting factor as the TV is much worse in this case.

The sharp cut-off at half the sampling frequency results in the omission of the small (dashed) tail of the overall MTF curve. An appraisal of the importance of the small tail can be quantified by examining the area under the overall MTF curve. As will be explained later, the area under the MTF curve is a measure of image sharpness.

Planimetric examination of the curves in Fig.10.6 reveals that removing of the dashed tails reduces the area under the overall curves by respectively 1.4\%, 4.9\% and 5.1\% in the 6.5", 10" and 14" modes. The effect on the image sharpness will thus be negligible.

10.7 Relation of edge gradient to MTF.

It can be understood intuitively that one measure of image sharpness is the steepness or gradient of the edge-response function. For most optical effects the spatial response functions are symmetrical (no phase errors) and the first derivative of the edge-response function is a symmetrical line-spread function LSF. The LSF and MTF of a system are related by a Fourier transformation:

\[
\text{LSF}(x) = \int_{-\infty}^{\infty} \text{MTF}(f) e^{i 2\pi f x} df
\]  

(10.9)

This integral can be solved analytically for only a small category of MTF expressions. A simplification occurs if we are only interested in the maximum of the edge-gradient which occurs at \(x = 0\). In that case:

\[
\text{LSF}(0) = 2 \int_{0}^{\infty} \text{MTF}(f) df
\]  

(10.10)

So it can be concluded that, irrespective of the exact shape of the MTF curve, the area underneath it is a useful measure of image quality as this area is proportional to the steepness of the edge-response.

10.8 Equivalent aperture approach.

The configuration with 14" II, XQ 1072 camera tube and a 512\(^2\) matrix as discussed in section 10.5 is just one permutation of a multitude of possible combinations.

The calculation and plotting of the overall MTF curves is a very tedious approach to drawing conclusions about the resolution
capabilities of a complete system. Now that we have succeeded in capturing the separate MTF curves in a few numbers (LFD, f_c and n), we would also like to have a straightforward simple method to estimate roughly the overall performance.

Several proposals in this direction have been put forward in the past. Most of the proposals (Albrecht,1962; Schade,1975; Wagner,1974) ascribe a certain blur or equivalent aperture dimension δ_i to each of the MTFs involved and calculate the overall blur by:

$$\delta_{\text{tot}} = (\delta_1^2 + \delta_2^2 + \delta_3^2 + \ldots)^{\frac{1}{2}}$$  \hspace{1cm} (10.11)

According to Schade(1975), the equivalent blur diameter δ_i of an imaging component is:

$$\delta_i = \frac{1}{2} \int_0^\infty \text{MTF}(f) \, df$$  \hspace{1cm} (10.12)

This more-or-less arbitrary definition correlates well with the subjective impression of sharpness, as for MTF curves with device index, n, between 1 and 2, δ closely matches the width of an edge transition between the 10% and 90% level (Johnson,1973).

Substitution of the Johnson curve Eq(10.2) in the blur diameter formula Eq(10.12) leads to a blur diameter which is slightly dependent on the MTF shape. The calculated relation between δ and f_c is presented in the following table.

Table 10.5: Relation between δ and f_c

<table>
<thead>
<tr>
<th>n</th>
<th>δ·f_c</th>
</tr>
</thead>
<tbody>
<tr>
<td>Exponential MTF</td>
<td>1.0</td>
</tr>
<tr>
<td>1.2</td>
<td>0.69</td>
</tr>
<tr>
<td>1.4</td>
<td>0.60</td>
</tr>
<tr>
<td>Practical range</td>
<td>0.6</td>
</tr>
<tr>
<td>0.8</td>
<td>0.25</td>
</tr>
<tr>
<td>Gaussian MTF</td>
<td>2.0</td>
</tr>
</tbody>
</table>

The blur diameter of an imaging component can now be determined by either the application of Eq(10.12) or using Table 10.5 when f_c and n are already known.

Although Eq(10.11) is quite generally used in practice, the formula is actually valid only for Gaussian MTF curves and corresponding Gaussian line-spread functions; in Johnson's terminology, only for device index n=2. In practice, however, the imaging components tend to have an index in the neighbourhood of n = 1.5 and the results of Eq(10.11) do not correlate very well with the width of the edge transition response as we have defined δ.

If all the imaging components involved had the same device index, n, a mathematically correct algorithm would be:

$$\delta_{\text{tot}} = (\delta_1^n + \delta_2^n + \delta_3^n + \ldots)^{\frac{1}{n}}$$  \hspace{1cm} (10.13)

In practice various device indexes occur in the chain and we have to resort to the use of a sort of average device index. This spoils the mathematical accuracy, but at least produces better results then Eq(10.11) which leads to too-optimistic values. In the following we will use n = 1.5 when Eq(10.13) is applied.

With the equivalent aperture approach just explained, each imaging component can be represented in a simplified way by a single quality figure 6 . The tables below show this figure derived from the previously-given data. These δ figures are much more satisfying than the generally used MTF curves for the process of understanding the physics of a complete system.

Table 10.6: Equivalent apertures of image intensifiers

<table>
<thead>
<tr>
<th>mode</th>
<th>f_c</th>
<th>n</th>
<th>δ - in entrance plane</th>
</tr>
</thead>
<tbody>
<tr>
<td>14&quot; II</td>
<td>1.2 lp/mm</td>
<td>1.55</td>
<td>0.72 mm</td>
</tr>
<tr>
<td>10&quot; II</td>
<td>1.4</td>
<td>1.48</td>
<td>0.63</td>
</tr>
<tr>
<td>6.5&quot; II</td>
<td>1.6</td>
<td>1.48</td>
<td>0.55</td>
</tr>
<tr>
<td>9&quot; II</td>
<td>1.5</td>
<td>1.50</td>
<td>0.59</td>
</tr>
<tr>
<td>6&quot; II</td>
<td>1.7</td>
<td>1.50</td>
<td>0.52</td>
</tr>
</tbody>
</table>

Table 10.7: Equivalent apertures of TV camera tubes.

<table>
<thead>
<tr>
<th>low I_s</th>
<th>f_c</th>
<th>n</th>
<th>low I_s</th>
<th>high I_s</th>
</tr>
</thead>
<tbody>
<tr>
<td>XQ 1072</td>
<td>20 lp/mm</td>
<td>1.50</td>
<td>0.044 mm</td>
<td>0.059 mm</td>
</tr>
<tr>
<td>XQ 1073</td>
<td>22</td>
<td>1.30</td>
<td>0.042</td>
<td>0.056</td>
</tr>
<tr>
<td>45 XQ</td>
<td>22</td>
<td>1.53</td>
<td>0.039</td>
<td>0.052</td>
</tr>
</tbody>
</table>
Table 10.3: Equivalent apertures of pixel matrices.

<table>
<thead>
<tr>
<th>Field</th>
<th>δ-px</th>
<th>δ-pix</th>
<th>δ-pix</th>
</tr>
</thead>
<tbody>
<tr>
<td>14&quot;</td>
<td>1.36 mm</td>
<td>0.68 mm</td>
<td>0.34 mm</td>
</tr>
<tr>
<td>10&quot;</td>
<td>0.99</td>
<td>0.48</td>
<td>0.24</td>
</tr>
<tr>
<td>6.5&quot;</td>
<td>0.64</td>
<td>0.32</td>
<td>0.16</td>
</tr>
</tbody>
</table>

As discussed before, the aperture figures of the TV camera need to be multiplied by the applicable optical magnification factor.

The overall resolution capability, previously approached by the overall MTF curve (section 10.6), can now be reconsidered by using the equivalent aperture method. The results of a configuration discussed earlier are presented in the following table.

Table 10.9: Equivalent apertures of an imaging system with: 512\(^2\) matrix, 14" II and XQ 1072 camera tube.

