DYNAmore GmbH

Biomechanical Material Models in LS-DYNA

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Information Day: Biomechanics with LS-DYNA
12 November 2013, Stuttgart
Overview

- Human Models (THUMS)
- Material Models
  - 1-d Material Models
    - Muscles, tendons
  - 3-d Material Models
    - Cartilage
    - Tendons
    - Brain
    - Muscles
Human Models

- Based on Multi-Body Systems
  - Easy to set up
  - Numerically cheap
  - No field functions (stress, strain, etc.)
  - Usually no failure prediction possible

- Based on Finite Element Models
  - Difficult to set up
  - Numerically expensive
  - Includes field functions
  - Failure prediction under research

[www.tass-safe.com]  [www.anybody.com]  [THUMS®  www.dynamore.de]
THUMS™ – Total HUman Model for Safety

- Detailed human model for numerical crash test simulation
- Sitting occupant & standing pedestrian model
- THUMS is developed by
  - Toyota Central R&D LABS
  - Wayne State University

[courtesy of Daimler AG]
THUMS Model Versions 1.x and 3.0

- Mostly based on literature data (geometry and material properties)
- Simple materials (mostly elastic, elastic-plastic, viscoelastic)

Versions 1.4/1.6 (2004-06)
- Kinematical model (skeletal structure, joints, flesh, simplified organs, simple head model)

Version 3.0 (beginning of 2008)
- Refined head model (based on CT-scans)
- Also: material adaptations, slight geometrical changes
- Theoretically head injury simulations possible
THUMS Model Version 4 (since end 2010)

- Geometry obtained from medical CT scans
  - Basis: 39 year-old male (173cm, 77.3kg, BMI 25.8)
  - Scaled to AM50 model (178.6cm, 74.3kg) → realistic geometry
  - High detailing of joints, internal organs, head, ...

- Model parameters
  - Element size 3-5mm, 1.8 Mio elements, 630,000 nodes
  - Mainly hexahedrons/tetrahedrons and some shell elements

occupant upper body  pedestrian thorax  THUMS 4 occupant and pedestrian models
Pedestrian frontal impact with THUMS V4

Internal loads during impact
Possibility to impose movement in human models

- Inverse kinematics
  - motion is captured and prescribed
  - muscle forces are computed as a reaction due to the imposed movement

- Forward kinematics
  - muscle forces are measured and prescribed
  - motion is computed

Posture and motion prediction

- forces and motion are unknown
- control theory used to predict muscle forces
- motion is computed

[courtesy of Prof. Syn Schmitt]
Material Models

- Out of more than 280 available material models for
  - 1-d discrete elements and finite element
    - *MAT_SPRING_*
    - *MAT_SPRING_MUSCLE*
    - *MAT_CABLE_DISCRETE_BEAM*
    - *MAT_MUSCLE*
  - 3-d finite elements
    - *MAT_OGDEN*
    - *MAT_MOONEY_RIVLIN*
    - *MAT_QUASILINEAR_VISCOELASTIC*
    - *MAT_LUNG_TISSUE*
    - *MAT_BRAIN_LINEAR_VISCOELASTIC*
    - *MAT_SOFT_TISSUE(_.VISCO)*
    - *MAT_HEART_TISSUE*
    - *MAT_TISSUE_DISPERSED*
1-d Material Models

- Discrete elements
  - *ELEMNET_DISCRETE & *SECTION_DISCRETE
    - Springs
    - Dampers

- Available material models

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<td>*MAT_S15:</td>
<td>*MAT_SPRING_MUSCLE</td>
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* Available for tendons, fixators, bracelets, other passive structures
* Available for muscles (active & passive)
**MAT_SPRING_MUSCLE (*MAT_S15)**

- Hill-type muscle model
- Rheological model
  - CE: contractile element
  - PE: parallel elastic element
  - SEE: serial elastic element (optional)
- Muscle force computation

\[
F^M = F^{PE} + F^{CE}
\]

\[
F^{SEE} = F^{CE}
\]

---

**MAT_SPRING_MUSCLE**

```
$------------------1------------------2------------------3------------------4------------------5------------------6------------------7------------------8$
$\# \text{mid} \text{ lo} \text{ vmax} \text{ sv} \text{ a} \text{ fmax} \text{ tl} \text{ tv}$
1 0.000 0.000 1.000000 0.000 0.000 1.000000 1.000000
$\# \text{fpe} \text{ lmax} \text{ ksh}$
0.000 0.000 0.000
```

