Multi-Mode C-Arm Fluoroscopy, Tomosynthesis, and Cone-Beam CT for Image-Guided Interventions: From Proof of Principle to Patient Protocols

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ABSTRACT

High-performance intraoperative imaging is essential to an ever-expanding scope of therapeutic procedures ranging from tumor surgery to interventional radiology. The need for precise visualization of bony and soft-tissue structures with minimal obstruction to the therapy setup presents challenges and opportunities in the development of novel imaging technologies specifically for image-guided procedures. Over the past ~5 years, a mobile C-arm has been modified in collaboration with Siemens Medical Solutions for 3D imaging. Based upon a Siemens PowerMobil, the device includes: a flat-panel detector (Varian PaxScan 4030CB); a motorized orbit; a system for geometric calibration; integration with real-time tracking and navigation (NDI Polaris); and a computer control system for multi-mode fluoroscopy, tomosynthesis, and cone-beam CT. Investigation of 3D imaging performance (noise-equivalent quanta), image quality (human observer studies), and image artifacts (scatter, truncation, and cone-beam artifacts) has driven the development of imaging techniques appropriate to a host of image-guided interventions. Multi-mode functionality presents a valuable spectrum of acquisition techniques: i.) fluoroscopy for real-time 2D guidance; ii.) limited-angle tomosynthesis for fast 3D imaging (e.g., ~10 sec acquisition of coronal slices containing the surgical target); and iii.) fully 3D cone-beam CT (e.g., ~30-60 sec acquisition providing bony and soft-tissue visualization across the field of view). Phantom and cadaver studies clearly indicate the potential for improved surgical performance – up to a factor of 2 increase in challenging surgical target excisions. The C-arm system is currently being deployed in patient protocols ranging from brachytherapy to chest, breast, spine, and head and neck surgery.

Keywords: flat-panel detector, fluoroscopy, tomosynthesis, cone-beam CT, 3D imaging, imaging performance, radiation dose, image-guided surgery, navigation, head and neck surgery, orthopedic surgery, brachytherapy

1. INTRODUCTION

The development and availability of high-performance flat-panel detectors (FPDs) has facilitated a broad range of applications in dynamic and tomographic imaging. Because such applications involve the collection of multiple x-ray projections (e.g., 100 or more), there are at least two key performance characteristics required of the FPD: i.) high detective quantum efficiency (DQE), providing quantum-limited imaging performance at a low dose per frame; and ii.) full-frame readout at high frame rates (1 fps or greater). Other performance characteristics that are particularly important in tomographic applications include: large-area format (sufficient to cover the volumetric region of interest and minimize lateral truncation artifacts); low image lag and ghosting effects; and stable dark- and flood-field characteristics (providing high-quality dark-flood correction). Several currently available FPD designs satisfy these basic performance requirements and have augmented a broad base of research ranging from diagnostic imaging (e.g., 3D breast imaging) to image-guided procedures. Developments on the horizon – e.g., higher-performance FPDs and fabrication on flexible substrates – promise an even broader scope of research, development, and clinical implementation.

For image-guided interventions, including interventional radiology, surgery, and radiation therapy, the development of novel, high-performance dynamic and tomographic imaging offers an important opportunity for increased therapy.
Figure 1. Illustration of the C-arm prototype and example multi-mode images.
(a) Photograph of the C-arm in the laboratory. 3D image acquisition typically involves a semi-circular orbit of the x-ray tube and detector, as shown in relation to a cadaveric head. Integration with a tracking and navigation system [e.g., based on a stereoscopic infrared camera (NDI Polaris)] provides real-time visualization of surgical tools within the context of intraoperative image data.
(b-d) Example images of a cadaveric head acquired with the C-arm, including: (b) a single lateral projection from a fluoroscopic sequence (1 fps; 2.6 mA); (c) a sagittal slice from a 3D tomosynthesis reconstruction (~34 projections acquired over an orbit of ~30°); and (d) a sagittal slice from a 3D cone-beam CT reconstruction (~200 projections acquired over 178°).

precision. Such capability offers to: i.) increase the performance of existing interventional techniques (e.g., tumor ablation and normal tissue avoidance); ii.) expand the application of interventional techniques to conventionally “untreatable” disease; iii.) support innovation in advanced interventional techniques (particularly those that require increased geometric precision and/or integration of multiple imaging modalities); and iv.) help to uncover the fundamental factors determining treatment outcome, particularly in cases where geometric precision is a limiting factor.

Over the past ~5 years, our research has focused on the development and application of an intraoperative imaging system based on a mobile isocentric C-arm (Siemens PowerMobil) capable of real-time fluoroscopy, tomosynthesis, and cone-beam CT (CBCT). The system was originally developed in collaboration with Siemens Medical Solutions and is under investigation in a broad range of image-guided procedures, including image-guided surgery (e.g., orthopedic, spine, head and neck, lung, and breast surgery) and brachytherapy. These investigations have provided valuable testing and evaluation of system performance, identified strategies for improving system design, and demonstrated the potential for improving surgical performance and precision.

This paper reports on a number of important technical challenges in system development and summarizes early progress in translating the technology from phantom and cadaver experiments to clinical trials conducted under research protocol in various interventional procedures. Technical aspects include: i.) modifications to the C-arm required for 3D imaging capability; ii.) implementation of a high-performance FPD with multiple gain modes; iii.) a robust method for geometric calibration; and iv.) investigation of image quality and radiation dose. Clinical applications currently under investigation include: i.) orthopedic surgery (advanced fracture reduction techniques); ii.) spine surgery (e.g., guidance of vertebroplasty and photodynamic therapy); iii.) head and neck surgery, including skull base surgery and anterior approach to pituitary and brain stem lesions; iv.) lung surgery (resection of sub-palpable nodules); v.) breast surgery (resection of sub-palpable masses); and vi.) brachytherapy (image guidance auxiliary to transrectal ultrasound).
2. C-ARM PLATFORM FOR INTRA-OPERATIVE 3D IMAGING

2.1 Modifications to the PowerMobil C-Arm Platform

The C-arm prototype is illustrated in Fig. 1. The main modifications to the PowerMobil platform include: i.) incorporation of a high-performance FPD (PaxScan 4030CB, Varian Imaging Products, Palo Alto CA); ii.) a collimator with expanded field of view; iii.) added x-ray filtration appropriate to CT imaging (2 mm Al + 0.1 mm Cu); iv.) a servo drive for C-arm rotation; v.) a method for geometric calibration (detailed in Reference 13 and summarized below); and vi.) integration with a computer control system (Dell Precision650, dual 2.0 GHz CPU, 3 GB RAM) for synchronized x-ray exposure, image readout, and 3D reconstruction. Software for image acquisition, processing, and reconstruction was based on a system initially developed for CBCT-guided radiation therapy,4 adapted to the C-arm system.

2.2 Flat-Panel Detector with Multiple Gain Settings

As described by Roos et al.,14 the FPD (PaxScan 4030CB) allows image readout in fixed, dual, and dynamic gain modes, requiring slightly different processing techniques and offering extended dynamic range. The fixed-gain mode is identical to the conventional method of FPD readout, providing a single gain value irrespective of signal level. The dual-gain mode investigated below involves alternating rows read at low- and high-gain, where the latter switches a second feedback capacitor in series with the first. The dynamic gain mode employs an adjustable threshold voltage to which the pixel signal is compared, switching between low- and high-gain depending on the signal level.

CBCT images of an anthropomorphic head phantom containing simulated soft-tissue structures and natural skeletal anatomy were acquired at various dose levels in each of the gain modes. As shown in Fig. 2, image quality in fixed-gain mode was degraded at high dose levels due to artifacts arising from pixel saturation, and dual-gain mode was compromised at the lowest dose levels due to increased image noise. Dynamic gain mode provided the best image quality at all dose levels. Table I summarizes basic characteristics associated with the multiple gain modes.

![Figure 2](image-url)

Figure 2. CBCT images of a ~(8 x 8) cm² region of interest within an anthropomorphic head phantom, illustrating (a) soft-tissue visualization and (b) bony detail in images acquired at various dose levels and in each gain mode.

<table>
<thead>
<tr>
<th>Gain Mode</th>
<th>Pixel Format</th>
<th>Max fps</th>
<th>Analog Gain</th>
<th>Pixel Dark Noise (ADU)</th>
<th>Gain (ADU/mR)</th>
<th>Saturation (mR)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fixed</td>
<td>1024×768</td>
<td>30</td>
<td>2</td>
<td>7.4</td>
<td>70,400</td>
<td>13,200</td>
</tr>
<tr>
<td>Fixed</td>
<td>2048×1536</td>
<td>7.5</td>
<td>2</td>
<td>5.0</td>
<td>24,400</td>
<td>14,500</td>
</tr>
<tr>
<td>Dual-Gain: High-Gain</td>
<td>1024×1536</td>
<td>15</td>
<td>1</td>
<td>3.8</td>
<td>24,000</td>
<td>14,200</td>
</tr>
<tr>
<td>: Low-Gain</td>
<td></td>
<td></td>
<td></td>
<td>2.8</td>
<td>3,200</td>
<td>14,100</td>
</tr>
<tr>
<td>Dynamic: High-Gain</td>
<td>1024×768</td>
<td>30</td>
<td>1</td>
<td>4.3</td>
<td>47,900</td>
<td>n/a</td>
</tr>
<tr>
<td>: Low-Gain</td>
<td></td>
<td></td>
<td></td>
<td>2.2</td>
<td>6,300</td>
<td>14,300</td>
</tr>
</tbody>
</table>

Table I. Summary of readout, noise, and saturation characteristics in various multiple gain modes.
2.3 Geometric Calibration

The geometric calibration method of Cho et al.\textsuperscript{13} was adapted to the C-arm system. The method provides a direct analytical (non-iterative) solution of all nine degrees of freedom in source and detector geometry [i.e., focal spot shifts in $(x,y,z)$, detector shifts in $(x,y,z)$, and detector rotation in $(\theta,\phi,\eta)$]. The method employs a phantom incorporating two rings of tungsten BBs within an acrylic cylindrical shell as shown in Fig. 3. From the centroid position of each BB, connected to that of the diametrically opposed BB in the opposite ring, the piercing point of the imaging system (along with all nine degrees of freedom and other derived system parameters, such as SDD, SAD, etc.) are evaluated by a direct analytic algorithm.\textsuperscript{13} As shown in Fig. 3, the geometric non-idealities of the source-detector orbit are significant, with departures from a semicircle of up to ~15 mm due to mechanical flex in the C-arm. Previous studies\textsuperscript{9} indicate that the source-detector orbit is highly reproducible, allowing these geometric non-idealities to be fully corrected in 3D reconstruction.

![Figure 3](image)

**Figure 3.** Geometric calibration. (a) Illustration of system geometry, showing the two-circle BB phantom. The coordinates $(x_w,y_w,z_w)$ identify the world reference frame. (b) Example projection image of the calibration phantom, illustrating the relationship between diametrically opposed BB locations and the piercing point. (c) Derived geometric parameters, including the piercing point in detector coordinates $(u_o,v_o)$ and source-to-detector distance, SDD. (d-f) Nine degrees of freedom in source and detector geometry as a function of projection angle. Note the significant departures from a semicircular orbit in the $(x,y)$ positions of the source and detector due to mechanical flex in the C-arm gantry.

### 3. Imaging Performance and Radiation Dose

#### 3.1 Multi-Mode Image Acquisition: Fluoroscopy, Tomosynthesis, and Cone-Beam CT

The C-arm platform offers multi-mode imaging capability for fluoroscopy, tomosynthesis, and cone-beam CT for interventional guidance. Such capability provides a useful spectrum of functionality to be utilized according to the imaging task (e.g., 3D localization, soft-tissue visualization, etc.) and clinical constraints (e.g., time and dose). Tomosynthesis and cone-beam CT represent a continuum in which image quality requirements (e.g., visualization of bone or soft-tissues, spatial resolution, etc.) should be considered in relation to the time available for image acquisition and reconstruction as well as the radiation dose delivered over the course of the procedure. As summarized in Table II, tomosynthesis acquisition from a limited arc offers reduction in time and dose, with tradeoffs in image quality as discussed below.
1.0
0.8
0.6
0.4
::-30 -20 -10 0
10 20 30
y (mm)
Acquisition, tacq (s) Processing, tproc (s) Reconstruction, trecon (s)
Dynamic Gain Full Volume Sub-Volume

<table>
<thead>
<tr>
<th>$\theta_{tot}$</th>
<th>$N_{proj}$</th>
<th>$D_{center}$ (mGy)</th>
<th>D$_{vol}$ (pixels)</th>
<th>Processing, $t_{proc}$ (s)</th>
<th>Recontruction, $t_{recon}$ (s)</th>
<th>Sub-Volume</th>
</tr>
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<tbody>
<tr>
<td>10$^o$</td>
<td>12</td>
<td>0.6</td>
<td>$(1024 \times 768)$</td>
<td>3</td>
<td>3</td>
<td>HiRes</td>
</tr>
<tr>
<td>20$^o$</td>
<td>22</td>
<td>1.1</td>
<td>$(512 \times 384)$</td>
<td>2</td>
<td>2</td>
<td>LoRes</td>
</tr>
<tr>
<td>30$^o$</td>
<td>34</td>
<td>1.6</td>
<td>$(256 \times 192)$</td>
<td>5</td>
<td>5</td>
<td>HiRes</td>
</tr>
<tr>
<td>60$^o$</td>
<td>68</td>
<td>3.3</td>
<td>$(4 \times 4)$</td>
<td>14</td>
<td>14</td>
<td>LoRes</td>
</tr>
<tr>
<td>90$^o$</td>
<td>102</td>
<td>4.9</td>
<td>$(2 \times 2)$</td>
<td>30</td>
<td>30</td>
<td>HiRes</td>
</tr>
<tr>
<td>120$^o$</td>
<td>134</td>
<td>6.5</td>
<td>$(20 \times 2)$</td>
<td>40</td>
<td>40</td>
<td>LoRes</td>
</tr>
<tr>
<td>178$^o$</td>
<td>200</td>
<td>9.7</td>
<td>$(60 \times 30)$</td>
<td>39</td>
<td>39</td>
<td>HiRes</td>
</tr>
</tbody>
</table>

Table II. Summary of time, dose, and reconstruction characteristics of 3D tomosynthesis and cone-beam CT. The terms “HiRes” and “LoRes” refer to voxel sizes 0.4 mm and 0.8 mm (isotropic), respectively.

3.2 Imaging Performance

The fluoroscopic imaging performance of the FPD has been previously reported, including characterization of image lag (first-frame lag ~2-5%) and detective quantum efficiency (DQE(0) >60%). Similarly, the image quality in cone-beam CT reconstructions has been a subject of considerable investigation, including spatial resolution, noise and noise-power spectrum, image artifacts, and qualitative evaluation of image quality. 16-22 For tomosynthesis, numerous reconstruction techniques may be considered, including shift-and-add (SAA), filtered backprojection (FBP), iterative, and algebraic reconstruction. Tomosynthesis reconstructions herein are based upon 3D FBP of projection data acquired over a limited angle, designated $\theta_{tot}$, with further improvements in limited-angle projection weighting, filtering, and noise reduction under investigation. Results below illustrate the spatial resolution, image quality, and radiation dose ranging from a limited arc ($\theta_{tot}$ ~10$^o$) to a source-detector trajectory equivalent to cone-beam CT ($\theta_{tot}$ ~178$^o$).

