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Accepted Article

Analysis of the applicability of two-dimensional detector arrays in terms of sampling rate and detector size to verify scanned intensity modulated proton therapy plans

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Abstract

Purpose: The introduction of advanced treatment techniques in proton therapy, such as intensity modulated proton therapy, leads to an increased need for patient specific quality assurance, especially an accurate treatment plan verification becomes inevitable. In this study, signal theoretical analysis of dose distributions in scanned proton therapy is performed to investigate the feasibility and limits of two-dimensional detector arrays for treatment plan verification.

Methods: Two-dimensional detector arrays are characterized by two main aspects: the distance between the single detectors on the array or the sampling frequency; and the lateral response functions of a single detector. The analysis is based on single spots, reference fields and on measured and calculated dose distributions of typical intensity modulated proton therapy treatment plans with and without range shifter. Measurements were performed with Gafchromic EBT3 films (Ashland Speciality Ingredients G.P., Bridgewater, NJ, USA), the MatriXX PT detector array (IBA Dosimetry, Schwarzenbruck, Germany) and the OCTAVIUS detector array 1500XDR (PTW-Freiburg, Germany) at an IBA Proteus PLUS proton therapy system (Ion Beam Applications, Louvain-la-Neuve, Belgium). Dose calculations were performed with the treatment planning system RayStation 6 or 8 (RaySearch Laboratories, Sweden).

Results: The Fourier analysis of the data of the treatment planning system and film measurements show maximum frequencies of 0.06/mm for the plan with range shifter and 0.083/mm for the plan without range shifter. According to the Nyquist theorem this corresponds to minimum required sampling distances of 8.3 mm and 6 mm, respectively. By comparison, the sampling distances of the arrays of 7.6 mm (MatriXX PT) and 7.1 mm (OD1500XDR) are sufficient to reconstruct the dose distributions adequately from measurements if range shifters are used, whereas some fields of the plans without range shifter violated the Nyquist requirement. The lateral dose response functions of the single detectors within the arrays have clearly higher frequencies than the treatment plans and thus the volume effect only slightly influences the measurements. Consequently, the array measurements show high gamma passing rates with at least 96 % and a good agreement between the investigated line profiles.

Conclusion: The results indicate that the detector dimensions and sampling distances of the arrays are in most studied cases adequate not to substantially influence the measurement process when they are used for analysing typical intensity modulated proton therapy treatment plans. Nevertheless, clinical conditions have been identified, for instance treatment plans without range shifter, under which the Nyquist theorem is violated such that a full representation of the dose distributions with the measurements is not feasible. In these cases, analysis of measurements is limited to pointwise comparisons.

Key words: 2D-array, patient plan verification, frequency analysis, intensity modulated proton therapy, volume effect, detector sampling

1.0 Introduction

In photon radiotherapy, the verification of intensity modulated radiotherapy (IMRT) and volumetric arc therapy (VMAT) treatment plans by measurements with two-dimensional detector arrays or radiochromic films is a standard procedure to ensure precise treatment delivery. To do so, the treatment plans are irradiated on a measurement device in a solid water or water phantom and measurements are then compared to the planned dose distributions e.g. by gamma index analysis.¹⁻⁴ The comparably higher complexity of film dosimetry usually favours the use of detector arrays due to their easier handling and analysis of measured data.⁵⁻⁷ Nevertheless, in comparison to films, detector arrays have worse signal theoretical properties such as a reduced sampling rate and an expanded size of the single detectors in the array.⁷⁻⁹ Poppe et al.⁸ showed that for typical intensity modulated dose distributions in photon therapy, the signal theoretical detector properties are sufficient. In the recent years, several new detectors have been introduced and tested for IMRT or VMAT dose distributions as well as stereotactic treatments.^{6,8-14}

Common clinically used two-dimensional planar detector arrays are based on air-filled ionization chambers (MatriXX, IBA Dosimetry, Schwarzenbruck, Germany, OCTAVIUS 1500 and 729, PTW Freiburg, Germany), liquid filled ionization chambers (OCTAVIUS 1000SRS, PTW Freiburg, Germany) or diodes (MapCheck, Sun Nuclear, Melbourne, FL, USA).

For intensity modulated proton therapy (IMPT), several studies have investigated the use of these two-dimensional detector arrays, especially the MatriXX PT detector array (IBA dosimetry, Schwarzenbruck, Germany).¹⁵⁻¹⁹ These studies mainly deal with the general applicability of the detectors for plan verification measurements. A detailed analysis of the applicability of these arrays in intensity modulated proton therapy by studying their signal theoretical properties has not been performed so far.

Meanwhile, IMPT has become the standard clinical treatment approach for proton therapy treatments. IMPT treatment plans are characterized by simultaneous optimization of weight of the single spots for all fields over the target volume. This can be performed either by single field or multiple field optimization. In contrast to passively scattered techniques, these IMPT plans are superior in sparing organs at risk but are more susceptible for uncertainties regarding positioning errors or range uncertainties.²⁰⁻²² Due to the complexity of IMPT plans the patient specific quality assurance, which includes plan verification with detector arrays, is of high significance.^{23,24}

From the viewpoint of signal theoretical considerations, detector arrays are characterized by two properties influencing their capability to detect errors during plan verification measurements: the sampling rate of the detectors within the array and the volumetric or lateral dose response function of the single detectors. In photon therapy, these aspects have been analyzed by Poppe et al.^{8,11} for clinical intensity modulated dose distributions. They have shown that the volume of the single detectors, reflected in the shape of the lateral dose response function, could cause volume perturbation effects in highly modulated plans, but in some cases, such as dynamic IMRT or VMAT techniques, the effect is sufficiently small without introducing large deviations to the planned dose distribution due to the measurement process. Furthermore, the detector sampling rate was analyzed, and they concluded that the sampling frequency needs to be adapted for different treatment plans. In situations with rapidly changing dose gradients, the high frequency components of the signals in the Fourier space become more pronounced. This results in higher Nyquist frequencies and therefore lower minimum sampling frequencies that are required to fully represent the original dose distributions to be verified. However, the sampling rates of the investigated detectors (one measurement point every 8 mm or 10 mm for the typical arrays) were not sufficient to fulfil the Nyquist theorem that requires approximately one measurement point every 5 mm for the highly modulated treatment plan analyzed in the study. As a workaround, the detector array can be shifted and the dose distribution re-measured resulting in merged measurements with higher resolutions.

