Technical assessment of a prototype cone-beam CT system for imaging of acute intracranial hemorrhage

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Purpose: A cone-beam CT scanner has been developed for detection and monitoring of traumatic brain injury and acute intracranial hemorrhage (ICH) at the point of care. This work presents a technical assessment of imaging performance and dose for the scanner in phantom and cadaver studies as a prerequisite to clinical translation.

Methods: The scanner incorporates a compact, rotating-anode x-ray source and a flat-panel detector (43 × 43 cm$^2$) on a mobile U-arm gantry with source-axis distance = 550 mm and source-detector distance = 1000 mm. Central and peripheral doses were measured in 16 cm diameter CTDI phantoms using a 0.6 cm$^3$ Farmer ionization chamber for various scan techniques and as a function of longitudinal position, including out of field. Spatial resolution, contrast, noise, and image uniformity were assessed in quantitative and anthropomorphic head phantoms. Two reconstruction protocols were evaluated, including filtered backprojection (FBP) for high-resolution bone imaging and penalized weighted least squares (PWLS) reconstruction for low-contrast soft tissue (ICH) visualization. A fresh cadaver was imaged with and without simulated ICH using the scanner as well as a diagnostic multidetector CT (MDCT) scanner using a standard head protocol. Images were interpreted by a fellowship-trained neuroradiologist for imaging tasks of ICH detection, gray-white-CSF differentiation, detection of midline shift, and fracture detection.

Results: The nominal scan protocol involved 720 projections acquired over a 360° orbit at 100 kV and 216 mAs, giving a dose (weighted CTDI) of 22.8 mGy (~1.2 mSv effective dose). Out-of-field dose decreased to <10% within 6 cm of the field edge (approximate to the thyroid position). Image uniformity demonstrated <1% variation between the edge of the field (near the cranium) and center of the image. The high-resolution FBP reconstruction protocol showed ~0.9 mm point spread function (PSF) full-width at half-maximum (FWHM). The smooth PWLS reconstruction protocol yielded ~1.2 mm PSF FWHM and contrast-to-noise ratio exceeding 5.7 in ~50 HU spherical ICH, resulting in conspicuous depiction of ICH down to ~2 mm (the smallest diameter investigated). Cadaver images demonstrated good differentiation of brain and CSF (sufficient, but inferior to MDCT, recognizing that the CBCT dose was one-third that of MDCT), excellent visualization of cranial sutures and fracture (potentially superior to MDCT), clear detection of midline shift, and conspicuous detection of ICH.

Conclusions: Technical assessment of the prototype demonstrates dose characteristics and imaging performance consistent with point-of-care detection and monitoring of head injury—most notably, conspicuous detection of ICH—and supports translation of the system to clinical studies.

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1. INTRODUCTION

Intracranial hemorrhage (ICH) is associated with a number of neurological diseases and injuries, including traumatic brain injury (TBI), hemorrhagic stroke, aneurysm, hypertensive intracerebral hemorrhage, and postsurgical hemorrhage. There are estimated to be $1.7 \times 10^8$ visits to the emergency department annually for TBI—with a 10%–15% mortality rate for cases of severe traumatic brain injury—and ~800 000 cases of stroke annually, with about 13% of those hemorrhagic. Non-intrastroment (NC) multidetector computed tomography (MDCT) is the most prevalent front-line imaging modality for detection, diagnosis, and monitoring of ICH in its acute stage, offering speed and high sensitivity to the presence of fresh blood in the brain.

Acute ICH typically presents as a hyperattenuating lesion in NC-MDCT. Common types of ICH include epidural hematoma (EDH), subdural hematoma (SDH), subarachnoid hemorrhage (SAH), and intraparenchymal hemorrhage (IPH) and exhibit an evolution in contrast that initially increases during the hyperacute and acute stages (hyperdense, ~40–80 HU contrast within ~3 days) and subsequently decreases to subacute (isodense) and chronic (hypodense, −5 to 20 HU contrast in ~10–20 days or longer). MDCT also provides some utility in detecting certain types of acute mild-to-moderate TBI associated with inertial (shear force) brain injuries—e.g., diffuse axonal injury (DAI) presenting with macroscopic hemorrhage in ~20% of cases. For stroke imaging, NC-MDCT is widely used for exclusion of ICH, with CT perfusion subsequently providing diagnosis, prognosis, and direction of therapeutic course. Increasing interest in “one-stop” stroke diagnosis and intervention is motivated by the need for rapid diagnosis and treatment in emergent, ambulatory, and/or critical care units.

Magnetic resonance imaging (MRI) has traditionally been the gold standard for the detection and monitoring of chronic-stage ICH, and recent advances in image acquisition speed, functional perfusion imaging, and susceptibility weighted imaging show that MRI may also be well suited to acute-stage ICH imaging. Recognizing barriers of cost, access, and time, such capability offers a potentially important advance in detection of acute ICH in the future (e.g., in pediatrics).

Patients in the neurosciences critical care unit (NCCU) present a particular need for reliable imaging of ICH, with intracranial bleeding accounting for 1/3 of perioperative mortality after intracranial procedures and symptomatic hemorrhage occurring in 2.1% of patients after functional neurosurgery. For this reason, patient monitoring routinely includes head CT, typically with one scan acquired at 24 h postoperation and additional scans ordered as warranted for suspected emergence or progression of ICH that can be triggered by neurological complications.