<table>
<thead>
<tr>
<th>δ-II</th>
<th>δ-TV</th>
<th>δ-pix</th>
<th>δ-total</th>
</tr>
</thead>
<tbody>
<tr>
<td>14&quot;</td>
<td>0.72 mm</td>
<td>1.42 mm</td>
<td>0.68 mm</td>
</tr>
<tr>
<td>10&quot;</td>
<td>0.63</td>
<td>0.99</td>
<td>0.48</td>
</tr>
<tr>
<td>6.5&quot;</td>
<td>0.55</td>
<td>0.66</td>
<td>0.32</td>
</tr>
</tbody>
</table>

From one quick glance at this table one can see the relative importance of the three contributing components. In all three cases the II is better than the TV and the discrepancy becomes larger when we go to the larger field sizes. Exactly the same conclusions can be drawn from Fig.10.6-a, b and c. The approximation of the resolution capability by the δ figures is so close that a δ-total, directly calculated from the f\(_c\) values (MTF = 37%) in Fig.10.6, agrees within a few percent with the values from Table 10.9. Or conversely, the δ-total figures from Table 10.9 could be used to synthesize a most-likely MTF curve given by Eq(10.2) with \(n = 1.5\) and \(f_c = 0.88/δ\)-total.

Of course the MTF curve so synthesized still has to have the sharp cut-off at \(f = 1/δ\)-pix.

The whole concept of equivalent aperture has been developed in order to be able to quickly review other configurations. As seen above, the worst component of the three considered is the TV camera tube and the overall system will gain the most if this weakest link is improved. The effect of the application of the high performance camera tube 45XQ is shown in the following table.

Table 10.10. Equivalent apertures of an imaging system with: 512\(^2\) matrix, 14" II and 45 XQ camera tube.

<table>
<thead>
<tr>
<th>δ-11</th>
<th>δ-TV</th>
<th>δ-pix</th>
<th>δ-total</th>
</tr>
</thead>
<tbody>
<tr>
<td>14&quot;</td>
<td>0.72 mm</td>
<td>0.73 mm</td>
<td>0.68 mm</td>
</tr>
<tr>
<td>10&quot;</td>
<td>0.63</td>
<td>0.51</td>
<td>0.48</td>
</tr>
<tr>
<td>6.5&quot;</td>
<td>0.55</td>
<td>0.35</td>
<td>0.32</td>
</tr>
</tbody>
</table>

The improvement with this tube is considerable, improvement factors of δ-total range from 28% to 35% in comparison with Table 10.9. In addition the overall LPD figure is also improved somewhat from 24% to 19%.

In this configuration the II is now the weakest link. If in this situation a further improvement is pursued by increasing the matrix size to 1024\(^2\), only a small improvement may be expected. The figures are shown in the next table.

Table 10.11: Equivalent aperture of an imaging system with: 1024\(^2\) matrix, 14" II and 45 XQ camera tube.

<table>
<thead>
<tr>
<th>δ-II</th>
<th>δ-TV</th>
<th>δ-pix</th>
<th>δ-total</th>
</tr>
</thead>
<tbody>
<tr>
<td>14&quot;</td>
<td>0.72 mm</td>
<td>0.73 mm</td>
<td>0.34 mm</td>
</tr>
<tr>
<td>10&quot;</td>
<td>0.63</td>
<td>0.51</td>
<td>0.24</td>
</tr>
<tr>
<td>6.5&quot;</td>
<td>0.55</td>
<td>0.35</td>
<td>0.16</td>
</tr>
</tbody>
</table>

The improvement factors of δ-total in comparison with Table 10.10 are respectively 12%, 14% and 17% for 6.5", 10" and 14" field size. The performance of the 14" field size is benefited the most by the application of the 1024\(^2\) matrix.

The whole concept of equivalent aperture has been developed in order to be able to quickly review other configurations. As seen above, the worst component of the three considered is the TV camera tube and the overall system will gain the most if this weakest link is improved. The effect of the application of the high performance camera tube 45XQ is shown in the following table.
Table 10.12: Equivalent apertures of an imaging system with: 14" II, 45XQ camera and various matrix sizes

<table>
<thead>
<tr>
<th>Infinite matrix</th>
<th>1024² matrix</th>
<th>512² matrix</th>
<th>256² matrix</th>
</tr>
</thead>
<tbody>
<tr>
<td>14&quot;</td>
<td>1.15 mm</td>
<td>100%</td>
<td>1.27 mm</td>
</tr>
<tr>
<td>10&quot;</td>
<td>0.91</td>
<td>100</td>
<td>0.99</td>
</tr>
<tr>
<td>6.5&quot;</td>
<td>0.72</td>
<td>100</td>
<td>0.77</td>
</tr>
</tbody>
</table>

10.9. Usefulness of geometric magnification.

In the preceding section the focal spot blurring has been intentionally kept out of the discussion in order to review the capabilities of the detector system as a whole. Now that we know the blurring of the detector system in the entrance plane, we can add the influence of the spot size by calculating the combined blurring in the object plane.

\[
\delta_{\text{obj}} = \left(\frac{1}{M_{\text{det}}} + \frac{1}{M_{\text{foc}}}\right)^{1.5}
\]

Depending on the situation, the magnification factor \( M \) can improve or deteriorate the overall performance. From Eq(10.14) it follows that if \( M=1 \), then \( \delta_{\text{obj}} = \delta_{\text{det}} \) and if \( M \gg 1 \), then \( \delta_{\text{obj}} = \delta_{\text{foc}} \). In the intermediate magnification range there will be gradual transition from the first to the second situation, and in general it can be stated that this transition will go in a desirable direction only if \( \delta_{\text{det}} > \delta_{\text{foc}} \).

Detailed results of Eq(10.14) for practical configurations are presented in Fig.10.7 and Fig.10.8. From these figures it could be concluded that the 14" mode gains the most from geometrical magnification, but this conclusion is of no practical benefit. If a 2x magnification were used with the 14" mode, the resolution capability of this mode would improve considerably indeed, but the field of view would also reduce to 14" : 2 = 7". With a field size this small, the choice of the 6.5" mode and lesser magnification gives a still better performance as can be concluded from the figures.
The intentional use of a greater than standard magnification \( (M = 1.2 - 1.5) \) thus makes sense only in the smallest mode of the image intensifier and additionally only when \( \delta_{	ext{FOC}} < \delta_{	ext{DET}} \) is fulfilled. A 1.2 mm focal spot does not meet this requirement. When a 0.6-mm focal spot is used a magnification \( M=2 \) will help to improve the configuration with the 1072 camera tube.

The configuration with the 45 XQ camera tube already is so much better that the use of a 0.6 mm focal spot and a magnification of 2x hardly has any further influence; this configuration needs a 0.3 mm focal spot to obtain noticeable improvements.

10.10. Conclusions.

The overall resolution performance of a digital X-ray imaging system has been examined.

Each of the imaging components has an MTF which can be characterized by three parameters: (1) low-frequency drop LFD; (2) frequency constant, \( f_c \), and (3) device index, \( n \). These parameters have been established for the various practical imaging components. Comparisons of these data are much more useful and quantitative than comparing MTF curves.

A still-more useful approach is the method of approximating the spatial resolution performance of an imaging component by a single blur figure, \( \delta \), which represents the width of the edge response function.

Overall blur caused by cascaded imaging components can be calculated mathematically with a simple formula, but the summation law is different from other summation laws which have been proposed in the past. The reason for this is that the previously-proposed method supposed Gaussian responses (device index \( n = 2 \)) for each component, whereas in practice most imaging components comply with \( n = 1.5 \). The consequence is a more-conservative blur-summation algorithm.

