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Enhanced x-ray imaging for a thin film cochlear implant with metal artefacts using phase retrieval tomography

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Abstract

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Keywords

phase, artefacts, metal, implant, cochlear, tomography, retrieval, film, x, thin, imaging, ray, enhanced

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Enhanced x-ray imaging for a thin film cochlear implant with metal artefacts using phase retrieval tomography

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Phase retrieval tomography has been successfully used to enhance imaging in systems that exhibit poor absorption contrast. However, when highly absorbing regions are present in a sample, so-called metal artefacts can appear in the tomographic reconstruction. We demonstrate that straightforward approaches for metal artefact reconstruction, developed in absorption contrast tomography, can be applied when using phase retrieval. Using a prototype thin film cochlear implant that has high and low absorption components made from iridium (or platinum) and plastic, respectively, we show that segmentation of the various components is possible and hence measurement of the electrode geometry and relative location to other regions of interest can be achieved. © 2012 American Institute of Physics. [<http://dx.doi.org/10.1063/1.4724343>]

I. INTRODUCTION

Phase retrieval tomography is used in many research applications^{1,2} to image weakly absorbing structures that are difficult to image using the conventional absorption contrast mode. Phase contrast imaging^{3–6} is an imaging modality using an x-ray phase shift produced by the object. Such phase shifts result from refraction of the x-ray beam through the sample and produce a modified intensity image. Phase retrieval is an inversion step which retrieves the phase change produced by an object from the measured intensity image. This can be done by a variety of techniques^{7,8} including, interferometry,⁹ analyser-based methods,¹⁰ grating methods,¹¹ and propagation-based methods. Propagation-based methods developed include those suitable for the inversion of data measured in the near Fresnel regime,¹² the intermediate Fresnel regime^{13,14} and, more recently, in the far field.^{15,16} Methods that use an assumption of sample homogeneity¹⁷ and methods that incorporate polychromaticity in the source have also been developed.¹⁸ In phase retrieval (PR) tomography, the projected intensity data in each angular position may be subjected to phase retrieval, which in turn may be reconstructed to obtain the three dimensional distribution of the sample phase.^{2,14} Alternatively, direct algorithms that combine phase retrieval and filtered back-projection steps into a single operation have been demonstrated.^{19,20} This direct method has been applied also to image real samples.²¹ Apart from the ability to image low density objects, phase retrieval tomography is also useful when imaging objects that exhibit almost uniform absorption properties but possesses a significant difference in phase shift properties.²²

While phase contrast and phase retrieval are used to deal with cases where there is not enough absorption, metal artefact reduction (MAR) is used to deal with the case where there is too much. Highly absorbing regions in a sample can exceed the dynamic range of the detection system when

imaging a sample reducing the information available for the 3D reconstruction. For x-ray based laboratory sources as used in this article, simply increasing the tube voltage typically does not provide enough penetration in the beam to ameliorate the situation. Accordingly, many different processing approaches have been developed to reduce this effect. One approach replaces the high absorbing region (commonly referred to as the metal, although it need not be actually metal) with a lower attenuation coefficient object.²³ The most popular method incorporated into metal artefact reduction software uses a post-reconstruction image processing algorithm²⁴ that treats the metal as an opaque object. There are also corrections in the form of an iterative process involving projection and reconstruction images.²⁵

Combination of a metal artefact reduction process with the phase retrieval step has the potential to enrich the imaging methods available for 3D imaging by making it possible to image objects that possess regions of both high and low absorption materials.

One such class of object is medical implants that are surgically implanted into the body to restore body function. Implants inside the body may possess regions where metal or bone and plastic or human tissue are juxtaposed. It can often be important to image implants either *in vivo* or *in situ* post-mortem so that positioning and interaction of the implant with the host material can be measured. In this demonstration, we analyse a prototype cochlear implant that uses a thin film plastic substrate to incorporate highly absorbing iridium electrodes.²⁶ Of interest here is the geometry of the components in the implant and its relationship to the anatomy of the cochlea.