For patients in the NCCU or alternatively, the intensive care unit (ICU) or postanesthesia care unit (PACU) as well as remote locations such as field hospitals, transport to the MDCT or MRI suite carries significant safety considerations. Especially in the critical care setting, patient transport is associated with alarmingly high moribidity and mortality. For example, a retrospective study indicated that adverse effects occur in up to 70% of all patient transports, with 8% of such events being potentially life-threatening. A prospective study of 125 intrahospital transports of ICU patients indicated a 1 in 3 rate of adverse events, with 75% occurring while the patient was in the radiology department. Ott et al. described a spectrum of contributing factors affecting patient safety in transport to radiology, including risk of dislodging tubes and lines, movement from the patient transport bed to the scanner couch (and back), the potential need for sedation for relief of claustrophobia, and separation from their usual caregivers. Despite such risks, however, the benefits gained in diagnostic confidence and direction of proper therapy justify the significant role of imaging in managing critically ill patients. For example, 45% of cases experience a change in management based solely on radiological findings (e.g., detection of new lesions).

The importance of imaging in the diagnosis and treatment of ICH combined with the risks of patient transport to the scanner suite motivates the development of point-of-care imaging for monitoring and management of patients with brain injury. A variety of commercially available portable imaging systems merit investigation for such application, including mobile C-arms for cone-beam CT (CBCT), and portable CT scanners. CBCT systems designed specifically for head imaging portable CT scanners, and even portable MRI. Some systems exhibit varying degrees of imaging performance and logistical compatibility with the critical care environment.

Cone-beam CT offers relative simplicity and flexibility in system design, with numerous embodiments emerging over the last two decades for Interventional and diagnostic imaging applications. Such systems can offer mechanical simplicity, portability, capability for 2D radiographic/fluoroscopic as well as 3D CBCT imaging, and relatively low cost. However, imaging performance in the current state of the art is typically not suitable for diagnostic-quality visualization of soft tissue. Notable exceptions, of course, include breast CBCT and extremity CBCT. For image-guided interventions, CBCT image quality may be sufficient for the task of localizing known targets and adjacent vital anatomy, but for diagnostic head imaging, such systems tend to be limited to high-contrast imaging of bone. For neuroimaging in particular, identification of natural anatomical landmarks [e.g., gray and white matter and cerebrospinal fluid (CSF)] requires contrast resolution better than ~20 HU. Such capability is primarily challenged by image noise and nonuniformity (artifacts) to which CBCT systems are particularly susceptible. Achieving contrast resolution sufficient for reliable detection of ICH at the point of care requires a significant advance in CBCT imaging performance without compromising other advantageous characteristics of portability, open geometry, ease of use, and cost.

Recent work has pursued the development of a CBCT system for imaging of head injury at the point of care, including the modeling and design of the scanner.
development of artifact correction algorithms,\textsuperscript{48} and development of model-based image reconstruction (MBIR) techniques.\textsuperscript{49} Such advances have guided the development of a prototype CBCT scanner specifically for point-of-care imaging of ICH. The work reported below presents the technical assessment of dose and imaging performance for the scanner prototype as a prerequisite to clinical studies.

2. METHODS AND MATERIALS

2.A. CBCT scanner prototype

Figure 1(a) shows the prototype CBCT system, and a summary of system characteristics is listed in Table I. The main hardware components are as follows.

2.A.1. Mobile gantry

The system includes a mobile U-arm gantry with a maximum rotation rate of 24°/s, allowing a half-scan orbit (210°) in 9 s or a full 360° rotation in 15 s. Positioning controls on each side of the gantry are used to position the U-arm (raise, lower, or rotate). The entire system can be powered from a standard (110 V AC, 20 A) North American outlet.

2.A.2. X-ray source and generator

The x-ray source is a rotating tungsten anode x-ray tube (Monobloc, IMD, Grassobio, Italy) with 150 kJ heat capacity and 17° anode. The system has 5 kW maximum power, pulse duration ranging 2–40 ms, tube current ranging 5–40 mA, and tube potential ranging 70–120 kV at 0.6 FS focal spot size. Specified inherent filtration for the x-ray window is 1.4 mm Al equivalent at 75 kV. Based on a previous analysis,\textsuperscript{47} the initial prototype did not include a bowtie filter.

2.A.3. Collimator

A manual square-field collimator (R72, Ralco srl, El Dorado Hills CA, USA) is attached to the face of the x-ray source. The collimator area ranges up to 43 × 43 cm\textsuperscript{2} at 1000 mm from the source. The system also features a light-field and a laser crosshair for positioning. The light-field mirror specifies 1 mm Al equivalent filtration. A filter slot includes an additional 0.2 mm Cu added filtration.

2.A.4. Detector

The system features a flat-panel detector (FPD, Paxscan 4343CB, Varian, Palo Alto CA, USA) with a 43 × 43 cm\textsuperscript{2} field of view (FOV) and native 0.139 × 0.139 mm\textsuperscript{2} pixel pitch, which can be hardware binned to 2 × 2, 3 × 3, and 4 × 4, resulting in isotropic pixel sizes of 0.278, 0.417, and 0.556 mm, respectively. The FPD readout gain mode can be varied between low gain (LG, 3–4 pF integrating capacitance), high gain (HG, 0.5 pF integrating capacitance), and dual gain (DG, 0.5 pF HG and 4 pF LG) using 2 × 2 or 4 × 4 pixel binning at 16-bit depth. The maximum frame rate of the detector in LG or HG mode is 30 frames/s (fps) for 4 × 4 binning, 25 fps for 3 × 3 binning, 15 fps for 2 × 2 binning, and 4 fps for 1 × 1 pixel binning. For DG readout, the frame rate is halved—for example, nominal 2 × 2 DG readout and ∼8 fps. The nominal frame rate, pixel binning, and gain modes are further detailed in Sec. 2.C. Based on a previous analysis,\textsuperscript{46} the current system configuration did not include an antiscatter grid.

