Modeling of Moving organs

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Outline

- Context: moving organs
- Challenge and Objective
- Methodology and why the biomechanical modeling patient specific can help?
- Results and validation
- Conclusion and future work
The delivered dose can be limited to the target, hence sparing the healthy tissues surrounding the tumor.

Context: Hadron therapy:
- Advanced radiotherapy technique for cancer treatment that uses charged particles like light ions or protons instead of photons.
- Requires some appropriate quality assurance procedures.
Context: **Hadron therapy:**

- Advanced radiotherapy technique for cancer treatment that uses charged particles like light ions or protons instead of photons.

- Requires an appropriate quality assurance procedures.

The position of the Bragg peak depends on the density of the matter traversed by the beam.
Context: moving organs

- **Ballistics problem:**
  - The position of the lung tumor
  - Calculate and optimize the dose delivery

  \[\text{Geometric data}\]

- **Specific problem of Hadrontherapy:**
  - Knowledge about the organs traversed by the beam
  - The position of the Bragg peak depends on the density of the matter traversed by the beam

  - Geometric data = insufficient !!!

  \[\text{Density of matter}\]
1- Imaging techniques: Target localisation techniques

- Imaging the tumour, if possible !!!
- Imaging of anatomical structures rigidly bound to the tumour (ex. bony landmarks)
- Detecting artificial fiducials implanted in or near the tumour (invasive, infection risk, pneumothorax...)

Techniques based on imaging

Existing solutions on tracking:
- Fiducial Markers
- CyberKnife system
- Online Pet Scan

assume a reproducible motion of the respiratory system.

intrusive or invasive the irradiation of the patient
2- Monitoring surrogate organs moving in synchronism with the lung tumour.

Correlation diaphragm movement vs pancreas and lung tumour

Correlation between internal motion lungs and external motion thorax

Respiratory instability

\[ E_{\text{Resp}}(t) = F(\phi(t)) \]
Non-reproducibility of the movements

Lung tumor inferior/superior vs left/right motion

Internal marker: exhalation & inhalation positions

Non-reproducibility of the movements (chaotic?)

H. Shirato et al. 2006
Breathing motion Inter/intra fractional uncertainties

Inter-fractional setup errors and tumor shrinkage

Intra-fractional respiratory motion

• Variability in motion trajectory (hysteresis loop)
• Inter/intra-fraction changes in respiratory parameters (baseline, amplitude and frequency)
Anatomy of the respiratory system.

Breathing is non-reproducible

- Contraction of the diaphragm muscle.
- Intercostal muscles of the ribcage.

The breathing cycle is not regular, varies in amplitude and in phase from one cycle to another.
A biophysical approach

- Approach based on the Biomechanical modeling of respiratory system allowing to:
  1. Take non reproducible aspects of lung motions into account
  2. Establish the biomechanical model from patients’ geometrical and physical data
  3. Be monitored by external sensors during the treatment
  4. Simulate a “virtual 4D-Scan”
  5. 4D dose distribution and online imaging control
External surrogates and sensors track internal organs motion based biomechanical model during treatment.
From CT scan images to Biomechanical Simulation

**Pipeline Modeling based CAD modeling for FEM simulation**
Semi-Automatic Segmentation of multi organs

Automatic and semi-automatic Segmentation of multi organs (respiratory system)
CAD reconstruction

Lungs

Thorax

Diaphragm

Respiratory system CAD
Mesh generation is a critical and important step in the finite element analysis. The mesh affects the **accuracy**, **convergence** and **speed** of the computational process.

- **Criteria**
  - Geometry: Distortion, aspect ratio, minimum angle, maximum angle, …
  - FE-based: jacobian

- **Influence:**
  - Low quality = bad mesh convergence
  - Large stress field discontinuities
  - Some elements may « lock » for high aspect ratio
  - round-off errors & singularities
  - if jacobian is negative!
Mesh quality

**Drawbacks**
- Non-robust
- High treatment time
- Quality elements!

**Advantages**
- Robust
- Fast
- High quality tetrahedrons

Pipeline Modeling for finite element simulation
Mesh treatment vs CAD modeling
CAD & Meshing:

In CAD:
- Create CLEAN parts for FEA:
  - Avoid creating small surfaces & edges
  - Avoid « tangent » connections (very small angles)
  - Try to minimize the number of faces present in the model

- Remove unsignificant geometric details:
  - Typical details: fillets / chamfers, small holes, unsignificant components (bolts & nuts, rivets)
Biomechanical modeling of the diaphragm and thorax
Diaphragm modeling

Semi-automatic Segmentation of the diaphragm

Segmentation and 3D mesh generation
Biomechanical modeling:

Type of the nonlinearities (geometric or material nonlinearities)?

Two nonlinear hyperelastic models:
The Saint-Venant Kirchhoff and Mooney-Rivlin models.
Biomechanical modeling of the diaphragm

We chose two simple hyperelastic models, Saint-Venant Kirchhoff and Mooney-Rivlin hyperelastic

For an isotropic elastic or hyperelastic material the elastic energy, noted $W$, can be written as:

$$ W(E) = \frac{\lambda}{2} (tr\ E)^2 + \mu (tr\ E^2) $$  \hspace{1cm} (1)

where $E$ is the Green-Lagrange strain tensor $\lambda$ and $\mu$ are the Lame’s coefficients

$$ E = \frac{1}{2} (F^T \cdot F - I) $$

$$ = \frac{1}{2} \left( \text{grad}\ U + \text{grad}^T\ U + \text{grad}^T\ U \cdot \text{grad}\ U \right) $$  \hspace{1cm} (2)
Biomechanical modeling of the diaphragm

For small deformations, the Green-Lagrange strain tensor is linearized into the infinitesimal strain tensor:

\[ \epsilon = \frac{1}{2} (\nabla U + (\nabla U)^T) \]

The relation between the stress tensor and the strain tensor (Hooke’s law), for isotropic material and for linear deformation, can be written:

\[ \varepsilon = \frac{1 + \nu}{E} \sigma - \frac{\nu}{E} tr(\sigma) I_d \]

\[ \mu = \frac{E}{2(1 + \nu)} \quad \lambda = \nu \frac{E}{(1 - 2\nu)(1 + \nu)} \]
Biomechanical modeling of the diaphragm

The Saint-Venant Kirchhoff law extends the Hooke’s law for large displacement. the Green-Lagrange:

\[ S = (\lambda (tr \mathbf{E}) \mathbf{I} + 2\mu \mathbf{E}) + \eta \left( \lambda (tr \mathbf{\dot{E}}) \mathbf{I} + 2\mu \mathbf{\dot{E}} \right) \]

The strain energy of the second hyperelastic Mooney Rivlin model can be written as:

\[ W(C) = c_1 \left( \tilde{I}_1(C) - 3 \right) + c_2 \left( \tilde{I}_2(C) - 3 \right) + \frac{K}{2} \left( J(C) - 1 \right)^2 \]

where \( c_1, c_2 \) are material parameters and \( K \) is the Bulk modulus. The quantities \( \tilde{I}_1 \) and \( \tilde{I}_2 \) are the isochoric invariants of the Cauchy-deformation tensor \( C : \tilde{I}_1(C) = J(C)^{-2/3} I_1(C), \tilde{I}_2(C) = J(C)^{-4/3} I_2(C) \) where \( I_1(C) = trC, I_2(C) = \frac{1}{2}((trC)^2 - trC^2) \) and \( J \) is the Jacobian:
\[ J(C) = \det \mathbf{F}. \]
1) **The diaphragm behavior** is considered as a compressible solid with an non-linear elastic, and as a heterogeneous material with the muscles in its peripheral part converging into a central tendon.

