CCD cameras for medical X-ray imaging

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ABSTRACT

A $1k^2$ CCD camera for medical X-ray imaging is studied, using a dual sensor concept where an optical interlacing with a mechanical accuracy better than 1 µm is obtained. Due to the fact that post-processing such as MTF restoration, edge enhancement and image subtraction (making only the blood vessels visible by subtracting images with and without X-ray contrast medium) has to be applied, an overall *"channel mismatch correction"* with an accuracy of about **0.2 %** is required and is realised. An experimental $2k^2$ acquisition system was built to evaluate a $2k^2$ CCD for medical applications. Important sensor parameters w.r.t. image quality such as MTF, noise and sensitivity were determined. The results of the experiments are presented.

INTRODUCTION

A wide range of CCD's are commonly used in consumer and professional imaging, in the area of camcorders, security, multi-media and broadcast cameras. In scientific imaging such as astronomy, CCD technology also has proven its merits. In medical X-ray imaging, where image intensifiers together with pick-up tube cameras or CCD cameras (fig. [1]) are being used, the requirements for a CCD based X-ray camera are high and diverse. Depending on the application there is a need for

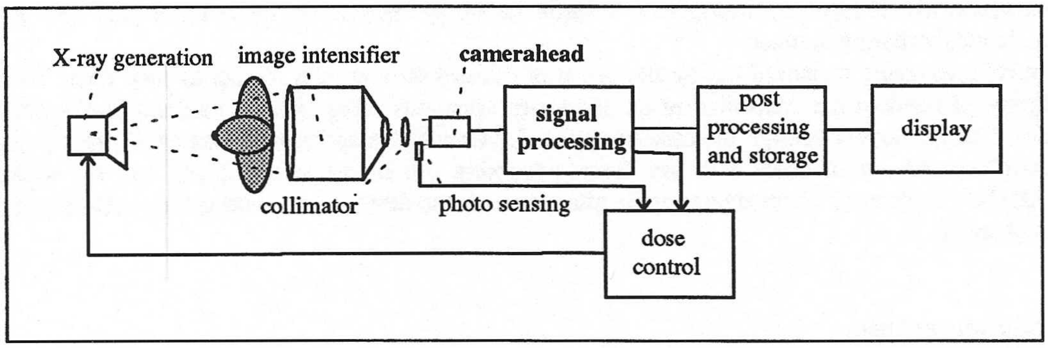

fig 1: Main functions of a IITV based X-ray system

various image formats such as standard or high resolution and low, normal or high frame rates (up to 60 fps), horizontal and vertical image reversal and progressively scanned or interlaced acquisition and display. With tube technology these image formats and sizes can be changed very easily by means of changing the horizontal and vertical scan frequencies, scan directions and scan magnitudes. Concerning the latter, CCD technology does not provide that flexibility. Nevertheless, the advantages of CCD's are such that the mentioned restrictions could not prevent the introduction of the solid state sensor in some fields of medical X-ray diagnostic imaging. A very important aspect in an X-ray imaging system is the performance in terms of *sensitivity, dynamic range, signal to noise ratio, MTF* and *smear.* E.g. in low dose applications (fluoroscopy) a high device sensitivity and low readout noise must ensure *X-ray quantum limited* operation.

1K² CCD CAMERA ARCHITECTURE

General:

As can be seen in fig.[2] the camera system consists of a small camera head and a separate camera control and correction unit. The camera head, which is mounted on the examination stand, incorporates two optically interlaced Philips FT12 /ref.1/ CCD-sensors (1024(h) x 512(v) pixels), iris and ND-filter, analog noise reduction circuits and an analog multiplexer to switch between the transmission and the reflection channel. The camera head is connected through a cable with a length of about 30 meters to the control and correction unit. The opto mechanics part and the correction part of the camera will be discussed

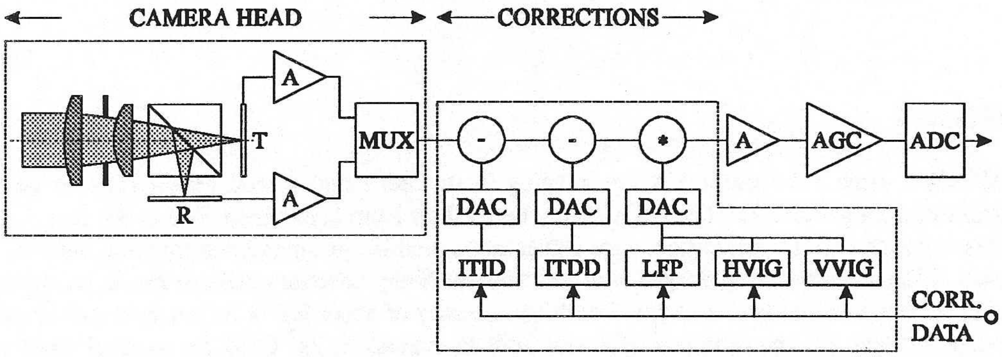

fig 2 Camera head and corrections.

Opto mechanics

The basis of the detector is provided by a CCD camera head equipped with two image sensors by which the optical image is de-multiplexed by geometrical beam splitting in the horizontal direction of the sensors.(fig [2]) The images are detected by the two FT12 sensors of 1024 pixels of horizontal resolution and 512 pixels in the vertical direction. A vertical shift of a half pixel pitch provides proper interleaving. The potentials of the four phase charge transport system of the CCD, a frame transfer CCD, are set such that during signal integration the height of the pixel is virtually halved, providing meaningful modulation at the Nyquist frequency of the resulting vertical 1024 pixels. After integration, the charges are transported towards the storage part of the CCD's and read out line by line in an alternating manner.

Optical interlacing is carried out by alignment of the two sensors with respect to each other. The 12 degrees of freedom are controlled in an alignment apparatus using fiducial markers on the CCD's, which can be viewed through the camera optics. Sensor rotation and registration (6 degrees) can be controlled with an accuracy of 1 μm. Sensor focusing can be set within .5 μm and tilt within 5 ARCMs. (6 degrees) These figures apply after securing the devices with joining rings containing UV cured epoxy.

Image corrections:

Image corrections have to be carried out very accurately due to the fact that long CCD integration times are used and extensive post-processing is done (MTF restoration, edge enhancement, zoom and image subtraction): The required (and achieved) accuracy is **0.2%.** Using the two sensor concept **no** pixel or column defects have to be accepted and corrected.

A special architecture has been chosen with analog image corrections and digital measuring functions and correction memories. In this way the complete ADC range is always available for the medical information and no ADC range is spoiled with dark current (especially at long integration times) and sensitivity differences.

All corrections use the same calibration method. The calibration process is iterative with an increasing K-factor during the correction calculation for noise reduction. Due to the iterative process tolerances on transfer functions are also compensated.

Following real time (36 MHz) image corrections are done (refer to *fig 2*):

1. Integration time independent dark current correction (ITID). Offset correction on pixel basis. Corrects for dark current from the CCD storage part and offsets in the analog electronics. Runs automatically between X-ray examinations so it also compensates for temperature drifts from the dark current.
2. Integration time dependent dark current correction (ITDD). Offset correction on pixel basis. Corrects for dark current from the CCD image part and is scaled for the actual integration time. Runs automatically between X-ray examinations so also compensates for temperature drifts from the dark current.
3. Light fixed pattern correction (LFP). Gain correction on pixel basis. Corrects for CCD sensitivity non-uniformity and gain differences between the two CCD's..
4. Horizontal vignetting differences correction (HVIG). Gain correction on line basis and dependent on iris aperture. Corrects for differences in horizontal vignetting between the two channels.
5. Vertical vignetting differences correction (VVIG). Gain correction on column basis and dependent on iris aperture. Corrects for differences in vertical vignetting between the two channels.
6. Temperature dependent gain correction. Gain correction on image basis. Corrects for drift in the CCD sensitivity as a function of temperature and wavelength. Partly based on information from the medical examination. Runs automatically between X-ray examinations.

Corrections 1 and 2 are realised in hardware with high gain settings,
Correction 3 is realised in hardware with image shift functions and measuring and calculation routines in software
Corrections 4, 5 and 6 are realised in hardware with measuring and calculation routines in software.