II. METHODS

A. Phase retrieval algorithm

In this paper, we use a phase retrieval approach^{17,27} that has been adapted for use with a polychromatic source¹⁸ and

which has been demonstrated to be suitable for experimental use due to its stability in the presence of noise.² The so-called polychromatic single-plane phase retrieval step can be written as

$$\varphi(\mathbf{r}) = -k_{poly}\delta_{poly}\mathbb{F}^{-1}\left(\frac{1}{\mu_{poly} + z\delta_{poly}\mathbf{u}^2}\mathbb{F}\left[\frac{I_{poly}^z(\mathbf{r})}{I_{poly}^{in}(\mathbf{r})} - 1\right]\right), \quad (1)$$

where $\mathbb{F}$ is the Fourier transform operator, $\mathbf{u}$ is the Fourier variable conjugate to the position coordinates, $\mathbf{r}$, μ_{poly} , and δ_{poly} are the spectrally weighted attenuation coefficient and the spectrally weighted decrement of the real part of the refractive index, respectively, where in the narrow band approach²⁸ we can use the effective energy for μ and δ , I^z is the intensity at a distance z , I^{in} is the intensity entering the sample, k_{poly} is the weighted wavenumber, and φ is the retrieved phase. It is readily seen that phase retrieval can be described as a filtering process with a filter term of $\frac{1}{\mu_{poly} + z\delta_{poly}\mathbf{u}^2}$. For typical experimental values, this is a low-pass filter. However, it should be noted that the phase retrieval filter is particular to the physics of diffraction. The phase retrieval filter term acts to remove fringes from the phase contrast intensity image and maps these back to the phase of the sample. It is readily seen that a naive application of a generic low-pass filter, familiar from the literature of image processing, will act in most cases simply to smooth the intensity image. Figure 1 shows the difference between phase retrieval and a common low-pass filter. In this figure, a phase contrast image (Fig. 1(a)) has been treated with both the low-pass Butterworth filter (Fig. 1(b)) and the phase retrieval filter (Fig. 1(c)). While the Butterworth filter, $B(\mathbf{u}) = \frac{1}{1 + (\frac{\mathbf{u}}{c})^{2n}}$, has a form that is the same as the phase retrieval filter it is obvious that unless the physics is known it is highly unlikely that an appropriate choice of the constants, c and n will be made.

B. Metal artefact correction

The PR step (Eq. (1)) that we use assumes the sample is homogeneous. It has also been demonstrated that high-quality segmentation can be achieved for non-homogenous samples.² By undertaking a MAR step, which consists of replacing high absorption regions in the measured intensity with lower absorption, before the PR step we create a data set that is similar to that produced by a homogenous sample. Accordingly, we expect this approach to create phase retrieved images that are readily segmented. Conversely, if the PR step is performed before the MAR step, then the data conform more poorly to the homogenous sample assumption and we expect the phase retrieval to produce additional artefacts. This assertion was tested as part of our experimental analysis as described below.

III. EXPERIMENTAL

A. Material

A thin film cochlear implant wrapped with plastic fibres was used in this study. The plastic fibres surrounding the cochlear implant simulate the presence of soft tissue. The rat animal model electrode array contains 32 individually addressable microelectrodes (Fig. 2(a) item B) to allow more discrete populations of neurons to be targeted than the current dominant commercial human device which consists of 22 electrodes²⁹ (Fig. 2(a) item A). The electrodes are thin layers of dense iridium sandwiched between two thin polymer sheets. The iridium is formed into tracks with round electrode pads exposed via through holes. Scanning electron microscopy was attempted to image the sample but was seen to damage the plastic at 15 kV accelerating voltage, as shown in Fig. 2(b). The most accurate distance measurements can be expected using 3D micro computed tomography (CT) imaging, which is also capable of measurements *in vivo*.

