Quantitative MR Imaging and MRI-based Radiomics in Radiation Therapy: Applications and Challenges

John Chetley Ford
Department of Radiation Oncology
University of Miami

- Quantitative MR imaging
- Potential MRI-based biomarkers for lesion detection and predicting treatment response
- Topics:
  - Diffusion-weighted MRI, Apparent diffusion coefficients (ADC) maps
  - Dynamic contrast enhanced (DCE) MRI
  - Blood oxygenation level dependent (BOLD) MRI
  - Radiomic texture analysis of MR images
    - Repeatability, reproducibility and validity
Diffusion-weighted MRI

• Background
  • Why diffusion-weighted MRI?
  • What is it and how does it work?
  • How does it relate to cancer biology?

• Applications to cancer treatment
  • Lesion detection - How good is DW-MRI?
  • Sensitive to treatment response?
Diffusion-weighted MRI (DW-MRI)

• Measures water self-diffusion, random, Brownian motion
• Weighting determined by amplitude and timing of diff Gradients
  • b-value = amount of D-weighting
  • Time on the order of 10 msec
• Freely diffusing protons (i.e., CSF) appear dark on DW-MRI
Diffusion-weighted MRI

- Einstein equation: rms displacement of freely diffusing water molecules is proportional to $\sqrt{\text{time}}$; $R^2 = 6Dt$
- MR signal in presence of diffusion encoding gradients:
  - $S(b) = S_0 \exp(-bD)$
  - $b$-factor $= (\gamma\delta G)^2(\Delta-\delta/3)$ (see below)
  - $D$ is the diffusion coefficient, usually called ADC (Apparent Diffusion Coefficient)
Diffusion-weighted MRI
VCU Phantom study

- 10 ml test tubes filled with water, acetone, ethanol, corn oil
- 1.5T Siemens MRI using novel lung DW-MRI protocol (fat sat, resp gated EPI)
- Log of MR signal vs b-value is linear
- ADC (slope) agrees with literature: water at room temperature (2.1 μm²/ms)
Diffusion-weighted MRI (DW-MRI)

- Water phantom upper left
- Mineral oil phantom lower right

T2-weighted image

\[ b = 300 \text{ sec/mm}^2 \]

ADC map
DW-MRI is a Probe of Tissue Microstructure

- Einstein equation becomes non-linear when diffusion is restricted by structures on the microscopic scale, such as cellular membranes.
DW-MRI is a Probe of Tissue Microstructure

- **Interstitial** diffusion: $D \approx 3 \, \mu m^2/ms$ (unrestricted)
- **Intracellular** diffusion: $D \approx 1 \, \mu m^2/ms$ (restricted)
- ADC decreases with increased intra/extracellular volume
  - E.g.: ADC decreases in ischemia, cancer
Diffusion in Stroke

- Dog model of MCA occlusion
- Ischemia causes water to move from interstitial space (high ADC) to intracellular space (lower ADC due to restricted diffusion inside of cell)
Diffusion Anisotropy

- Weight-drop model of spinal cord injury
- MRI of formalin-fixed rat specimens
  - 1.9T, 6mm bore 2-turn saddle coil, 4 cm long
  - 80 µm resolution
  - Made T2, ADC, MT maps

First echo  T2  Longitudinal ADC  Transverse ADC  MTC
Anisotropic Diffusion

- Longitudinal ADC in rat spinal cord injury
  - Diffusion sensitivity parallel to cord
  - $b = 0, 240, 480 \text{ s/mm}^2$, no mixing with image gradients
  - Cell swelling causes ADC decrease
    - Relative increase in intracellular water
    - ADC inside cell lower due to intracellular structures
Anisotropic Diffusion

- Transverse ADC
  - Diffusion sensitivity perpendicular to cord
  - $b = 0, 240, 480 \text{ s/mm}^2$, no mixing with image gradients
  - Myelin breakdown causes increase in tADC
Rat Spinal Cord Imaging