---

1-d Material Models
Background information

- Skeletal muscles and their activation
  - Sarcomeres are the contractile or functional unit of the muscle
  - Muscle force depends on the sarcomere length

[Graphs are courtesy of Benjamin Cummings]
*MAT_SPRING_MUSCLE (*MAT_S15)

- Contractile element CE

\[ F_{CE} = a(t) F_{max} f_{TL}(L) f_{TV}(V) \]

\[ L = \frac{L^M}{L_0} \]

\[ V = \frac{V^M}{V_{max} S_V(a(t))} \]

- L0 Initial muscle length \( L_0 \)
- VMAX Maximum CE shortening velocity \( V_{max} \)
- SV* Scale factor for \( V_{max} \) vs. active state \( a(t) \)
- A* Activation level \( a(t) \)
- FMAX Peak isometric force \( F_{max} \)
- TL* Active tension vs. length \( f_{TL}(L) \)
- TV* Active tension vs. velocity \( f_{TV}(V) \)

Parameters*:
- < 0: absolute value gives load curve ID
- > 0: constant value of 1.0 is used
**MAT_SPRING_MUSCLE (**MAT_S15**)

- Passive Element PE
  - FPE  Force vs. length function for PEE
    - $<0$: absolute value gives load curve ID
    - $>0$: constant value of 0.0 is used
    - $=0$: exponential function is used

\[
F_{PE}^{PE} = \begin{cases} 
  \frac{F_{PE}}{F_{max}} = 0 & , \forall L \leq 1 \\
  \frac{F_{PE}}{F_{max}} = \frac{1}{\exp(K_{sh}) - 1} \left[ \exp \left( \frac{K_{sh}}{L_{max}} (L - 1) \right) - 1 \right] & , \forall L > 1 
\end{cases}
\]

- LMAX  Relative length when FPE reaches FMAX
- KSH  Constant governing the exponential rise of FPE
Note: *ELEMENT_DISCRETE is no longer being developed and extended

Instead: Use *ELEMENT_BEAM & *SECTION_BEAM

- Pin-jointed elements with 3 degrees of freedom at each node
- Axial force depends on $l_0$, $l$, $A$, and the material model

![Beam Diagram](image)

$l_0, l, A, \sigma$

- Beam type 3: Truss element with 6 material models
  - *MAT_001: Elastic
  - *MAT_003: Elastic-plastic
  - *MAT_004: Elastic-plastic thermal
  - *MAT_027: Mooney-Rivlin rubber
  - *MAT_098: Simplified Johnson-Cook
  - *MAT_156: Hill’s muscle model

- Beam type 6: Discrete beam/cable with 1 material model
  - *MAT_071: Non-linear elastic
*MAT_MUSCLE (*MAT_156)
- Based on Hill-type muscle model *MAT_SPRING_MUSCLE
- Available for discrete and truss beam elements
- Extended by damper element (DE)
- Rheological model
  - CE: force generation by the muscle
  - PE: energy storage from muscle elasticity
  - DE: muscular viscosity
- Muscle force computation

\[ \sigma_{CE} = \sigma_{\max} \cdot a(t) \cdot f(\varepsilon) \cdot g(\dot{\varepsilon}) \]
\[ \sigma_{PE} = \sigma_{\max} \cdot h(\varepsilon) \]
\[ \sigma_{DE} = D \cdot \varepsilon \cdot \dot{\varepsilon} \]

\[ \sigma = \sigma_{CE} + \sigma_{PE} + \sigma_{DE} \]
**MAT_MUSCLE (*MAT_156)**

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**Parameters***:
- `< 0`: absolute value gives load curve ID
- `> 0`: constant value of 1.0 is used

- **RO**: Density
- **SNO**: Initial stretch ratio \( l_0/l_{\text{orig}} \) (nodal distance / original length)
- **SRM**: Maximum strain rate
- **PIS**: Peak isometric stress
- **SSM**: Strain when the stress in SSP is maximal
- **CER**: Exponential rise of SSP (if SSP=0)
- **DMP**: Damping constant
- **ALM***: Activation level vs. time
- **SFR***: Scale factor for strain rate maximum vs. stretch ratio \( l_0/l_{\text{orig}} \)
- **SVS***: Active dimensionless tensile stress vs. the stretch ratio \( l_0/l_{\text{orig}} \)
- **SVR***: Active dimensionless tensile stress vs. the normalized strain rate \( \Delta l / \Delta t \)
- **SSP**: Isometric dimensionless stress vs. the stretch ratio \( l_0/l_{\text{orig}} \)
*MAT_MUSCLE (*MAT_156)

Example of an activation (Prof. Syn Schmitt & Julian Blaschke, INSPO, Uni-Stuttgart)
1-d Material Models

FEM model of bent muscles or tendons which are guided by bones around joints

Examples: Ankle, elbow, knee,

Quantities of interest:
- friction, angle, forces, ...