In this study, two different detector arrays, the MatriXX PT detector array (IBA dosimetry, Schwarzenbruck, Germany) and the OCTAVIUS detector array (OD) 1500XDR (PTW Freiburg, Germany) have been investigated in the context of IMPT. Both detectors are air-filled plane-parallel ionization chamber arrays, but with different detector dimensions, shapes, and arrangements.

A signal theoretic analysis of these detector arrays for the plan verification process in typical IMPT dose distributions has been performed. Fourier components of different treatment plans, single fields as well as of single pencil beam spots have been analyzed to determine the adequate sampling frequency for these situations. Furthermore, the influence of the size and shape of the single detectors on the array was studied. The MatriXX PT and the OD1500XDR measurements were then compared to Gafchromic EBT3 films, serving as a high-resolution measurement and the calculated dose distributions from the treatment planning system.

2.0 Material and Methods

2.1 Signal theory

Based on a convolution model^{25,26}, the measured signal of a detector $M(x,y)$ when measuring in a fluence profile $\Phi(x,y)$ within a water phantom valid for a certain energy and depth, is given by the convolution (in the following indicated by “ * ”)

$$M(x,y) = \Phi(x,y) * K_M(x,y) \quad (1)$$

where the convolution kernel $K_M(x,y)$ is the so-called lateral fluence response function of the detector. Thereby, the lateral fluence response function is influenced on the one hand by geometrical properties of the detector and on the other hand by charged particle transport in the surrounding medium.

The dose profile $D(x,y)$ under the same conditions as above within the water phantom but without the presence of a detector is described by a similar approach but this time associated with the dose deposition kernel describing the radiation transport when passing a medium:

$$D(x,y) = \Phi(x,y) * K_D(x,y) \quad (2)$$

Lastly, the influence of the detector's volume-averaging effect and the disturbance of the local particle fluence by the detector can be described by the lateral dose response function $K(x,y)$ that serves as the convolution kernel transforming the dose profile $D(x,y)$ into the measured signal $M(x,y)$:

$$M(x,y) = D(x,y) * K(x,y) \quad (3)$$

Nevertheless, the lateral dose response function can only be determined indirectly by deconvolution of the fluence response function of the detector $K_M(x,y)$ with the dose deposition kernel $K_D(x,y)$ as given when combining equations (1) to (3):

$$K_M(x,y) = K(x,y) * K_D(x,y) \quad (4)$$

For photon therapy, the influence of the depth and the energy on the lateral dose response function has been studied^{26,27}, but for proton therapy a detailed analysis has not been performed so far. For different two-dimensional detector arrays for photon therapy, including those that are investigated in this study and are used for proton therapy, the fluence response functions are already determined by photon pencil or slit beam measurements and by Monte Carlo simulations and the typically bell-shaped lateral dose response functions have been calculated.^{28,29} Furthermore, the perturbation of the associated volume effect will lead to a broadening of the penumbra and reduction of the measured value in the centre of small fields according to Equation 3.^{7,30,31}

On the other hand, the adequate detector sampling rate for the dose distribution under investigation can be assessed by applying the Nyquist theorem, which states that the investigated dataset can only be fully

reconstructed from the measurements if the sampling frequency is at least two times the maximum frequency that is present in the original data. The frequency components of a dose distribution of proton therapy plans can be analyzed either by using high-resolution measurements, such as those obtained with films, or by using the calculated treatment planning system (TPS) data in the Fourier domain. The maximum frequency is often determined by considering only frequency components above the 1%-limit of the maximum magnitude of the frequency spectrum as suggested by Bortfeld et al.³² (2000) and Poppe et al.¹¹ (2007).

In cases where the sampling frequency is not sufficient to achieve a full reconstruction, that is, the sampling frequency is lower than twice the highest frequency component, only point-wise analysis of the actual measured data is feasible.¹¹ Data interpolation should be avoided in these under-sampled measurements during comparison with high-resolution reference data, such as the calculated dose distribution from TPS, as this may create erroneous data points and consequently lead to a misinterpretation of the measurement results.

As a first estimation, the lateral dose response functions are modelled as ideal two-dimensional step functions with the chamber cross-sections. Their suitability has been discussed in Section 4.2.

2.2 Detector arrays

The OD1500XDR consists of 1405 cubic ionization chambers arranged in a checkerboard setup in an area of 27 x 27 cm² with a side length of 4.4 mm and a height of 3 mm. The centre-to-centre distance of the chambers is 7.1 mm in the diagonal direction and 10 mm parallel to the crossplane and inplane axes (see Figure 1 a)). The reference point of the array is 7.5 mm below the surface of the array.

The IBA MatriXX PT detector array consists of 1020 plane parallel air-vented ionization chambers arranged in a 32 x 32 grid on 24.4 x 24.4 cm² with a centre-to-centre distance of 7.62 mm. The ionization chambers are cylindrical in shape with a height of 2 mm and a diameter of 4.2 mm (see Figure 1 b)), and the effective point of measurement is 6 mm below its surface.

2.3 Film calibration

Gafchromic EBT3 films (Ashland Speciality Ingredients G.P., Bridgewater, NJ, USA) were calibrated in a solid water phantom in 2 cm water equivalent depth with 5 cm support material, the surface of the phantom was positioned in the isocentre plane. Film pieces of 4 x 5 cm² were placed in the centre of a 10 x 10 cm² field of 160 MeV proton beams and irradiated with seven dose points (10.58, 21.15, 42.3, 84.6,

112.8, 169.2 and 423 cGy). Each dose point was measured three times. Furthermore, three unexposed reference films were used to obtain the background at zero dose.

The films were scanned with an Epson 10000XL flatbed scanner (Seiko Epson Corp., Suwa, Japan) at a resolution of 72 dpi and 48 bit. The calibration was performed using the single channel (red) method.^{33 34} All film pieces were scanned in the portrait orientation positioned in the centre of the scanner area three times consecutively. The mean pixel values (PV) within a region of interest of 2.5 x 2.5 cm² was obtained. The corresponding optical density (OD) was calculated according to Equation 5 for each channel separately.

$$OD = \log \left(\frac{2^{16}}{PV} \right) \quad (5)$$

The calibration curve was then fitted by a third-degree polynomial function.^{34–36}

2.4 Array measurements

All measurements were performed at an IBA Proteus PLUS proton therapy system (Ion Beam Applications, Louvain-la-Neuve, Belgium) with active scanning proton beams in the energy range of 70 MeV ($\sigma = 6.5mm$, range: 4.1 cm) to 225 MeV ($\sigma = 3.3mm$, range: 31.8 cm), the energy range can be extended towards lower energies using a range shifter. The measurement devices were placed in a solid water phantom (Solid Phantom SP34, IBA Dosimetry, Schwarzenbruck, Germany).

Both arrays are cross calibrated against water phantom measurements in a 20 x 20 cm² field, defined at the isocentre plane on the phantom surface, in 4.4 cm water equivalent depth. The OD1500XDR was placed above 3 cm solid water, whereas the MatriXX PT was placed directly on the table. The arrays were aligned to the laser markers and the setup was verified by kV-imaging. For all subsequent measurements, the detector arrays were positioned with their effective points of measurement at the selected measurement depth.

2.5 Scanned proton pencil beam plans

Single pencil beam spots

To obtain the maximum frequency that could occur in proton therapy plans, the dose profiles of single pencil beam spots were measured with EBT3 films. Three different energies (70 MeV ($\sigma = 6.5mm$, range: 4.1 cm), 150 MeV ($\sigma = 4.3mm$, range: 15.8 cm), 225 MeV ($\sigma = 3.3mm$, range: 31.8 cm), were selected and each energy was measured in three different depths in a solid water phantom according to Table 1. Additionally, the 150 MeV pencil beam spot was re-measured with a range shifter (water equivalent

thickness: 40.3 mm) at 1.9 and 7 cm solid water depth to demonstrate and evaluate exemplarily the influence of the range shifter in terms of the broadening of the spot (width and steepness) and the change in frequency contents of the pencil beam.

To evaluate the pencil beams, transversal line profiles through the spots were used. The spot properties such as the full-width at half-maximum (FWHM) and the penumbra (80% -20% width ($w_{80/20}$)) were calculated from these profiles. Furthermore, the measured pencil beam spot dose distributions were transformed into the Fourier domain to determine the upper limit of the frequency that could occur in plans by superposition of these pencil beam spots. The maximum frequency is defined as the frequency above which all frequency components are below 1 % of the maximum magnitude in the Fourier spectrum.^{11,32}

10 x 10 cm² reference field

Profile measurements of a homogenous 160 MeV 10 x 10 cm² field with films and both arrays were performed to evaluate the penumbra broadening due to the volume effect of the single ionization chamber in the arrays. To increase the nominal sampling rate, the arrays were scanned across the field by shifting it in 1 mm steps. The single measurements were then merged into one high-resolution measurement for evaluation.⁸

The film measurements were evaluated by averaging 72 profiles (2.54 cm) around the central axis of the field to minimize the noise. All measurements were performed in 2 cm water equivalent depth.

Intensity modulated treatment plans

Treatment plan verifications were performed for three different clinical IMPT plans with EBT3 films, the MatriXX PT and OD1500XDR detector arrays. Two of the plans were head and neck cases with range shifter consisting of four single fields while the third plan was a three fields base of skull plan without range shifter. The treatment plans were created using the TPS RayStation 6 or 8 (RaySearch Laboratories, Sweden). Dose calculations were performed with a voxel size of 1x1x1 mm³. Each field was verified individually from 0-degree gantry angle in a solid water phantom containing the measurement device. Measurements were performed in three depths: 1 cm water equivalent depth and two plan specific depths in the vicinity of the prescription volume.

Frequency analysis of the treatment fields was performed by transforming the whole fields as well as the line profiles extracted from the measured distributions and the TPS data into the Fourier domain.¹¹ Furthermore, gamma index analysis was performed by comparing the raw data of the array

measurements to TPS calculated dose distributions using gamma index criteria of 3 mm/3 % and 2 mm/2 %.³⁷ The evaluation was differentiated between global gamma index analysis (dose difference is calculated from the maximum dose of the investigated field) and local gamma index analysis (dose difference is calculated from the local dose at the point of evaluation).³⁸ All dose values below 10 % of the maximum dose were suppressed.

Fourier analysis of two-dimensional TPS dose distributions

To be able to generalize the results obtained for the measured IMPT treatment plans, a set of clinical TPS dose distributions that include diverse entities aiming to cover different possible clinical scenarios has been additionally analyzed in the Fourier domain in terms of their frequency components. This set of TPS data contains three base of skull plans (of which two plans are with range shifter each consisting of two fields and one plan without range shifter consisting of four fields), three head and neck plans (all consisting of four fields, two plans are with range shifter and one plan has two fields with range shifter and two without) and six lung cases without range shifter with three fields each. All plans were analyzed in three depths, whereas the lung plans were analyzed in 1 cm depth, which corresponds to the measurement depths that would have been chosen during clinical plan verification. Altogether, the data provide a set of 14 fields with range shifter and 24 fields without range shifter.

3.0 Results

Pencil beam spot profiles

Figure 2 a) shows the central dose profiles of selected pencil beam spot measurements. The corresponding frequency components in the Fourier domain are shown in Figure 2 b). With decreasing energy and increasing depth, the FWHM of the spot increases. The use of range shifters also leads to a further increase of the pencil beam's width. Due to the close resemblance of the spot profiles to Gaussian functions, larger FWHM of the single pencil beam spots are also associated with an increase of the penumbra widths. Consequently, the Fourier spectra of pencil beam spots with larger FWHM also consist of less high frequency components as demonstrated in Figure 2 b).

The results for all measured pencil beam spots, together with the Nyquist frequency (f_{Nyq}) and the corresponding minimum sampling distance (s_{min}) are summarized in Table 2. The 225 MeV pencil beam spot in 2 cm water equivalent depth has the smallest FWHM and a maximum frequency of 0.239/mm. With increasing spot widths, the maximum frequency decreases, whereas most of the spots investigated possess frequency components less than 0.15/mm. The Nyquist frequencies range from 0.146/mm to

0.479/mm, resulting in required sampling distances between 6.84 mm and 2.09 mm which are all below the sampling distances of the detector arrays with 7.1 mm (OD1500XDR) and 7.6 mm (MatriXX PT). Therefore, the Nyquist theorem is not fulfilled for all the single spots investigated.

10 x 10 cm² reference field

The half profile of the 10 x 10 cm² field is shown in Figure 3 a), with the corresponding Fourier transform in Figure 3 b). The measured penumbra widths $w_{80/20}$ obtained using the MatriXX PT (7.6 mm $\pm$ 0.24 mm) and the OD1500XDR (7.9 mm $\pm$ 0.16 mm) are higher than those obtained with EBT3 films (7.3 mm $\pm$ 0.2 mm). The maximum frequency component derived from the film measurement is 0.064/mm leading to a Nyquist frequency of 0.128/mm and a required minimum sampling distance of 7.8 mm.

Intensity modulated treatment plans

In Figure 4 a), the profiles of one field of a head and neck treatment plan measured in 1 cm depth are shown along one axis at a position, at which both detector arrays have single chambers arranged within $\pm$ 0.5 mm along the profile. The measurements were compared to the calculated TPS data. The corresponding Fourier transform evaluated using the TPS data and the film measurements are shown in Figure 4 b).