- Modeling water self-diffusion in white matter
- Monte Carlo simulation on light microscopic image, 70 µm FOV

\[
D_m = \frac{1}{\frac{1}{D} + \frac{|c \theta|}{ip a}}
\]
Rat Spinal Cord Imaging

- Modeling water self-diffusion in white matter
- Simulated results very close to measured ADC values

![Diagram showing diffusivity over time](image-url)
Rat Spinal Cord Imaging

- In vivo MRI
  - 1.9T, 80x80 µm x 1mm

TR/TE 2500/35  tADC map  IADC map  Control, Long TR, Short TE  Injured, Long TR, Long TE
T2 MRI Is Routinely Used to Visualize the Prostate

Dynamic contrast enhanced (DCE-MRI)

- Intravenous injection of Gd-chelate extracellular contrast agent
- Serial T1-weighted images taken every 3-60 sec over several min
- Change in T1 is proportional to [Gd]
DCE-MRI

- $K^{\text{trans}}$, forward transfer constant plasma->EES, initial slope (“wash-in”)
- $v_e$, volume of EES per volume of tissue
- $k_{ep} = K^{\text{trans}} / v_e$, reverse rate constant, “wash-out”
DCE-MRI is a Probe of Tissue Microstructure

- DCE-MRI derived kinetic parameters measure:
  - Microvascular density and capillary permeability ($K_{\text{trans}}$)
  - Relative extravascular, extracellular space ($v_e$)
- Cancer characterized by elevated density of leaky capillaries and increased cellularity
- Therefore, rapid wash-in and wash-out are indicators of cancer
How Well Does DCE-MRI Detect Lesions in Prostate Cancer?

- Detectability calculated by taking mean over published 14 studies, weighted by number of cancer samples:
  - Sensitivity = 89% +/- 9%
  - Specificity = 92% +/- 11%

- A negative MRI is as reassuring for absence of cancer as is a negative repeat sextant biopsy. Kirkham, et al., Eur Urol 50, 1163 (2006)

- DCE-MRI detects recurrence following prostatectomy:
  - Specificity = 88%, Specificity = 100% Casciani, et al., AJR 190, 1187 (2006)

- DCE-MRI is able to detect recurrence following radical prostatectomy even before it can be detected by biopsy Alonzi, et al., Eur J Rad 63, 335 (2007)
Combination of DW-MRI and DCE-MRI Yields Improved Lesion Detection
Quantifying BBB Permeability

- Freezing blood-brain barrier (BBB) injury in rat cerebral cortex
  - Necrotic core surrounded by edematous zone
Quantifying BBB Permeability

- Injected intravascular Gd-based contrast agent
  - Does not cross intact BBB
- Developed model relating post-contrast MR signal changes over time to:
  - BBB permeability
  - Leakage volume
    - (extracellular space)
  - Gadodiamide relaxivity
Quantifying BBB Permeability

- Nephrectomized rats:
  - Constant plasma Gd level
  - Quantify Gd relaxivity
- Gadodiamide relaxivity differed in necrotic and edematous tissue
ADC and DCE both probe relative intra/extracellular volume

- Quantitative diffusion measurements of lesion
  - Increasing diffusion weighting, left to right (below)
  - Necrotic becomes progressively darker than edematous region
    - Higher ADC in necrotic zone
ADC and DCE both probe relative intra/extracellular volume

- Dynamic imaging can provide information on nature of pathology
  - BBB permeability from MR signal change following Gd
  - Leakage space (relative extracellular space)
    - Model-derived ECS correlates with ADC
DW-MRI is a Probe of Tissue Microstructure

- **ADC decreases in cancerous tissue**
  - (increased cellularity)

Diffusion-weighted MRI (DW-MRI)

- MR signal depends on product of $b$ and ADC
  - ADC (apparent diffusion coefficient)
- MR images + $b$ => computer algorithm => ADC map

DW-MRI: Cancer detection

- **Breast cancer**
  - Benign ADC ≈ 1.4-1.7
  - Malignant ADC ≈ 0.8-1.3
  - Cutoff ≈ 1.1-1.6 sens/spec = 80-95%/70-90%