AN EXPERIMENTAL $2K^2$ CCD CAMERA

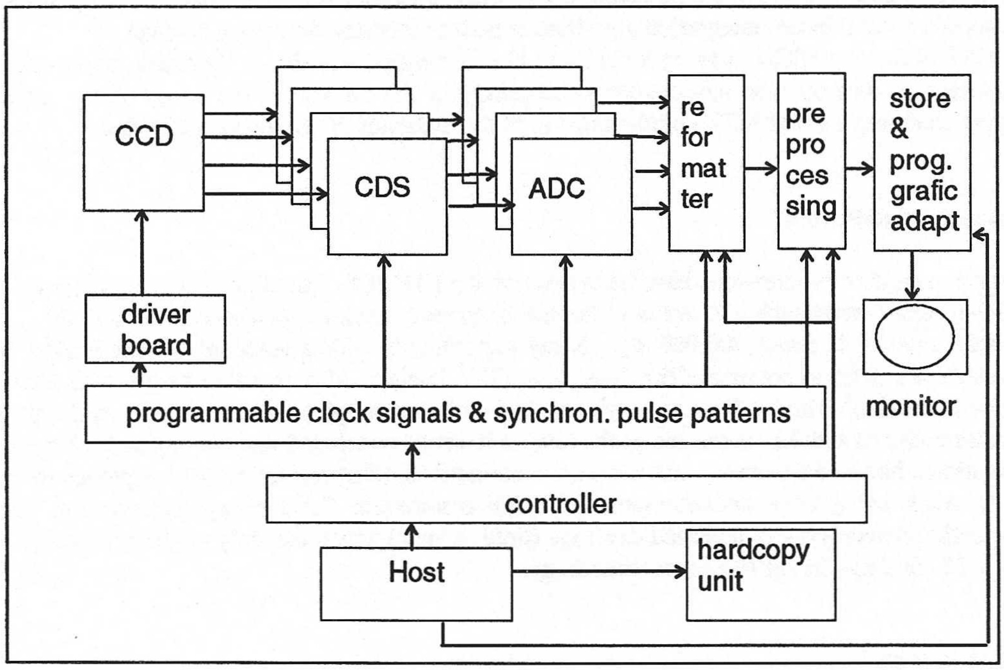

Fig. 3: Architecture of the digital camera system

A digital camera (fig.3) for experimental use has been designed /ref.2/. The camera has been used to evaluate various high resolution sensors. One of the sensors which was evaluated was the FTF2020 /ref.3/ from Philips Professional Imaging. This sensor has four output channels. Four CDS circuits and four ADC's are applied. In some applications the channels must be read out in parallel to have a high frame rate. In that case the reformatter takes care of the image reconstruction from the four sub images. Clock signals can be programmed in software so that various readout modes can be generated.

SENSOR REQUIREMENTS

$1k^2$ camera.

As mentioned before two modes of operation can be distinguished: on the one hand fluoroscopy, or low dose continuous imaging, where the patient X-ray dose should be minimal and a high intensity exposure on the other hand, in which the dose per image is set such that a minimal input quantum signal to noise ratio is guaranteed at an acceptable patient dose level. These two tasks require specific properties of the image sensor: In order to maximise the signal during fluoroscopy, the quantum efficiency should be high and preferably be matched to the spectral content of the source, i.e. the output phosphor of the image intensifier. The phosphor used, P20, has its maximum emission at 530 nm and fits the spectral sensitivity of the frame transfer CCD very well. Maximising the flux through the optical system is achieved by opening the aperture of the camera lens to an F number close to 1.2. With this relative aperture a trade off between correction of lens aberrations and optical throughput is set for this particular design. A large back focus provides space to accommodate a beam splitting prism . The multi layer coating between the two prism halves is designed such that minimal illumination differences between the reflection and transmission images exist, while maintaining a low absorption level. Anti reflection coatings on all the glass air transitions complete the actions to end up with a high optical throughput. The low dose of 1 µRö / frame then results in 1500 generated electrons/pixel per frame. As a S/N ratio of minimum 34 dB referred to the Nyquist bandwidth is required, the total read out noise should be below 30 electrons, which has been achieved indeed. In case of an exposure, the dose / image is more than two orders of magnitude higher. X-ray quantum limited detection requires then a sufficient number of generated electrons per absorbed X-ray quant in the scintillator. Usually a factor of 10 will do. For these high doses we are dealing with about 1500 absorbed quanta / pixel per frame for the average signal magnitude. For peak values one should take into account a factor of 4 with respect to this average value: 6000 absorbed quanta. The CCD should be able to generate more than 60000 electrons then. The maximum linear range being 70000 for a FT 12, the condition of more than 10 electrons/absorbed quantum has been met. For very high dose exposures a special image integration algorithm is used to increase the dynamic range.
The MTF of the total CCD imaging system has been measured and shows noticeable improvement in comparison to pick-up tube based systems especially in the corners of the image field. The total imaging chain has a lower MTF mainly because of the influence of the image intensifier.

$2k^2$ sensor experiments:

Sensor parameter measurements have been done on the FTF2020: Quantum Efficiency amounts 26 % for green light (wavelength 530 nm) and the full bandwidth readout noise amounts about 60 e_- . The full-well capacity is about 200.000 e_-. X-ray experiments with a standard 38 cm Philips Image Intensifier are done to compare $(2k)^2$ images to $(1k)^2$ images. Also standard screen-film have been compared to $(2k)^2$ images. Exposures of technical and medical phantoms have been analysed. From the measurements and the evaluation of the images it can be concluded that for typical high-resolution applications, like small vessels - and skeleton examinations, $(2k)^2$ resolution only improves the image quality when using edge enhancement and MTF restoration. Several applications (e.g. skeleton examinations) may require increased coverage (field of view) which can only be obtained using larger II's (> 38 cm diameter) or flat panel technology.

REFERENCES

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2. H. Reiter, A.J.C. Bruijns, N. Jung, A. Morgenstern, G. Spekowius: "A digital high-resolution, high-performance CCD camera with 2048 * 2048 pixels for dynamic applications" *SPIE Electronic Imaging 1997*

3. FTF20 data sheets, Philips Professional Imaging