Both plans with range shifter (plan 1 and 2) exhibit negligible frequency components above 0.058/mm, leading to a maximum Nyquist frequency of 0.116/mm. Therefore, the minimum required sampling distance for these plans is 8.68 mm.

Due to the absence of the range shifter the gradients in the base of skull treatment plan (plan 3) become steeper and the single fields contain frequencies up to 0.083/mm leading to a minimum required sampling distance of 6.02 mm (Figure 5). The maximum frequencies together with the required minimum sampling distances for adequate reconstruction are summarized in Table 3 for all fields and the measurements in different depths of the investigated plans.

The sampling process of the measurements leads to a repetition of the frequency components according to the sampling frequencies (1/7.6 mm⁻¹ for the MatriXX PT and 1/10 mm⁻¹ for the OD1500XDR). If the minimum required sampling distance is exceeded, the repetition of the measurement leads to overlapping frequencies, also known as aliasing. Figure 6 a) shows the Fourier transform of the TPS data and Fourier transforms of sampled TPS data according to the sampling distances of the detector arrays of a profile through one field of the base of skull plan. Due to the violation of the Nyquist theorem of both arrays (minimum required sampling for this case is 6.02 mm), the measurement process leads to

overlapping of the replications of the Fourier transforms. Since the sampling frequency of the MatriXX PT along a row of chambers is higher, the aliasing effect is also less prominent than for the case of the OD1500XDR. Nevertheless, the attempt to reconstruct the original dose profile from the Fourier spectrum introduces errors as shown in Figure 6 b), especially in the low dose regions. One important clinical consequence in cases where the Nyquist theorem is not fulfilled is that an attempt to reconstruct the original dose profiles using the limited number of measurement points, including interpolations between the data points, may lead to erroneous data points. This is shown exemplarily in Figure 6 c) where linear interpolation between the measurement points has been performed. Although the original measurement points show good agreement to the TPS calculation at these positions, the same does not always apply to the artificial data points added through the interpolation process (indicated by arrows) and incorporation of these additional points into the gamma index evaluation may falsify the result from the verification process.

Figure 7 a) shows the comparison between the Fourier spectrum of the line dose profile of Figure 5 and that of the lateral dose response functions of the detector arrays. The response function approximations are in this case two-dimensional step functions of the size of the single detectors: a cylindrical shaped step function with a diameter of 4.2 mm (MatriXX PT) and a squared step function with 4.4 mm side length (OD1500XDR). In Fourier space, the convolution in Equation 3 is given by the multiplication of the Fourier transform of the response function with the Fourier transform of the dose distribution. Since the width of the frequency spectrum of the lateral response functions clearly exceeds that of the dose distributions, the signal components of the dose distribution remain nearly uninfluenced in the measurement process as it can be seen in Figure 7 b). It is noteworthy that the frequency distributions of the array measurements are only shown up to half of the sampling frequency of the arrays, due to the evaluation parallel to the axes the cut off frequencies of the arrays are 0.05/mm for the OD1500XDR and 0.066/mm for the MatriXX PT.

However, when comparing the measured data of the arrays with those of the TPS, slight deviations can still be seen. After the convolution of the TPS dose distribution with the two-dimensional lateral dose response function, the agreement between the measurement and calculation are improved as shown exemplarily in Figure 8 for the base of skull plan measured with the OD1500XDR. This shows that the volume effect of the chambers can be adequately accounted for using the first estimates of the response functions.

Frequency analysis of TPS dose distributions of spot scanned proton therapy plans

The maximum frequencies obtained from analysing different treatment plans underline the findings obtained with the detailed analysis of the measured treatment plans presented in the previous section. All fields with range shifter show a maximum frequency of 0.06/mm with a corresponding minimum required sampling distance of 8.39 mm. Therefore, none of the analyzed fields with range shifter exceeds the Nyquist frequencies of the detector arrays. Generally, despite the broadening of the single pencil beams with depth, the maximum frequencies in the treatment plans do not seem to decrease with increasing depth but are rather more dependent on the degree of modulation of the treatment plans. On the other hand, the plans without range shifter show a maximum frequency of 0.077/mm and a corresponding minimum required sampling distance of 6.49 mm. From the 24 investigated fields without range shifter, five exceeded the Nyquist frequencies of the investigated detector arrays.

Gamma index analysis

Overall, the gamma index passing rates achieved with both detector arrays are comparable, independent on the used criteria. The minimum passing rate for the head and neck plan is 96 % for the MatriXX PT detector array and 98 % for the OD1500XDR using the local 2 mm/2 % criterion. For the base of skull plans, the minimum passing rates are 99 % and 100 % for the MatriXX PT and the OD1500XDR, respectively. As expected, the global gamma evaluation leads to slightly higher gamma passing rates compared to the local gamma evaluation due to the overall large dose deviation allowance when the maximum dose point is used for the normalisation. Overall, the passing rates are above 96% for all the evaluated plans, which would be considered as clinical acceptable. For all gamma evaluations, especially for the base of skull plan, mainly single points fail the analysis in all of the cases.

4.0 Discussion

4.1 Frequency analysis of proton fields

The evaluation of the pencil beam spot profiles depicted in Figure 1 shows an increase of the spot width with decreasing energy due to the increased angular dispersion of multiple-Coulomb scattering with decreasing energy of the proton pencil beams.³⁹ The same is true for multiple-Coulomb scattering for the pencil beam spots with increasing measurement depth in water.⁴⁰ In general, it is shown that the dose distributions of higher energy pencil beam spots and at shallow depths possess more high frequency components. The maximum frequency in this study was 0.239/mm, obtained with a 225 MeV pencil beam spot in 2 cm depth. Consequently, due to multiple Coulomb scattering, the dose distribution will be more smoothed out at larger depths, so that the minimum sampling frequency required at these depths is also lower.

Since multiple spots are usually superposed in clinical intensity modulated treatment plans to form conformal dose distributions, such as exemplarily shown by the 10 x 10 cm² reference field measured with EBT3 films depicted in Figure 3, the shape of a scanned field is also influenced by the initial pencil beam profiles of the single spots. An increase in energy may lead to a sharper penumbra for a modulated field and consequently to higher frequency contents. In larger depths, beam broadening may lead to lower maximum frequencies of these scanned fields. On the other hand, the penumbra regions are formed by overlapping single pencil beams. Consequently, the gradient in the final composed dose is also less pronounced with a wider penumbra than a single pencil beam spot in the corresponding depth and hence has less high frequency components.