- Similar for liver, prostate, cervical
  - Cutoff values among anatomical sites

- Cutoff values vary among institutions
  - Use different b-values
DW-MRI: Treatment Response

- Increase in ADC noted in several sites:
  - Breast, liver, bone sarcoma, brain
- ADC increases within:
  - 3-7 days (chemo), 24-72 hr (radiation)
- Strong correlation between ADC changes and tumor size (r =0.93)

Challenges in using DW-MRI

- Motion management
- Fast imaging, gating
Challenges in using DW-MRI

• VCU MRI lung cancer study

ADC map

PET
Challenges in using DW-MRI

- VCU MRI lung cancer study
  - We used all 8 b-values to calculate ADC

- Intra-patient variation is very good!

- Why do CSF and Cord ADC vary so much among patients?
  - Perhaps due to contamination by “perfusion” component
  - IVIM model (intravoxel incoherent motion)
  - \[ S = S_0 (f e^{-bD^*} + (1 - f) e^{-bADC}) \]
  - \( f \) is the perfusion fraction
  - \( D^* \) is \( \approx 10x \) ADC

- Choice of b-values is very important

<table>
<thead>
<tr>
<th></th>
<th>CSF</th>
<th>Spinal Cord</th>
<th>Lung Tumor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean (25 scans N=7)</td>
<td>2.47 µm²/ms</td>
<td>1.02 µm²/ms</td>
<td>1.34 µm²/ms</td>
</tr>
<tr>
<td>CV (inter-patient)</td>
<td>18%</td>
<td>13%</td>
<td>20%</td>
</tr>
<tr>
<td>CV (intra-patient)</td>
<td>3%</td>
<td>2-4%</td>
<td>*4-8%</td>
</tr>
</tbody>
</table>

* scans on same day

J Chetley Ford, PhD
Monoexponential model using typically two or three b-values is used in clinical DW-MRI to compute the ADC due to computational simplicity and reduced post-processing times.

A consensus is needed to establish correct selection of limited number of b-values, typically 2 b-values, in DW-MRI to ensure comparability across clinical ADC data.

Using 40 DW-MRI scans in a previous study, active tumor monoexponential ADC values using 250 and 1000 μs/μm² b-value were not significantly different from ADC$_{IVIM}$ values.

(K Karki et al. Estimation of Optimal b-value Set for Obtaining Apparent Diffusion Coefficient Free from Perfusion in Non-small Cell Lung Cancer, 2015 AAPM Meeting.)
Rigidly registered monoexponential ADC maps (obtained using all eight b-values) of lung tumor region (shown by the arrows in the left panel) of a slice at 0, 3 and 6 weeks (from left to right) during radiochemotherapy indicating the increase of ADC value and decrease of tumor volume.
Longitudinal change of lung tumor

![Graph showing ADC values over time](image)

- 0 week: 
- 3 weeks: 
- 6 weeks: 

Statistical significance:
- $p = 0.0023$
- $p = 0.0335$
An example of the effect of the signal intensity (S) vs b-value data at or near noise floor. The ADC value is the negative of the slope. Panel D has much lower ADC than panel A. Similarly, panel D has lower $R^2$ than panel A. The data represented by the open circles in panel B are almost at the noise floor.
Potential Role of BOLD MRI in Discrimination of Aggressive Tumor Habitat in Prostate Cancer

- John Chetley Ford
- Christopher Lopez
- Yohann Tschudi
- Adrian Breto
- Kyle Padgett
- Alan Pollack
- Radka Stoyanova
Potential Role of BOLD MRI in Discrimination of Aggressive Tumor Habitat in Prostate Cancer

• Outline:
  • Hypothesis
    • Hypoxia = ↓O\textsubscript{2} => ↓T2*
  • Methods
    • Retrospective prostate MRI study
  • Results
    • T2* association with tumor environment
  • Conclusions and Future Work
Human in vivo MR Microscopy

