Abstract
This study investigates the exposure parameters and required x-ray tube output when performing neurological procedures on a hybrid Angio-MR concept system proposed by Siemens Healthineers. The x-ray part of this system uses a longer source to detector distance than conventional (C-arm) systems and will have a fixed amount of filtration. Additionally, as the x-ray source is situated inside a magnetic field, the focal spot size and shape may be slightly distorted. In order to compare the Angio-MR system to a typical C-arm system, the exposure parameters of 60 thrombectomy procedures, performed in our hospital over the course of one year, were investigated in detail and a set of median values was determined. An analytical simulation platform was then developed to calculate the required tube voltage, tube current and pulse length to reach similar spatial frequency dependent signal difference to noise ratio (SDNR(u)) values as a conventional C-arm angiography system. These simulations were performed for a variety of focal spot sizes for the Angio-MR system. Results show that a standard current x-ray tube has sufficient power to reach similar SDNR(u) values as obtained in a conventional system if the focal spot size between both systems is comparable.

1. Introduction
Interventional radiology procedures have evolved greatly over the years. As these procedures have gained in sophistication, so has the complexity of the x-ray angiography systems used to perform the examinations. The strength and principal clinical utility of interventional angiography systems lies in their ability to deliver finely detailed, dynamic sequences of images of anatomical structures. These aid the interventional radiologist in the positioning and manipulation of guidewires and devices such as stents during therapeutic interventions. The principal aim is that of diagnosing and treating a wide range of diseases and one option to further improve the efficacy of radiology interventions is to integrate x-ray angiography imaging with magnetic resonance (MR) imaging. This can be either as two separate modalities located in the same imaging suite (Vogl et al 2002) or as a hybrid device (Fahrig et al 2008). This brings together functional imaging—provided by the MR unit and useful for diagnosis—with dynamic x-ray imaging used in the treatment stages.

Combining these imaging modalities is expected to especially beneficial in the treatment of stroke, where rapid diagnosis and treatment is essential (Khatri et al 2009). The difficulty in performing fast and efficient emergency procedures for patients suffering from a stroke lies precisely in this difference between the imaging methods used for diagnosis and treatment. Diagnosis is typically made using angiographic imaging via magnetic resonance imaging (MRI) or by computed tomography (CT) to establish the presence of a blood clot or whether a vessel is bleeding in the brain. Once diagnosed, x-ray guided neuro-angiography is used to treat stroke in a number of ways. Medication can be delivered at or near the clot site to dissolve clots, balloons and stents can be used to improve blood flow while embolization coils are used to control or stop
bleeding. The use of two modalities, perhaps in different departments or sections of the hospital, and different healthcare teams can lead to delays in patient treatment. The main benefit of a hybrid system would be to reduce the time between diagnosis and treatment significantly, which is essential in the recovery of the patients (Khatri et al 2009).

The European Institute of Innovation and Technology (EIT) has funded the ‘P3-Stroke’ project with the aim of improving image-guided, minimally invasive treatment of stroke and cardiac arrhythmia, two diseases that affect a large population group. Flat-panel angiography is to be combined with MR imaging into a single hybrid imaging modality to guide minimally invasive treatment. P3-Stroke comprises an international and interdisciplinary consortium including radiologists, cardiologists, computer science engineers, medical physicists and engineers from University Clinics Erlangen, Bordeaux University Hospital, Friedrich-Alexander University Erlangen-Nuremberg, KU Leuven, Coimbra Health School and Siemens Healthineers. These teams have undertaken to develop and deploy new medical imaging technology with two aims. The first is to speed up the clinical workflow for endovascular treatment in acute stroke management and second to increase the safety and effectiveness of radiofrequency ablation treatment for cardiac arrhythmia.

When integrating the x-ray imaging system into the MR system we are faced with a number of challenges, such as the correct operation of the x-ray tube and x-ray detector within the strong magnetic field of the MR device (Fahrig et al 2008). Another important aspect relates to the x-ray factor selection of the automatic dose rate control (ADRC) system of the angiography unit (Lin 2009, AAPM 2012), which plays a crucial role in setting the balance between image quality and patient dose (Jones et al 2014, Dehairs et al 2017). The ADRC selects the appropriate x-ray control parameters (for example, tube voltage, tube current, pulse length, focus size, spectral pre-filtration) in response to changes in patient anatomical thickness and x-ray projection angle. There are a number of differences between a conventional C-arm angiography system used for interventional work and the proposed Angio-MR x-ray system setup expected to influence the ADRC/x-ray tube operation. These include the geometry, the filtration of the x-ray beam and the potential influence of the magnetic field on the x-ray focal spot.

Figure 1 gives a graphical representation of a standard C-arm and a proposed geometry for the Angio-MR device. The specified source to object distance (SOD) is assumed to be 136 cm while source to detector distance (SID) is 176 cm for the Angio-MR system. Typical values of SOD and SID for standard C-arm device are respectively 75 cm and 105 cm, depending on patient thickness and current angulation/view. The added filtration of current C-arm systems is often flexible and typically consists of Cu and/or Al. The Angio-MR system, on the other hand, will most likely use either less or no additional Cu; the spectral shaping will come from the entrance and exit windows in the MRI magnet that will have to be created to allow the x-rays to reach the patient and detector. Additionally, since the x-ray tube of the Angio-MR system
is located inside the magnetic fringe field, the focal spot, where electrons are hitting the anode, will be distorted to some degree. The degree of this distortion will depend on the local magnetic field at the position of the x-ray tube and the interaction of the tube electron beam with this magnetic field (Wen et al 2007).

The goals of this study were therefore (1) to calculate the exposure parameters that the hybrid Angio-MR system would have to use in order to achieve the same technical image quality as a conventional C-arm angiography system, (2) to assess whether a current x-ray tube would be able to produce the necessary output to reach this level of image quality and (3) simulate the effect of the increased focus size on a measure of technical image quality. This was done via a simple narrow beam simulation platform of both imaging systems. Calculations in this study focused on neurological procedures, the main clinical application investigated in the P3-Stroke project.

2. Simulation platform

2.1. Modelling the x-ray system

A simulation platform was created using the programming language code Python, which gave the signal difference to noise ratio (SDNR) of a small metal object (to simulate a catheter or stent) and the incident air kerma rate (K_{air}) at the entrance of the patient. SDNR is defined using the following equation:

\[
\text{SDNR} = \frac{|I_{\text{obj}} - I_{\text{bkg}}|}{\sigma_{\text{bkg}}}
\]  

where \(I_{\text{bkg}}\) and \(I_{\text{obj}}\) are respectively the signal intensities measured in a region of interest within the background and within the object of interest, and \(\sigma_{\text{bkg}}\) is the standard deviation measured within a region in the background.

The first step was to obtain a set of measured reference air kerma data for tube voltages from 50 kV to 120 kV and copper pre-filtration ranging from 0.0 mm to 0.9 mm for an angiography system. The reference system used was a Siemens Artis Q angiography system at our hospital, with a measured inherent tube filtration of 3.3 mm of Al. A calibrated Piranha lead-backed solid-state detector (RTI Electronics AB, Malmö, Sweden) was used for these measurements, with a reference point defined at a distance (d_{ref}) of 86 cm from the source (30 cm from the detector cover for a conventional C-arm with a 120 cm SID).