2) **The complete thorax behavior** with musculoskeletal structure is modeled as incompressible solid with an elastic linear behavior including the ribcage kinematics model using the finite helical axis method.
3D Reconstruction of the Thorax

- Thoracic vertebra
- Body of sternum
- Cartilage Costal margin
- All ribs
- Diaphragm: tendon + muscles

Fixed vertices
- Linear elastic and kinematic behavior
- Linear elastic behavior
- Non linear hyperelastic behavior

= Thorax ribs + diaphragm
Boundary conditions

Enhancement of diaphragm model

- Based on rib kinematics & muscle tension
- Heterogeneous modeling of diaphragm (tendon & muscles) and ribcage kinematic

Finite helical axis method principle

Tendon
Muscles
Attachment to ribs
Tension

The boundary conditions
Diaphragm simulation

Table 1: Mechanical properties of the human diaphragm and thorax, where $E$: Young's modulus, $v$: Poisson's coefficient.

<table>
<thead>
<tr>
<th>Tissues</th>
<th>Young modulus (MPa)</th>
<th>Poisson's coefficient</th>
<th>References</th>
</tr>
</thead>
<tbody>
<tr>
<td>Diaphragm muscle</td>
<td>5.32</td>
<td>0.33</td>
<td>Behr et al. 2006, Pato et al. 2011</td>
</tr>
<tr>
<td>Diaphragm tendon</td>
<td>33</td>
<td>0.33</td>
<td>Behr et al. 2006, Pato et al. 2011</td>
</tr>
<tr>
<td>Ribs</td>
<td>5000</td>
<td>0.3</td>
<td>Kimpara et al. 2005</td>
</tr>
<tr>
<td>Cartilage costal margin</td>
<td>49</td>
<td>0.4</td>
<td>Abe et al. 1996, Yamada 1970</td>
</tr>
<tr>
<td>Body of sternum</td>
<td>11500</td>
<td>0.3</td>
<td>Kimpara et al. 2005</td>
</tr>
<tr>
<td>Thoracic vertebra</td>
<td>9860</td>
<td>0.3</td>
<td>Kimpara et al. 2005</td>
</tr>
</tbody>
</table>

Iterative process to minimize the displacement errors between simulation $U_{sim}$ and initial mesh $U_{exp}$. The comparison is measured by the following equation:

$$\text{Dist} = \frac{1}{n} \sum \frac{|U_{exp} - U_{sim}|}{|U_{exp}|}$$
Diaphragm simulation: quantitative and qualitative analysis

Displacement (U) and Stress (S) for the diaphragm

Maximum displacement Right Posterior (RP) and Left-Posterior (LP) sides,
Slightly larger (RP) side motion than the (LP) side motion

This corresponds to the anatomical reality.
Simulation results

View cut of the diaphragm and ribs

Diaphragm motion → Inspiration → Expiration → Ribs motion

Diaphragm behavior ~15mm ~15mm

Simulation results
4D Biomechanical & Kinematical modeling

Movie
Fig. 9. 3D distance map error between the FE simulation and the reference real mesh at inspiration including or no rib kinematics. Color coding ranges from 0.0mm (blue) to 16.0mm (red), the vertical slices (showing the view from the cross sections) between the reference mesh and FE at expiration. Zone 1 presents the cross sections of interest region (contact with lungs). Zone 2 presents the peripheral contact between the diaphragm and ribcage.

Average error measurement $\varepsilon$ and standard deviation (SD): applied to different scenarios.