- Custom built RF coil
- High resolution 3D pulse sequences
- Analyze in vivo trabecular microstructure
  - Differentiate bone from marrow
- Clinical result:
  - Structural parameters significantly different in osteoporosis
    - Mean trabecular thickness, spacing
  - Can apply FEM directly to image to calculate bone strength and fracture risk
Background and Hypothesis

• **BOLD: Blood Oxygenation Level Dependent**
  - \( \frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2'} \) or \( R_2^* = R_2 + R_2' \)
  - \( T_2' \) (and hence, \( T_2^* \)) is shortened by increased magnetic field inhomogeneity arising from presence of de-oxyhemoglobin in capillaries.

• **Hypothesis: \( T_2^* \) is a measure of hypoxia, and is associated with tumor aggressiveness.**
  - \( \downarrow O_2 \Rightarrow \downarrow T_2^* \)
Clinical trials of Prostate Radiotherapy

- MR studies in ongoing trials utilize fusions between:
  - Diagnostic MRI and Planning MRI (#1)
  - Planning MRI and Planning CT (#2)
  - Via #1 and #2, diagnostic MR is fused to planning CT (#3)

- Fusion #2 is fiducial-based
  - Planning MRI uses gradient-echo MRI
  - T2*-weighted image
  - Good fiducial visualization
  - MERGE
MR Imaging
GE Discovery MR750 3T

• Diagnostic MRI
  • T1, T2, DCE
  • Diffusion-weighted
    • EPI, fat-suppressed, TR/TE=9500/53
    • 2.5x2.5x2.5 mm
    • $b = 50, 500, 1000$ s/mm²
    • Apparent Diffusion Coefficient (ADC) maps

• Planning MRI
  • T2*
    • MERGE, TR/TE/flip=30/12.8/12
    • 1.25x1.25x2.5 mm
Patients

- Among patients enrolled in several ongoing prostate clinical trials:
  - 31 had ADC and MERGE (T2*) on GE 3T
  - Identical MR protocol
- Time between diagnostic and planning MRI:
  - 1.7 ± 1.3 months (mean ± sd)

<table>
<thead>
<tr>
<th>Patients’ clinical characteristics</th>
</tr>
</thead>
<tbody>
<tr>
<td># of Patients</td>
</tr>
<tr>
<td>Age</td>
</tr>
<tr>
<td>Median</td>
</tr>
<tr>
<td>Range</td>
</tr>
<tr>
<td>Gleason Score</td>
</tr>
<tr>
<td>6</td>
</tr>
<tr>
<td>7(3+4)</td>
</tr>
<tr>
<td>7(4+3)</td>
</tr>
<tr>
<td>8</td>
</tr>
<tr>
<td>9</td>
</tr>
<tr>
<td>Tumor Category</td>
</tr>
<tr>
<td>T1</td>
</tr>
<tr>
<td>T2</td>
</tr>
<tr>
<td>T3</td>
</tr>
<tr>
<td>PSA</td>
</tr>
<tr>
<td>Median</td>
</tr>
<tr>
<td>Range</td>
</tr>
</tbody>
</table>

Abbreviations: PSA = Prostate Specific Antigen
Analysis

• **Structure delineation**
  • Tumor volumes defined by ADC on Diagnostic MRI
    • **High, Medium, Low** risk: < 800, 800-1000, 1000-1200 µm²/s
  • Normal-appearing tissue in *peripheral* and *transition* zones contoured by physician on T2 Diagnostic MRI

• **Image registration (rigid)**
  • Diagnostic (T2) and Planning (T2*) MRIs in MIM

• **Signal normalization**
  • T2*-weighted signals in delineated volumes were all normalized to signal in nearby *obturator internus muscle*
Analysis

• We analyzed dependence of normalized T2*-weighted signal on:
  • Normal-appearing tissue vs. tumor
  • ADC level
  • Gleason Score
BOLD signal discriminates tumor from non-tumor in prostate

$p = 0.02$
BOLD signal appears to be related to ADC level and normal prostate zone
BOLD signal is related to Gleason Score

Each error bar is constructed using 1 standard error from the mean.
Conclusions

• T2* signal is related to Gleason Score and ADC level
  • Possibly a surrogate for hypoxia
  • Possibly provides information about tumor environment orthogonal to GS and/or ADC
  • Potential biomarker for prostate cancer

• Strengths of study:
  • ADC-based tumor delineation
  • Number of patients

• Weaknesses:
  • ADC and MERGE in different scans
  • Presence of fiducials
Future Work

• Prospective MRI Study
  • Acquire ADC and BOLD in the same diagnostic study
    • Eliminate registration errors
    • Eliminate effect of fiducial markers
  • Acquire quantitative T2’ maps
    • Acquire T2* images with multiple gradient echoes
    • Mitigate possible effects of T2 and T1
    • Obviate need for normalization to muscle signal
  • Correlate BOLD with patient outcome (biochemical failure, survival, etc.)
Challenges of MRI-based Radiomics

- Fei Yang
- Nesrin Dogan
- Radka Stoyanova
- John Chetley Ford
Radiomic Texture Analysis of MRI

<table>
<thead>
<tr>
<th>A. mpMRI</th>
<th>B. SEGMENTATION</th>
<th>C. FEATURE EXTRACTION</th>
<th>D. DATA INTEGRATION</th>
<th>E. DATA MINING</th>
</tr>
</thead>
<tbody>
<tr>
<td>T2w</td>
<td>Prostate Structures</td>
<td>Volume/Shape Features</td>
<td>Clinical Genomic</td>
<td>Radiogenomics</td>
</tr>
<tr>
<td>DWI</td>
<td>Prostate, PZ, Urethra</td>
<td>Histogram Features</td>
<td>Metabonomic</td>
<td>Predictive/prognostic models</td>
</tr>
<tr>
<td>ADC</td>
<td>Normal Appearing Tissues</td>
<td>Texture Features</td>
<td>Proteomic</td>
<td>Diagnostic models</td>
</tr>
<tr>
<td>DCE-MRI</td>
<td>Regions of Interest</td>
<td>Transform Analysis</td>
<td>Metabonomic</td>
<td></td>
</tr>
</tbody>
</table>

Stoyanova et al., Transl Cancer Res 2016;5(4):432-447
Texture Analysis

- **GLIHM**
  - Variance, skewness, kurtosis characterizing the shape of intensity histogram: dispersion, symmetry, peakedness

- **GLCOM**
  - These features capture local spatial properties of the image through joint occurrence probability of one gray-level value relative to another at a specified linear displacement (one voxel)

- **GLZSM**
  - These features depict regional spatial properties of the image content through the spatial frequency of contiguous regions that encompassed voxels sharing identical gray-level values

- **GLNDM**
  - These exploit visual perceptual property of textures by discerning the spatial details within an image in terms of the gray-level difference between image voxels and their local neighborhoods (3x3x3)