Because many more imaging components are involved in a digital X-ray imaging system than in a conventional film-screen system, many more factors each have a degrading influence and the overall result will be affected accordingly.

Film-screen detector systems usually have effective blur sizes of about 0.3, 0.45, and 0.70 mm for fine, mid-speed and high-speed categories. II-TV systems as discussed in this chapter have effective blur sizes which come close to the 0.70 mm of the high-speed screens only in the 6.5" mode of the intensifier. Larger field size modes even have a much larger blur dimension (1.4 mm for 14").
The superficial idea that a $1024^2$ matrix must be about twice as good as $512^2$ for image quality is refuted by the equivalent aperture calculations. Employment of a $1024^2$ matrix instead of a $512^2$ matrix helps only a little; the equivalent aperture size improves by about 12-17%.

Conditions are examined where geometrical magnification of say 2x is helpful. Because the effective blur diameter of an II-TV system is larger than that of a film-screen system, the conditions for positive effects are more easily fulfilled than with conventional radiography; a 0.6 mm focus is sometimes good enough to use with magnification.

In spite of the much larger overall effective blur dimension in the 14" mode compared with the 6.5" mode, the surprising effect has been noticed that the displayed 14" images appear much crisper to the viewer than the 6.5" mode images of the same object. This is a subjective effect caused by the minification of the image. A sensible explanation for this paradoxical effect is that the width of an edge response (or for medical users: the edge of a large vessel) expressed in pixels or in a percentage of image diameter on the monitor, is lower in the 14" case than in the 6.5" case. Electronic zoom of a fraction of the 14" image up to the same size as an original 6.5" image will prove that the 6.5" mode has better inherent resolution capability.

Sometimes investigators include the MTF of the video CRT and the MTF of the human eye in the overall appreciation. Just because the distance can be varied at will, and also because electronic zoom is a real feature of a digital image processing apparatus, neither factor is an actual physical limiting factor and is not considered in this chapter.

Other image processing functions such as two-dimensional MTF corrections (also called spot correction or edge enhancement or unsharp masking) are feasible features of a digital system, which are definitely required to bridge the gap between the detector performance characteristics of screen-film and II-TV systems. A couple of years ago the author proved that a high scan line II-TV system with analogue spot-correction circuits can successfully compete in image quality with 70-mm spot-film exposures.

References.


Johnson C.B.: Point spread functions, line spread functions and edge response functions associated with MTF's of the form $\exp\left(-\frac{f}{f_c}\right)^n$. Applied Optics 12, 1973, 1031-1033.


11. OVERALL SYSTEM PERFORMANCE.

11.1. Introduction.

In chapter 5 it was shown that there is a theoretical relationship between the size of a detail and the required contrast of that detail. Small details need a higher contrast to surpass the quantum noise level. In addition to this basic quantum noise limitation, imperfections of the X-ray equipment further degrade the low contrast detectability performance.

The main imperfections are: (1) limited detection efficiency, (2) limited spatial resolution capability, and (3) additional noise sources. The results of previous chapters will now be recapitulated, in order to assess the influence of each of the factors involved.

As a final result, the important medical question about the smallest detectable detail size can be answered. The required dose of contrast agent required and the X-ray dose to be used can also be established.

11.2. Limited detection efficiency.

Limited detection efficiency (DQE <1), as discussed in more detail in chapter 6, is caused mainly by two effects: (1) attenuation of the beam before it reaches the detector device, and (2) transparency of the detector device.

The antiscatter grid and the II input window together attenuate about 40% of the primary beam in the relevant part of the X-ray spectrum.

The thinness of the X-ray detection layer, which is needed to obtain an acceptable spatial resolution performance, makes this layer partly transparent for the X-ray photons.

In total the effective utilization of the input beam is only about 35% in the relevant part of the X-ray spectrum.

Due to this effect, the exposure dose needed is about three times the theoretically derived values as shown in the contrast-detail curves Fig.4.3.

This factor seems dramatic, but there is really not much that can be done about it; it is the state-of-the-art in diagnostic X-ray imaging.

More efficient detection could be obtained with a thicker detection layer, but the MTF of such a detector would be degraded by the greater lateral spreading light of in this layer. The optimum layer thickness is a result from a trade-off between MTF and DQE performance.
One could imagine that different application areas need a different trade-off. This philosophy has resulted in the availability of different intensifying screens (high-speed, mid-speed and high-resolution) to be used in combination with film. With image intensifiers this choice is not so broad, as it is much more difficult, in manufacturing as well as in the medical application, to cope with such a broad range of types. The currently available IIs could be classified as comparable to the high-speed screens.

Removal of the grid also helps more-efficient detection, but the information in the image will become so degraded by the scatter that it is not advisable to do so. Usually the characteristics of grids and detection devices such as screen-film combinations or IIs are considered completely separately in the literature. As far as the X-ray exposure to the patient is concerned, the grid should strictly speaking be considered as an integral part of the detection process. In order to conform with usual practice, however, the dose level of an exposure will be defined behind the grid at the input of the II. With this definition one should remember that the grid attenuates the primary beam by some 30%, and that this attenuation has to be compensated by correspondingly higher dose to the patient.

11.3 Criterion for negligible additional noise.

As X-ray noise is not the only noise source in a total DSA system, the effect of the additional noise source has to be considered in relation to the X-ray noise. In a good system design the additional noise sources are negligible; the quantum sink is in the X-ray beam impinging on the detector. The criterion for a second noise source to be negligible is that the overall signal-to-noise ratio is degraded negligibly by the addition of the second noise source. An example, the combined effect of two uncorrelated noise currents \( i_{n1} \) and \( i_{n2} \) is considered. The total noise current is:

\[
I_{n,\text{tot}} = \sqrt{i_{n1}^2 + i_{n2}^2}
\]

When the signal current is \( I_s \), then:

\[
\frac{S}{N_1} = \frac{I_s}{I_{n1}} \\
\frac{S}{N_2} = \frac{I_s}{I_{n2}}
\]

\[
\frac{(S/N)_{\text{tot}}}{S/N_1} = \frac{1}{\sqrt{i_{n1}^2 + i_{n2}^2}}
\]

When \( S/N_2 \) is better than \( S/N_1 \), the overall signal-to-noise ratio is determined mainly by \( S/N_1 \) as is demonstrated in the following table.

**Table 11.1: S/N degradation by a second noise source.**

<table>
<thead>
<tr>
<th>( S/N_2 )</th>
<th>( \frac{(S/N)_{\text{tot}}}{S/N_1} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.71</td>
</tr>
<tr>
<td>2</td>
<td>0.89</td>
</tr>
<tr>
<td>3</td>
<td>0.95</td>
</tr>
<tr>
<td>4</td>
<td>0.97</td>
</tr>
</tbody>
</table>

As is shown in this table, a 3-times better \( S/N_2 \) is in practice sufficient to maintain a virtually unaffected \( S/N_1 \). In the following sections each of the additional noise contributions will be checked against this criterion.

11.4. Contribution of TV shot noise.

For large signal values the TV shot noise is the largest of the additional noise sources; see Fig. 7.3. When a peak camera-current of 2000 nA is used, the signal-to-shot-noise ratio is of the order of 1000:1; see Fig. 11.1.