2.A.5. System geometry

The system has a source-axis distance (SAD) of 550 mm and source-detector distance (SDD) of 1000 mm. The FPD is approximately centered on the FOV, giving a 25° fan and cone.

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Fig. 1. (a) Prototype CBCT system developed for ICH imaging. Dose was assessed in (b) a stack of two 16 cm diameter CTDI phantoms as a function of position. Image quality was assessed in (c) an anthropomorphic head phantom containing a natural skeleton and tissue-equivalent plastic (18 HU) embedded with spheres of varying size (2–12 mm) and attenuation (−30 to 900 HU) and (d) a simple cylindrical phantom as well as a cadaver (not shown).
angle. The resulting 3D image FOV is $23.7 \times 23.7 \times 23.7$ cm$^3$. Previous analysis shows this geometry to be nearly optimal with respect to tradeoffs among focal spot blur, geometric magnification, air gap (x-ray scatter rejection), and detector exposure level. According to Zhuang and Bradtmiller, it is also sufficient to give full axial coverage of the head in 99% of the population.

### 2.A.6. Control console

The control console and workstation are located behind a mobile shield wall. The workstation used for CBCT image acquisition in this work was a HP Z620 workstation (Hewlett Packard, Palo Alto CA, USA). Although 3D image reconstruction (details below) can be performed on the acquisition computer, for this study, 3D images were reconstructed on a Precision T7910 (Dell, Round Rock, TX, USA) offering superior GPU (GeForce GTX Titan X, NVIDIA, Santa Clara CA, USA).

### 2.B. Image acquisition

#### 2.B.1. Beam quality

Beam quality was characterized by measuring the half-value layer (HVL) using a backshielded silicon diode (Accu-Dose, RadCal Corp., Monrovia CA, USA) placed at the FPD with varying thickness of Al filtration (placed at the exit face of the collimator) across a broad range of x-ray tube potentials. The HVL was determined by exponential fit of exposure vs Al thickness and computing the thickness required to reduce

| Hardware parameter         | Value                  |
|----------------------------|------------------------|
| Power (max)                | 5 kW                   |
| Focal spot size            | 0.6 FS                 |
| Anode Rotating (W)         | 17°                    |
| Tube voltage               | 70–120 kV              |
| Total filtration           | 2.4 mm Al equiv. inh. filtration |
| 0.2 mm Cu added filtration |
| HVL (80 kV)                | 6.1 mm Al              |
| HVL (90 kV)                | 6.8 mm Al              |
| HVL (100 kV)               | 7.4 mm Al              |
| HVL (110 kV)               | 7.7 mm Al              |
| FPD pixel pitch ($\alpha_{\text{pix}}$) | 0.139 mm |
| Scintillator               | 0.6 mm CsI:Tl          |
| Gantry rotation rate (max) | 24°/s                  |
| SAD                        | 550 mm                 |
| SDD                        | 1000 mm                |
| FOV at isocenter           | $23.7 \times 23.7 \times 23.7$ cm$^3$ |

| Acquisition parameter      | Nominal scan technique | Clinical scan protocol |
|-----------------------------|------------------------|------------------------|
| Orbital extent              | 360°                   |                        |
| Detector readout mode       | Dual gain (0.5 pF HG/4 pF LG) |                        |
| kV                          | 100                    |                        |
| Total mAs                   | 216                    |                        |
| mAs/projection              | 0.3                    | 0.48                   |
| Number of projections/scan  | 720                    | 450                    |
| Scan time                   | 90 s                   | 30 s                   |
| Pixel binning               | $2 \times 2$ (hardware) | $4 \times 4$ (hardware) |
|                            | $2 \times 2$ (software) | $1 \times 1$ (software) |
| Frame rate                  | 8 fps                  | 15 fps                 |
| Central dose, $D_0$         | 20.8 mGy               |                        |
| Peripheral dose, $D_P$      | 23.8 mGy               |                        |
| Weighted dose, $D_w$        | 22.8 mGy               |                        |

| Reconstruction parameter    | Sharp bone protocol    | Smooth soft-tissue protocol |
|-----------------------------|------------------------|-----------------------------|
| Algorithm                   | FBP                    | PWLS                        |
| Filter, cutoff frequency    | Hann, $k_{\text{filt}} = 1$ | n/a                         |
| Penalty, regularization     | n/a                    | Huber, $\beta_R = 10^{2.6}$, $\delta = 10^{-4}$ mm$^{-1}$ |
| Voxel size                  | $0.3 \times 0.3 \times 0.3$ mm$^3$ | $0.5 \times 0.5 \times 0.5$ mm$^3$ |
| Slice thickness (display)   | 0.3 mm                 | 2.5 mm                      |
exposure to 50% of the bare-beam value. These data were also
used to parameterize an x-ray spectral model using the spekr
3.0 (Ref. 51) implementation of TASMICS. The spectral
model was subsequently used in correction of scatter and
beam hardening artifacts (below).

2.B.2. Radiation dose

Dose was measured using a Radcal electrometer (Accu-
dose, Radcal Corp., Monrovia CA, USA) and a 0.6 cm³
Farmer ionization chamber placed at the central and peripheral
locations of a 16 cm diameter CTDI phantom. As shown in
Fig. 1(b), two 16 cm diameter CTDI phantoms were placed
end to end and centered along the axis of rotation, with one
end of the stack aligned at the superior edge of the x-ray
field and the other end of the stack extending 6.5 cm beyond
the inferior edge of the beam, roughly approximating the
diameter of the head and neck and sufficient to capture the dose
associated with long x-ray scatter tails. The central dose ($D_\text{c}$)
was measured as the absorbed dose (mGy) by translating the Farmer chamber along the $z$
axis and measuring the absorbed dose (mGy) in the center of
the beam) by translating the Farmer chamber along the $z$
axis in approximately 2 cm intervals, providing a longitudinal dose
profile including that outside the inferior aspect of the FOV
in approximately 2 cm intervals, providing a longitudinal dose
profile including that outside the inferior aspect of the FOV.