Problem of deflecting forces

Crossing point (pulley)
Classical modeling technique

- Truss elements with *MAT_MUSCLE and *CONTACT_GUIDED_CABLE

Problems
- Non-smooth contact
- Non-uniform axial forces
Keyword *ELEMENT_BEAM_PULLEY [Erhart 2012]

- Pulleys allow continuous sliding of truss elements through a sharp change of angle
- Available for *MAT_ELASTIC, *MAT_CABLE_DISCRETE_BEAM, *MAT_MUSCLE

*ELEMENT_BEAM_PULLEY

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- PUID: Pulley ID.
- BID1: Truss beam element 1 ID.
- BID2: Truss beam element 2 ID.
- PNID: Pulley node, NID.
- FD: Coulomb dynamic friction coefficient.
- FS: Optional Coulomb static friction coefficient.
- LMIN: Minimum length.
- DC: Decay constant

\[ \mu_c = FD + (FS - FD)e^{-DC \cdot |v_{rel}|} \]

Slip condition: \[ T_2 \leq T_1 e^{\mu \theta} \]

*Euler-Eytelwein Equation*
New modeling technique
- Truss elements with *MAT_MUSCLE and *ELEMENT_BEAM_PULLEY

Problems solved
- Smooth contact
- Uniform axial forces
3-d Material Models

- Passive isotropic material models
  - MAT_OGDEN_RUBBER
  - MAT_MOONEY_RIVLIN
  - MAT_QUASILINEAR_VISCOELASTIC
  - MAT_LUNG_TISSUE
  - MAT_BRAIN_LINEAR_VISCOELASTIC

- Passive anisotropic material models
  - MAT_SOFT_TISSUE(_VISCO)

- Active anisotropic material models
  - MAT_HEART_TISSUE
  - MAT_TISSUE_DISPERSED
*MAT_QUASILINEAR_VISCOELASTIC

- Based on a one-dimensional model by Fung 1993
- Quasi-linear, isotropic, viscoelastic material
- For solid and shell elements
- Old Formulation (FORM = 0)
  - Instantaneous elastic response and convolution integral with relaxation to zero stress

\[
\sigma_e(\varepsilon) = \sum_{i=1}^{k} C_i \varepsilon^i \\
\sigma_V(t) = \int_{0}^{t} G(t - \tau) \frac{\partial \sigma_e[\varepsilon(\tau)]}{\partial \varepsilon} \frac{\partial \varepsilon}{\partial \tau} d\tau \\
G(t) = \sum_{i=1}^{n} G_i e^{-\beta t}
\]

alternative via load curve

- New Formulation (FORM = 1)
  - Split into hyperelastic and viscous contribution
  - Hyperelastic part based on *MAT_SIMPLIFIED_RUBBER assuming incompressibility
  - Relaxation to hyperelastic stress

\[
\sigma(\varepsilon, t) = \sigma_{SR}(\varepsilon) + \sigma_V(t) \\
\sigma_V(t) = \int_{0}^{t} G(t - \tau) \frac{\partial \varepsilon}{\partial \tau} d\tau
\]
*MAT_LUNG_TISSUE

- Hyperelastic model for heart tissue Vawter 1980
  - Isochoric and volumetric strain-energy function

\[
W(I_1, I_2) = \frac{C}{2\Delta} e^{\alpha(I_1^2 + \beta I_2)} + \frac{12C_1}{\Delta(1 + C_2)} [A^{(1+C_2)} - 1]
W_H(J) = \frac{K}{2} (J - 1)^2
\]

\[
A^2 = \frac{4}{3} (I_1 + I_2) - 1
\]

- Linear viscoelasticity based on Christensen 1980
  - Convolution integral with relaxation to the zero stress state
  - Maximum of 6 terms in the Prony series

\[
\sigma_{ij} = \int_0^t g_{ijkl}(t - \tau) \frac{\partial \varepsilon_{kl}}{\partial \tau} d\tau
\]

\[
g(t) = \sum_{i=1}^{n} G_i e^{-\beta_i t}
\]

- Optionally prescribed relaxation of *MAT_GENERAL_VISCOELASTIC via *DEFINE_CURVE

3-d Material Models
*MAT_BRAIN_LINEAR_VISCOELASTIC

- Simple material model for solid elements only

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- