Internal Forces of the Femur: An Automated Procedure for Applying Boundary Conditions Obtained From Inverse Dynamic Analysis to Finite Element Simulations

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Simulate Interaction of ‘Medical Device’ and internal body structures

…but first
Finite Element Analysis (FEA) => an accepted practice and well-used tool in traditional engineering fields (i.e. aerospace, mechanical)

Has not yet achieved widespread acceptance in the areas of human performance, ergonomics, and biomedicine (Anderson et al. 2007, Post 2004)

Results from FEA applications in such fields have even been characterized as “inherently false, and thus discourage any clinical recommendation based on these methods” (Viceconti et al. 2005)
A Finite Element Analysis Review (highly abridged)

Geometry

Mesh
A Finite Element Analysis Review (highly abridged)

Material Properties and Boundary Conditions
A Finite Element Analysis Review (highly abridged)

Solve (usually involves waiting)

Post Process
Similar Steps for Musculoskeletal Systems (Geometry)

Previous approaches required approximated geometry (Silver-Thorn et al. 1996), traditionally developed from cadaver data or commercially available substitutes (Cristofolini et al. 1996, Heiner 2001).
Computed Tomography (CT) scans of individual patients have been used to reconstruct 3D geometry acceptable for use with FEA problems (Keyak et al. 2001, Viceconti et al. 2003, Bitsakos et al. 2005, Lee et al. 2007).

The CT data set has a resolution of 0.684mm and provides slices at 1.5mm intervals.
Similar Steps for Musculoskeletal Systems (Geometry)

The femur is 41 cm long with a maximum width of 5.4 cm at the condyles.

A mesh consisting primarily of ~2mm tetrahedral elements with excluded mid-side nodes was imported into ANSYS Workbench.

The complete model of the human femur consisted of 89,891 tetrahedral elements (Ansys Element Type, Solid185 – 4 node linear tet) with 18497 nodes.
Similar Steps for Musculoskeletal Systems (Materials)

Commercially available software packages with tomographic reconstruction capabilities (Mimics, Analyze, Osiris) can also be used to define material properties (isotropic) suitable for FEA => using Hounsfield Units relationships

$$\text{HU} = \frac{\mu_X - \mu_{H_2O}}{\mu_{H_2O} - \mu_{air}} \times 1000$$

HU are normalized units associated with CT image scans
- based on the linear attenuation coefficient ($\mu$)
- based on scale -1000 (air) : + 1000 (bone), 0 (water)

The material property of each tetrahedral element was defined using a procedure similar to that used by Peng et al. (2006).
Similar Steps for Musculoskeletal Systems (Materials)

**Density**

The Hounsfield Units (HU) of each voxel in the CT scan indicates the radiodensity of the material, distinguishing the different bone tissue types. *There exist an approximate linear relationship between apparent bone density and HU* (Rho et al. 1995).

The maximum HU of the CT scan, 1575, was defined to be the hardest cortical bone of density \(2000 \text{ kg/m}^3\) and the HU value of 100 was defined to be the minimum density of cortical bone \(100 \text{ kg/m}^3\).
Elastic Moduli

There exist an approximate power relationship between bone material properties and apparent densities (Wirtz et al. 2000).

Elements were assigned elastic moduli calculated from apparent densities using axial loading equations developed by Lotz et al. (1991):

- HU \geq 801, cortical bone (E = 2065\rho^{3.09} \text{ MPa})
- HU \leq 800, cancellous bone (E = 1904\rho^{1.64} \text{ MPa})
- HU < 100, intramedullar tissue (E = 20 \text{ MPa})

A Poisson's ratio of 0.30 was used for all materials.
Similar Steps for Musculoskeletal Systems (Materials)

Methods allow for patient specific information to be utilized in simulation models where previous approaches required approximated geometry (Silver-Thorn et al. 1996), traditionally developed from cadaver data or commercially available substitutes (Cristofolini et al. 1996, Heiner 2001).

What does this all mean?

Geometry → Mesh → Material Properties → Boundary Conditions

Solve → Post Process
Boundary Conditions

Musculoskeletal AnyBody Model

Rigid-body femur (right) consisted of 28 connected ‘via-point’ muscles and one wrapped muscle that was used to model the Iliopsoas muscle group

Data from Vaughan et al., 1992
Musculoskeletal AnyBody Model

Input
- Muscles
- Bones
- Joints
- Drivers: Motion of joints or markers
- Loading on model boundary conditions

Inverse Dynamics Analysis

Biomechanical model

Output
- Muscles: forces, activity, power
- Joints: reaction forces, motion
- And much more!
Boundary Conditions

Gait Cycle Points of Interest

<table>
<thead>
<tr>
<th>% Gait Cycle</th>
<th>Notes</th>
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<th>Notes</th>
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</thead>
<tbody>
<tr>
<td>9%</td>
<td>First Peak in GRF</td>
<td>47%</td>
<td>Second Peak in GRF</td>
</tr>
<tr>
<td>55%</td>
<td>Max Muscle Activity</td>
<td>82%</td>
<td>Swing Phase</td>
</tr>
</tbody>
</table>
Boundary Conditions (Geometry)

Problem with using AnyBody Data ‘As-Is’ - Geometry

Geometry (i.e. muscle insertion points)

AnyBody Model (Original)

Patient Specific

Overlay
Boundary Conditions (Geometry)

Match Geometry

A manual iterative process was utilized in which the *FE model was,

1) uniformly scaled to match selected nodal distances of the rigid body model, and
2) reoriented to align the corresponding nodal positions of both models. Several iterations using different nodal sets were performed.

*FE model was ‘fit’ to the AnyBody joint and select landmark positions.
Due to the differences in anatomical definitions, the AnyBody nodes that were used to define the points of force application could not be perfectly aligned with the nodes or surface of the FE geometry.

A second automated procedure was then used to re-define the AnyBody nodal positions of the musculoskeletal model to the closest (by Euclidean distance) FE model surface node.
Boundary Conditions (Mass Properties)

Problems with using AnyBody Data ‘As-Is’

Rigid Body Definition in AnyBody

\[
\text{AnySeg Thigh} = \{
\]

\[
r0=\text{.HipNodeRef.sRel*.Trunk.SegmentsLumbar.PelvisSeg.Axes0'}+.\text{.Trunk.SegmentsLumbar.PelvisSeg.Axes0'} \]

\[
\text{#include "./DrawSettings/Nodes.any"}
\]

\[
\text{#include "./DrawSettings/SegmentAxes.any"}
\]

\[
\text{AnyFunTransform3D &Scale = .Scaling.GeometricalScaling.Thigh.ScaleFunction;}
\]

\[
\]

\[
\text{//AnyVar MassStandard = 0.1*BodyParRef.BodyMass; // Winter}
\]

\[
\text{AnyVar MassStandard = StandardParameters.Thigh.Mass;}
\]

\[
\text{Mass = MassS*MassStandard}
\]

\[
\text{//AnyVar LengthStandard = ( (-0.025)^2 + (0.197+0.2580)^2 + (-0.028+0.038)^2 )^{0.5}}
\]

\[
\text{AnyVar Length = ( (KneeJoint.sRel[0]-HipJoint.sRel[0])^2 + (KneeJoint.sRel[1]-HipJoint.sRel[1])^2)}
\]

\[
\text{AnyVar Radius = (Mass/(3.1416*Length*.StandardParameters.Thigh.Density))^{0.5};}
\]

\[
\text{AnyVar Ixx = 0.25*Mass*Radius*Radius + 1/12*Mass*Length*Length;}
\]

\[
\text{AnyVar Iyy = 0.5*Mass*Radius*Radius;}
\]

\[
\text{AnyVar Izz = Ixx;}
\]

\[
\text{Jii = \{Ixx, Iyy, Izz\};}
\]
**Boundary Conditions (Mass Properties)**

**ANSYS Mass Properties (from geometry)**

<table>
<thead>
<tr>
<th><strong>TOTAL MASS (X,Y,Z)</strong></th>
<th>0.27563</th>
<th>0.27563</th>
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</table>

**Moments and Products of Inertia Tensor (I) About Origin (Global Cartesian):**

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<thead>
<tr>
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<th>0.52654E-02</th>
<th>-0.24316E-03</th>
<th>0.20062E-02</th>
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**Center of Mass (X,Y,Z):**

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<th>0.13436E-01</th>
<th>0.18810E-01</th>
<th>-0.48539E-01</th>
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</table>

**Moments and Products of Inertia Tensor (I) About Center of Mass (Global Cartesian):**

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<th>0.18265E-02</th>
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<td>0.52305E-02</td>
<td>0.40709E-03</td>
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<tr>
<td>0.18265E-02</td>
<td>0.40709E-03</td>
<td>0.87017E-03</td>
<td></td>
</tr>
</tbody>
</table>