2.B.3. CBCT image acquisition technique

Initial studies identified three CBCT image acquisition
protocols for the system: (1) DG readout mode with $2 \times 2$ pixel
binning (0.278 x 0.278 mm² pixels), 720 projections over 360° at 0.3 mAs/pulse (216 mAs/scan) with a total scan time of 90 s; (2) LG readout mode with $3 \times 3$ pixel binning, 720
projections over $360^\circ$ and $0.3$ mAs/pulse (216 mAs/scan)
with a total scan time of 28 s; and (3) DG readout mode with $4 \times 4$ pixel binning, 450 projections over $360^\circ$ and 0.48 mAs/pulse (216 mAs/scan) with a total scan time of 30 s.
All protocols use 100 kV with total filtration as in Table I.
The standard scan protocol in the studies below was protocol
(1)—identified as the “nominal” protocol in Table I—unless
otherwise specifically mentioned. Protocol (2) was used in
phantom studies to investigate the advantages of DG readout.
Protocol (3) was defined following the technical assessment
detailed below as part of the translation to clinical studies,
providing the same dose as protocol (1) with faster scan speed.

2.C. Image reconstruction

2.C.1. Data pre-processing

Fifty dark (offset) scans were acquired immediately before
each scan. Two hundred flood-field projections were acquired
in air with the gantry stationary at the 90° position shown in
Fig. 1(a). Two field-flood data sets were acquired for DG
\correction: low mAs (0.04 mAs/pulse) for the HG channel
and high mAs (0.25 mAs/pulse) for the LG channel. After
acquisition, the HG and LG channels of each DG projection
were separately offset- and gain-corrected, and each HG
projection was normalized to the value in a $100 \times 100$ pixel
region of the unsaturated region of the corresponding LG
region. The two projection data sets were then recombined
such that HG pixels >40% of the HG signal saturation value
were replaced with the corresponding LG pixel values. Pixel
defects were corrected by $3 \times 3$ median filtering for isolated
defective pixels ($3 \times 5$ filtering for line defects) identified in
a map of pixels demonstrating anomalous dark signal or
\correction: gain characteristics. For soft-tissue reconstructions (below),
the projection data were further binned (additional factor of
$2 \times 2$, giving pixel pitch = 4x and $a_{\text{pix}} = 0.556$ mm).

2.C.2. Lag correction

The projection data were then corrected for temporal lag
as described in Sisniega et al. by deconvolution with the lag
kernel, $L(k)$,

$$L(k) = \sum_{n=1}^{N} b_n (1 - e^{-a_n})^{-1} u(k), \quad (1)$$

where $k$ represents the frame number, $u(k)$ is the unit step
function, and six exponential terms ($N = 6$) were used to
characterize the falling-edge step response function, with $a_n$ = [1.96, 0.01, 0.07, 0.99, 3.90, 31.46] and $b_n$ = [0.029, 0.003, 0.006, 0.02, 0.20, 0.75].

2.C.3. Glare correction

Correction of low-frequency “glare” effects arising from
detector veiling glare and/or off-focal radiation was applied by
deconvolution of the projection data with the glare spread
function, GSF(r),

$$\text{GSF}(r) = \frac{c_1}{\pi c_2} \left( \frac{1}{1 + r^2/c_2^2} \right), \quad (2)$$

where $r$ represents radial distance (dimensionless pixel
number), and $c_1$ and $c_2$ are 0.1 and 14.59, respectively (also
dimensionless).

2.C.4. Beam-hardening correction

Two-pass beam hardening correction was applied ac-
cording to the algorithm described by Joseph and Spital53
and implemented in Sisniega et al. Both the water and
bone corrections were based on the spectral model derived
in Sec. 2.A. A first-pass water correction was performed by
replacing the measured attenuation values by those in
a pregenerated look-up table (LUT) that matched the
“ideal” (prehardened) attenuation values to beam-hardened
attenuation values, assuming an object composed entirely
of water (with varying density). A 3D image was then
reconstructed, and voxels with value greater than 200 HU were segmented as “bone” and forward projected, giving a bone-only projection. A second LUT for bone attenuation was used to correct the bone-only projections, and the corrected bone-only projections were recombinated with the water-corrected projections for reconstruction of a final beam-hardening corrected 3D image.

2.C.5. X-ray scatter correction

X-ray scatter correction was integrated with the beam-hardening correction as described in Sisniega et al. A Monte Carlo (MC) estimation of scatter fluence was computed for each projection based on the beam-hardening corrected 3D volume. The computation time of scatter fluence estimation was reduced using variance reduction, GPU parallelization, and sparse sampling in the angular domain. The scatter fluence was subtracted from each projection, and all projections were individually renormalized to the bare-beam signal prior to reconstruction. Application of the beam-hardening correction to the scatter-corrected image from iteration \(i\) providing input to the beam-hardening correction of iteration \((i + 1)\), and so on for four iterations.

2.C.6. Geometric calibration

The system geometry was calibrated using a cylindrical phantom with tungsten BBs embedded in a spiral pattern at precisely known locations. A 360° scan of the phantom was acquired, and the center of all BBs in each projection was computed. A projection matrix describing the pose of the x-ray source and detector according to nine degrees of freedom is computed for each view based on the projected BB locations, similar to the procedure described in Navab et al. The view angle for each projection in the calibration scan was recorded from the gantry motor encoders. For subsequent imaging scans, the projection matrices were interpolated from the calibration data according to the encoder values in each view (trilinear interpolation of the six translational components and quaternion average of the three rotational components).