Image reconstructions of a uniform polyurethane cylinder (~16 cm diameter) containing a 6.4 mm diameter sphere (~165 HU contrast to background) are shown in Fig. 4. Visualization of the sphere within the coronal (x,z) plane is maintained at $\theta_{tot}$ down to ~10$^o$, and the associated spread function in the x and z directions is unchanged (FWHM consistent with the sphere diameter). In the y direction, however, as evident in the axial (x,y) and sagittal (y,z) views, out-of-plane blur degrades the spread function significantly for smaller tomosynthetic arcs. As shown in Fig. 4(c), the FWHM in the y-direction degrades by a factor of ~5 between $\theta_{tot}$ of 178$^o$ and ~30$^o$. Thus as expected, tomosynthesis is seen to provide excellent in-plane visualization and spatial resolution (in this case, in coronal planes), but the choice of tomosynthesis angle must be weighed against the criticality of spatial resolution in y (i.e., depth resolution) in the imaging task.

Example 3D images of a cadaveric head reconstructed as a function of tomosynthesis angle are shown in Fig. 5. For image-guided interventions such as endoscopic sinus surgery, head and neck surgery, and skull base surgery, relevant imaging tasks include both high-contrast features (e.g., sinus air cells, turbinates, and clivus) and soft-tissue structures (e.g., optic nerves, carotid, and pituitary gland). The ability of CBCT to satisfy such guidance tasks has been previously reported and found to be of significant value in guiding soft-tissue resections adjacent to critical structures. 6-8 For tomosynthesis, the results suggest adequate visualization of high-contrast features at a fairly limited arc. For example,
visualization of the clivus (skull base) in sagittal views is exquisite down to $\theta_{\text{tot}} \sim 30^\circ$, with minimal interference from out-of-plane structures [compared, for example, to the lateral projection in Fig. 2(b)].

<table>
<thead>
<tr>
<th>$\theta_{\text{tot}}$</th>
<th>Coronal</th>
<th>Sagittal</th>
<th>Axial Dose Map</th>
</tr>
</thead>
<tbody>
<tr>
<td>$10^\circ$</td>
<td>![Coronal Image]</td>
<td>![Sagittal Image]</td>
<td>![Dose Map Image]</td>
</tr>
<tr>
<td>$30^\circ$</td>
<td>![Coronal Image]</td>
<td>![Sagittal Image]</td>
<td>![Dose Map Image]</td>
</tr>
<tr>
<td>$60^\circ$</td>
<td>![Coronal Image]</td>
<td>![Sagittal Image]</td>
<td>![Dose Map Image]</td>
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<td>$90^\circ$</td>
<td>![Coronal Image]</td>
<td>![Sagittal Image]</td>
<td>![Dose Map Image]</td>
</tr>
<tr>
<td>$178^\circ$</td>
<td>![Coronal Image]</td>
<td>![Sagittal Image]</td>
<td>![Dose Map Image]</td>
</tr>
</tbody>
</table>

**Figure 5.** Example 3D reconstructions of a cadaveric head, showing coronal and sagittal planes reconstructed at various tomosynthesis angles, ranging from $10^\circ$ to $178^\circ$ (the last equivalent to the cone-beam CT orbit). The radiation dose (mGy) imparted within the central axial plane is shown in the right column.

### 3.3 Radiation Dose

The radiation dose associated with intraoperative CBCT has been previously reported. As shown in Fig. 5, the dose imparted within the axial plane was computed based on the system geometry and normalized to the dose measured at the...
center of a 16 cm cylindrical dosimetry phantom and at the 4 cardinal locations ~1 cm within the phantom. Dose values are shown corresponding to a tube mA (2.6 mA) for which CBCT images reconstructed from 200 projections (~160 mAs total, 9.7 mGy at isocenter) exhibit soft-tissue visualization sufficient for image guidance. Dose to isocenter decreases with the number of projections (i.e., with \( T_{\text{tot}} \)), since the dose per projection is fixed, falling from ~9.7 mGy for full CBCT to ~1.5 mGy for \( T_{\text{tot}} \sim 30^\circ \). Combined with the image quality considerations discussed above, these results point to the use of intraoperative tomosynthesis in a manner that minimizes the total radiation dose delivered over the course of a procedure. Specifically, CBCT could be employed at milestones within the procedure requiring high image quality and soft-tissue visualization – e.g., at the beginning of a procedure (for target delineation), at critical intraoperative junctures (to visualize normal anatomy with respect to residual tumor), and at the conclusion (to verify surgical product). Intermittently, tomosynthesis could be employed for purposes of fast, high-contrast feature visualization at low dose. Together, C-arm tomosynthesis and CBCT would allow repeat intraoperative imaging for guidance and verification with total dose delivered over the course of the procedure less than that of a typical diagnostic CT scan.

