Ultrasound Guided Procedures

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MICCAI 2011 Tutorial on Image-Guided Interventions

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Agenda

- Challenges of 3D Freehand Ultrasound
- Image-based CT-Ultrasound Registration
- Real-time respiratory motion compensation
- Visual Computing / GPU Algorithms
  - Visualization
  - 3D Freehand Ultrasound Reconstruction
  - Ultrasound Simulation
- Clinical Applications
  - Liver RFA
  - Prostate Biopsy
  - Cardiac EP Procedures
- Conclusion
About Myself

- Ph.D. student from 2004-2006 at Computer Aided Medical Procedures (CAMP), Prof. Nassir Navab, Technische Universität München (TUM)
- March 2006 – April 2010: Research Scientist at Siemens Corporate Research (SCR), Princeton, NJ USA
- Since May 2010: CTO at White Lion Technologies AG, München, Germany

→ The work presented in the following was done at CAMP and SCR
3D Freehand Ultrasound

Calibration Problem: Recover transformation $T_c$ between ultrasound image and tracking sensor/target coordinates

Registration Problem: Recover transformation $T_R$
3D Freehand Ultrasound Calibration

- Phantoms with geometric features:
  - Plane (Single-Wall calibration, Prager et al)
  - Spheres / Lines (Cross-Wire, N-Wire, Trobaugh et al, Prager et al)
  - Wedges (Boctor et al)

- Other Phantoms:
  - MI-based calibration of Agar phantom (Blackall et al)

- Phantom-less:
  - Tracked tool submerged in water (Muratore et al, Khamene et al)

- Image-based in-vivo:
  - Monitoring of calibration (Boctor et al)
  - Alignment of interleaved slices (Wein et al)
CT-Ultrasound Registration - Overview

- Manual vs. automatic registration
- Feature-based vs. intensity-based registration
- B-mode US vs. secondary imaging modes

Liver vasculature derived from contrast-CT and doppler ultrasound; defined landmarks

After registration (courtesy of Papenberg et al., SPIE MI 2008)
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Automatic CT-US Registration – General Approach

- Image-based registration difficult:
- Two physically very different modalities → different representation of anatomy
- Limited quality, artifacts etc.

Crucial components:
Extraction of US information from CT
Assessing alignment

CT Scan → Preprocessing → Intermediate CT Volume

Ultrasound Image(s) → Preprocessing → Intermediate Ultrasound

Rendering → Simulated Ultrasound → Comparison

Transformation → Optimization → Similarity Value
Reflection and transmission at an interface depend on angle and acoustic impedances:

\[
\Delta r(Z_1, Z_2, \theta) = (\cos \theta)^n \left( \frac{Z_2 - Z_1}{Z_2 + Z_1} \right)^2 \\
t(Z_1, Z_2) = 1 - \left( \frac{Z_2 - Z_1}{Z_2 + Z_1} \right)^2 = \frac{4Z_2Z_1}{(Z_2 + Z_1)^2}
\]

Estimation from CT:

\[
\Delta r(\vec{x}, \vec{d}) = \left( \vec{d}^T \frac{\nabla \mu(\vec{x})}{|\nabla \mu(\vec{x})|} \right)^n \left( \frac{|\nabla \mu(\vec{x})|}{2\mu(\vec{x})} \right)^2
\]

for \( n = 1 \):

\[
\Delta r(\vec{x}, \vec{d}) = \left( \vec{d}^T \nabla \mu(\vec{x}) \right) \frac{|\nabla \mu(\vec{x})|}{(2\mu(\vec{x}))^2}; \quad t(\vec{x}) = 1 - \left( \frac{|\nabla \mu(\vec{x})|}{2\mu(\vec{x})} \right)^2
\]

Ray-Tracing approach: Following each ultrasound scanline, updating reflected and transmitted intensity

\[
I(\vec{x}) = I_0 \exp \left( -\int_0^{\lambda x} \left( \frac{|\nabla \mu(\vec{x}_0 + \lambda \vec{d})|}{2\mu(\vec{x}_0 + \lambda \vec{d})} \right)^2 d\lambda \right) \left( \vec{d}^T \nabla \mu(\vec{x}) \right) \frac{|\nabla \mu(\vec{x})|}{(2\mu(\vec{x}))^2}
\]

Log-compression:

\[
r(\vec{x}) = (\log(1 + aI(\vec{x}))) \cdot (\log(1 + a))
\]
Ultrasound simulation from CT (2)