The same can be seen for the treatment plans consisting of single fields each formed by composite pencil beam spots. As shown in Figures 4 to 8, Table 3 and in the analysis of the TPS data, the maximum frequencies present in these fields do not exceed 0.083/mm. Furthermore, fields without range shifter are also shown to possess higher frequency components. The results from analysing the different TPS dose distributions further underline the results obtained with measurements. Nevertheless, it was found that not only the size of the single pencil beams is affecting the maximum frequencies, but also the degree of modulation in the single fields. This is the reason why some of the dose distributions in larger depths show higher frequencies than the ones in 1 cm depth as shown in Table 3 and is also seen in the analysis of the TPS data. In larger depths, fewer single pencil beams are overlapping, and the irradiated area becomes smaller. Additionally, in larger depths, dose is deposited by pencil beams with higher initial energies. As seen in the evaluation of the single pencil beams in Figure 2, these high energy pencil beams are still sharper in larger depths than the lower energy pencil beams in shallow depths.

Poppe et al.^{11,12} have investigated the maximum frequency components of typical step-and-shoot IMRT plans in photon therapy and found that 0.1/mm is the highest frequency in those dose distributions. For the proton therapy plans investigated in this study, the maximum frequency is 0.083/mm, which is lower than for the photon case (compare to Figures 4 to 8 and Table 3). However, if collimated proton fields are considered, lateral dose gradients have been observed in several studies.^{41–44} Indeed, dynamic techniques in photon therapy, such as sliding window IMRT and VMAT will also lead to lower gradients in the dose distributions.^{11,12} For example, Poppe et al. (2007)¹¹ investigated those plans and found maximum frequencies around 0.06/mm, which is then comparable to the results obtained in this work for intensity modulated proton therapy plans.

4.2 The role of the detector's volume effect

In this study, the lateral dose response functions of the two detector arrays are approximated as ideal step functions. As shown in previous studies, in which these lateral dose response functions have been studied for high energetic photon beams,^{26,28,45} the functions characterize on one hand the volume-averaging effect due to the extensive dimension of the single detectors and on the other hand the charged particle fluence perturbation due to the presence of the non-water equivalent components of the arrays, including the air cavities of the chambers. In high energetic proton beams, the average energy of the secondary electrons in water is lower than in photon beams and thus their range in the medium and the detector's material is limited. For example, the lateral range of secondary electrons in 6MV photon beams can be larger than 1.5 cm⁴⁶, whereas in proton therapy the total range of electrons produced by 200 MeV protons is 2 mm in maximum⁴⁷ and the range of the lateral displacement of protons is in the range of a few mm⁴⁸. Furthermore, in contrast to photon beams, where secondary electrons are responsible for the local energy deposition, the dose contribution from these secondary electrons is low and the dose is mainly deposited by primary protons. The local fluence of the protons will not be largely perturbed by the presence of the detectors due to their large mass and high energy. Nevertheless, the energy and depth dependence of the detectors' lateral dose response functions in proton beams still need to be further investigated in future work. But as already shown in Figure 8, their approximations using ideal geometrical based response functions could provide first estimations on the influence of the volume effect on the measurements.

Figure 7 demonstrates that the Fourier transforms of the detectors' lateral dose response functions, here estimated as step functions that correspond to the cross-sections of the chambers, consist of frequency components that clearly exceed those of the plans. Therefore, it can be concluded that the latter is only slightly suppressed by the detector's volume effect (compare Figure 7 b)) during the measurement process. In other words, the detector sizes of the two arrays can be deemed sufficiently small to verify the intensity modulated proton therapy plans investigated in this study. This agrees with the observations by Furukawa et al. (2013)⁴⁹ who stated that the low gradients in the proton therapy plans make the measurement nearly independent on the measurement process in penumbra regions such that a direct comparison between measurement and calculated TPS data can be performed without convolving the response functions with the TPS data.

4.3 The role of the array's sampling distance

According to the frequency analysis of the treatment plans, the maximum frequency evaluated was 0.116/mm for the plans with range shifter and 0.166/mm for the plan without range shifter. These values correspond to a minimum sampling rate of 8.68 mm and 6.02 mm, respectively. The Matrixx PT and the

OD1500XDR have sampling frequencies of 0.13/mm and 0.14/mm, respectively which fulfil the Nyquist theorem for the plans with range shifter but fail for some of the fields without range shifter. Therefore, for some of these plans without range shifter, an accurate reconstruction of the dose distribution would not be possible for every field. In these cases, the sampling process of the detector arrays leads to a reproduction of the frequency content in the Fourier space leading to overlapping frequency contents or aliasing effects. In other words, the dose distributions cannot be fully represented by the measurements and attempting to do so will lead to erroneous results as shown in Figure 6.

Therefore, in cases when the sampling frequency is not sufficient for the measured dose profile, only pointwise analysis using the actually measured data points is possible.¹¹ If the evaluation of the treatment plan measurements is performed with software packages that interpolate the measurements to obtain two dimensional dose distributions before comparing them to the data of the treatment planning system, the interpolation of the data may lead to larger deviations or a misinterpretation of the measurement data. This behaviour has been demonstrated in Figure 6 c) where the interpolated data points show larger deviations from the TPS calculations at these positions.

Additionally, with the introduction of collimators or apertures in proton therapy to enhance the penumbra sharpness, higher frequencies will occur and therefore, higher detector sampling rates might become necessary.^{41–43,50}

4.4 Gamma index analysis

The gamma index analysis shows gamma index passing rates of over 96% for the investigated fields in all depths. Due to the overall high gamma index passing rates, an increased gamma index criterion of 2 mm/2 % as well as a use of the local gamma index criterion to increase the sensitivity to detect treatment errors seems to be recommendable. The influence of the detector's volume effect on the measurement may in some cases compromise the results of gamma index analysis. A common clinical practice to overcome this issue, especially in the field of photon therapy, is to match the TPS dose grids to the detector's dimensions or to include the response functions of the detectors in this analysis. Nevertheless, it has been demonstrated in this study that the influence of the detector's volume effect on the measured signal and hence the gamma index analysis in proton therapy is negligible in most cases. For small target volumes, due to the limited number of evaluated measurement points and eventually high gradients, the gamma index should not be used as the sole quality assurance metric but the analysis should be complemented by comparisons of line profiles, especially at those measurement points that lie in the

vicinity of critical structures. In cases where the sampling rate of the arrays is not sufficient, film measurements can also be performed.