3.4 Image Artifacts

CBCT is subject to a host of image artifacts – many common to transaxial CT, such as beam-hardening, and others amplified and more onerous for FPD-based CBCT, such as x-ray scatter, and lateral truncation. X-ray scatter artifacts and methods for correction have been a subject of considerable research. For the C-arm system, truncation of the object in projection data is common, since the lateral field of view is ~20 cm, insufficient to cover large anatomical sites. The effect on CBCT reconstructions is shown in Fig. 6 in axial views of the lung and liver in an anthropomorphic phantom. Lateral truncation results in severe cupping of the image bounded by a bright artifact about the circle of reconstruction. A straightforward correction of the artifact is achieved by extrapolation of the projection edges – e.g., by constant, linear, and/or exponential extrapolation of pixel values across a specified range. Results indicate a significant reduction in the cupping artifact without degradation in soft-tissue contrast-to-noise ratio.

4. INTEGRATED IMAGING, NAVIGATION, REGISTRATION, AND DISPLAY

4.1 Navigation System and Target Registration Accuracy

Real-time tracking and navigation has become an important component of the surgical arsenal over the past 20 years, with proliferation in neurosurgery, orthopedics, spine surgery, and radiotherapy. Essential to a navigation system is the registration method, which provides the mapping of the real-world coordinate system at the time of surgery to the imaging coordinate system (conventionally in preoperative CT and MR). Current methods triangulate the position of passive or active markers affixed to surgical tools using stereoscopic cameras (e.g., NDI Polaris, shown in Fig. 7).
A development under investigation for CBCT-guided surgery involves an automated registration process to replace the conventional (manual) process. The automatic registration involves 3 or more IR markers (spheres) affixed to the patient or head-frame with CBCT-compatible mounts. The relative positions of the spheres are defined to create a “tool” containing all spheres. From any intra-operative CBCT scan with IR markers in the FOV, in-house navigation software is employed to automatically locate the spheres in the CBCT image using a template-matching algorithm. The two point sets (real-world positions determined by the camera, and image-space positions determined by CBCT) are used to compute the registration by rigid body transformation. This allows real-time update to the registration to account for patient rotation / translation, and the geometric registration is automatically updated with each intraoperative CBCT scan. The target registration error (TRE) for the manual and automatic procedures was measured, with 4 fiducials in each case. TRE for the manual process was ~2.3 (±1.1) mm, comparable to values in the literature. Initial implementation of the auto-registration process provided TRE of ~2.4 (±0.4) mm. The results suggest approximately equivalent TRE between the two techniques, with improved reproducibility for the automated approach.

4.2 Deformable Image Registration (DIR)

DIR is critical to the advancement of IG procedures, since deformation limits the precision of integrating multiple images by rigid registration alone. Tissue deformation can result from the intervention itself (e.g., tissue excision and response of surrounding structures) and changes in patient position between pre- and intra-operative imaging. Thus, the development of fast, accurate methods for DIR is essential to extending surgical guidance techniques beyond the current state of the art. Methods for DIR include: similarity measures [e.g., mean square difference in image intensity, cross correlation (CC), mutual information (MI), and surface alignment]; and interpolation methods [e.g., thin plate and B-splines, fluid and optical flow, and finite element modeling (FEM)]. FEM has many advantages in the context of interventional procedures, and numerous studies have investigated its use in surgical simulation – particularly in brain deformation. FEM operates upon a biomechanical model of the organ to accommodate complex interactions between organs and overcomes the limitation of techniques driven by image intensity similarity (e.g., increased sensitivity of one imaging modality compared to another). The creation of a geometrically resolved view via model-based (rather than image intensity-based) registration can be critical in comparing structures in different imaging modalities or at different times in the intervention. In addition, FEM utilizes the biomechanical nature of the organs undergoing deformation and can therefore predict deformations likely to occur in subsequent intervention.