The X-ray signal-to-noise ratio \( S/N_{X-ray} \) per pixel is:

\[
S/N_{X-ray} = \left\{ \frac{D_{II} I_d \text{pix}^2}{DOE_{II} (d/D)} \right\}^{\frac{1}{2}}
\]
Signal-to-noise ratio of a total system with 2000 nA/1000 μR sensitivity setting

\[ \text{S/N}_{\text{x-ray}} = \frac{D_{\text{II}} \cdot \text{DQE}_{\text{II}} \cdot (\varphi/D)}{\text{DQE}_{\text{M}} \cdot (\varphi/D) \cdot \text{d}_{\text{pix}} \cdot \text{ conversion factor from X-ray exposure to particle fluence; see Fig.4.2.}} \]

The X-ray noise is determined by both the exposure dose \( D_{\text{II}} \) and the image field size (which sets \( d_{\text{pix}} \)). For a given exposure level the 14" field size has a better S/N than the 6.5" field size (\( d_{\text{pix}} \) is larger). Therefore the TV noise shall be compared with the "best-case" X-ray noise at 14".

When 1000 μR is taken as a realistic maximum applicable exposure dose per image, substitution in Eq(11.2) together with previously-discussed parameter values gives:

\[ \text{S/N}_{\text{x-ray}} = \left( 10^{-5} \times 0.098 \times 0.35 \times 2.2 \times 10^{-10} \times R_{\text{pix}} \right) \frac{1}{R_{\text{pix}}} = 190 \]

The variation of S/N\(_{\text{x-ray}}\) with exposure dose is also indicated in Fig.11.1 (see full line).

S/N\(_{\text{TVshot}}\) is much higher than S/N\(_{\text{x-ray}}\) and as both curves have the same slope (slope 0.5) on the log scale, this relationship holds for all the exposure levels of the sensitivity selection presented (sensitivity 2000 nA/1000 μR).

TV shot noise is of a similar nature as the X-ray noise, as it is caused by the limited number of electrons used in the TV camera to build up the charge pattern of the image. In the above case the number of electrons per pixel is thus much higher (about 70x) than the number of detected X-ray photons per pixel.

A more accurate analysis must also take into account that the X-ray noise will be somewhat reduced by the blurring of the II and TV camera, while the TV shot noise is unaffected by this blurring. Instead of calculating the X-ray quantum noise per pixel area, the area of the point-spread function is more appropriate as a basis for X-ray noise calculations (noise equivalent aperture approach). Using this method gives:

\[ \text{S/N}_{\text{x-ray}} = \left( D_{\text{II}} \cdot \text{d}_{\text{pix}} \cdot \text{DQE}_{\text{II}} / (\varphi/D) \right) \frac{1}{\text{d}_{\text{pix}} \cdot \text{DQE}_{\text{M}} / (\varphi/D)} \]

\[ = \left( D_{\text{II}} \cdot \text{d}_{\text{pix}} \cdot \text{DQE}_{\text{II}} / (\varphi/D) \right) \frac{1}{\text{d}_{\text{pix}} \cdot \text{DQE}_{\text{M}} / (\varphi/D)} \] (11.3)
Substituting \( d = 1.48 \text{ mm} \) (see Table 10.10) and further using the same values as used above we find:

\[
S/N_{\text{X-ray}} = 190 \left( \frac{1.48}{0.68} \right) = 410
\]

The result is that the TV shot noise contribution is still negligible; see dashed line in Fig. 11.1.

If a more conservative video peak current of 200 nA were used (sensitivity setting 200 nA/1000 µR), then the TV shot noise would no longer be negligible any more. This is the main reason for employing the unusually high camera signal current with DSA. TV camera improvements with lower preamplifier noise would not have been helpful in this respect.

The above considerations are applicable for the "best case" X-ray noise. In practice a somewhat lower exposure level (e.g. 600 µR) is usually used for 14" fieldsize. For 6.5" field size 1000 µR is frequently used, but \( d_{\text{pix}} \) is smaller here. In both cases more X-ray noise is present than in the calculations given above, which makes the TV shot noise even more negligible.

11.5. Contribution of preamplifier noise.

For small signal values the preamplifier noise dominates the additional noise sources; see Fig. 7.3. With a typical preamplifier noise current of 1 nA, the signal-to-preamp-noise ratio increases linearly with signal level up to \( S/N_{\text{preamp}} = 2000 \) at \( I_s = 2000 \text{ nA} \); see Fig. 11.1. The intersection of this \( S/N_{\text{preamp}} \) line with the "blurred" \( S/N_{\text{X-ray}} \) line occurs at 120 nA or at 120:2000 = 6% of peak video level. Below this level the preamplifier noise dominates. At about 200 nA or 10% of video peak the preamplifier noise is already negligible compared with the X-ray noise.

In medical applications a video level below 10% of video peak will seldom occur as most medical images will cover a dynamic range which is less than 10:1.

The importance of appropriate exposure control can be appreciated now. If the exposure is too low, by for instance a factor of two, the video peak will be at 50% and the lowest video level of a 10:1 dynamic range object will be at 5% or 100 nA. In this example the lowest video level is so low that the preamplifier noise is no longer negligible.

For field size selections smaller than 14" and/or exposure levels lower than 1000 µR the preamplifier noise is even more negligible for reasons similar to those discussed in the previous section.

For large-area contrast differences the fine-grained preamplifier noise is subjectively less significant, which also decreases the significance of the preamplifier noise.


In section 9.2 it has been shown that digital quantization noise is negligible if the quantizing step size is smaller than twice the rms noise amplitude of the input signal. In section 9.5 the choice of the appropriate ADC accuracy was based on the presence of TV noise only. If the criterion that quantization noise is negligible compared with TV noise is met, the quantization noise is certainly negligible in comparison with the X-ray noise, as can be concluded from the previous sections of this chapter.

11.7. Theoretical predictions of minimal acceptable \( S/N \) and exposure dose for DSA.

Due to the contrast enhancement in the DSA processor, the iodinated blood vessels of interest are better visualized than before, but the noise in the image is also enhanced by the same factor. An experienced radiologist, who knows how much image noise is associated with a given dose level, will be deceived by the subtraction and enhancement procedures. A dose level which is known to produce good image quality with unsubtracted imaging will produce noisy to very noisy pictures, depending on the selection of the enhancement factor.

Although the appreciation of noise in an image is a subjective matter, some theoretical estimations can be made about the minimum required \( (S/N)_{\text{min}} \) for DSA imaging with a certain enhancement factor.

When the noiseband, superimposed on the input signal, is expressed in dB, then a transposition of this noiseband to the selected display range will make it possible to indicate how much the fractional noise will be in this selected display range. The following table shows the results of such calculations.
Table 11.2. Noise in DSA display for various input S/N ratios.

<table>
<thead>
<tr>
<th>(S/N)_{in}</th>
<th>range of rms noise beyond signal level</th>
<th>range of rms noise after subtraction</th>
<th>noise fraction ( \text{in} \ +2 \ dB ) display range</th>
<th>expected subjective image quality</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>0.928 dB</td>
<td>1.17 dB</td>
<td>0.04</td>
<td>very bad</td>
</tr>
<tr>
<td>30</td>
<td>0.285</td>
<td>0.402</td>
<td>0.01</td>
<td>bad</td>
</tr>
<tr>
<td>100</td>
<td>0.0854</td>
<td>0.122</td>
<td>0.001</td>
<td>good</td>
</tr>
<tr>
<td>300</td>
<td>0.0289</td>
<td>0.0409</td>
<td>0.001</td>
<td>excellent</td>
</tr>
<tr>
<td>1000</td>
<td>0.00868</td>
<td>0.0123</td>
<td>0.001</td>
<td>excellent</td>
</tr>
</tbody>
</table>

The selected display range of \(+2 \ dB\) is the mostly used during practical DSA examinations.

The last column of Table 11.2 estimates the corresponding image quality impression. The first row with \( (S/N)_{in} = 10 \) has the appreciation "very bad" because the r.m.s. noise level of 30% of the display range will have a peak-to-peak noise level which exceeds the display range, i.e. blacker than black and whiter than peak white.