Next, the method described by Boone and Seibert (Boone and Seibert 1997) was used to generate tungsten (W) anode spectra for a given tube voltage (kVp) to give an initial spectrum defined as \(\varphi_0(E)\) photons/mAs/mm^2. The Lambert–Beer law was used to apply the 3.3 mm Al inherent filtration to this initial spectrum. This spectrum was then scaled such that the air kerma (K_{air}) calculated from this spectrum matched the air kerma measured at the reference point for the reference system, as follows. First, the curve describing the exposure in air (units mR) per x-ray photon (Boone and Seibert 1997) was divided by a factor of 8.76 to give a curve describing the air kerma (units \(\mu\)Gy) per x-ray photon \(\kappa(E)\) (International Atomic Energy Agency (IAEA) 2007). This was then multiplied by the number of photons \(\varphi_0(E)\) and integrated over the energy spectrum to give the air kerma K_{air}:

\[
K_{\text{air}} = \int_{E=0}^{E_{\text{max}}} \varphi_0(E) \kappa(E) \, dE. \tag{2}
\]

K_{air} calculated at a given tube voltage using formula (2) was divided by the measured air kerma and a second order polynomial curve fitted through these ratios as a function of kVp was determined. This resulted in a calibration function CF for the K_{air} as a function of tube voltage. This calibration function gives the calibration factor that scales the spectrum \(\varphi_0(E)\) at each kVp such that the calculated K_{air} matches the measured air kerma. Applying the CF to the generated spectrum then gave the final spectrum exiting the x-ray tube, \(\varphi_1(E)\):

\[
\varphi_1(E) = \varphi_0(E) \times \text{CF}(kVp) \tag{3}.
\]

Comparison between measured reference K_{air} and simulated values showed good agreement (< 10% deviation) for low Cu thickness values (< 0.3 mm) and higher (≤ 20%) deviations for the thicker Cu values. However, the accuracy of the simulated reference K_{air} was sufficient for the (low) Cu thickness values used in this work.

In the following step, the effect of any additional Cu (\(x_{Cu}\)) and or Al (\(x_{Al}\)) pre-filtration on \(\varphi_1(E)\) was calculated using the linear attenuation coefficients of copper (\(\mu_{Cu}\)) and aluminium (\(\mu_{Al}\)) (Boone and Chavez 1996) and the Lambert–Beer law:

\[
\varphi_2(E) = \varphi_1(E) \times e^{-(\mu_{Cu}(E)x_{Cu})} \times e^{-(\mu_{Al}(E)x_{Al})}. \tag{4}
\]
In addition to inherent 3.3 mm Al tube filtration, further sources of filtration were added including the patient table (1 mm Al equivalent), a 20 cm poly(methyl) methacrylate (PMMA) phantom and in the case of the Angio-MR system: the magnet, radio-frequency (RF) body coil and local receive coils to calculate the total attenuation in the background of the image. Based on estimates for x-ray transparent vacuum windows, the filtration of the x-ray path penetrating through the magnet windows of the MR system and the local and body coil was estimated to be an aluminium equivalent thickness of 7.4 mm. However, it should be noted that the actual thickness and distribution of different materials determining the pre-patient filtration is not yet fixed. Of importance is the composition of these materials, as this could influence the SDNR results. Materials in the beam will include aluminum itself (in the magnet/windows) plus some materials containing low atomic number elements such as plastics, polycarbonates and composites such as Kevlar. There may be a very thin layer of copper but this is likely to be just a few microns thick. We therefore consider that using aluminium equivalence is a reasonable approach for the SDNR modelling. In order to calculate a measure of image quality, a 0.36 mm thick iron object (diameter of a typical guidewire) was added to calculate an SDNR value, resulting in the spectra:

\[ \varphi_3(E) = \varphi_2(E) \ e^{-\left(\mu_{\text{bkg}}(E) x_{\text{bkg}}\right)} \]  

(5a)

\[ \varphi_4(E) = \varphi_2(E) \ e^{-\left(\mu_{\text{bkg}}(E) x_{\text{bkg}} + \mu_{\text{Fe}}(E) 0.36\right)} \]  

(5b)

with \( \mu_{\text{bkg}} \) and \( x_{\text{bkg}} \) the linear attenuation coefficients and thicknesses of the materials in the background and \( \mu_{\text{Fe}} \) the linear attenuation coefficient of iron. In the next step, the spectrum absorbed by the detector was calculated. The detector was defined as a Caesium Iodide (CsI) layer with a thickness \( x_{\text{CsI}} \) of 600 \( \mu m \), a density of 4.51 \( gcm^{-3} \) and a packing fraction (PF) of 0.8, giving an effective density of 3.6 \( gcm^{-3} \) (Zhao 2004, Yorkston 2007). Additionally, the spectra were scaled to the correct distance to the detector, using the inverse square law: \( (d_{\text{ref}}/\text{SID})^2 \). This gave the following spectra for photons absorbed by the detector:

\[ \varphi_5(E) = \varphi_3(E) \left( \frac{86}{\text{SID}} \right)^2 \left(1 - e^{-\left(\mu_{\text{CsI}}(E) x_{\text{CsI}} PF\right)}\right) \varepsilon(E) \]  

(6a)

\[ \varphi_6(E) = \varphi_4(E) \left( \frac{86}{\text{SID}} \right)^2 \left(1 - e^{-\left(\mu_{\text{CsI}}(E) x_{\text{CsI}} PF\right)}\right) \varepsilon(E) \]  

(6b)

with \( \varepsilon(E) \) the absorption probability (fraction of the attenuated photons that gets absorbed) of the primary x-ray photons in the CsI layer.

Finally, the SDNR was calculated using the energy absorbed in regions of the detector behind and adjacent to the object (i.e. the background). The energy in the region in the background \( (E_{\text{bkg}}) \) and object region \( (E_{\text{obj}}) \) were calculated using:

\[ E_{\text{bkg}} = \int_{E=0}^{E_{\text{max}}} \varphi_5(E) \ E dE \]  

(7a)

\[ E_{\text{obj}} = \int_{E=0}^{E_{\text{max}}} \varphi_6(E) \ E dE \]  

(7b)

while the variance in the background region is given by (Zamenhof 1982):

\[ \sigma_{\text{bkg}}^2 = \int_{E=0}^{E_{\text{max}}} \varphi_5(E) \ E^2 dE. \]  

(7c)

The SDNR was then finally estimated using:

\[ \text{SDNR} = \frac{|E_{\text{bkg}} - E_{\text{obj}}|}{\sigma_{\text{bkg}}} . \]  

(8)
2.2. Defining image quality and system efficiency

The standard definition of SDNR contains only a measure of large area signal difference or signal contrast (Chakraborty 1989) and does not include any information on contrast transfer as a function of spatial frequency. When imaging small structures such as vessels and interventional equipment such as guidewires and stents, which may or may not be moving, the spatial and temporal resolution of the imaging system are important components of the final image quality. Standard SDNR is not sensitive to the influence of focus unsharpness and motion blurring and therefore a version of SDNR adjusted for the influence of these blurring sources with a method proposed by Bernhardt et al (Bernhardt 2005, Dehairs et al 2017) was used.