5.0 Conclusion

This study has demonstrated that the MatriXX PT and the OCTAVIUS 1500XDR detector arrays are suitable for plan verification measurements in pencil beam scanning proton therapy in most cases from a signal theoretical perspective. The single chamber dimensions of the single detectors in these arrays are sufficiently small not to suppress the frequency contents of the treatment plans considerably. In other words, the volume effect introduced by the measurement process is only very small or negligible. The detector sampling rates of the arrays fulfil the Nyquist theorem for the plans with range shifter so that reconstruction of the dose distribution from the measurements is feasible. For plans without range shifter, high frequencies might occur and therefore the sampling rate may not be sufficient for dose reconstruction anymore and the measured points do not provide a full representation of the dose distribution. These scenarios are still comparable to typical clinical situations in photon therapy and in these cases the analysis is only valid at the points of measurement.¹¹ To overcome this limitation, the sampling distance can be increased for these plans by shifting or rotating the array and performing multiple measurements.^{11,17} It is noteworthy that appropriate sampling rate and detector size are the prerequisites for these two-dimensional arrays to guarantee that the measurement process does not introduce additional errors to the measured dose distributions. The actual capability of these detector arrays to detect errors in the treatment delivery process still needs to be investigated in future work based on the results obtained in this study.

- Accepted Article
1. Low DA, Harms WB, Mutic S, Purdy JA. A technique for the quantitative evaluation of dose distributions. *Med Phys*. 1998;25(5):656-661. doi:10.1118/1.598248
 2. Ceberg C. A note on the interpretation of the gamma evaluation index. *J Phys Conf Ser*. 2013;444:012082. doi:10.1088/1742-6596/444/1/012082
 3. Winiecki J, Morgaś T, Majewska K, Drzewiecka B. The gamma evaluation method as a routine QA procedure of IMRT. *Rep Pract Oncol Radiother*. 2009;14(5):162-168. doi:10.1016/S1507-1367(10)60031-4
 4. Hussein M, Clark CH, Nisbet A. Challenges in calculation of the gamma index in radiotherapy – Towards good practice. *Phys Med*. 2017;36:1-11. doi:10.1016/j.ejmp.2017.03.001
 5. Buonamici FB, Compagnucci A, Marrazzo L, Russo S, Bucciolini M. An intercomparison between film dosimetry and diode matrix for IMRT quality assurance. *Med Phys*. 2007;34(4):1372-1379. doi:10.1118/1.2713426
 6. Saminathan S, Manickan R, Chandraraj V, Supe SanjayS. Dosimetric study of 2D ion chamber array matrix for the modern radiotherapy treatment verification. *J Appl Clin Med Phys*. 2010;11(2):116-127. doi:10.1120/jacmp.v11i2.3076
 7. Low DA, Moran JM, Dempsey JF, Dong L, Oldham M. Dosimetry tools and techniques for IMRT: IMRT dosimetry tools. *Med Phys*. 2011;38(3):1313-1338. doi:10.1118/1.3514120
 8. Poppe B, Blehschmidt A, Djouguela A, et al. Two-dimensional ionization chamber arrays for IMRT plan verification: Dose-plan verification with 2D-ARRAYS. *Med Phys*. 2006;33(4):1005-1015. doi:10.1118/1.2179167
 9. Hussein M, Adams EJ, Jordan TJ, Clark CH, Nisbet A. A critical evaluation of the PTW 2D-ARRAY seven29 and OCTAVIUS II phantom for IMRT and VMAT verification. *J Appl Clin Med Phys*. 2013;14(6):274-292. doi:10.1120/jacmp.v14i6.4460
 10. Jursinic PA, Nelms BE. A 2-D diode array and analysis software for verification of intensity modulated radiation therapy delivery. *Med Phys*. 2003;30(5):870-879. doi:10.1118/1.1567831

- Accepted Article
11. Poppe B, Djouguela A, Blechschmidt A, Willborn K, Rühmann A, Harder D. Spatial resolution of 2D ionization chamber arrays for IMRT dose verification: single-detector size and sampling step width. *Phys Med Biol*. 2007;52(10):2921-2935. doi:10.1088/0031-9155/52/10/019
 12. Poppe B, Stelljes TS, Looe HK, Chofer N, Harder D, Willborn K. Performance parameters of a liquid filled ionization chamber array. *Med Phys*. 2013;40(8):082106. doi:10.1118/1.4816298
 13. Markovic M, Stathakis S, Mavroidis P, Jurkovic I-A, Papanikolaou N. Characterization of a two-dimensional liquid-filled ion chamber detector array used for verification of the treatments in radiotherapy: 2D ion chamber detector array for plan verification. *Med Phys*. 2014;41(5):051704. doi:10.1118/1.4870439
 14. Markovic M, Narayanasamy G, Stathakis S, et al. Clinical Evaluation of a Two-dimensional Liquid-Filled Ion chamber Detector Array for Verification of High Modulation Small Fields in Radiotherapy. *J Med Phys*. 2019;44(2):91-98. doi:10.4103/jmp.JMP_4_19
 15. Chan MF, Chen C-C, Shi C, et al. Patient-Specific QA of Spot-Scanning Proton Beams Using Radiochromic Film. *Int J Med Phys Clin Eng Radiat Oncol*. 2017;06(02):111-123. doi:10.4236/ijmpcero.2017.62011
 16. Lin L, Kang M, Solberg TD, et al. Use of a novel two-dimensional ionization chamber array for pencil beam scanning proton therapy beam quality assurance. *J Appl Clin Med Phys*. 2015;16(3):270-276. doi:10.1120/jacmp.v16i3.5323
 17. Lin Y, Bentefour H, Flanz J, Kooy H, Clasié B. Design of a QA method to characterize submillimeter-sized PBS beam properties using a 2D ionization chamber array. *Phys Med Biol*. 2018;63(10):105007. doi:10.1088/1361-6560/aabd89
 18. Arjomandy B, Sahoo N, Ding X, Gillin M. Use of a two-dimensional ionization chamber array for proton therapy beam quality assurance. *Med Phys*. 2008;35(9):3889-3894. doi:10.1118/1.2963990
 19. Varasteh Anvar M, Attili A, Ciocca M, et al. Quality assurance of carbon ion and proton beams: A feasibility study for using the 2D MatriXX detector. *Phys Med*. 2016;32(6):831-837. doi:10.1016/j.ejmp.2016.05.058