- soft tissue window from CT maps to relative echogeneity
- for CTA scans, an intensity mapping has to be applied
We don’t know the relative contribution of simulated echogeneity and reflection to the ultrasound intensity!
The LC² Similarity Measure

(1) Generalized Correlation Ratio:

\[ CR = 1 - \frac{\sum_{x \in \Omega}(U(x) - f(\mu(T(x))))^2}{|\Omega| \text{Var}(U)} \]

(2) We define the intensity mapping \( f \) as:

\[ f(x_i) = \alpha p_i + \beta r_i + \gamma \]

(3) The parameters \( \alpha, \beta \) and \( \gamma \) minimize:

\[ \left\| M \begin{pmatrix} \alpha \\ \beta \\ \gamma \end{pmatrix} - \begin{pmatrix} u_1 \\ \vdots \\ u_n \end{pmatrix} \right\|^2 \]

with \( M = \begin{pmatrix} p_1 & r_1 & 1 \\ \vdots & \vdots & \vdots \\ p_n & r_n & 1 \end{pmatrix} \)

(4) we get

\[ \begin{pmatrix} \alpha \\ \beta \\ \gamma \end{pmatrix} = (M^T M)^{-1} M \begin{pmatrix} u_1 \\ \vdots \\ u_n \end{pmatrix} = \left( \begin{pmatrix} p_i^2 & p_i r_i & p_i \\ p_i r_i & r_i^2 & r_i \\ p_i & r_i & n \end{pmatrix} \right)^{-1} \begin{pmatrix} \sum p_i u_i \\ \sum r_i u_i \end{pmatrix} \]

Linear correlation of a linear combination: LC²

- Simultaneous estimation of simulation weights and registration parameters
- Weights & similarity can be computed per sweep, per frame or locally
Comparison with standard measures

- Randomized registration study
- Global/frame/local LC^2 and Mutual Information, Correlation Ratio, Higher-Dimensional MI

<table>
<thead>
<tr>
<th>Outliers (%)</th>
<th>Median FRE (mm)</th>
<th>Capture Range (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>global</td>
<td>frame</td>
<td>local</td>
</tr>
<tr>
<td>0</td>
<td>1</td>
<td>35</td>
</tr>
<tr>
<td>8.7</td>
<td>8.5</td>
<td>4.6</td>
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<tr>
<td>15.8</td>
<td>13.7</td>
<td>6.7</td>
</tr>
<tr>
<td>statistical</td>
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<td></td>
</tr>
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<td>global</td>
<td>frame</td>
<td>local</td>
</tr>
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<td>0</td>
<td>1</td>
<td>35</td>
</tr>
<tr>
<td>29</td>
<td>7</td>
<td>100</td>
</tr>
<tr>
<td>8.1</td>
<td>8.7</td>
<td>25.3</td>
</tr>
<tr>
<td>3.8</td>
<td>10.1</td>
<td>0</td>
</tr>
</tbody>
</table>
## Automatic Registration Algorithm Evaluation

**Automatic Registration statistics in 25 patients**

<table>
<thead>
<tr>
<th>Registration Method</th>
<th>Quartile 1 (mm)</th>
<th>Median (mm)</th>
<th>Quartile 3 (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>FRE</td>
<td>TRE</td>
<td>FRE</td>
</tr>
<tr>
<td>Point-based (fiducial)</td>
<td>3.5</td>
<td>4.9</td>
<td>4.4</td>
</tr>
<tr>
<td>Point-based (fid. + target)</td>
<td>4.0</td>
<td>3.3</td>
<td>4.8</td>
</tr>
<tr>
<td>Automatic Rigid Registration</td>
<td>6.3</td>
<td>5.8</td>
<td>10.9</td>
</tr>
<tr>
<td>Automatic Affine Registration</td>
<td>6.4</td>
<td>4.9</td>
<td>10.7</td>
</tr>
</tbody>
</table>

FRE: Fiducial Registration Error  
TRE: Target Registration Error

Automatic – 40 sec.  
Manual – 10 min.

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3D Ultrasound Mosaicing

- Extending common similarity measures for multivariate registration
- Automatic registration, considering artifacts, overlap independency

C. Wachinger, W. Wein, and N. Navab.
*Three-dimensional ultrasound mosaicing.* MICCAI 2007 conference, Brisbane, Australia
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Real-time Motion Compensation

- Dealing with moving targets for cancer treatment
- In Radiotherapy:
  - Gating techniques are established (on LINACs)
  - Adaptive RT techniques (fluoro-based) emerging (e.g. Accuray CyberKnife)

- Our goal: adaptive guidance for Interventional Oncology!
- Based on tracked 3D freehand ultrasound as real-time modality

Adjust pre-operative plan & imaging wrt. patient respiration
→ Ad-hoc, without motion model
Related Work

- Blackall et al ’05: 4D MRI to 3D freehand ultrasound registration
- Khamene et al ’04: 2D+t cine MRI correlated with surrogate marker
- Nakamoto et al ’06: 4D motion model from laparoscopic tracked US

Problems:
- 4D motion model needed → often not clinically feasible
- Breathing irregularities not covered

Our approach:
- Online motion compensation without any motion model, only a surrogate sensor
Our Approach - Overview

1. Record breath-hold sweep

2. Register to CT (as shown before)

3. Attach surrogate sensor, run our method

Compensation Algorithm

CT Volume / Patient

Tracked Ultrasound

Surrogate Signal

Real-time registered CT!
Slice-to-Volume Registration

- Optimizing affine deformation of breath-hold volume wrt. image similarity to live US

\[ \tilde{\alpha}_k = \arg \max_{\alpha_k} \mathcal{M} \left\{ (I_R(\mathbf{M}_{a_k}(\tilde{p})), I_{U_k}(\tilde{p})), \tilde{p} \in \Omega_{I_U} \right\} \]

- \( \mathbf{M}_{a_k} \) is an affine transformation comprising of

\[ \mathbf{M}_{a_k}(\tilde{p}) = \mathbf{A}_k^{-1} \mathbf{T}_R \mathbf{T}_k \mathbf{T}_C \tilde{p}; \quad \tilde{p} = (x, y, 0, 1)^T \]

- We can’t optimize the parameters \( \mathbf{A}_k \) of every frame → constraint needed!
Incorporating surrogate signal

- Assumption: Last n frames lie on a path along the transformation space.

- The location along the path is derived from the surrogate signal:

\[
\tilde{a}_{k-n+i} = \tilde{a}_{k-n} + \frac{s(k - n + i) - s(k - n)}{s(m) - s(k - n)} (\tilde{a}_k - \tilde{a}_{k-n}); \quad i \in [1 \ldots n]
\]

where

\[
m = \arg \max_i |s(i) - s(k - n)|
\]
Parameterization & Registration algorithm

- Affine transformation has 12 DOF, however abdominal motion usually can be represented by a few modes → PCA-based parameter space reduction to 3 DOF

- Transformation interpolation with Lie groups:

  \[ A_{k-n+i} = A_{k-n} \exp \left( \frac{s(k-n+i) - s(k-n)}{s(m) - s(k-n)} \log(A_{k-n}^{-1} A_k) \right) \]

- Local Normalized Cross-Correlation (LNCC) similarity metric
- Powell-Brent optimizer on 3 PCA modes, maximum 3 sets of line minimizations
Experiments

- Siemens Sequoia, abdominal 6C2 probe, magnetic tracking
- Acquisition on four volunteers (regular & irregular breathing)

Mid-exhale

Mid-inhale

First PCA vs. surrogate signal
Results

- Similarity measure results consistent
- Computation time:

<table>
<thead>
<tr>
<th></th>
<th>CPU LNCC</th>
<th>CPU SAD</th>
<th>GPU SAD</th>
</tr>
</thead>
<tbody>
<tr>
<td>time / evaluation</td>
<td>4.8 ms</td>
<td>1.9 ms</td>
<td>1.2 ms</td>
</tr>
<tr>
<td>evaluations / frame</td>
<td>43</td>
<td>38</td>
<td>41</td>
</tr>
<tr>
<td>time / frame</td>
<td>210ms</td>
<td>65ms</td>
<td>49ms</td>
</tr>
</tbody>
</table>
Results (2)

- Visual results convincing, even for highly irregular breathing

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Contextual Visualization

- Shows live ultrasound within dense CT volume, important vasculature emphasized in front & opaque behind ultrasound plane

GPU-accelerated Freehand Ultrasound Reconstruction

- Directly compute oblique cross-sections → improved quality
- Drawing 2D frames as quads onto reconstruction slice(s)
- Mix between forward/backward compounding

- Computes entire $256^3$ reconstruction volume from dense tracked 3D ultrasound sweep in 340ms (on NVidia GeForce 8800 GTX)
- Arbitrary MPR through US sweep ~10ms