To calculate SDNR(u) (the spatial frequency dependent SDNR) the approach of (Bernhardt 2005) was followed in which presampling modulation transfer function (MTF) based weighting factors are used to describe the influence of blurring from the x-ray focal spot (MTF\textsubscript{focus}) and object motion (MTF\textsubscript{motion}) on SDNR. The MTF-based weighting factors are evaluated at a spatial frequency relevant to the task under consideration noted as \(u\text{\textsubscript{task}}\), resulting in a spatial frequency dependent SDNR(u). The frequency is calculated from the relationship \(u\text{\textsubscript{task}} = \frac{1}{2d}\) where \(d\) is a characteristic length in the object. For example, for a guidewire of diameter of 0.36 mm, \(u\text{\textsubscript{task}}\) is defined to be \(1/(2 \times 0.36 \text{ mm})\) which is approximately \(1.4 \text{ mm}^{-1}\) in the reference plane of the object. The use of a large focus or long exposure pulses gives low MTF values at \(u\text{\textsubscript{task}}\), due to focal spot blurring and motion blurring, reducing SDNR(u) and penalizing the use of inappropriate focus size or pulse length in ADRC factor selection. Depending on the shape of the focal spot, two main functions are used to model focal blurring. For Gaussian shaped foci, a Gaussian function is used (the Fourier transform of a Gaussian function remains Gaussian), while for a more rectangular shaped focus, a sinc function is appropriate (Fourier of a rectangular function is a sinc function) (Doi 1982). Figure 2 shows two examples of 1-dimensional line profiles of typical focal spots. The focus on the left can be modeled using a Gaussian function while the focus on the right is much more rectangular shaped. The 1D version of SDNR(u) is then given by:

\[
SDNR(u) = SDNR \cdot MTF\textsubscript{focus}(u) \cdot MTF\textsubscript{motion}(u) \tag{9}
\]

where SDNR is the standard large area signal contrast, which is calculated using equation (8) and \(u\) is the spatial frequency. The spatial frequency has to be defined in a specific plane, this is typically the detector plane, denoted \(u_d\), or the object plane, \(u_o\). Spatial frequencies defined in the object plane can be referred to the detector plane by \(u_o = \gamma u_d\). The focal blurring can be described using either a Gaussian function (10) or a sinc function (11):

\[
MTF\textsubscript{focus}(u_o) = e^{-1.3\left(\frac{(\gamma - 1)}{\gamma}\right)^2 u_o^2 a_o^2} \tag{10}
\]

\[
MTF\textsubscript{focus}(u_o) = \text{sinc}\left( a_o \frac{(\gamma - 1)}{\gamma} u_o \right) \tag{11}
\]

and motion blurring can be described as:

\[
MTF\textsubscript{motion}(u_o) = e^{-1.3(b_x u_o)^2} \tag{12}
\]

\[
MTF\textsubscript{motion}(u_o) = \text{sinc}(b_x u_o) \tag{13}
\]
where \( a_x \) is the frequency of interest defined in the object plane, \( y \) is the magnification factor: (SID/SOD), \( b_x \) is the focal spot size (in one direction), \( v_x \) is the object velocity (in one direction) and \( t \) is the pulse length.

The derivation and use of these MTF weighting factors is described in (Dehairs et al 2017).

The incident air kerma at the entrance of the patient was calculated using equation (2), after correcting for the appropriate filtration and distance (the spectrum is only calibrated correctly at the reference distance: 86 cm). For the C-arm system, filtration (Cu/Al) and table attenuation were included and the distance was adjusted such that the entrance of the 20 cm thick PMMA phantom was 65 cm from the source. For the Angio-MR modelling, the spectrum was attenuated by the magnet windows and coils but not by the patient table, as an over-couch configuration is proposed.

Using SDNR\((u)\) and the incident air kerma, a figure of merit (FOM) was calculated giving a measure of system efficiency (Zamenhof 1982):

\[
FOM = \frac{SDNR^2(u)}{EntranceAirKerma}.
\]  
(14)

Scattered radiation was excluded from this simulation and this was justified as follows. A number of parameters control the fraction of scattered radiation relative to primary radiation (scatter fraction, SF), at the exit of the scattering object. The thickness of the object has a strong influence on SF (Dick et al 1978, Aichinger et al 2012), however we are mainly concerned with projections around the head and we assume an approximately isotropic thickness. Second, the field of view (FOV) affects SF magnitude, although above ~200 cm\(^2\) SF only varies by approximately 10% (Dick et al 1978, Aichinger et al 2012), at typical diagnostic radiology energies. x-ray energy has a small effect on SF, introducing a variation of roughly 5% as tube voltage increases from 60 kV to 120 kV (Aichinger et al 2012). The FOV and object thickness are fixed and the energy is varied in these simulations and thus it is reasonable to assume that the SF at the exit of the phantom will not change dramatically as energy changes. Furthermore, we are assuming that an antiscatter grid is in place and this will limit the quantity of scattered radiation contributing to the image. A typical grid for a C-arm used for interventional radiology at our centre has strip density of 80 lines cm\(^{-1}\), and a ratio of 15:1. The data of Mizuta et al (Mizuta et al 2012) give a value of 0.060 for the scatter transmission (T\(_s\)) of a grid with strip density of 60 cm\(^{-1}\) and ratio 14:1, for a 20 cm PMMA phantom. We can therefore expect ~6% of the scattered radiation exiting the phantom to pass through the grid to the image. In order to test these assumptions, SDNR was measured on the Artis Q system as a function of x-ray tube voltage, tube current, the spectral pre-filtration (i.e. Al and Cu thickness) and SID, using a method described previously (Dehairs et al 2017). An additional set of measurements were carried out to measure PMMA thickness was varied from 7.5 cm to 37.5 cm and the x-ray control parameters were set by the ADRC. The measured SDNR was then compared to the simulated values.

3. Materials and methods

To verify the simulation platform, images were acquired of a small iron sheet (2 by 2 cm), placed at the centre of 20 cm of PMMA (i.e. 10 cm PMMA—object—10 cm PMMA). The images were obtained on a Siemens Healthineers Artis Q C-arm system, located at the interventional radiology centre in our hospital. The following parameters were varied: tube voltage, tube current, pulse length, Cu filtration, Al filtration, source to detector distance and phantom thickness, and the SDNR was measured for each of the images. The anti-scatter grid was used during all the measurements. A linear look up table was set and no clinical image processing—just the standard detector offset, gain and defect pixel corrections were applied (Dehairs et al 2017). Afterwards, the setups of these measurements were simulated and the calculated SDNR values were compared to the measured results.

After the verification steps, the setup of the conventional C-arm system and the hybrid Angio-MR system were simulated and the exposure parameters required by the Angio-MR to achieve the same SDNR\((u)\) as the C-arm setup were calculated. To determine a typical set of exposure parameters for a conventional system, the acquisitions of 60 thrombectomy procedures were studied in detail using the online dose-monitoring platform installed in our hospital (Qaelum NV ). Thrombectomy procedures are routinely performed on a Philips Allura Clarity FD20, an up to date system considered to deliver excellent image quality by the interventional radiologists at our site. Exposure parameters for each image in the fluoroscopy and acquisition sequences taken during the 60 procedures were summarized in histograms. From this, median exposure parameter sets were determined. These parameters were then set on the Siemens Healthineers Artis Q system and the SDNR of the small iron object was measured for the C-arm setup. Next, the MTF-based weighting factors were applied (equations (8)–(12)) on the standard SDNR. In this study, the frequency of interest was fixed at 1.4 cycles mm\(^{-1}\), derived from the dimension of a typical guidewire: 1/(2 \times 0.36 mm) (Bernhardt 2005). This study focuses on neurological procedures in which the patient must not move during the DSA procedures.
acquisitions of the head otherwise there are strong misregistration artefacts generated in the subtraction image. This in turn implies that temporal blurring from motion of arterial structures or guidewire/devices is not a limiting factor for neurological procedures. In the modelling of the task for the system we have therefore assumed a velocity of 1 mm s\(^{-1}\). As mentioned before, a 20 cm thick PMMA phantom was simulated to represent a patient head (Ng et al 2018). Finally, two focal spot sizes were assessed: one scenario with a small focal spot of 0.7 mm and one with a large focal spot of 1.4 mm. These sizes are based on typical focal spots with nominal sizes of 0.4 and 0.8 for the small and large focal spot respectively, which according to IEC 60 363 corresponds to: F0.4 = 0.60 × 0.85 mm\(^2\) and F0.8 = 1.20 × 1.60 mm\(^2\) (International Electrotechnical Commission (IEC) 1993).

From the starting exposure parameters and the target SDNR(u) value for the conventional c-arm system, an algorithm was created that systematically varied the tube voltage, tube current and pulse length in order to calculate SDNR(u) for the Angio-MR setup. For a given tube voltage, the tube current is systematically increased. If the target SDNR(u) is not achieved before the maximum permitted tube current is reached, the pulse length is increased, the tube current is reset to the starting value and then systematically increased again. If the target SDNR(u) is not met below the limit set for tube current-exposure time product (i.e. mAs) then the tube voltage is increased and the tube current and pulse length reset to initial values. The process is then repeated. This is a simple algorithm that searches the tube voltage, tube current and pulse length space for the lowest set of exposure parameters that deliver the requested SDNR(u), prioritizing low tube voltage. Increasing the tube current first, prior to tube voltage increase, keeps image quality high as increasing the tube voltage reduces object contrast (Lin 2009, AAPM 2012).

The maximum value of mAs that can be set will vary depending on the specific tube power rating and the required frame rate for the application. Therefore, several limits for tube current multiplied by pulse length were examined, as this affects the final tube voltage and the required tube output. There are a number of effects that may occur when the x-ray tube and the flat panel x-ray detector are operated in the magnetic field of the MRI system. First, there is an alternating current flowing through the tube filament, which in the presence of a strong magnetic field, may cause the filament to vibrate (Fahrig et al 2005), leading to a short circuit if the filament touches the focusing cup. The magnetic field could potentially deflect the electron beam, missing the anode track and causing misalignment (Fahrig et al 2005). There is also the potential for distortion of the focal spot shape and an increase in the size of the focus, just in a single direction or in both the horizontal and vertical directions (Fahrig et al 2001, 2005). Early work, albeit for a different geometry, has shown that there was no motion of the focal spot and the focus area increased by ~17% (Fahrig et al 2001). Later work on another system found an increase of ~100% in focus size (Ganguly et al 2005). The main concern of the system designers at Siemens is the increase of the focus size and the influence this will have on system sharpness. The electron beam will be shielded in order to control the distortion of the size and shape. Initial measurements by Siemens Healthineers have verified that the mean focal spot diameters increase by a maximum of 10%, provided measures are introduced to shape the stray field of the magnet at the position of the electron beam in the x-ray tube. Therefore, this study has considered increases in x-ray focal length from 14% to 130% in the SDNR(u) calculation, which should cover the range of focus size change expected, depending on the degree of shielding possible in the final design.

For the conventional system, the added filtration was set at 0.1 mm Cu + 1 mm Al for acquisition imaging and 0.4 mm Cu + 1 mm Al for fluoroscopy imaging (standard values of the Philips system used in our hospital), while for the Angio-MR the added filtration was set at 7.4 mm Al (no added Cu). Once a set of exposure parameters was found with the same SDNR(u) as the target value, those parameters were recorded and a check was made whether the tube output could be achieved for a standard x-ray tube. Table 1 gives the serial load rating of a typical Siemens Healthineers x-ray tube (taken from the Megalix x-ray source user manual). In order to see which limiting values had to be used in the simulation, frame rates of 2fr/s and 10fr/s for acquisition and fluoroscopy were chosen respectively (typical values on the system used in our hospital). In some cases, the target SDNR(u) could not be reached with the set mAs limit, even at the limiting tube voltage of 125 kV. For these cases, the target SDNR(u) was decreased in steps of 5% until the value could be reached. This gave an idea of the reduction or compromise that may have to be made in switching the Angio-MR geometry compared to the C-arm unit.

4. Results

Figure 3 shows the correlation between the measured SDNR and the calculated value for different parameters. There are large absolute differences in the magnitude of the values, probably due to the absence of scatter in the simulation and the rather simple detector model assumed. However there is a strong linear correlation between the measured and simulated values. Looking at the relative changes in SDNR, the average deviation between the measured change and calculated change was 2.6% with a maximum value of
Table 1. These tables contain the limiting pulse power values in kW for serial loading of a Megalix x-ray tube. Top: values for a small focal spot. Bottom: values for a large focal spot. Data is taken from the Megalix x-ray source user manual.

| Series Duration | Small Focus | 1 s | 2 s | 4 s | 6 s | 10 s | 16 s | 20 s | 25 s | 40 s | 63 s | 100 s | 120 s | 600 s |
|-----------------|-------------|-----|-----|-----|-----|------|------|------|------|------|------|-------|-------|-------|
| Pulse length (ms) x fr/s (s^{-1}) | 50 | 30.0 | 30.0 | 30.0 | 30.0 | 30.0 | 30.0 | 30.0 | 30.0 | 29.9 | 29.3 | 26.0 |
|                 | 100 | 30.0 | 29.9 | 29.8 | 29.7 | 29.6 | 29.3 | 29.2 | 29.0 | 28.4 | 27.6 | 26.5 |
|                 | 150 | 28.9 | 28.8 | 28.6 | 28.5 | 28.3 | 27.9 | 27.7 | 27.4 | 26.6 | 25.6 | 24.1 |
|                 | 200 | 27.9 | 27.4 | 26.8 | 26.4 | 25.8 | 25.1 | 24.8 | 24.4 | 23.3 | 22.0 | 20.3 |
|                 | 250 | 27.0 | 25.8 | 24.5 | 23.8 | 23.6 | 22.7 | 22.2 | 21.7 | 20.4 | 19.0 | 17.2 |
|                 | 300 | 26.2 | 24.7 | 23.1 | 22.2 | 21.6 | 20.5 | 20.0 | 19.4 | 18.1 | 16.5 | 14.7 |

| Series Duration | Large Focus | 1 s | 2 s | 4 s | 6 s | 10 s | 16 s | 20 s | 25 s | 40 s | 63 s | 100 s | 120 s | 600 s |
|-----------------|-------------|-----|-----|-----|-----|------|------|------|------|------|------|-------|-------|-------|
| Pulse length (ms) x fr/s (s^{-1}) | 50 | 86.0 | 86.0 | 86.0 | 85.5 | 85.5 | 85.2 | 85.2 | 82.9 | 79.7 | 75.6 | 73.3 | 50.0 |
|                 | 100 | 86.0 | 86.0 | 86.0 | 84.3 | 83.8 | 80.3 | 78.6 | 74.0 | 68.9 | 62.5 | 61.1 | 30.0 |
|                 | 150 | 86.0 | 86.0 | 86.0 | 84.5 | 82.9 | 77.2 | 73.0 | 71.1 | 66.0 | 60.5 | 50.2 | 45.9 | 20.0 |
|                 | 200 | 79.7 | 77.4 | 74.8 | 73.0 | 73.0 | 66.6 | 64.7 | 62.6 | 57.2 | 50.9 | 38.4 | 32.9 | 15.0 |
|                 | 250 | 76.9 | 73.1 | 69.2 | 67.0 | 67.0 | 59.0 | 56.9 | 54.6 | 49.0 | 43.0 | 31.2 | 26.4 | 12.0 |
|                 | 300 | 74.0 | 68.7 | 66.5 | 66.5 | 60.0 | 53.2 | 51.0 | 48.6 | 43.0 | 37.2 | 26.3 | 21.6 | 10.0 |

Figure 3. Correlation between the measured SDNR on the x-axis and the calculated SDNR on the y-axis for varying several exposure parameters: kV, mAs, SID, Cu filtration and Al filtration. The graph on the bottom right contains all the measurements and indicates a slope of approximately 3. The open red circles show the data where the PMMA thickness was varied (along with the kV, mA and Cu filter).

8.7%. For example, the measured change in SDNR when increasing Al filter thickness from 0 mm to 6 mm was a factor of 0.71, while the calculated change was 0.69 giving a deviation of 3.2%. Changes in mAs, SID and filtration all had similar or smaller margins of error. Changes in tube voltage however were somewhat less accurate, e.g. the measured change in SDNR when varying the tube voltage from 60 to 90 was a factor of 1.46, while the calculated change was 1.34 giving a deviation of 8.7%. All the SDNR measurements are collected in the final graph in figure 3, with the open circles (shown in red) indicating data points where the PMMA thickness is varied (along with tube voltage, tube current and spectral filter according to the ADRC selection). This shows that the simulation can predict relative SDNR, even where the SF at the exit of object is changing strongly and is consistent with the grid limiting the quantity of scattered radiation in the image. The ratio between calculated and measured SDNR is almost constant at approximately 3. When this factor is accounted for, the deviation of the SDNR with the measured values is on average ~4%, with a maximum value of 14%. These results are consistent with scatter only having a small influence on relative changes estimated for SDNR as acquisition factors are changed.
Figure 4 shows the histograms of tube voltage, tube current and pulse length acquired from all the fluoroscopy and acquisition series of 60 thrombectomy procedures performed on the Philips Allura Clarity FD20. From these results, typical exposure parameter sets were defined which were used for calculating the SDNR(u) of the conventional C-arm and which served as a starting point for the Angio-MR exposure parameters. The Philips Allura Clarity also has nominal focal spot sizes of 0.4 and 0.7 for the small and large focus respectively, which according to the IEC coincides with permissible values of the focal spot dimensions between 0.6–0.85 mm and 1–1.5 mm respectively, being similar to the typical values chosen for this study. For acquisition imaging, the resulting starting set was: 75 kV, 285 mA and 45 ms while for fluoroscopy the starting point was: 80 kV and 60 mA (noted as normal dose) or 120 mA (noted as high dose) and 4 ms.

Looking at the average exposure time for fluoroscopy and acquisition, the analysis of the 60 patients showed an average total fluoroscopy time of approximately 900 s (15 min), with ten procedures close to 30 min and 2 cases over 60 min, and an average acquisition time per scan of 12 s, with the patients typically receiving around 10 acquisition series per procedure.

Table 2 shows the Angio-MR exposure parameters for an acquisition series starting from a small focus (2a) and a large focus (2b) assuming a rectangular shaped focal spot, and therefore equations (11) and (13) to describe the focus and motion blurring were used. Because the algorithm increases the tube current and pulse length values first to their respective maximum allowed values before increasing the tube voltage, several scenarios with different maximum values were investigated. The table contains the results for three scenarios of limiting mAs values to investigate the effect this has on the necessary exposure parameters and the resulting tube power. These scenarios were chosen theoretically. Additionally, the entrance air kerma rate (µGy/fr), the FOM and the necessary tube power (kW) are given. Table 3 shows the corresponding results.
assuming a Gaussian focus, using equations (10) and (12). Tables 4 and 5 show the results for fluoroscopy imaging for two different mA values: a typical value and a high one (double the typical value).

Figures 5(a) and (b) show the MTF weighting factors for focus size and motion for the Angio-MR system. Increasing focus size from 0.7 mm to 1.6 mm takes the MTF$_{\text{focus}}$ factor from 0.92 to 0.63 for a rectangular focus evaluated at 1.4lp/mm, which is a reduction of ~30%. This is a hypothetical worst-case scenario, where the difference between the conventional focal spot and the Angio-MR focal spot is extremely large. Figure 6 shows the effect of filtration and distance on the photon fluence for both systems. The effect of the different filtrations can be found in figures 6(a) and (c) for fluoroscopy and acquisition respectively. The effect of distance is added in figures 6(b)) and (d)).

5. Discussion

The simulations of the required exposure parameters for fluoroscopy (tables 4 & 5) show that for a normal dose image (simulation for 60 mA) using the conventional setup, the required power per frame for the Angio-MR system to match the SNDR(u) is approximately 9–10 kW for both focus shapes. Using 10 frames per second as a typical frame rate and 5 ms as a typical pulse length, the limiting output values should be taken from the first row of tables 1(a) or (b) (i.e. fr/s × ms = 50). Table 1 shows that this limiting value is 26 kW for a small focus and 50 kW for the large focus. Assuming all the focal spots below 1 mm can be placed in the ‘small’ category and everything above 1 mm in the ‘large’ focal spot category, even for a fluoroscopy run of 600 s the limiting value is not reached in the Angio-MR setup. Since the average total fluoroscopy time is approximately 900 s, but this is acquired over many fluoroscopy runs over an extended period of time, the Angio-MR system will be able to match the SDNR(u) of the conventional system for a normal dose fluoroscopy run. Looking at the results for the high dose fluoroscopy run, the required tube output lies closer to 20 kW. As the limit for a long fluoroscopy run is at least 26 kW, the Angio-MR should again be able to produce the necessary exposure parameters under these conditions. Even the calculation using the largest focal spot of 1.6 mm has a value below 30 kW. This calculation compares a scenario with a focal spot more than 2 times the size of the conventional system, which has a strong, negative impact on SDNR(u), and must be compensated by a much higher tube output. In a first step, this is done by increasing the tube current and pulse length, but if the set mAs limit is reached, the system must increase the tube voltage. Increasing the tube voltage while keeping the mAs fixed will lower the contrast, but increases the number of photons mm$^{-2}$ at the x-ray detector, which in turn reduces the relative quantum noise in the image. Assuming that the x-ray detector has sufficient dynamic range to cope with the higher signal level, signal to noise ratio is increased and therefore SDNR is increased (Dehairs et al 2019).

Figure 5(a)) demonstrates the effect of focus size and pulse length on the MTF weighting factors. The effect of focus size can clearly be seen, on the other hand changing the pulse length from 10 to 100 ms has only a small impact. Figure 5(b)) shows that the geometry of the hybrid system has a small advantage regarding the MTF weighting factors. Since the magnification (SID/SOD) is smaller for the hybrid system (1.29 compared to 1.4 for the conventional system), the MTF factors are higher for the same focus size, and have a smaller detrimental impact on SDNR(u), meaning the requirements on the exposure parameters of the Angio-MR system are less stringent. This is clear when comparing the necessary tube power between two identical Angio-MR focal spot sizes in tables 2(a) and (b), starting from the small or large conventional C-arm focus size. Additionally, the graph also shows that using the Gaussian blurring model results in a smaller penalizing effect of the focus size. At the larger focal spot sizes the sinc function decreases more rapidly than the Gaussian function, meaning the Angio-MR system would need to compensate for this by increasing the tube output. This can be seen in tables 2–5 where the power requirement for the Angio-MR system when assuming the Gaussian blur is lower compared to the sinc function blur.

The results for the acquisition runs are given in tables 2 and 3. Using a typical frame rate of 2 fr s$^{-1}$ and a pulse length between 50 and 100 ms, the limiting values can be found in rows 2 to 4 of table 1. As mentioned before, the average duration of an acquisition run was found to be approximately 12 s. Assuming some variability in this duration, the limiting values of interest when using a small focus lie between ~30 kW, for a short run using shorter pulses and ~24 kW, for a longer run (25 s) using long pulses, and values between ~86 kW and ~62.6 kW when using the large focus. The results clearly show that the use of a small focal spot will not be possible for acquisition imaging. None of the calculated values lies below 30 kW, which is the upper limiting value for a small focus. This immediately eliminates all of the results in the first three rows, with focus sizes: 0.7, 0.8 and 1.0 mm. A focus size of 1.2 mm could already be considered as ‘large’. When allowing the mAs to increase by factor of 3, no calculated exposure sets exceed the limiting values. However, as can be seen by the $\Delta$SDNR(u) column, using the largest focal spots means the target SDNR(u) cannot be achieved. In the scenario where the mAs can increase by a factor of 4, only the calculations using the largest focus sizes struggle to provide the tube output for the longer acquisition runs. Finally, allowing the
Table 2. Calculated exposure parameters for an acquisition series on the Angio-MR setup with a rectangular focal spot for 3 different limiting mAs values and focal spot sizes. The exposure settings of the conventional system can be found at the top of each table. The deviation from the target SDNR(u) is also given. Additionally, the entrance air kerma rate (μGy/fr), the FOM and the necessary tube power (kW) are shown. Table a): scenario 1, conventional system uses a small focal spot. Table b): scenario 2, conventional system uses a large focal spot. Blurring is modeled using sinc functions. Power values indicated in green lie below the limiting value (for that scenario), values indicated in red lie above the limit.

### Rectangular Focal Spot

| Focal spot (mm) | kV  | mA  | ms  | K<sub>ae</sub> (μGy/fr) | FOM  | ∆ SDNR(u) | Power (kW) | kV  | mA  | ms  | K<sub>ae</sub> (μGy/fr) | FOM  | ∆ SDNR(u) | Power (kW) | kV  | mA  | ms  | K<sub>ae</sub> (μGy/fr) | FOM  | ∆ SDNR(u) | Power (kW) |
|----------------|-----|-----|-----|--------------------------|------|-----------|------------|-----|-----|-----|--------------------------|------|-----------|------------|-----|-----|-----|--------------------------|------|-----------|------------|
|                |     |     |     |                          |      |           |            |     |     |     |                          |      |           |            |     |     |     |                          |      |           |            |
|                | 0.7 | 75  | 285 | 4575                     | 3.04 | -         | 21.4       | 0.7 | 91  | 420 | 87  | 609 | 3.32 | <1%         | 38.2       | 0.7 | 75  | 420 | 89  | 659 | 3.07 | <1%         | 39.1       |
|                | 0.8 | 93  | 420 | 89  | 659 | 2.47 | <1%         | 42.9       | 1.0 | 101 | 425 | 89  | 821 | 1.79 | <1%         | 48.3       | 1.2 | 115 | 420 | 89  | 1128 | 1.32 | <1%         | 52.5       |
|                | 1.2 | 115 | 420 | 89  | 1128 | 1.32 | <1%         | 52.5       | 1.4 | 125 | 420 | 89  | 1408 | 1.32 | <1%         | 52.5       | 1.6 | 125 | 420 | 89  | 1424 | 1.04 | 15%         | 52.5       |
|                | 1.4 | 125 | 420 | 89  | 1424 | 1.04 | 15%         | 52.5       |      |     |     |     |     |     |            |            | 1.5 | 79  | 420 | 89  | 415 | 1.98 | <1%         | 33.2       |
|                | 1.6 | 83  | 425 | 89  | 493 | 1.67 | <1%         | 35.3       | 1.7 | 91  | 425 | 87  | 616 | 1.33 | <1%         | 38.7       | 1.8 | 101 | 425 | 87  | 802 | 1.02 | <1%         | 42.9       |
|                | 2.0 | 123 | 425 | 87  | 1332 | 0.57 | 5%          | 52.3       |      |     |     |     |     |     |            |            | 2.0 | 123 | 425 | 87  | 1332 | 0.57 | 5%          | 52.3       |

### Conventional C-arm

| Focal spot (mm) | kV  | mA  | ms  | K<sub>ae</sub> (μGy/fr) | FOM  | ∆ SDNR(u) | Power (kW) | kV  | mA  | ms  | K<sub>ae</sub> (μGy/fr) | FOM  | ∆ SDNR(u) | Power (kW) | kV  | mA  | ms  | K<sub>ae</sub> (μGy/fr) | FOM  | ∆ SDNR(u) | Power (kW) |
|----------------|-----|-----|-----|--------------------------|------|-----------|------------|-----|-----|-----|--------------------------|------|-----------|------------|-----|-----|-----|--------------------------|------|-----------|------------|
|                |     |     |     |                          |      |           |            |     |     |     |                          |      |           |            |     |     |     |                          |      |           |            |
|                | 1.4 | 75  | 285 | 675 | 3.04 | -         | 21.4       | 1.5 | 79  | 420 | 87  | 415 | 1.98 | <1%         | 33.2       | 1.6 | 83  | 425 | 89  | 493 | 1.67 | <1%         | 35.3       |
|                | 1.7 | 91  | 425 | 87  | 616 | 1.33 | <1%         | 38.7       | 1.8 | 101 | 425 | 87  | 802 | 1.02 | <1%         | 42.9       | 2.0 | 123 | 425 | 87  | 1332 | 0.57 | 5%          | 52.3       |

Angio-MR (limit: 1.5 × mA = 420, 2 × ms = 90) | Angio-MR (limit: 2 × mA = 570, 2 × ms = 90) | Angio-MR (limit: 3 × mA = 855, 2 × ms = 90)
Table 3. Calculated exposure parameters for an acquisition series on the Angio-MR setup with a Gaussian focal spot for different limiting mAs values and focal spot sizes. The exposure settings of the conventional system can be found at the top of each table. The deviation from the target SDNR(u) is also given. Additionally, the entrance air kerma rate (µGy/fr), the FOM and the necessary tube power(kW) are shown. Table a): scenario 1, conventional system uses a small focal spot. Table b) scenario 2, conventional system uses a large focal spot. Blurring is modeled using Gaussian functions. Power values indicated in green lie below the limiting value (for that scenario), values indicated in red lie above the limit.

| Focal spot (mm) | kV | mA | ms | K_{ae} (µGy/fr) | FOM | Δ SDNR(u) | Power (kW) | Δ SDNR(u) | Power (kW) |
|----------------|----|----|----|----------------|-----|-----------|-----------|-----------|-----------|
| a) Conventional C-arm | | | | | | | | | |
| 0.7 | 75 | 285 | 45 | 675 | 3.25 | - | 21.4 |
| 0.7 | 77 | 565 | 89 | 530 | 4.06 | <1% | 43.5 |
| 0.8 | 79 | 570 | 87 | 563 | 3.82 | <1% | 45.0 |
| 1.0 | 83 | 570 | 87 | 647 | 3.32 | <1% | 47.3 |
| 1.2 | 89 | 570 | 87 | 780 | 2.77 | <1% | 50.7 |
| 1.4 | 97 | 560 | 89 | 978 | 2.2 | <1% | 54.3 |
| 1.6 | 109 | 560 | 89 | 1311 | 1.64 | <1% | 61.0 |
| 1.4 | 75 | 855 | 61 | 509 | 4.23 | <1% | 64.1 |
| 1.5 | 75 | 835 | 65 | 529 | 4.06 | <1% | 62.6 |
| 1.6 | 75 | 845 | 71 | 585 | 3.68 | <1% | 63.4 |
| 1.7 | 77 | 850 | 87 | 779 | 2.76 | <1% | 65.4 |
| 1.8 | 83 | 845 | 89 | 981 | 2.20 | <1% | 70.1 |