FIG. 1. Phase retrieval and image smoothing with the Butterworth low-pass filter. (a) A phase contrast image for a specific geometry and wavelength, (b) the Butterworth low-pass filtered image, (c) phase retrieved image, (d) a plot showing the value of both filters.

FIG. 2. (a) Optical microscopy of a (A) typical animal model cochlear implant and (B) the new thin film cochlear implant prototype. The optical image was obtained using a BX61TRF microscope (Olympus). (b) Scanning electron micrograph of the thin film cochlear implant displaying severe charging artefacts. Scanning electron micrographs were obtained on a JSM 6340 F (Jeol) at 15 kV accelerating voltage.

B. X-ray micro-computed tomography instrument

The CT imaging was undertaken in the Department of Physics, La Trobe University, using the x-ray micro computed tomography instrument (Xradia, Inc., USA). An x-ray source with a closed tube and a tungsten target was operated at 60 kV tube voltage and a power of 10 W. The source size of this laboratory-based x-ray tube system is sufficiently small (in our case $7\ \mu\text{m}$) to produce phase contrast imaging. The thin film cochlear implant was placed 100 mm from the source and 30 mm from the detector. The imaging detector was a CCD camera coupled with a scintillator system and objective lenses. The CCD camera (Andor Technology) has 2048×2048 pixels with a physical pixel size of $13.5\ \mu\text{m}$. A $10\times$ magnification objective lens was used. By taking the geometric magnification, due to the distances between the source, sample, and detector, into account the effective pixel size for this set-up was $1.03\ \mu\text{m}$. Each projection image was recorded in 60 s. Each image was corrected for the dark current image and for the non-uniform illumination in the imaging system, determined by taking a reference image of the beam without the sample present. In order to collect the three dimensional data-set, a large number of projection images are obtained by rotating the sample. The sample was scanned by acquiring 721 projections at equal angles through an angular range of 180° . A filtered back projection algorithm (TXM Reconstructor, Xradia, Inc.) was applied to reconstruct the three dimensional image of the sample. $2\times$ binning was used in the three dimensional reconstruction to reduce the size of the reconstructed data. The total reconstructed volume contains $1024 \times 1024 \times 1024$ voxels, with a voxel size of $2.06\ \mu\text{m}$.

IV. RESULTS AND DISCUSSION

Figure 3(a) shows one of the intensity projection images used in this study. The black circular features are the high absorbing iridium electrodes, while the surrounding plastic fibres are more transparent to x-rays. The plastic substrate and the fine contacts in the implant are not visible in the projection image. Figures 3(c), 3(e), 3(g), and 3(i) show the end results of interchanging the MAR and PR steps. It is observed that Fig. 3(c) is the closest to the expected phase retrieved image where the phase contrast effects seen in Fig. 3(a) are mapped back to objects without fringing. In the case of Fig. 3(e), the incorrect filter kernel acts to smooth the image and does not remove fringes. When the PR step is

performed first as is the case in Fig. 3(g) it can be seen that the departure from homogeneity is such that the kernel mismatch between the high and low absorption regions distorts the imaging of the electrodes. Finally, in the case of Fig. 3(i), it can again be seen that the incorrect choice of filter kernel means that no effective remapping of fringes for the fine features takes place.

Initially, we present the three dimensional reconstructed image with no metal artefact correction and no phase retrieval, as shown in Fig. 4. The effect of the metal artefact, due to the presence of the high density iridium, is to distort the true implant geometry and results in the grey-scale range of dissimilar features overlapping, thus preventing the possibility of segmentation based on the reconstructed grey-scale. Additionally, the metal artefact removes information at the metal/environment interface. This can be seen in XZ plane of Fig. 4 that the fine contact looks unconnected to the round electrode. This artefact would prevent accurate measurement of the electrode/neuron distance when used for *in vivo* testing.