**Principal Centroidal Moments of Inertia:**

<table>
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<tr>
<th></th>
<th>0.52777E-02</th>
<th>0.52664E-02</th>
<th>0.74961E-04</th>
</tr>
</thead>
</table>

**Orientation Vectors (Global Cartesian) for the Principal Axes:**

- (0.850, -0.423, 0.313)
- (0.363, 0.902, 0.234)
- (-0.382, -0.086, 0.920)

**Angles (XY, YZ, XZ) of the Principal Axes:**

-21.90, 13.54, -18.80
Muscle force directions, relative magnitudes, and application points (labeled) for the right femur at the time of the first peak in GRF (9% in the gait cycle).
Solve

1. ANSYS APDL script
2. Re-oriented (and scaled) geometry file to selected time step
Results

1. Max-Principal Surface Strains
2. Von-Mises Stresses
3. Deformation

=> Comparisons w/ Previously Published Findings
The strain data for each side was calculated on paths manually defined along the edges between elements in the FEM mesh.
Duda et al. (1998) => The maximum principal strains showed similar trends between the two studies along the medial and anterior surfaces with the peak strains being located along the medial surface in the subtrochanteric region for both studies.
Strains

Taylor et al. (1996) reported maximum medial strains in the axial direction below -1500με with the majority of bone having less than -1000με. The medial surface strain results presented here were slightly higher, possibly due to the maximum principal strains not being directly aligned with the axial direction.
Strains

Speirs et al. (2007) => The anterior surface strains were similarly the lowest in average magnitude across all the surfaces for both studies. Similar trend (increasing in a linear manner from the distal diaphyseal to the intertrochanteric level) for lateral surface strain, however the magnitude in this study was approximately twice as large suggesting the femur presented here exhibited more bending in the coronal plane.
The von Mises stress gradients between the medial/lateral and anterior/posterior surfaces were not uniform as suggested should be the case for physiological loading conditions by Taylor et al. (1996).

Taylor et al. (1996) observed similar results for initial loading cases and was able to generate more uniform stress distributions by increasing the angle of the applied hip reaction force to 20 degrees from vertical.
Deformation

Smaller deflections were reported than for similar FE models with constraints not explicitly defined to be physiological (Taylor et al. 1996, Speirs et al. 2007). However, a greater magnitude of deflection (at the mid-bone) than corresponding ‘physiologically’ constrained models.

Gait Cycle

9%  47%  55%  82%
Max: 9.2mm  Max: 9.9mm  Max: 7.5mm  Max: 0.3mm
Deformation

Speirs et al. (2007) explicitly observed a lateral bow of the femur, similar to the deformation presented here.

The elastic modulus used by Taylor et al. (1996) and Speirs et al. (2007), particularly to model the cortical bone (17000 MPa), was significantly higher than the range of values used here (ranging from 1850 to 16737 MPa).

=> This would result in a stiffer femur than the one used here and potentially explain the smaller observed overall deflections in those respective studies.
Limitations

Definition of material properties

The relationship between the elastic modulus and the Hounsfield Units (Peng et al. 2006) developed by linearly relating HU to apparent density seems to have resulted in particularly low values of elastic moduli when compared to other models in literature (Wirtz et al. 2000). The range of HU used in this study yields an elastic modulus an order of magnitude smaller at lower HU values when compared to other models (Yosibash et al. 2007).

Inability to directly validate the developed FE model with a physical test specimen

The capacity to directly measure femur strains, stresses, or deformations for individual subjects performing dynamic tasks is highly prohibitive.

Geometry Alignment

• The procedure for displacing the rigid body defined nodes to be coincident with the bone geometry was assumed to not significantly affect the anatomical definitions validated in the musculoskeletal model in a negative fashion. => Muscle Insertion Points from MRI?

• The manual procedure for aligning the bone geometry to the un-displaced node positions was performed in an ad-hoc manner and was assumed to not have introduced significant errors associated with mis-aligned geometry, although as previously mentioned, may have resulted in higher femur stresses.
Summary

With the increasing cost-effectiveness of computational performance, the impediment for accurate FE models of select biological systems is not the capability to solve those complex models, but the capacity to define them in meaningful ways.
Questions?