| Focal spot (mm) | kV | mA | ms | K_{ae} (µGy/fr) | FOM | Δ SDNR(u) | Power (kW) | Δ SDNR(u) | Power (kW) |
|----------------|----|----|----|----------------|-----|-----------|-----------|-----------|-----------|
| b) Conventional C-arm | | | | | | | | | |
| 1.4 | 75 | 285 | 45 | 675 | 1.76 | - | 21.4 |
| 1.4 | 71 | 560 | 75 | 410 | 2.85 | <1% | 42.0 |
| 1.5 | 75 | 565 | 81 | 446 | 2.62 | <1% | 42.4 |
| 1.6 | 75 | 560 | 89 | 486 | 2.40 | <1% | 42.0 |
| 1.7 | 79 | 560 | 87 | 553 | 2.11 | <1% | 44.2 |
| 1.8 | 83 | 570 | 85 | 632 | 1.85 | <1% | 47.3 |
| 2.0 | 91 | 570 | 89 | 845 | 1.38 | <1% | 51.9 |
| 1.4 | 75 | 845 | 49 | 404 | 2.89 | <1% | 63.4 |
| 1.5 | 75 | 845 | 53 | 437 | 2.68 | <1% | 63.4 |
| 1.6 | 75 | 855 | 57 | 475 | 2.46 | <1% | 64.1 |
| 1.7 | 75 | 845 | 63 | 519 | 2.25 | <1% | 63.4 |
| 1.8 | 75 | 850 | 69 | 572 | 2.04 | <1% | 63.8 |
| 2.0 | 75 | 855 | 85 | 709 | 1.65 | <1% | 64.1 |
Table 4. Calculated exposure parameters for a fluoroscopy series on the Angio-MR setup with a rectangular focal spot for different starting mAs values and focal spot sizes (a normal mA value on the left and a high mA value on the right). The exposure settings of the conventional system can be found at the top of the table. The deviation from the target SDNR(u) is also given. Additionally, the entrance air kerma rate (µGy/fr), the FOM and the necessary tube power(kW) are shown. Blurring is modeled using sinc functions.

| Focal spot (mm) | Conventional C-arm | Conventional C-arm | Rectangular Focal Spot | Conventional C-arm |
|-----------------|--------------------|--------------------|------------------------|--------------------|
|                 | kV     | mA    | ms  | $K_{d,e}$ (µGy/fr) | FOM | $\Delta$ SDNR(u) | Power (kW) | kV     | mA    | ms  | $K_{d,e}$ (µGy/fr) | FOM | $\Delta$ SDNR(u) | Power (kW) |
| 0.7             | 80     | 60    | 4   | 6.6   | 3.23 | -            | 4.8        | 80     | 120   | 4   | 13.2  | 3.23 | -            | 9.6        |
| 0.7             | 80     | 115   | 4   | 5.4   | 3.93 | <1%          | 9.2        | 80     | 225   | 4   | 10.6  | 3.93 | <1%          | 18.0       |
| 0.8             | 80     | 120   | 4   | 5.7   | 3.73 | <1%          | 9.6        | 80     | 240   | 4   | 11.3  | 3.73 | <1%          | 19.2       |
| 1.0             | 80     | 110   | 5   | 6.5   | 3.29 | <1%          | 8.8        | 80     | 220   | 5   | 12.9  | 3.29 | <1%          | 17.6       |
| 1.2             | 80     | 110   | 6   | 7.8   | 2.80 | <1%          | 8.8        | 80     | 215   | 6   | 15.2  | 2.80 | <1%          | 17.2       |
| 1.4             | 84     | 120   | 6   | 9.7   | 2.20 | <1%          | 10.1       | 84     | 235   | 6   | 19.0  | 2.20 | <1%          | 19.7       |
| 1.6             | 96     | 120   | 6   | 13.8  | 1.52 | <1%          | 11.5       | 96     | 240   | 6   | 27.5  | 1.52 | <1%          | 23.0       |

Angio-MR (limit: 2 × mA = 120, 1.5 × ms = 6)

Angio-MR (limit: 2 × mA = 240, 1.5 × ms = 6)
Table 5. Calculated exposure parameters for a fluoroscopy series on the Angio-MR setup with a Gaussian focal spot for different starting mAs values and focal spot sizes (a normal mA value on the left and a high mA value on the right). The exposure settings of the conventional system can be found at the top of the table. The deviation from the target SDNR(u) is also given. Additionally, the entrance air kerma rate (μGy/fr), the FOM and the necessary tube power (kW) are shown. Blurring is modeled using Gaussian functions.

| Focal spot (mm) | Conventional C-arm | Gaussian Focal Spot | Conventional C-arm | Angio-MR (limit: 2 × mA = 120, 1.5 × ms = 6) | Angio-MR (limit: 2 × mA = 240, 1.5 × ms = 6) |
|----------------|-------------------|-------------------|-------------------|---------------------------------|---------------------------------|
|                | kV | mA | ms | K_{a,r} (μGy/fr) | FOM | Δ SDNR(u) | Power (kW) | kV | mA | ms | K_{a,r} (μGy/fr) | FOM | Δ SDNR(u) | Power (kW) |
| 0.7            | 80 | 60 | 4  | 6.6            | 3.43 | -        | 4.8       | 80 | 120 | 4   | 13.2          | 3.43 | -        | 9.6 |
| 0.7            | 80 | 120| 4  | 5.7            | 4.08 | <1%      | 9.6       | 80 | 235 | 4   | 11.1          | 4.08 | <1%      | 18.8 |
| 0.8            | 80 | 100| 5  | 5.9            | 3.92 | <1%      | 8         | 80 | 240 | 4   | 11.3          | 3.92 | <1%      | 19.2 |
| 1.0            | 80 | 110| 5  | 6.5            | 3.57 | <1%      | 8.8       | 80 | 215 | 5   | 12.7          | 3.57 | <1%      | 17.2 |
| 1.2            | 80 | 120| 5  | 7.1            | 3.18 | <1%      | 9.6       | 80 | 240 | 5   | 14.1          | 3.18 | <1%      | 19.2 |
| 1.4            | 80 | 115| 6  | 8.1            | 2.77 | <1%      | 9.2       | 80 | 230 | 6   | 16.3          | 2.77 | <1%      | 18.4 |
| 1.6            | 86 | 120| 6  | 10.3           | 2.21 | <1%      | 10.3      | 86 | 235 | 6   | 20.3          | 2.21 | <1%      | 20.2 |
Figure 5. a) MTF weighting factors for focus blurring (solid lines) and motion blurring (dashed lines) using a rectangular focus as a function of spatial frequency for the Angio-MR system. The spatial frequency of interest is marked at 1.4 mm\(^{-1}\). MTFs are given for three focus sizes and two pulse lengths (velocity was assumed to be 1 mm s\(^{-1}\)). b) comparison of focus MTF between using a sinc function or Gaussian function to model the focal blurring. The spatial freq. was fixed at 1.4 mm\(^{-1}\).