From the table it can be estimated that \( (S/N)_{in} = 30 \) is the bare minimum as regards acceptable image quality. Substitution of this minimum S/N ratio in Eq(11.3) gives the corresponding dose levels at the input of the image intensifier. Results are presented in the following table.

Table 11.3. Minimum exposure dose for DSA with \(+2 \ dB\) display range.

<table>
<thead>
<tr>
<th>Field size</th>
<th>( D_{II} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>14&quot;</td>
<td>5 ( \mu R )</td>
</tr>
<tr>
<td>10&quot;</td>
<td>9</td>
</tr>
<tr>
<td>6.5&quot;</td>
<td>16</td>
</tr>
</tbody>
</table>

The minimum exposure dose just calculated is needed for those parts of the image which correspond to the densest parts of the object. More transparent parts will produce a higher exposure level on the X-ray detector of course.

11.8. Image quality evaluation.

In addition to the S/N calculations and their predicted image quality performance some measurements are also in order to support the theoretical predictions. The experiments are separated in to two categories:

(1) X-ray noise alone

(2) TV noise alone.

In the experiments of the first category, the II is exposed to various uniform dose levels, while the iris of the TV camera is adjusted to obtain a full strength video signal even at the lower dose levels. By this method the S/N ratio is determined by the X-ray noise alone.

As noise appreciation is subjective, the best way of presenting the results of the experiments is by showing the images; see Photo 1-12. All the images were made with the \(+\ 2 \ dB\) display range selection.

It may be concluded from the photos that the minimum acceptable dose levels predicted in preceding section \( (16\, \mu R \text{ for } 6.5" \text{ and } 5\, \mu R \text{ for } 14") \) are in agreement with reality. Good image quality starts to appear at dose levels which are about 5 times the quoted minima \( (80\, \mu R \text{ for } 6.5" \text{ and } 25\, \mu R \text{ for } 14") \).

How far one should go with maximum dose level depends on the expected dynamic range of the object. For instance, \( 1000\, \mu R \text{ on } 6.5" \) gives excellent to good image quality in the range \( 1000\, \mu R - 80\, \mu R \), which is a dynamic range of \( 1000 : 80 = 12.5 \). This range is good enough for most applications.

An identical dynamic range for 14" field size requires \( 300\, \mu R - 25\, \mu R \) of exposure level for similar image quality.

When X-ray collimation is used during an examination, in theory no radiation should reach the II in the shuttered area. In practice a minute amount of radiation still hits the shuttered areas because of afocal radiation and/or scattered X-rays. The amount of dose is so small here that these areas will show the worst case S/N ratio; usually a noise pattern similar to Photo 5 and 12 will be visible.

A frequently-encountered misconception is that the shuttered areas receive no radiation at all and so the noise in these areas represents the inevitable system noise mainly caused by the TV. This line of thinking is false of course. Complaints about the noise level in the shuttered areas have nothing to do with the basic system noise; it is still the unexpected X-ray fluence which causes the noise there.

The result of an interesting experiment is shown in Photo 13, where the 14" - 4.8\( \mu R \) image is electronically zoomed by a factor of 2x. It appears as if the S/N has become worse although the noise amplitude has remained exactly the same. Only the spatial distribution in the presentation has been changed and it is the coarser noise pattern that is experienced as more disturbing.
The fact that the coarser noise pattern appears more disturbing is only a cosmetic effect; for the detection of low contrast details the detail size-matched filtering theory applies which gives equal results in the zoomed and unzoomed image. Both noise and detail are enlarged with zooming and the zoom feature has a beneficial effect only if either the human eye or the display monitor does not have enough spatial resolution to resolve the fine details in the original unzoomed image. Usually the zoom feature is a superfluous gadget.

The experiments of the second category with TV noise alone were carried out with no X-rays, a uniform illumination to the TV camera being supplied by means of a lamp. At high video levels (much light) no noise at all is discernible in the subtraction images. At low video levels (dimmed light) the preamplifier noise starts to become slightly visible due to the steep slope of the logarithmic amplification in this region. Photo 14 is a subtraction image from a situation with 100 nA (= 5% of video peak) of camera signal current. The image appreciation is in accordance with the predictions from Table 9.3. In general it can be concluded from the photos that TV noise is indeed negligible compared with the X-ray noise.

11.9. Usefulness of frame integrations.

As discussed in section 7.8 the application of frame-integrations will result in a reduction of the relative contribution of the TV noise. However the X-ray quantum noise will increase by the more inefficient utilization of the applied dose. It has been concluded already that four integrations may be worthwhile, as only this number gives a reasonable compromise between dose utilization and exposure time.

We concluded in section 11.4 and 11.5 that TV noise is negligible, provided that enough camera signal current is used. Therefore any improvement in the TV noise is bound to be negligible also in those circumstances. On the other hand the reduction of the dose efficiency to 4/5 = 80% will immediately degrade the total S/N ratio by some 10%.

When a DSA system is equipped with a TV camera that can not provide the high signal currents required, the frame integration method can be beneficial. Integration of 4 frames is equivalent to using a signal current 16 times higher.

Another advantage of the frame integration method is that instrumental deficiencies of the DSA system will be more or less concealed if they occur. Two examples of this concealment have been noticed in practice.

Horizontal structured noise caused by instabilities of the vertical deflection (section 7.6) nearly disappears when 4-frame integration is used instead of single sequential-scan read-out of the TV camera tube.

Further it has been noticed that low video signal levels are particularly susceptible to deficiencies of the AD converter. These errors show up most conspicuously when no X-ray noise is involved, as in the experiments described in section 11.8, where the low video level is generated by low-level light from a lamp.

Any non-monotonicities of the ADC shows as noisy contour lines in the difference image. Application of 4-frame integration partly covers up this error by smoothing it over a broader area in the image. In practice the presence of X-ray noise will also help to mask errors of the AD converter.

Both of the DSA system deficiencies mentioned should really be cured at their origin. These deficiencies should be no reason to use the frame integration method for solving the problem.

11.10. Contrast-detail curves.

Contrast-detail curves as derived in chapter 4 (Fig.4.3) need adaptation in two respects:

1) in general a higher exposure level is required due to the subtraction trick itself (combining the noise of two images) and due to the limited detection efficiency.

2) smaller details require extra contrast to compensate for the blurring.

When analyzing the small-detail performance of a system, the MTF of the system is normally used as this directly correlates with measurements which can be done with a line pair test object. This method of analysis shows how the system performs at a particular frequency, but does not relate unambiguously to its response to a single circular detail of a given diameter. For instance, a spatial frequency higher than half the pixel density is impossible to visualize, while single details smaller than a pixel can in principle be visualized. This smaller detail then gives a relatively smaller amplitude and will
A more accurate analysis of this basic idea takes the PSF (point spread function) of the system into account. Any small detail in the object plane which is much smaller than the PSF will be displayed spatially with at least the PSF dimensions.

As discussed in section 10.7, the PSF dimensions do not correlate with a particular point of the MTF curve, but the shape of the whole curve is involved instead.

The influence of blurring on the amplitude of a contrast difference can be quantified by using Schade's equivalent aperture theory (Schade, 1971). According to this theory the contrast amplitude of a circular object is reduced by a factor which is proportional to the ratio of the areas of the original and the blurred circular object:

\[ C_{\text{blur}} = C_{\text{in}} \frac{d_{\text{obj}}}{d_{\text{eq}}^2} \]

where

- \( C_{\text{in}} \) = input contrast
- \( C_{\text{blur}} \) = contrast of blurred image
- \( d_{\text{obj}} \) = diameter of object in detector plane
- \( d_{\text{eq}} \) = equivalent aperture diameter in detector plane

Visualizing details smaller than \( d_{\text{eq}} \) is thus possible, but due to the blurring the contrast will be much lower.

The X-ray noise is also reduced by the blurring, so an adapted version of the Rose contrast-detail equation becomes:

\[ C_{\text{in}} \frac{d_{\text{obj}}}{d_{\text{obj}}^2 + d_{\text{eq}}^2} \geq \frac{k_T \sqrt{2}}{\left\{ D_{\text{II}} \left( d_{\text{obj}}^2 + d_{\text{eq}}^2 \right) D_{\text{GII}} (d/D) \right\}^{\frac{1}{2}}} \]

or:

\[ C_{\text{in}} \geq \frac{k_T}{d_{\text{obj}} \left( D_{\text{II}} (d/D) \right)^{\frac{1}{2}} \sqrt{D_{\text{GII}}}} \cdot \sqrt{2} \cdot \left( 1 + \frac{d_{\text{eq}}^2}{d_{\text{obj}}^2} \right)^{\frac{1}{2}} \]

The first factor in this equation is identical to the original Rose equation.

The second factor takes the limited detection efficiency and the noise addition of two images into account.

**Fig 11.2-a** Contrast-detail curves 6.5°

**Fig 11.2-b** Contrast-detail curves 10°
The last factor represents the increase in input contrast required to overcome the reduction caused by the blurring.

The more-sophisticated contrast-detail curves which follow from Eq(11.5) are shown in Fig.11.2. As \( d_{eq} \) is dependent on the selected field size mode of the II, the three situations 14", 10" and 6.5" are presented separately. The \( d_{eq} \) values are taken from Table 10.10. Asymptotes of the curves have slope -1 for \( d_{obj} > d_{eq} \) and slope -2 for \( d_{obj} < d_{eq} \).

11.11. Required iodine concentration.

Contrast-detail curves, as discussed in the preceding section, show the image contrast required to make a certain detail visible. Of much more practical relevance to the doctor in the hospital is the required iodine concentration in the vessels required to make vessel details visible.

As shown in chapter 6, the primary generated contrast by the iodine-filled vessel is:

\[
C = \left| \frac{\mu}{\rho_i} \right| \rho_i t
\]

where \( \left| \frac{\mu}{\rho_i} \right| \) = effective mass attenuation coefficient of iodine for the spectrum used (cm\(^2\)/g).

\( \rho_i \) = mass concentration of iodine (g/ml).

\( t \) = inner diameter of the vessel (cm).

The effective attenuation coefficient varies with object thickness and selected kVp, has been comprehensively discussed in chapter 6; see Fig.6.8. In the centre of the kVp range typically used at 67 kVp the effective attenuation coefficient is at 15 cm\(^2\)/g. This value will be used now, but it must be kept in mind that large deviations from this centre value are possible.

Three levels of iodine concentration, 40, 80 and 160 mgI/ml are presented in Fig.11.2. These concentration levels more-or-less represent typical values for intravenous, non-selective arterial and selective arterial injections; see Table 5.5.

Scatter influence is not included in the lines drawn, although we do know from chapter 6 that scatter spoils the contrast by a factor of 0.6 - 0.9; see Table 6.1.

From Fig.11.2 it can be concluded that for larger arteries (>5 mm) the vessel contrast is well above the noise threshold, even with the
intravenous injection method. For smaller vessels the visualization becomes more critical.

The intersections of the iodine concentration lines with the contrast-detail threshold curves clearly indicate how much is to be gained when selective injections are used instead of intravenous injections. If, in the 6.5" mode, selective injection is used instead of intravenous injection, the size of the smallest visible detail decreases from about 0.65 mm to about 0.4 mm.

Capillary vessels which have a diameter of about 8 μm are usually so crowded in the capillary bed, that instead of visualizing single vessels, the whole conglomerate shows up as a "blushing" of the tissue. Due to this effect the perfusion of, for example the kidneys, can be examined very effectively; partial perfusion defects are very well demonstrated by the DSA technique.

The contrast of a perfused area is also given by Eq(11.6) with the remark that thickness, t, is now the thickness of the tissue involved (thus much larger than a single artery), while the iodine concentration is much lower due to the capillary vessel - tissue volume ratio. The two effects counterbalance in such a way that iodine in the kidneys is about equally as visible as iodine in medium-sized arteries.

Due to the influence of the vessel-tissue volume ratio, any areas with hypervascularization, such as some tumours, are clearly revealed by the DSA technique; clinical experience has proved that the DSA technique is superior in this respect to the "gold standard" of the conventional angiographic technique.


The performance level of image intensifiers is a compromise between MTF and DQE performance; currently available ITS can be classified as comparable with high-speed screens.

When radiographic exposure levels (500 - 1000 μR) are used, a 100 nA peak signal current in the TV camera tube is insufficient by far to ensure a negligible shot noise contribution. A much higher signal current is needed in this application.

The importance of a high signal current is not generally recognized, as most publications on DSA stress only the importance of the S/N ratio of the TV system (defined as max signal current divided by rms preamplifier noise) by requiring S/N \( \geq 1000 \).

S/N improvement by drastically reducing the preamplifier noise would not help to solve the problem, while S/N improvement by increasing the available signal current is a must.

In a worst-case situation the preamplifier noise becomes dominating below 1/17 of the maximum signal level. When proper exposure control is used, this signal range will not be penetrated in practice.

For practical exposure dose levels (1000 μR on 6.5" and 500 μR on 14") the preamplifier noise is even more negligible.

Frame integration is useful only to mask some deficiencies of the system and/or when the available TV camera is not able to supply enough signal current in a single scan.

Due to the high contrast sensitivity of the DSA system more quantum noise will also be visible in the images. This can only partly be offset by using more exposure dose because noise-free DSA images would require a prohibitively large X-ray dose.

For cosmetic reasons a certain minimum exposure dose is required at a particular enhancement setting.

The maximum exposure dose on the detector depends on how far the dynamic range of the object is above the minimum exposure. A homogeneous object such as an abdomen does not require as much peak exposure dose as inhomogeneous objects such as the chest and neck.

Noise experiments have proved that TV noise is negligible when enough signal current is used, even when the very high radiographic exposure dose of 1000 μR is used.

More sophisticated contrast-detail curves are presented which also take MTF degradations (blurring) into account.

Arteries smaller than the pixel can be visualized, provided that enough contrast is available.

The just-visible size of vessels varies with the iodine concentration to the power - 1/3; selective injections can visualize vessels about 1.6x smaller than those shown by intravenous arm injections.
12. ARTEFACTS.

12.1. Introduction.

As discussed in the previous chapter, a very basic limitation with regard to the usefully applicable enhancement factor is the increased visibility of the X-ray noise. A second, very practical, limitation factor is caused by image artefacts. Image artefacts are undesired disturbances in the image which come in addition to the undesired fluctuation noise. More enhancement results in correspondingly more disturbance by the artefacts.

In the DSA technique most of the artefacts are caused by movement of the images to be subtracted. The origin of the movement may be either in the object or the apparatus itself. Other artefacts are caused by instabilities of the image signal amplitude.

12.2. Patient movement.

Although each patient is asked to remain as immobile as possible, even holding his breath during the acquisition run, many patients are unable to do so. The involuntary movement is initiated by the fright reaction to the sudden sensation of heat caused by the injection of contrast medium. Most patients feel an inclination to swallow, which produces artefacts in the images of a carotid artery examination.

Involuntary movements of the head preclude the production of clean intracerebral DSA images; the bony structures do not subtract properly due to misregistration and a noisy overlay appears in the image. Mechanical head-immobilizing aids can be helpful in this respect. DSA examinations in the chest suffer from artefacts caused by misregistration of the pulmonary vasculature. ECG gating of the X-ray exposures, preferably during the end-diastolic period, have proved to be very successful in this respect. The gating pulses generated by ECG equipment usually coincide with the QRS complex in the ECG curve. A certain delay is needed to produce the end-diastolic exposure gating pulse from the QRS pulse. The required delay is in the range 0.50-0.75 s depending on the heart rate of the patient. Instead of manually adjusting this delay for each patient, the microprocessor of the DSA system monitors the heart frequency and automatically generates a time delay of 75% of the heartbeat period after each QRS complex.