Next, we present the three dimensional reconstructed image with only metal artefact correction and no phase retrieval, as shown in Fig. 5. The reduction of the high absorption regions has been done in each intensity projection image before the 3D reconstruction algorithm is applied. In Fig. 5, XY, YZ, and XZ planes are chosen in the same location as in Fig. 4. It can be seen that the distortions and grey-scale confusion due to the metal artefact are both reduced but as no phase retrieval has been performed, the low-density features are not well recovered for segmentation.

Finally, we present the three dimensional reconstructed image which includes both the metal artefact correction and phase retrieval in the process, as shown in Fig. 6. The data set of each intensity projection was subjected to both the MAR and PR processes as detailed in Sec. II. The resulting Fig. 6 shows that it is obvious that the metal artefact is significantly reduced and the contrast of the plastic fibres is greatly enhanced in comparison with Figs. 4 and 5. A loss of detail (dark regions) around the iridium electrode initially seen in the XY and XZ planes is also corrected. The unconnected fine contact in Fig. 4 (XZ plane) now appears connected. It allows an accurate measurement to be done around the electrode-environment interface. The position of the XY, YZ, and XZ planes in Figure 6 is exactly the same as shown in Figs. 4 and 5. The poorly defined image of the fibres seen in Figs. 4 and 5 is significantly improved. The image in

FIG. 3. Various sequences for the image processing steps. (a) intensity projection as raw data, (b) after intensity reduction of the high absorption regions followed by (c) the phase retrieval step using the material's properties of the low absorption region to define the PR filter term (kernel). (d) The same as (b) followed by (e) the same as (c) but using the high absorption material to define the filter kernel. In (f) and (g), the steps taken in (b) and (c) are interchanged and in (h) and (i) the steps taken in (d) and (e) are interchanged.

FIG. 4. 3D tomographic image of a thin film cochlear implant wrapped with plastic fibres. Different views of the XY, YZ, and XZ planes are presented. There is no MAR and no PR applied.

FIG. 5. 3D tomographic image with only MAR applied to the projection images but no PR.

FIG. 6. 3D tomographic image with MAR and PR applied to the projection images. The metal artefact is significantly reduced, and the contrast of the fibres is improved.

Fig. 6 can now be used for further analysis such as image segmentation.

While the homogenous sample approximation for the phase retrieval step of Eq. (1) is clearly violated, the MAR step means that the dissimilar components are sufficiently similar that qualitative data suitable for segmentation can be obtained.² As Fig. 3 demonstrates violating the homogenous sample approximation can result in geometric distortions. Further study may result in some rule of thumb for what level of similarity between materials will result in acceptable non-distorted phase retrieved images. In the meantime, it is possible to use *a priori* information about the sample to test the assertion that a given retrieval has produced a data set suitable for segmentation that will provide useful geometric data.

The segmentation process was undertaken with Avizo (Mercury Computer Systems) for 3D image analysis. Automated segmentation based on the grey scale was performed for the electrodes and the surrounded fibres. For the plastic substrate and fine contacts, a semi-automatic approach where the operator chose localised grey level thresholds between slices and then interpolated the resulting segmentations between slices was used. In principle, this approach could also be automated with more image-processing resources. Volume renderings of the cochlear implant from the segmented image are presented in Fig. 7. The ability of the approach to extract useful information from the sample is obvious in that the geometry of the electrodes is clearly identified, it can be seen that the fine contacts are connected to the electrodes and there is no missing information around the electrodes. Therefore, the electrodes and their distances to

FIG. 7. Volume rendering of the thin film cochlear implant after image segmentation.

other locations can now be easily measured. The surrounding plastic fibres are also recognized plainly.

V. CONCLUSIONS

We have developed a simple and practical phase retrieval algorithm for samples that suffer from metal artefacts in 3D imaging. The metal artefacts due to high absorbing materials are corrected and the contrast of weakly absorbing materials is enhanced by this algorithm. The final result allows for further image analysis and segmentation, which was not possible prior to metal artefact removal and phase retrieval.

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