20. Lomax A. Intensity modulation methods for proton radiotherapy. *Phys Med Biol*. 1999;44(1):185-205. doi:10.1088/0031-9155/44/1/014
21. Lomax AJ, Boehringer T, Coray A, et al. Intensity modulated proton therapy: A clinical example. *Med Phys*. 2001;28(3):317-324. doi:10.1118/1.1350587
22. Kooy HM, Grassberger C. Intensity modulated proton therapy. *Br J Radiol*. 2015;88(1051):20150195. doi:10.1259/bjr.20150195
23. Zhu X, Li Y, Mackin D, et al. Towards Effective and Efficient Patient-Specific Quality Assurance for Spot Scanning Proton Therapy. *Cancers*. 2015;7(2):631-647. doi:10.3390/cancers7020631
24. Lomax AJ, Böhringer T, Bolsi A, et al. Treatment planning and verification of proton therapy using spot scanning: Initial experiences. *Med Phys*. 2004;31(11):3150-3157. doi:10.1118/1.1779371
25. García-Vicente F, Delgado JM, Peraza C. Experimental determination of the convolution kernel for the study of the spatial response of a detector. *Med Phys*. 1998;25(2):202-207. doi:10.1118/1.598182
26. Looe HK, Stelljes TS, Foschepoth S, Harder D, Willborn K, Poppe B. The dose response functions of ionization chambers in photon dosimetry – Gaussian or non-Gaussian? *Z Für Med Phys*. 2013;23(2):129-143. doi:10.1016/j.zemedi.2012.12.010
27. Looe HK, Harder D, Poppe B. The energy dependence of the lateral dose response functions of detectors with various densities in photon-beam dosimetry. *Phys Med Biol*. 2017;62(3):N32-N44. doi:10.1088/1361-6560/aa54aa
28. Stelljes TS, Looe HK, Harder D, Poppe B. The “collimator monitoring fill factor” of a two-dimensional detector array, a measure of its ability to detect collimation errors. *Med Phys*. 2017;44(3):1128-1138. doi:10.1002/mp.12106
29. Gago-Arias A, Brualla-González L, González-Castaño DM, et al. Evaluation of chamber response function influence on IMRT verification using 2D commercial detector arrays. *Phys Med Biol*. 2012;57(7):2005-2020. doi:10.1088/0031-9155/57/7/2005
30. Laub WU, Wong T. The volume effect of detectors in the dosimetry of small fields used in IMRT. *Med Phys*. 2003;30(3):341-347. doi:10.1118/1.1544678

31. Pappas E, Maris TG, Papadakis A, et al. Experimental determination of the effect of detector size on profile measurements in narrow photon beams. *Med Phys*. 2006;33(10):3700-3710. doi:10.1118/1.2349691
32. Bortfeld T, Oelfke U, Nill S. What is the optimum leaf width of a multileaf collimator? *Med Phys*. 2000;27(11):2494-2502. doi:10.1118/1.1319524
33. Micke A, Lewis DF, Yu X. Multichannel film dosimetry with nonuniformity correction. *Med Phys*. 2011;38(5):2523-2534. doi:10.1118/1.3576105
34. Castriconi R, Ciocca M, Mirandola A, et al. Dose–response of EBT3 radiochromic films to proton and carbon ion clinical beams. *Phys Med Biol*. 2017;62(2):377-393. doi:10.1088/1361-6560/aa5078
35. Zhao L, Das IJ. Gafchromic EBT film dosimetry in proton beams. *Phys Med Biol*. 2010;55(10):N291-N301. doi:10.1088/0031-9155/55/10/N04
36. Khachonkham S, Dreindl R, Heilemann G, et al. Characteristic of EBT-XD and EBT3 radiochromic film dosimetry for photon and proton beams. *Phys Med Biol*. 2018;63(6):065007. doi:10.1088/1361-6560/aab1ee
37. Mackin D, Zhu XR, Poenisch F, et al. Spot-Scanning Proton Therapy Patient-Specific Quality Assurance: Results from 309 Treatment Plans. *Int J Part Ther*. 2014;1(3):711-720. doi:10.14338/IJPT-14-00017.1
38. Park JM, Kim J, Park S-Y, Oh DH, Kim S-T. Reliability of the gamma index analysis as a verification method of volumetric modulated arc therapy plans. *Radiat Oncol*. 2018;13(1):175. doi:10.1186/s13014-018-1123-x
39. Titt U, Mirkovic D, Sawakuchi GO, et al. Adjustment of the lateral and longitudinal size of scanned proton beam spots using a pre-absorber to optimize penumbrae and delivery efficiency. *Phys Med Biol*. 2010;55(23):7097-7106. doi:10.1088/0031-9155/55/23/S10
40. Pedroni E, Scheib S, Böhringer T, et al. Experimental characterization and physical modelling of the dose distribution of scanned proton pencil beams. *Phys Med Biol*. 2005;50(3):541-561. doi:10.1088/0031-9155/50/3/011

41. Bues M, Newhauser WD, Titt U, Smith AR. Therapeutic step and shoot proton beam spot-scanning with a multi-leaf collimator: a Monte Carlo study. *Radiat Prot Dosimetry*. 2005;115(1-4):164-169. doi:10.1093/rpd/nci259
42. Winterhalter C, Lomax A, Oxley D, Weber DC, Safai S. A study of lateral fall-off (penumbra) optimisation for pencil beam scanning (PBS) proton therapy. *Phys Med Biol*. 2018;63(2):025022. doi:10.1088/1361-6560/aaa2ad
43. Dowdell SJ, Clasie B, Depauw N, et al. Monte Carlo study of the potential reduction in out-of-field dose using a patient-specific aperture in pencil beam scanning proton therapy. *Phys Med Biol*. 2012;57(10):2829-2842. doi:10.1088/0031-9155/57/10/2829
44. Bäumer C, Janson M, Timmermann B, Wulff J. Collimated proton pencil-beam scanning for superficial targets: impact of the order of range shifter and aperture. *Phys Med Biol*. 2018;63(8):085020. doi:10.1088/1361-6560/aab79c
45. van't Veld AA, van Luijk P, Praamstra F, van der Hulst PC. Detector line spread functions determined analytically by transport of Compton recoil electrons. *Med Phys*. 2001;28(5):738-751. doi:10.1118/1.1366665
46. Palmans H, Andreo P, Huq MS, Seuntjens J, Christaki KE, Meghzifene A. Dosimetry of small static fields used in external photon beam radiotherapy: Summary of TRS-483, the IAEA–AAPM international Code of Practice for reference and relative dose determination. *Med Phys*. 2018;45(11):e1123-e1145. doi:10.1002/mp.13208
47. Newhauser WD, Zhang R. The physics of proton therapy. *Phys Med Biol*. 2015;60(8):R155-R209. doi:10.1088/0031-9155/60/8/R155
48. Mustafa AAM, Jackson DF. Small-angle multiple scattering and spatial resolution in charged particle tomography. *Phys Med Biol*. 1981;26(3):461-472. doi:10.1088/0031-9155/26/3/009
49. Furukawa T, Inaniwa T, Hara Y, Mizushima K, Shirai T, Noda K. Patient-specific QA and delivery verification of scanned ion beam at NIRS-HIMAC: QA and verification for scanning delivery. *Med Phys*. 2013;40(12):121707. doi:10.1118/1.4828845

50. Wang D, Smith BR, Gelover E, Flynn RT, Hyer DE. A method to select aperture margin in collimated spot scanning proton therapy. *Phys Med Biol.* 2015;60(7):N109-N119. doi:10.1088/0031-9155/60/7/N109

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Conflict of interest

The authors have no conflicts of interest or financial disclosures to state.