<table>
<thead>
<tr>
<th>Category</th>
<th>Feature</th>
</tr>
</thead>
<tbody>
<tr>
<td>Based on Gray-level Intensity Histogram (GLIHM)</td>
<td>Variance (VARI)</td>
</tr>
<tr>
<td></td>
<td>Skewness (SKEW)</td>
</tr>
<tr>
<td></td>
<td>Kurtosis (KURT)</td>
</tr>
<tr>
<td>Based on Gray-level Co-occurrence Matrix (GLCOM)</td>
<td>Contrast (CONTR)</td>
</tr>
<tr>
<td></td>
<td>Correlation (CORR)</td>
</tr>
<tr>
<td></td>
<td>Dissimilarity (DISSI)</td>
</tr>
<tr>
<td></td>
<td>Energy (ENGY)</td>
</tr>
<tr>
<td></td>
<td>Entropy (ENTR)</td>
</tr>
<tr>
<td></td>
<td>Homogeneity (HOMO)</td>
</tr>
<tr>
<td></td>
<td>SumAverage (SUMAVG)</td>
</tr>
<tr>
<td></td>
<td>Variance (VARI)</td>
</tr>
<tr>
<td>Based on Gray-level Zone Size Matrix (GLZSM)</td>
<td>Short Zones Emphasis (SZE)</td>
</tr>
<tr>
<td></td>
<td>Large Zones Emphasis (LZE)</td>
</tr>
<tr>
<td></td>
<td>Low Gray-level Zones Emphasis (LGZE)</td>
</tr>
<tr>
<td></td>
<td>High Gray-level Zones Emphasis (HGZE)</td>
</tr>
<tr>
<td></td>
<td>Short Zones Low Gray-level Emphasis (SZLGE)</td>
</tr>
<tr>
<td></td>
<td>Short Zones High Gray-level Emphasis (SZHGE)</td>
</tr>
<tr>
<td></td>
<td>Large Zones Low Gray-level Emphasis (LZLGE)</td>
</tr>
<tr>
<td></td>
<td>Large Zones High Gray-level Emphasis (LZHGE)</td>
</tr>
<tr>
<td></td>
<td>Gray-Level Non-uniformity (GLN)</td>
</tr>
<tr>
<td></td>
<td>Zone Size Non-uniformity (ZSN)</td>
</tr>
<tr>
<td></td>
<td>Zone Size Variance (ZSV)</td>
</tr>
<tr>
<td></td>
<td>Gray-Level Variance (GLV)</td>
</tr>
<tr>
<td></td>
<td>Zone Percentage (ZP)</td>
</tr>
<tr>
<td>Based on Gray-level Neighborhood Difference Matrix (GLNDM)</td>
<td>Coarseness (COAR)</td>
</tr>
<tr>
<td></td>
<td>Contrast (CONTR)</td>
</tr>
<tr>
<td></td>
<td>Busyness (BUSY)</td>
</tr>
<tr>
<td></td>
<td>Complexity (CPLX)</td>
</tr>
<tr>
<td></td>
<td>Strength (STRG)</td>
</tr>
</tbody>
</table>
Radiomic Texture Analysis of MRI: Challenges

- Like CT, MRI texture influenced by filtering, noise level, voxel size
- Unlike CT, MRI in general does not reflect physical parameters
  - Somewhat mitigated by acquiring quantitative maps (T1 map, $B_0$ map, ADC, $k_{trans}$, etc.)
- Unlike CT, MRI has many acquisition parameters that can influence texture:
  - TR/TE/flip angle
  - Pulse sequence
  - RF coil selection and placement
  - Reconstruction algorithm (parallel imaging)
Radiomic Texture Analysis of MRI: Challenges

- Are MRI-based texture features:
  - Repeatable? (i.e., consistent within a patient, test/retest)
  - Reproducible? (i.e., consistent across machines and/or pulse sequences)
  - Valid? (i.e., consistent with ground truth)
Repeatability of Texture Features in T1- and T2-weighted MR Images

R. N. Mahon¹, E. Weiss¹, J. Ford², K. Karki¹, G. D. Hugo¹

¹Department of Radiation Oncology, Virginia Commonwealth University, Richmond, VA. ²University of Miami Miller School of Medicine, Miami, FL
## Repeatability of Texture Features in T1- and T2-weighted MR Images