2.C.7. 3D image reconstruction

Two reconstruction protocols were used: A sharp reconstruction based on 3D filtered backprojection (FBP) for bone visualization and a smooth reconstruction based on penalized weighted least squares (PWLS) for low-contrast soft-tissue and ICH visualization. FBP reconstructions were performed using a Hann apodization window with variable cutoff frequency, \(k_{\text{filt}}\). Voxel size in FBP reconstructions was nominally \(0.3 \times 0.3 \times 0.3 \text{ mm}^3\); alternatively, \(0.5 \times 0.5 \times 0.5 \text{ mm}^3\) for studies that compare to PWLS, below. PWLS reconstructions were performed by minimizing the objective function

\[
\hat{\mu} = \arg\min_{\mu} \| A_0 \mu - l \|_2^2 + \beta R(\mu),
\]

where \(\hat{\mu}\) is the image estimate, \(A_0\) is the linear projection operator, \(l\) represents the measured line integrals, and \(W\) are the data fidelity weighting terms optionally taken either as the data directly or modified to account for artifact corrections as in Dang et al. The penalty strength \(\beta\) was freely variable, with a nominal value of \(10^2\). The roughness penalty, \(R(\mu)\), was a Huber function that penalizes the difference between a voxel at index \(j\) and its \(N = 6\) nearest neighbors (at location \(k\)) in 3D as

\[
R(\mu) = \sum_j \sum_{k \in \mathcal{N}} \left[ \frac{1}{2\delta} (\mu_j - \mu_k)^2, |\mu_j - \mu_k| \leq \delta \right] + \delta \left[ |\mu_j - \mu_k| > \delta \right]
\]

to control the transition between quadratic smoothing (for small signal differences) and linear edge preservation (for large signal differences). The \(\delta\) term determines the threshold contrast (signal difference) between the quadratic and linear penalty, varying within the range \((1 \times 10^{-3} \text{ to } 5 \times 10^{-5}) \text{ mm}^{-1}\) as in Dang et al. The separable quadratic surrogates method with ordered subsets (OS-SQS) was used as in Erdogan and Fessler to perform the optimization in Eq. (3) with precomputed curvatures using 20 subsets and 100 iterations. A separable footprint forward- and back-projector was used with modifications as in Wang et al. using custom CUDA libraries for GPU acceleration. Voxel size in PWLS reconstructions was nominally \(0.5 \times 0.5 \times 0.5 \text{ mm}^3\) with option for displayed slice averaging.

2.D. Image quality assessment

2.D.1. Artifact corrections

The overall quality of the artifact corrections was first assessed in terms of uniformity and effect on contrast, noise, and contrast-to-noise ratio (CNR). An anthropomorphic head phantom (RANDO, The Phantom Lab, Greenwich NY, USA) was used, featuring a natural skeleton encased in tissue-equivalent plastic (~18 HU). Two arrangements of spheres were included within the cranial vault as illustrated in Fig. 1(c): the first contained 12 mm diameter spheres of varying contrast (~30 to 900 HU) and the second contained acrylic spheres (124 HU) of varying size (2–12 mm diameter). Image uniformity was measured in terms of the mean signal difference in a \(10 \times 10\) voxel region near the interior of the skull (\(\hat{\mu}_{\text{edge}}\)) compared to a region near the center (\(\hat{\mu}_{\text{center}}\)) normalized by the signal value in the edge region

\[
\text{CNR} = \frac{\hat{\mu}_{\text{edge}} - \hat{\mu}_{\text{center}}}{\hat{\mu}_{\text{edge}}}.
\]

Contrast was measured as the difference in mean signal between a \(7 \times 7\) voxel region within the largest acrylic sphere and a \(7 \times 7\) voxel region in the adjacent soft-tissue background, repeated for nine slices about the center of a sphere. Noise was measured as the mean standard deviation in voxel values in soft-tissue background. The CNR was evaluated with and without artifact corrections for the FBP and PWLS reconstructions.
2.D.2. Contrast, noise, and spatial resolution

Imaging performance was further evaluated (in FBP and PWLS reconstructions with artifact correction) in terms of contrast, noise, and spatial resolution using a 15 cm cylindrical phantom [Fig. 1(d)] containing tissue-simulating inserts (Gammex, Middleton WI, USA) and a 0.127 mm diameter tungsten wire in Styrofoam. The phantom also contained a custom gelatin formulation (Knox Gelatine, Kraft, Camden NJ, USA) mixed to simulate brain (∼47 HU of 47, relative to gelatin background) sphere. Reconstructions was characterized with respect to the low-contrast insert and background.

Images were acquired using the nominal protocol (1) specified in Sec. 2.B (100 kV with 720 projection images). CNR was measured as a function of dose by varying 72–346 mAs (adjusted by varying 0.10–0.48 mAs/projection). Similar to Sec. 2.D.1, contrast and noise were evaluated for the low-contrast inserts (10 mm ICH sphere) using 7×7 voxel regions of low-contrast insert and background.