Figure 1: Schematic representation of the detector arrangement of the OD1500XDR (a)) and the MatriXX PT detector array (b)).

Figure 2: Pencil beam spot profiles for selected energies and measurement depths and the use of a range shifter (RS) (a) and the corresponding frequency components (b).

Figure 3: a) Line profiles of a 10 x 10 cm² field measured with Film, MatriXX PT and OD1500XDR and corresponding Fourier transform of the film measurement (b)). For better visibility only half of the profile is shown.

Figure 4: a) Line profiles of one field of a head and neck treatment plan with range shifter in 1 cm water equivalent depth calculated with the TPS data and measured with films, the MatriXX PT and the OD1500XDR detector arrays. b) Fourier transform of the TPS data and the film measurement of the line profile.

Figure 5: a) Line profile of one field of a base of skull treatment plan without range shifter in 1cm water equivalent depth calculated with the TPS data and measured with films, the MatriXX PT and the OD1500XDR detector arrays. b) Fourier transform of the TPS data and the film measurement of the line profile.

Figure 6: a) Fourier transform of a TPS dose profile of the base of skull plan and for different sampled data of the TPS profiles with sampling rates according to the investigated detector arrays (MatriXX PT: 7.1 mm, OD1500XDR: 10 mm). The sampling of the dose distributions leads to replication of the frequency components according to the Nyquist frequencies of the arrays and therefore to overlapping frequency components due to violation of the Nyquist theorem. (b) Reconstruction of the dose profile out of the Fourier domain will then lead to erroneous data. (c) Linear interpolation between the measurement points commonly performed clinically may lead to artificial points that fail the gamma index analysis (indicated by arrows).

Figure 7: a) Fourier transform of the line profile of figure 5 and Fourier transforms of step functions as the first estimates of the response functions of the MatriXX PT and the OD1500XDR with the dimensions of the single detector. b) Fourier transform of the TPS data, the OD1500XDR data and the MatriXX PT data. The measurement process leads only to a slight suppression of the high frequency contents in the TPS data.

Figure 8: Line profile through TPS data of one field of the base of skull treatment plan (solid line), measurement points of OD1500XDR and convolution of rectangular response function of OD1500XDR with TPS data (dashed line).

Table 1: Energies and measurement depths in solid water of the single pencil beams

70 MeV	150 MeV	225 MeV
1 cm	1.9 cm ¹⁾	1.9 cm
1.9 cm	7 cm ¹⁾	15 cm
3.6 cm	15 cm	26 cm

¹⁾ Additional measurements with range shifter

Table 2: Evaluation of the spot measurement regarding their penumbra width $w_{80/20}$, full-width at half-maximum (FWHM), maximum frequency f_{max} , corresponding Nyquist frequency f_{Nyq} and minimum sampling distance s_{min} for different energies and measurement depths and the use of a range shifter (RS).

Energy	70 MeV			150 MeV					225 MeV		
Depth [cm]	1	2	3.6	2	7	15	2+RS	7+RS	2	15	26
$w_{80/20}$ [mm]	7.08 ± 0.21	8.18 ± 0.25	7.10 ± 0.21	6.73 ± 0.20	4.68 ± 0.14	5.75 ± 0.17	8.65 ± 0.26	9.50 ± 0.29	3.70 ± 0.11	4.40 ± 0.11	6.85 ± 0.13
FWHM [mm]	14.60 ± 0.44	15.25 ± 0.46	14.85 ± 0.45	10.35 ± 0.31	10.30 ± 0.31	12.05 ± 0.36	17.60 ± 0.53	17.60 ± 0.53	7.15 ± 0.53	8.25 ± 0.21	12.55 ± 0.38
f_{max} [1/mm]	0.087	0.073	0.073	0.150	0.157	0.122	0.073	0.073	0.239	0.132	0.150
f_{Nyq} [1/mm]	0.174	0.146	0.146	0.300	0.315	0.245	0.146	0.146	0.479	0.265	0.300
s_{min} [mm]	5.747	6.844	6.844	3.332	3.175	4.082	6.844	6.844	2.090	3.780	3.332

Table 3: Maximum frequencies $f_{\max}$ and minimum required sampling distances $s_{\min}$ of the three different plans evaluated for the single fields. Plan 1 and 2 are head and neck plans with range shifter, plan 3 is a base of skull plan without range shifter. Fields that exceed the sampling frequencies of the arrays (MatrixX PT: 0.132/mm, OD1500XDR: 0.141/mm) are printed in bold type.

	Field 1			Field 2			Field 3			Field 4		
Plan 1												
Depth [cm]	1	6	12	1	2.5	6	1	6	12	1	2.5	6
$f_{\max}$ [mm ⁻¹]	0.048	0.046	0.050	0.048	0.050	0.056	0.048	0.048	0.048	0.048	0.048	0.058
$s_{\min}$ [mm]	10.45	10.89	10.04	10.45	10.04	8.98	10.45	10.45	10.45	10.45	10.45	8.68
Plan 2												
Depth [cm]	1	8	13	1	8	13	1	8	13	1	8	13
$f_{\max}$ [mm ⁻¹]	0.042	0.046	0.048	0.056	0.054	0.044	0.046	0.042	0.044	0.038	0.050	0.040
$s_{\min}$ [mm]	11.91	10.89	10.45	8.98	9.319	11.38	10.89	11.91	11.38	13.13	10.04	12.49
Plan 3												
Depth [cm]	1	3	8	1	8	10.5	1	10.5	13			
$f_{\max}$ [mm ⁻¹]	0.065	0.065	0.067	0.083	0.077	0.075	0.075	0.069	0.065			

a)

b)

mp_14346_f1.tif

mp_14346_f2.tif

mp_14346_f3.tif

a)

b)

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a)

b)

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