<table>
<thead>
<tr>
<th>VIFR (T1-weighted)</th>
<th>VIBE (T1-weighted)</th>
<th>VIFR (T2-weighted)</th>
<th>VIBE (T2-weighted)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Wavelet Ratio</strong></td>
<td>0.5</td>
<td>0.6</td>
<td>1</td>
</tr>
<tr>
<td><strong>RN</strong></td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
</tr>
<tr>
<td><strong>GLM</strong></td>
<td>0.95</td>
<td>0.95</td>
<td>0.95</td>
</tr>
<tr>
<td><strong>GLS</strong></td>
<td>0.95</td>
<td>0.95</td>
<td>0.95</td>
</tr>
<tr>
<td><strong>ZN</strong></td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
</tr>
<tr>
<td><strong>Z</strong></td>
<td>0.96</td>
<td>0.96</td>
<td>0.96</td>
</tr>
<tr>
<td><strong>Energy</strong></td>
<td>0.97</td>
<td>0.97</td>
<td>0.97</td>
</tr>
<tr>
<td><strong>Entropy</strong></td>
<td>0.97</td>
<td>0.97</td>
<td>0.97</td>
</tr>
<tr>
<td><strong>Homogeneity</strong></td>
<td>0.91</td>
<td>0.91</td>
<td>0.91</td>
</tr>
<tr>
<td><strong>Median</strong></td>
<td>0.91</td>
<td>0.91</td>
<td>0.91</td>
</tr>
<tr>
<td><strong>Mean</strong></td>
<td>0.91</td>
<td>0.91</td>
<td>0.91</td>
</tr>
</tbody>
</table>

### Highly Repeatable Features
- RN
- GLM
- GLS
- ZN
- Z
- Energy
- Entropy
- Homogeneity
- Median
- Mean

### Potentially Repeatable Features
- VIFR (T1-weighted)
- VIFR (T2-weighted)
- VIBE (T1-weighted)
- VIBE (T2-weighted)

### Not Repeatable Features
- **Wavelet Ratio**
- **Comparision**
- **Contrast**
- **EntropyPS**
- **Euler**
- **Entropy**
- **Gabor**
- **Gabor2**
- **GrayLevel**
- **GrayLevel2**
- **Intensity**
- **IntensityPS**
- **MeanPS**

---

### Repeatability Index (R.I.)
- **R.I. = Wavelet Ratio / VIFR**

---

*J Chetley Ford, PhD*
Radiomic Texture Analysis of MRI: Challenges

- Are MRI-based texture features:
  - Repeatable? (i.e., consistent within a patient, test/retest)
  - Reproducible? (i.e., consistent across machines and/or pulse sequences)
  - Valid? (i.e., consistent with ground truth)
Radiomic Texture Analysis of MRI: Reproducibility

- Generated T1- and T2-weighted images based on 3D digital phantom
  - quantitative maps of human brain
  - Used SimuBloch
Radiomic Texture Analysis of MRI: Reproducibility

- Variance for spin-echo T2-weighted images can be substantial
  - TE varying from 60-120 ms, TR = 6400 ms
Radiomic Texture Analysis of MRI: Challenges

- Are MRI-based texture features:
  - Repeatable? (i.e., consistent within a patient, test/retest)
  - Reproducible? (i.e., consistent across machines and/or pulse sequences)
  - Valid? (i.e., consistent with ground truth)
Radiomic Texture Analysis of MRI: Validity

- Compared images to ground truth (analytical digital phantom)
  - Investigated texture dependence on reconstruction algorithm and noise level

<table>
<thead>
<tr>
<th>Recon Type</th>
<th>No added noise</th>
<th>Rician noise added</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Reconstructed Image</td>
<td>Error Map</td>
</tr>
<tr>
<td>CG</td>
<td><img src="image1.png" alt="Reconstructed Image" /></td>
<td><img src="image2.png" alt="Error Map" /></td>
</tr>
<tr>
<td>TV</td>
<td><img src="image5.png" alt="Reconstructed Image" /></td>
<td><img src="image6.png" alt="Error Map" /></td>
</tr>
<tr>
<td>WL</td>
<td><img src="image9.png" alt="Reconstructed Image" /></td>
<td><img src="image10.png" alt="Error Map" /></td>
</tr>
</tbody>
</table>

Radiomic Texture Analysis of MRI: Validity

- Vertical gray bar
  - Typical noise level in brain MRI
- Horizontal orange bar
  - Ground truth value
- Features vary with reconstruction algorithm
- Many do not approach ground truth, even with zero noise