Non-linear CT-US registration

- Dense-field deformable registration based on LC²
- Full GPU Implementation with re-simulation of ultrasound in every iteration

GPU-accelerated Ultrasound Simulation

- Dense ray-based simulation of US volume from CT in ~60 ms

Together with Athanasios Karamalis

- Ray-based simulation does not take into account: Diffraction, Scattering, Interference, Multiple reflection and refraction

- Solve explicitly or implicitly the wave equation for sound waves

- Current approaches are computational highly demanding

- Popular software “Field II” can take 12-24 hours on a 32 workstation cluster for simulating a single ultrasound image
Finite-difference Time-domain (FDTD) method

- Introduced and commonly used in computational electrodynamics for solving Maxwell’s time-dependent PDEs
- Further applications in computational fluid dynamics (CFD)

- Solve Partial Differential Equations:
  - Discretize domain
  - Take finite differences for partial derivatives
  - Substitute and solve for desired variable
The Westervelt Equation

- Full nonlinear wave equation for acoustic simulation
- Applications
  - Harmonic frequency simulation
  - Temperature fields simulation for focused ultrasound

\[ \nabla^2 p - \frac{1}{c_0^2} \frac{\partial^2 p}{\partial t^2} + \frac{\delta}{c_0^4} \frac{\partial^3 p}{\partial t^3} + \frac{\beta}{\rho_0 c_0^4} \frac{\partial^2 p^2}{\partial t^2} \]

- Ta: Linear lossless wave propagation
- Tb: Absorption due to heat conduction
- Tc: Nonlinear distortion
- p: Acoustic pressure
- c₀: Speed of sound
- ρ₀: Ambient density of mass
- β: Nonlinear coefficient
- δ: Diffusivity of sound
The Westervelt Equation (2)

- **Discretizing the 2D Westervelt equation**

  \[ \nabla^2 p \quad \text{Fourth order central differences} \]
  \[ \frac{1}{c_0^2} \frac{\partial^2 p}{\partial t^2} \quad \text{Second order central differences} \]
  \[ \frac{\delta}{c_0^4} \frac{\partial^3 p}{\partial t^3} \quad \text{Second order backward differences} \]
  \[ \frac{\beta}{\rho_0 c_0^4} \frac{\partial^2 p^2}{\partial t^2} \quad \text{Second order backward differences} \]

- **Results**
  - Explicit solution for the next time-step
  - Based on the 6 previous time-steps
  - Non-linear distortion and absorption
Westervelt FDTD - Results

- Performance:
  ~1290 time-step evaluations/s for 512x512 sampling grid, on NVidia GeForce 8800 GTX

- Linearly aligned point sources

- Manually segmented CT phantom dataset

- Medium speed variations 600, 1555, 1560 and 4080 [m/s]
Challenges and future work

- Absorbing boundary conditions for mesh boundaries
- Segmentation/labeling of datasets for property assignment
- Variety of excitation pulses (sinusoidal, phase delayed; currently Dirac impulse)
- Pipeline from RF data to final ultrasound image
- Validating simulation with real measurements

Possible Application – High Intensity Focused Ultrasound (HIFU)

- MR guided HIFU demonstrates potential
- Use simulation to guide and predict therapy
- Temperature field simulation in thermoviscous fluids was the first step

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Clinical Application – Interventional Fusion

- Biopsies and Radiofrequency Ablation in the Liver and Kidney
- Based on CT Diagnosis & Planning
- Interventional Navigation by CT or Ultrasound:

  CT: not real-time, interruption for imag.
  US: long training, no pre-op information

→ Combining US Navigation with planning information from CT!
Targeted Prostate Biopsy

- Project with MR/SP/CAS

### Setup

- TRUS probe with sensor
- Tracking field generator
- Workstation
- MRI-TRUS side-by-side

Clinical Application - ICE-CT Fusion

- Fusion of Intracardiac Echography (ICE) with C-Arm CT for Electrophysiological Procedures
- Tracking sensor integrated within ICE catheter

Y. Sun et al. in MICCAI 2007 conference

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Conclusion - Take-away Messages

- Ultrasound challenging, dynamic modality
- Precise modeling of imaging physics and uncertainties important
- Research frontiers: Tracking (sensor fusion), multi-modality registration, motion compensation, visualization, workflow
- Exploiting concurrency is mandatory
- Vision: Image and sensor data feeds are used in real-time to build and update patient-specific model
- Close feedback loop with physicians is crucial – details matter
Thanks!

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