Multi-organ FEM for DIR has been pursued extensively in radiotherapy and has begun adaptation to IGS. A commercially-available FEM package (HyperMesh, Altair Engineering, Troy, MI) and Finite Element Analysis (FEA) software (ABAQUS, ABAQUS Inc., Providence, RI) are used for model creation and analysis in a process developed at
Regions of interest (ROIs) are identified manually or by segmentation and converted to a 2D surface mesh (nodes connected to form volume elements), which in turn is used to generate a 3D volume mesh of tetrahedral elements. Multi-organ models are created by constructing surface interfaces between neighboring ROIs to govern interaction. The registration is driven by alignment of a subset of the ROI surfaces using a guided surface projection method, and the full deformation map relating two images is calculated using biomechanical models and solving the constitutive equations using FEA. The model-based approach allows DIR between any two images regardless of modality, contrast, etc., provided that common structures can be identified in each image. Near-real-time DIR is important to its implementation in IG interventions. Early results show substantial gains in efficiency (>order of magnitude) achieved at minimal loss in accuracy (<1 mm). Currently the multi-organ DIR can be performed in ~1 min (single CPU 2 GHz), with further reduction anticipated from improved computer processing speed. Image fusion and rendering for tri-planar views can be done essentially on the fly.

5. APPLICATION IN MINIMALLY INVASIVE THERAPIES: FROM PRE-CLINICAL EVALUATION TO PATIENT PROTOCOLS

Over the past ~3 years, the C-arm system has been deployed in pre-clinical investigation across a broad range of interventional procedures. In each case, the approach has proceeded from phantom studies (investigating task-specific imaging and guidance performance as well as logistical constraints) to cadaver or animal studies (furthering such investigation in a context that more completely considers clinical requirements) to pre-clinical research in patients (under voluntary informed consent, according to protocols consistent with institutional research ethics requirements, and in accordance with governmental authorization for investigational testing).

Figure 8. Illustration of CBCT image quality in pre-clinical studies of surgical guidance. (a) Volume rendering of a cadaveric skull, with target structures (ethmoid and sphenoid air cells) segmented. (b-d) CBCT images of a chest phantom, showing simulated bronchial and vascular structure in (b) axial, (c) coronal, and (d) tri-planar views.

Areas of pre-clinical investigation include: i.) orthopedic surgery – e.g., precise reduction of tibial plateau fractures, correction of femoral malrotation in longbone fractures, and guidance of hip / pelvis surgery; ii.) spine surgery – e.g., vertebroplasty and guidance of photodynamic therapy (PDT) of spinal metastases; iii.) sinus surgery – e.g., 3D visualization of critical structures auxiliary to endoscopy; iv.) head and neck surgery – e.g., visualization of target and normal structures in bone and soft-tissue tumors; v.) combined skull base and neurosurgery – e.g., guiding anterior approach and excision of pituitary and brain stem lesions; vi.) ear surgery – e.g., facial nerve avoidance and guidance / verification of cochlear implants; vii.) breast surgery – e.g., imaging of the breast directly in the treatment position,
facilitating resection of sub-palpable breast lesions; viii.) thoracic surgery – e.g., resection of sub-palpable lung nodules; ix.) interventional radiology – e.g., guidance and evaluation of stent placement; and x.) brachytherapy – e.g., guidance and verification of radioactive seeds in the prostate. Example images illustrating the quality of intraoperative 3D image data provided to the surgeon are shown in Fig. 8.

In each application, the C-arm platform and multi-mode imaging functionality could offer a potentially significant advance over existing image-guidance approaches. Key logistical advantages include the open geometry (allowing excellent patient access), low radiation dose (facilitating repeat intraoperative scanning), portability (allowing service to multiple operating rooms or throughout an ICU or ER), and low cost (relative to CT or MRI). Key imaging performance characteristics include sub-mm spatial resolution, soft-tissue visibility, and geometric accuracy. Remaining challenges include speed of image acquisition and reconstruction, management of image artifacts, and streamlined integration with real-time navigation and visualization systems.

6. ACKNOWLEDGMENTS

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