<table>
<thead>
<tr>
<th>Patients</th>
<th>St-Venant Kirchhoff</th>
<th>Mooney-Rivlin</th>
</tr>
</thead>
<tbody>
<tr>
<td>$S_1$ (Without RK)</td>
<td>$S_2$ (With RK)</td>
<td>$S_3$ (Without RK)</td>
</tr>
<tr>
<td>mean $\pm SD$ (mm)</td>
<td>2.1 $\pm$ 2.3</td>
<td>2.0 $\pm$ 2.2</td>
</tr>
</tbody>
</table>
Analysis of the diaphragm behavior

### Table III

| Patients | $\varepsilon_1 = \frac{I_1 - I_0}{I_0}$ | $\varepsilon_2 = \ln \left( \frac{I}{I_0} \right)$ | $\varepsilon_3 = \frac{1}{2} \left( \frac{I_{12}^2 - I_0^2}{I_0^2} \right)$ |
|----------|--------------------------------------|---------------------------------|---------------------------------
| P1       | 6.31%                                | 2.83%                           | 6.11%                           |
| P2       | 5.49%                                | 2.45%                           | 5.34%                           |
| p3       | 3.97%                                | 1.76%                           | 3.89%                           |
| P4       | 8.07%                                | 3.65%                           | 7.74%                           |
| P5       | 7.56%                                | 3.41%                           | 7.27%                           |
| P6       | 3.98%                                | 1.70%                           | 4.06%                           |
| P7       | 6.26%                                | 2.81%                           | 6.06%                           |
| P8       | 9.23%                                | 4.20%                           | 8.80%                           |
| P9       | 6.59%                                | 2.96%                           | 6.38%                           |
| P10      | 5.14%                                | 2.41%                           | 5.26%                           |

(Mean ± SD)% 6.38 ± 1, 31 2.86 ± 0.62 6.18 ± 1, 21

Small strains (with the large displacement) -> Saint-Venant Kirchhoff (linear elastic behavior with large displacement)
Biomechanical modeling
patient specific
Biomechnical Patient Specific

Patient specific of the respiratory system
3D reconstruction
3D mesh

- Intercostal muscles action
- Diaphragm action

Two independent actions
Boundary conditions

boundary conditions are inferred from the anatomy
Finite Element Patient Specific Simulation

Finite element Patient specific of the respiratory system

Deformation amplitude

Field deformation and orientation

Stress Von Mises
Experimental Validation

Pre-clinical validation 300 markers

Left Lung

Right Lung

Errors (mm)

L/R  S/I  A/P
Sup    Middle    Inf
Experimental Validation

Patient 6: Axial CT image of lungs showing tumor in left lung

Patient 10: Axial CT image of lungs showing tumor in right lung in contact with the diaphragm

Fig. 5. Mean errors ± standard deviation of lung tumor position during the whole cycle of breathing (10 phases between the EI and EE) between the trajectory calculated from 4D CT images compared to trajectory calculated from biomechanical finite element simulation without (blue curve) and with (red curve) the lung-pressure/diaphragm-force optimization for two patients P6 and P10.
4D Biomechanical simulation
Virtual 4D CT scan reconstruction & dose simulation

Transformation of biomechanical model into a 4D density map

Conversion of displacement simulation into CT scan

Dose simulations on the density map
Multiphysics continuous representation

**Tetrahedral structure**

- Density
- Masse
- Displacements
- Dose

*IEEE ISBI 2014, JCARS 2014*

Voxelisation

Cartography of dose deposit

- Original tumour image
- Reconstructed tumour image
- Original lung image
- Reconstructed lung image
Reconstruction of 4D-PET images

- Tissues implicit tracking
- Taking into account volume variations

Dose accumulated over Deforming tetrahedra
Radioactivity reconstructed on tetrahedral elements

PET Scanner

multiphysics model
Current works

1. Clinical validation of the model by external sensors:
   - Spirometer
   - Image processing
Current works

1. Clinical validation of the model by external sensors
   - Spirometer
   - Thorax movement tracking (RPM markers)

Realistic anthropomorphic phantom
Lung Cancer LuCas (PSI)
Conclusion

Extension of the current works towards:

- Tracking different internal organs from external surrogates in real time, based on patient specific data
- A potential Pre-Therapeutic tool for adaptive dose based on 4D patient specific,
- Contributing to the development of on-line control systems for Radio hadrontherapy (4D PET, Gamma prompt)
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