For FBP reconstructions, the spatial resolution was characterized in terms of a 2D Gaussian fit to the point spread function (PSF) measured in 100 adjacent axial slices of the tungsten wire. Each Gaussian fit was deconvolved with a symmetric (0.127 mm diameter) circle function representing the wire cross-section, and the ensemble average full width at half maximum (FWHM) was computed. As broadly appreciated, nonlinear reconstruction methods such as PWLS exhibit contrast-dependent spatial resolution characteristics; therefore, the spatial resolution for PWLS reconstructions was characterized with respect to the low-contrast (ΔHU of 47, relative to gelatin background) sphere. As previously described, an oversampled edge spread function (ESF) was generated by converting voxel locations to spherical co-ordinates, (r,θ,φ), with r = 0 corresponding to the center of the sphere. The sphere was then separated into 12 equiangular sectors, and for each sector a cumulative 1D Gaussian function, erf(r;μ,σ), was fit to the voxel values as a function of r for all angles in that sector. The equivalent PSF FWHM was computed from each fit, and the ensemble average was reported. The PSF FWHM was analyzed as a function of k_{fil} (for FBP) and as a function of β (for PWLS, with δ fixed to a value of 1 × 10^{-4} mm^{-1}). The results were further characterized in terms of the modulation transfer function (MTF) by Fourier transform (FT) of the spread functions: for FBP, the central slice of the 2D FT of the Gaussian PSF fit to the wire; and for PWLS, the 1D FT of the derivative of the erf fit to the low-contrast ESF.

2.D.3. Image quality in cadaver with simulated ICH

A human cadaver was imaged 9–15 h postmortem without preservation agents (to maintain the natural contrast of the brain as much as possible, recognizing that tissue decomposition in the brain proceeds within hours of death). Data were acquired at the nominal image acquisition technique (1) specified in Sec. 2.B. Sharp (FBP) and smooth (PWLS) images were reconstructed as specified in Sec. 2.C and Table I. Images were first acquired with the specimen in its natural state. Then, a custom solution was prepared with contrast simulating ICH (~64 HU contrast to brain) by mixing 12% sucrose, 4% NaCl, and 0.5% gelatin powder (all by weight) in solution with water. Two injections of the simulated ICH into the right frontal lobe were performed using a cannula and trochar (Jamshidi Needle, Becton Dickinson, Franklin Lakes NJ, USA)—first, a small (2 cm³) bolus, followed by an additional larger (6 cm³) bolus, with images acquired after each. A diagnostic CT scan was also acquired (12 h post-mortem, prior to the simulated ICH injection) using a standard head scan protocol (120 kV, 500 mAs, 59.4 mGy, 0.42×0.42×1 mm³ voxel size; Brilliance CT Big Bore, Philips Healthcare, Amsterdam, Netherlands).

Images were displayed on a 3D imaging workstation (VuePACS, Carestream Health, Rochester NY, USA) with diagnostic-quality monitors (MDCG-3221, Barco, Kortrijk, Belgium) and interpreted by a fellowship-trained neuroradiologist. Image features were qualitatively evaluated with respect to a variety of imaging tasks: (i) brain-CSF differentiation, (ii) ability to visualize midline shift, (iii) conspicuity of the ICH injection, (iv) gray-white matter differentiation, (v) spatial resolution characteristics in bone with respect to visualization of fracture, and (vi) overall image uniformity (including shading and streak artifacts that may confound visualization of ICH). For each task, the neuroradiologist assessed task performance as (1) conspicuous, (2) well visualized, (3) adequately visualized, (4) challenging, or (5) unidentifiable and provided free response regarding factors of contrast, noise, spatial resolution, and artifacts.

3. RESULTS

3.A. Dose

Figure 2 shows the central (D₀) and peripheral (D_p) dose measured as a function of tube potential and superior-inferior position, z. The dose increased with tube potential as shown in Fig. 2(a). The longitudinal variation in dose is shown in Fig. 2(b), showing that the central and peripheral dose peak near the central slice (z = 12 cm) at a value of D₀ = 22.8 mGy. The dose is reduced toward the superior and inferior edges of the FOV to values of D_w ~ 15.5 and 8.2 mGy, respectively. The distribution is asymmetric in z, and the peripheral dose is higher than the central dose across the superior portion of the FOV (toward z = 0) but lower near the inferior edge (z ~ 23 cm) as well as outside the beam due to asymmetry in the internal attenuation of x-ray scatter—i.e., increased and decreased scatter attenuation in the superior and inferior regions, respectively. The out-of-beam central and peripheral doses were 32% and 11% of the respective maximum values, giving D_w ~ 1.6 mGy at z ~ 30 cm (approximate location of the thyroid).

Absolute dose measurements were related to effective dose (E, mSv) as often reported in the literature according to $E = D_w \times L \cdot k_{head}$, where $D_w = 22.8$ mGy, the length of the scan, L, was taken as 23.67 cm, and the tissue weighting factor, $k_{head} = 0.0023$ (mSv/mGy)/cm, was taken from ICRP publication 103 for the head. There was no adjustment for size-specific dose estimation (SSDE) in this work. This
yielded an effective dose of 1.2 mSv, roughly one-third the dose of a typical MDCT head scan protocol.68

3.B. Artifact corrections

Figures 3(a) and 3(b) show FBP reconstructions with and without artifact corrections. Prior to artifact correction, the image exhibited severe shading ($t_{cup} = 13.5\%$), strong reduction in contrast ($\Delta HU = 39$ HU, compared to true difference in attenuation, $\sim 106$ HU between acrylic and Rando material), and modest CNR (5.4). The artifact correction provided a strong improvement in uniformity ($t_{cup} = -0.8\%$), boost in contrast (~92 HU) close to the true value, and an improvement in CNR (7.9) despite an increase in image noise ($\sigma \sim 7$ HU prior to correction, increased to $\sim 12$ HU after correction). A variety of other streak and shading artifacts were also visibly improved by the correction method—e.g., streaks attributed to beam hardening about the supraorbital ridge.