Abdominal examinations may suffer from artefacts due to moving intestines. Usually the patients for these examinations are premedicated by an injection of glucagon to stop this movement temporarily. Additionally an abdomen compression band is frequently used.

The artefacts during DSA examinations of the lower peripherals can be very misleading because the lateral movement of a leg may produce an artefact which very much looks like a vessel in the image along the bone-tissue boundary.

Additional problems arise during these lower peripheral examinations when the "stepping table" concept is used. As the field size of the image intensifier is limited, a fixed position of this imaging device can not visualize the whole length of a leg. Therefore the examining physician commands (interactively or by preprogramming) the patient support table to move longitudinally so as to follow the downstream course of the contrast medium. With unsubtracted conventional peripheral angiography this is already quite a complicated procedure. With DSA examinations, multiple pre-injection masks have to be made for each table position, and these must be called up in sequence during the live run. The chance of good image registration for subtraction is very rare with this type of examination.

As peripheral angiography is a frequently needed diagnostic procedure, experiments are going on to circumvent the misregistration problem by using a separate injection for each fixed-position view. Owing to the DSA's contrast enhancement, a very small amount of contrast medium is needed for each injection if the injection is made arterially (e.g. in the femoral artery). The injection rate required is so low that a catheter need not be introduced into the artery; instead a thin bore injection needle can be used. The damage to the vessel wall is so small that in spite of the arterial injection the examination is considered to be safe enough to avoid patient hospitalization.

12.3. Equipment registration instability.

In practice the most serious artefacts are caused by patient motion. Of smaller amplitude, but undeniably present, are also some movement artefacts caused by the equipment. This effect can be noticed, if instead of a real patient, a phantom object is placed in the X-ray beam path. Perfect subtraction of the phantom would yield a homogeneously grey DSA image. In practice some of the high contrast edges of the original always show up. This is apparently due to small spatial irreproducibilities of the image. The amplitude of the artefact depends on the amplitude and slope of the high-contrast edge in the unsubtracted image. From a figure similar to Fig.8.5 it can be
derived that the amplitude, $e$, of the artefacts is:

$$e = \frac{\Delta x}{x_s} E$$  \hspace{1cm} (12.1)$$

where $\Delta x =$ image shift (mm)

$x_s =$ spatial spread of transition (mm)

$E =$ contrast amplitude.

The highest amplitudes occur if highly absorbing objects, such as lead pills, screwdrivers, scissors, etc are placed in the beam. If no further scatter medium is present, the relative contrast amplitude $E$, expressed in dB, may be as high as 20.

For sharp-edged objects the spatial spread of the transition may be as low as 0.8 mm (see Fig.10.8). An image shift of only 0.1 mm (= 1/3 pixel in a 6° field) will give in this case:

$$e = 0.1 \text{ mm} \times 20 \text{ dB} = 2.5 \text{ dB}$$

which forms 62% of the display range when a ± 2 dB display range has been selected.

This example indicates that it is hardly possible to eliminate completely the misregistration artefacts in this worst-case imaging situation. In more practical imaging situations with patients the contrast amplitudes of, for instance, bone-tissue boundaries are about one order of magnitude lower. The artefacts are then still visible (64% of display range) but not excessively disturbing.

The spatial instability of the system can be caused by both the X-ray source and the X-ray detection system. Separation of these effects is possible by a suitable measuring set-up.

12.3.1. Focal spot drift.

The focal spot dimensions of X-ray tubes are usually measured by a so called pinhole camera (IEC recommendation 336). The relative position of the pinhole and the detector is so chosen that the focal spot intensity distribution at the detector is sufficiently magnified geometrically to be able to measure the magnified spot dimensions. With a large magnification factor the detector parameters such as blur and shift become irrelevant for the focal spot measurements.

During experiments with such a pinhole camera and the DSA equipment it has been found that during the first few exposures of a series a drift of the focus position occurs. Especially high-load exposures (e.g., 75 kVp, 75 mAs) show a drift which can amount to up to 40% of the focal spot dimension. This 40% figure has been observed for both the 0.6 mm and 1.2 mm focal spot of the tube examined. The main direction of the shift is parallel to the axis of the tube. The shift perpendicular to the axis of the tube is much less.

The influence of the focal spot drift in a clinical situation is reduced by the fact that the object is closer to the detector. Geometrical relationships similar to those shown in Fig.11.1-b result in an image shift $\Delta x_f$ in the detector plane which is equal to:

$$\Delta x_f = (M-1) x_r$$

where $M =$ geometrical magnification ratio.

$x_r =$ focal spot shift

For a typical situation with 1.2 mm focal spot and $M = 1$, the 40% spot shift leads to an image shift $\Delta x_f = 0.14 \text{ mm}$.

12.3.2. Spatial instability of the detector system.

Imaging of a simple object (e.g., a metal washer) which is placed as close as possible to the detector entrance plane shows mainly the spatial instability of the detector ($M = 1$ in Eq(12.2)).

In principle both the II and the TV camera could be sensitive for spatial instabilities, but so far no evidence has been found that the II gives a measurable contribution. The TV camera on the other hand is the main suspect in the detector chain.

Because of the mains-lock operation of the TV scan frequencies the stability of these frequencies could be doubted. The high power switching (100 kW) of the mains loading by the X-ray generator generates so much mains voltage disturbance that the Phase-Locked-Loop that couples the scan frequencies to the mains frequency may pick up some of it. The doubt about the scan frequency stability has been removed by an experiment where these scan frequencies were crystal locked; the spatial instability remained as before. Instability of the scan signal amplifiers and the influence of stray magnetic fields is currently under investigation in order to improve the camera performance further in this respect.
An important second-order effect of the spatial instability of the detector is the generation of extra noise in the DSA subtraction images. Each detector component, II and TV, has a certain inhomogeneity of the sensitivity over the imaging field. Slow variations are usually known as vignetting or shading. Medium to high-frequency local irregularities result in a multiplicative "fixed pattern noise" in the image which usually is negligible. Ideal subtraction of subsequent identical images would remove the shading and fixed pattern-noise. Spatial misregistration of the two images to be subtracted causes the fixed pattern noise to be reintroduced and it may become in a worst case even twice as high.

12.4. Image reregistration; pixel shift.

In the normal course of DSA examinations many of subtraction images have motion artefacts when the first image is used as the mask. During postprocessing another mask can be chosen from the series, which usually produces diagnostically sufficient image quality. In about 15% of the studies no useful image quality can be obtained, mainly because of too many motion artefacts. A certain percentage of these unsuccessful studies can be saved by applying a processing algorithm which horizontally and vertically shifts one of the images of the pair until a better subtraction result is obtained. In most cases the pixel shifting is able to improve the image only in a certain area while other parts of it even deteriorate. This is caused by the queer warped movement distribution over the image. Therefore it is advantageous to provide "pixel shift" only in a Region Of Interest (ROI) which can be defined by the user. In section 12.3 it has been calculated that shifts of even fractions of a pixelsize may introduce large amplitude artefacts. For this reason it has been found necessary to counteract accordingly with fractional pixel shifting. A shift of 1/10 of a pixel is obtained by linearly adding 10% of an unshifted image to 90% of an image which has been shifted by one pixel. In practice it turns out that the pixel shift feature offers the possibility of further improving images that were already diagnostically useful. Gross movement in diagnostically-unsuccessful images can usually not be removed sufficiently to save the study.