Figure 6. a) Spectrum arriving at the reference point (86 cm from source) free in air for both systems for fluoroscopy imaging assuming they are using identical tube voltage, tube current and pulse length. b) Spectrum as it is absorbed by the detector for both systems for fluoroscopy imaging again assuming identical exposure parameters. c) and d) are similar graphs but for acquisition imaging. The graphs on the left show the effect of the filtration, the graphs on the right add the effect of the distance.

mAs to increase by a factor of 6, means all the target SDNR(u) values can be reached, but the required tube output is above 60 kW for all focus sizes. This means this scenario will only be possible for shorter runs, and when the Angio-MR system will have a focus size close to that of the conventional system. As expected, increasing the allowed mAs significantly increases the resulting tube power, but also means all target SDNR(u) values could be reached. Another advantage of using higher mAs values is the increase in the FOM.
The results show that for fluoroscopy imaging it is possible to reach a higher efficiency (i.e., FOM) for the Angio-MR if the focal spot size is similar to the conventional system. This is due to the reduction in the entrance K_{Air} which is caused by the different geometry and different filtration of the MR system. When the Angio-MR uses larger focal spot sizes the tube voltage increases, also increasing K_{Air}, which leads to a reduction in imaging efficiency. The same can be seen for acquisition imaging. Compared to the conventional system using a small focus, the Angio-MR will have to operate at a lower efficiency, meaning relatively higher patient dose (skin dose) for the same image quality. If image quality requirements are such that a large focal spot can be used then the Angio-MR could potentially achieve higher imaging efficiency. With the larger SID, the use of the large focal spot is indeed more likely to be acceptable. It is thought that the MR device would not just provide information for diagnosis. For example the rotational CBCT scans that are taken to generate a roadmap of the patients arteries or a scan to check whether blood flow has been restored could potentially be performed using the MRI system. X-ray scans that can be replaced by MRI sequences will reduce both patient dose and the dose to medical personnel and this potential should be exploited where possible.

Some remarks regarding the starting exposure parameters: firstly, as can be seen in the histograms of figure 4, actual tube voltages for fluoroscopy cover a wide range. Values above 100 kV are regularly seen and there were even some cases where the maximum of 120 kV was used. If the Angio-MR system needs to match the SDNR(u) of a conventional system using maximum tube voltage, then this can only be accomplished by increasing the mAs. For fluoroscopy, this should not lead to problems in terms of tube output limits, when using the large focal spot. Secondly (from figure 4), the pulse length for acquisition also varies widely, with values up to 140 ms. Increasing the pulse length above 200 ms is not recommended, resulting in less room for the pulse length to increase, which has to be compensated for by a higher mA or tube voltage. Additionally, since the pulse length is now longer, the pulse length times pulse rate used for determining the tube load rating now lies around 400, meaning the limiting power is approximately 49 kW for a 10 s run and 39 kW for a 25 s run for the large focal spot. Assuming identical focal spots of 1.4 mm and an mA limit of a factor of 1.5, the necessary exposure parameters for the Angio-MR system in order to match the SDNR(u) of the conventional system using 75 kV, 285 mA and 140 ms is the following parameter set: 93 kV, 425 mA and 198 ms, leading to a tube power of 39.5 kW. Comparing this to the above limits, some drop in SDNR(u) would be required to maintain a safe working point for the longer acquisition runs. Using a different mA limit in this scenario would lead to power requirements that cannot be obtained. As expected, whether or not the Angio-MR will be able to match the SDNR(u) of the conventional C-arm, assuming they have the same tube and detector, will depend on the exposure parameter selection made by the conventional system.

The two reasons why the hybrid system must use much higher output is because of the large SID and different amounts of filtration. In the case of acquisition, the increase in SID is combined with an increase in filtration: 0.1 mm Cu + 1 mm Al for the conventional c-arm compared to 7.4 mm Al for the hybrid system. This can be seen in figure 6. For fluoroscopy, the filtration is reduced in the hybrid system compared to the 0.4 mm Cu + 1 mm Al for the conventional system. This also explains why the necessary increase in tube output is relatively lower for fluoroscopy than for acquisition, as can be seen in tables 2–5.

In these calculations, the goal was to reach identical SDNR(u), but as the results show, this becomes difficult when the assumed focal spot of the hybrid system is much larger than that of the conventional system. If the goal is just to match SDNR, the required tube output would be the values found when using identical focal spots between both systems, and these numbers lie well within the limiting power output values. SDNR(u), on the other hand, weights (penalizes) larger focal spot sizes strongly. It may well be that the reduction in SDNR(u) by for example a large focal spot, does not match the change in task performance of the radiologist. This is rather difficult to test for the imaging steps/tasks involved in complex procedures such as thrombectomy however, there is likely to be some task dependency and this should be the topic of further research.

There are obviously some uncertainties/limitations on the presented results. The verification measurements accurately described the change in SDNR for different amounts of filtration, tube current, pulse length and varying SID, but the accuracy when increasing the tube voltage was somewhat lower. The simulation underestimated the increase in SDNR by approximately 10% as tube voltage changed from 60 kV to 90 kV. In reality, the required tube outputs may therefore be lower than the results obtained in these calculations, meaning the required tube power could also be somewhat lower.

We are using a simple analytical model of the x-ray detector in order to calculate the SDNR(u). It is possible to implement much more detailed simulations of x-ray detectors, often using Monte Carlo techniques. These can be used to study interactions within the x-ray scintillator and subsequent light collection properties of the CsI needle structures (Badano 2003, Badano and Sempau 2006), the intrinsic sharpness of x-ray conversion layers (Hajdok et al 2008) and the propagation of signal and noise through the x-ray detector (Hajdok et al 2008, Star-Lack et al 2014). These simulations give great insight in the physical processes occurring within the x-ray detector however the emphasis here is on the parameter selection of the
x-ray tube and generator rather than detailed simulation of the detector. The detector model will certainly influence the selected x-ray parameters as this governs the energy distribution of the x-rays and thus the signal generated/collected in the x-ray detector. However, the simple analytical model described here has been shown to predict the relative changes in SDNR resulting from changes in x-ray parameters with reasonably accuracy, as seen in figure 3.

While this study has clearly focused on neuroradiological imaging used in stroke procedures, the general method could be applied to a range of imaging scenarios. In order to do this, the imaging task would have to be defined in terms of the size, composition and velocity of the critical structure(s). Additional information required includes the x-ray tube focus sizes and associated MTFs along with tables of the x-ray tube operating power and a description of the expected x-ray series (run length, frame rate etc). Finally, some basic information about the x-ray detector x-ray conversion material (composition, physical thickness) is needed. Once this information is included in the framework, x-ray tube parameters can be varied, within some constraints such as maximum tube current for a focus size, so that the required image quality level is achieved. In fact, one difficulty might be the definition of the required level of technical image quality. In this study, the required level was set by the standard C-arm angiography system and the aim was to establish the ADRC parameters required to match this.

This study focused on simulations of acquisition parameters used during stroke procedures. Of course the simulations and requirements of the Angio-MR system would be very different if other types of procedures would be considered. For example in cardiac applications, object motion plays a much more important role as the blood vessels are in continuous motion. Therefore, the requirements on the pulse length would be much stricter in order to avoid severe motion blurring. Additionally, because the objects of interest have relatively high velocities and the pulse lengths are short, the MTF_{motion} weighting factor will have a noticeable effect on SDNR(u), which again limits the pulse length the Angio-MR can use. When simulating abdomen procedures, patient thickness will start to become the limiting factor. In some scenarios, for thicker patients, conventional angiography systems are already operating at high power values. There is therefore little room to increase acquisition parameters and a loss of image quality and imaging efficiency can be expected.

6. Conclusion

Using an analytical simulation platform, this study investigated the required x-ray exposure parameters and tube output in order to perform neurological procedures on a proposed hybrid Angio-MR concept. The results show that a typical current x-ray tube should have sufficient power to reach similar SDNR(u) values as obtained in a conventional C-arm angiography system, assuming that the focal spot size of the hybrid system is comparable to conventional systems.

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