Figure 3(c) shows the same data reconstructed with PWLS after correction. The PWLS images exhibited comparable uniformity ($t_{cup} = 0.4\%$) and contrast (~86 HU) and improved CNR (12.9) by virtue of reduced noise ($\sigma \sim 7$ HU), with parameters selected to match the spatial resolution with respect to the width of the acrylic sphere ESF—viz., for FPD ($k_{filt} = 0.6$ and $a_{vox} = 0.5$ mm) and for PWLS ($\beta = 10^{2.4}$, $\delta = 10^{-4}$, and $a_{vox} = 0.5$ mm). The results illustrate the benefit of artifact correction (which improved uniformity but somewhat amplified noise) combined with PWLS reconstruction (to mitigate the increase in noise).

3.C. Contrast, resolution, and noise

Figure 4(a) shows the measured FWHM of the wire PSF for FBP (as a function of $k_{filt}$) and the FWHM associated with the low-contrast ESF for PWLS (as a function of $\beta$). The FBP reconstruction at the highest cutoff frequency ($k_{filt} > 0.8$) achieves slightly better spatial resolution than PWLS [with low regularization, $\log_{10}(\beta) < 1$] and supports the use of FBP for the sharp bone reconstruction protocol. Figure 4(b) shows the MTF for images reconstructed with the nominal FBP high-resolution ($k_{filt} = 1$, voxel size = 0.3 mm isotropic), FBP soft tissue ($k_{filt} = 0.5$, voxel size = 0.5 mm isotropic), and PWLS soft tissue ($\beta = 10^{2.6}$) protocols. As expected, the bone reconstruction protocol provides the highest MTF, while the nominal PWLS reconstruction (and soft-tissue FBP reconstruction, roughly spatial-frequency matched) exhibited reduced MTF associated with noise reduction.

Figure 4(c) shows the image noise measured as a function of cutoff frequency and regularization strength. The FBP

![Fig. 3. Comparison of (a) uncorrected and (b) artifact-corrected FBP image reconstructions. Note the reduction in cupping and restoration of contrast close to the true value (~92 HU for acrylic spheres) despite increase in noise. (c) Reconstruction using PWLS maintains the benefits of artifact correction and reduces noise (at matched spatial resolution).](image-url)
images exhibit a simple monotonic increase in noise with cutoff frequency as expected. The PWLS images, on the other hand, exhibit a distinctly nonlinear dependence of noise on regularization, with the strongest noise reduction occurring in the range \( \log_{10}(\beta) \approx 1.8-2.5 \) (for \( \delta \) fixed at \( 10^{-4} \) mm\(^{-1} \)).

Considering Figs. 4(a) and 4(c) together, the benefit of PWLS for ICH imaging becomes evident: for example, taking \( \log_{10}(\beta) = 2.6 \) gives noise \( \sim 6.4 \) HU and FWHM \( \sim 1.1 \) mm; however, achieving the equivalent noise level in FBP by setting \( k_{\text{filt}} = 0.5 \) gives a FWHM of \( \sim 1.4 \) mm. The improved noise-resolution tradeoff for PWLS supports its use for the smooth low-contrast ICH reconstruction protocol.

Figure 4(d) shows the effect of detector readout gain mode on CNR and spatial resolution. Previous work\(^{47}\) showed a \( \sim 15\%-20\% \) increase in CNR from use of DG readout (with FBP reconstruction). At matched spatial resolution (insets

![Figure 4](image.png)

**Fig. 4.** (a) Spatial resolution for FBP (FWHM of the wire PSF) and PWLS (FWHM associated with the low-contrast ESF) as a function of filter and regularization strength, respectively. (b) The modulation transfer function measured in image reconstructions for the nominal FBP high-resolution, soft tissue, and PWLS reconstruction protocols. (c) Image noise measured as a function of filter and regularization strength, illustrating the distinct noise-resolution tradeoff between the two reconstruction methods. (d) Dual gain readout mode shows up to 20\% improved contrast to noise ratio at matched spatial resolution compared to LG readout mode when using PWLS image reconstruction.

![Figure 5](image.png)

**Fig. 5.** Visualization of simulated ICH. (a) PWLS image reconstructions of the gelatin-ICH insert at various levels of dose and regularization. (b) Spatial resolution exhibits a steeper dependence on regularization strength at low dose. (c) Image noise similarly has a stronger dependence on regularization strength at the lower dose level.
showing $\beta = 10^{2.4}$ for DG and $\beta = 10^{2.6}$ for LG), a $\sim 20\%$ improvement in CNR can be observed in DG data compared to LG data using a PWLS reconstruction algorithm as well.

Figure 5 shows the effect of dose on spatial resolution and noise in PWLS image reconstructions. Figure 5(a) shows the region of the ICH-gelatin insert at low ($D_0 = 9.6$ mGy), nominal ($D_0 = 24.8$ mGy), and high ($D_0 = 38.2$ mGy) dose reconstructed with regularization ranging $\beta = 10^{1.6} - 10^{2.8}$. Operation at the nominal dose and reconstruction parameters achieves clear visualization of small, $\sim 50$ HU contrast ICH. Figure 5(b) shows the spatial resolution measured as a function of $\beta$ for the three dose levels. While the spatial resolution at low $\beta$ is fairly similar among the three dose levels, the degree of smoothing increases more steeply for the lower dose images as $\beta$ is increased. Similarly, Fig. 5(c) shows the image noise as a function of $\beta$ for the three dose levels, with the lower dose scenario again having a steeper tradeoff between noise and regularization.

### 3.D. Image quality: Anatomical features in cadaver

Assessment of the cadaver images (Figs. 6 and 7) by a neuroradiologist rated performance for task (v) (visualization of bone details, cranial sutures, and possible fracture) as “conspicuous,” citing clear visualization of sutures and other fine details in bone anatomy. Figure 6(a) shows an axial
slice through the petrous bone using the high-resolution FBP reconstruction, demonstrating clear visualization of the stapes and cranial sutures. Figure 6(b) shows a surface rendering of the skull to further illustrate the uniform image quality characteristics supporting visualization of the coronal, lambdoid, and sagittal cranial sutures. Overall, the visibility of fine bone details was considered excellent and superior to the MDCT image (recognizing that higher-resolution, higher-dose MDCT head scan protocols could certainly be employed).

The images were similarly evaluated with respect to imaging tasks (i)–(iv) for assessment of low-contrast imaging performance. Figure 6(c) shows a soft-tissue PWLS image reconstruction demonstrating visualization of fat, muscle, and glandular structures that was rated by the neuroradiologist as “well visualized,” although cone-beam artifacts arising from the occipital bone, mandible, and cervical vertebrae were potentially confounding in regions of the neck.

Figure 7 shows soft tissue reconstructions of the cadaver head (brain window, width = 90 HU). The neuroradiologist assessed the differentiation of CSF and brain [task (i)] and visualization of midline shift [task (ii)—e.g., mass effect] as well visualized. Visualization of the ICH [task (iii)—both small and large boluses] was rated as “conspicuous,” especially in the PWLS image. Differentiation of gray and white matter [task (iv)] was rated as “challenging,” but was attributed primarily to postmortem degradation of the tissue (also evident in the MDCT scan). To the extent that gray and white matter were distinguishable, visualization of the internal capsule (white matter surrounded by gray matter) was comparable to the reference MDCT scan. The neuroradiologist noted the presence of cone-beam artifacts (e.g., posterior to the orbits and at the superior-most extent of the cranial vault) which, although visually distracting, did not affect evaluation of the tasks assessed in this study. Additionally, the radiologist remarked on the higher level of image noise in the CBCT images (due in large part to the lower imaging dose, ~1/3 the dose of the MDCT). Overall, task performance was rated superior for the PWLS reconstructions in comparison to FBP, but the slightly unfamiliar texture of the image noise in the PWLS images was noted as potentially distracting.

4. DISCUSSION AND CONCLUSIONS

The imaging performance of a prototype CBCT system for point-of-care imaging of acute ICH was evaluated in terms of spatial resolution, contrast, noise, CNR, image uniformity, and dose, and image acquisition and reconstruction protocols were identified for high-quality imaging of bone and soft tissue. Nominal scan protocols carried a dose of 22.8 mGy, which is approximately 1/3 the dose of a standard head MDCT scan. Direct comparison of image quality and dose for the scanner prototype and diagnostic MDCT is the subject of ongoing and future work. The artifact correction methods addressed major factors of x-ray scatter, beam-hardening, image lag, and veiling glare, resulting in 0.4% nonuniformity in images of the head. High-resolution FBP reconstruction demonstrated ~0.9 mm PSF FWHM and excellent delineation of bone features (e.g., cranial sutures), and an edge-preserving PWLS method was shown to provide ICH CNR ~5.7 and conspicuous visualization of simulated ICH. A fresh cadaver imaged with and without simulated ICH demonstrated clear differentiation of CSF from gray-white matter and conspicuous delineation of hemorrhage.

The technical assessment agreed with expectations based on previous modeling and laboratory benchtop experiments and was prerequisite to translation of the prototype to clinical studies. The work was also essential to determining key parameters that affect dose and scan speed, which are in turn determined by both the technique factors (kV and mAs) and FPD readout (number of views, frame rate, and readout gain mode). With respect to the latter, the assessment showed that while DG readout can improve CNR by ~15%–20% at equivalent dose, the scan time (90 s for the nominal DG technique) is likely prohibitive for clinical use. We therefore worked with the FPD manufacturer to identify an alternative DG scan protocol with hardware binning at 4 × 4 pixels (cf., 2 × 2 hardware binning followed by another 2 × 2 in postprocessing) to give a frame rate of 15 fps (cf., 8 fps). Further reduction in scan time was obtained by reducing the number of projections to 450 (cf., 720), since previous work showed negligible increase in noise due to view sampling effects. The total dose was held constant by increasing the mAs per projection to 0.48 mAs (cf., 0.3 mAs), yielding scan time of 30 s, as shown in the clinical protocol in Table I. Future work will further evaluate image quality associated with a half-scan orbit (180° + fan angle) as a means to further increase scan speed. The analysis and resulting scan protocols supported translation of the system to clinical studies, now underway.

As discussed above, bringing CT imaging capability to the point of care could offer major benefits to critically ill patients in areas such as the NCCU, ICU, and PACU. Mobile MDCT systems such as the CereTom (Neurologica, Danvers MA, USA) have the potential to address this need as well, and while CBCT offers possible advantages of cost, open geometry, patient access, mechanical simplicity, and capability for radiography/fluoroscopy, both types of system warrant further development and clinical evaluation for this important clinical problem. While other mobile CBCT systems, such as the Vario C-arm (Ziehm Imaging, Nurnberg, Germany) and O-arm (Medtronic, Dublin, Ireland), may also be applicable, the imaging performance for such systems has been conventionally limited to high-contrast (bone) visualization. The system evaluated in this work overcame such conventional limitations through a combination of system design, artifact correction, and MBIR methods developed specifically for high-quality imaging of head injury.

Future work includes clinical studies and image quality with assessment by multiple expert readers. Additional areas of technical development include extension of the artifact correction framework to mitigate image truncation (e.g., due to the head holder) and patient motion. Furthermore, the MBIR method described above is being extended to
include spatially varying regularization for more uniform resolution and noise characteristics in a manner that is optimal to the ICH imaging task. New detector technologies also warrant investigation, including complementary metal oxide semiconductor (CMOS) detectors offering reduced electronic noise and faster frame rate.

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CONFLICT OF INTEREST DISCLOSURE

The authors have no COI to report.

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