12.5. Signal amplitude instability.

As mentioned earlier in section 9.7 the irreproducibility of the X-ray exposure dose for each image in the series causes a disturbance which precludes a smooth subsequent image display. The X-ray output change affects the detected signal multiplicatively. Due to logarithmic signal processing, the effect on the displayed subtraction image is just a d.c. brightness shift, as:

$$\log x - \log a = \log x - \log a - \log x = - \log a$$

When each subtraction image has its own brightness offset, the display of subsequent images is disturbed by a random jumping up and down of this offset value. Sophisticated X-ray generators have kV en mA stabilization circuits which stabilize the output sufficiently (better than a few percent). Less sophisticated generators with higher output-instability will produce DSA images with marked brightness instability, but each individual subtraction image can be corrected by applying the manual level control during viewing.

12.6 Conclusions.

The contrast resolution of the DSA equipment is so high that 0.1 mm misregistration of the images may cause a major artefact in a worst case situation.

Unsuccessful DSA studies (about 15%) are usually caused by too many movement artefacts.

Most of the movement artefacts are caused by involuntary patient movement. Several measures are possible to prevent this.

Equipment instabilities also produce some misregistration artefacts.

Fractional pixel shifting, which is a rather complicated image processing function, can be applied to reduce the misregistration artefacts.

Of the unsuccessful studies only a fraction can be saved by the pixel shift feature; the most successful postprocessing function is simple "remasking".

Less sophisticated X-ray generators produce DSA images with unpredictable brightness offsets.
Samenvatting.

Electronisch subtractie, waarbij voor het eerst digitale beeldtechnieken werden toegepast, is door Mistretta en zijn medewerkers sinds 1976 gepropageerd voor angiografische toepassingen. Dankzij de digitale beeldverwerking is het mogelijk om zodanig lage contrasten zichtbaar te maken dat volstaan kan worden met intraveneuze injecties van contrastmiddel.

In het proefschrift is geanalyseerd hoeveel verdunning van het geïnjecteerde contrastmiddel ontstaat bij een intraveneuze injectie. Het blijkt dat de injectiesnelheid van ondergeschikter belang is dan voorheen werd aangenomen.

Het opgewekte radiografische contrast hangt af van de opnamecondities. Via berekeningen met Röntgenstralen spectra blijkt dat lage buisspanningen over het algemeen de voorkeur verdienen. Er worden speciale eisen gesteld aan de toegepaste TV camera. De aspecten die de signaal-ruisverhouding van de TV camera bepalen zijn uitvoerig behandeld.

Bij het digitaliseren van videosignalen zijn twee aspecten van belang: de grootte van de beeldpuntenmatrix en de kwantiseringsnauwkeurigheid. In relatie met de andere beeldvormende factoren in een totaal systeem blijkt dat een $512^2$ matrix voldoende is.

De benodigde kwantiseringsnauwkeurigheid hangt af van de soort signaalbehandeling die toegepast wordt. Het toepassen van look-up tabellen voor logarithmische conversie vereist extra bitdiepte.

Bij de voor DSA gewenste contrast gevoeligheid instelling blijkt het nodig om een relatief hoge Röntgendosis toe te passen. Bij de gebruikelijke dosiswaarden is de ruis van het electronisch/digitaal gedeelte van het systeem verwaarloosbaar t.o.v. Röntgenruis. In een contrastdetail diagram is aangegeven hoeveel het ingangscontrast moet zijn om duidelijk boven de Röntgenruis uit te komen. Kleine details behoeven een groter ingangscontrast omdat er minder kwanten in het detailopgavevlak meedoen en omdat de MTF afval gaat meespelen. Tevens is aangegeven hoeveel contrastmiddelconcentratie er nodig is om het gewenste ingangscontrast op te wekken.

Afgezien van Röntgenruis vormen ook artefacten, die ontstaan tengevolge van een niet-ideale subtractie, een beperking voor de zinnig toepasbare contrastversterkingsfactor.

Curriculum Vitae.


In 1976 werd hij tijdelijk gedetacheerd in Amerika om daar te helpen bij de detectorelectronica van de eerste Philips Röntgenscanner. Daarna trad hij in dienst bij Philips Medical Systems om bij de afdeling "Conventioneel Röntgen" voor de TV techniek een breder toepassingsgebied te ontwikkelen.

Het Philips DVI-2 apparaat en ook dit proefschrift zijn het resultaat van de laatste activiteiten.
Dankbetuiging.

Voor het tot stand komen van dit proefschrift ben ik dank verschuldigd aan:

- de staf van het Antoniusziekenhuis voor de goede samenwerking.
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- Mijn gezin voor het dulden van mijn permanente andersgerichte prioriteitsstelling gedurende meerdere jaren.

Quantum noise in DSA images with ± 2dB display range and 6.5" field size.
Quantum noise in DSA images with ± 2 dB display range and 14" field size.

Quantum noise in a DSA image with ± 2 dB display range and 2x zoomed out of a 14" field size.

Noise in a DSA image with ± 2 dB display range and low level (5% of videopeak) light input.
1. In tegenstelling tot wat bij conventionele angiografie bekend is, heeft bij intraveneuze injectie een verhogen van de injectiesnelheid een beperkte invloed op de concentratie van het contrastmiddel in het arteriële vaten-systeem.

2. TV-uitlezing volgens een "slow scan" methode vermindert weliswaar de relatieve ruisbijdrage van de videovoorversterker, maar omdat tevens de beschikbare signaalstroom evenredig afneemt, is een netto voordelig effect hiervan nauwelijks te verwachten.

3. Bij het toepassen van een digitale "Look Up Table" (LUT) om logarithmische conversie van de signaalamplitudes te verkrijgen, is het nodig dat de uitgang van de tabel meer bits omvat dan de ingang van de tabel.

4. Toepassing van een $1024^2$ beeldpuntenmatix levert de duidelijkste verbetering van de spatiale resolutie voor de grootste beeldveldformaten, doch zelfs voor een 35 cm formaat is de te bereiken verbetering van de steilheid van een kantovergang minder dan 20%.

5. Wanneer een hoge Røntgendosis per beeld wordt toegepast, wordt meer verbetering van de signaal-ruis verhouding bereikt door een grotere camera-signaalstroom toe te passen, dan door vermindering van de versterkerruis na te streven.

6. Van DSA-onderzoeken kan niet beweerd worden dat de methode dosisbesparend is.
Het gemak waarmee elektronisch opgeslagen beelden van een anqjografieonderzoek weer kunnen worden uitgewist, is niet bevorderlijk voor een karig gebruik van Röntgenstraling.

8. Kwaliteitsvergelijking van beeldvormende systemen met het criterium "grensoplossend vermogen" heeft alleen zin indien de vorm van de MTF-curves identiek is.

9. Indien het maken van een "hardcopy" uit een elektronisch beeldarchief even eenvoudig wordt als het maken van een "Xerox copy", dan zal de elektronische beeldarchivering niet of nauwelijks tot filmkostenbesparing leiden.

10. De moeite die NEMA (National Electrical Manufacturers Association) zich getroost om tot een algemeen aanvaardbare norm te komen voor de uitwisseling van digitale beeldinformatie, gaat geheel voorbij aan de mogelijkheden die datareductie kan bieden.
   Gezien het grote aantal alternatieve datareductie procedures is dit voorlopig een eerechte houding.
   (ACR-NEMA Digital Imaging and Communications Standard; draft November 19, 1984)

11. De internationaal gevestigde behoefte om veel TV kanalen te kunnen kiezen, belemmert de doortraak van echte HiFi omroep-tv televisie.

12. Het gepraat en geschrijf over arbeidsduurverkorting verschaf veel mensen werk.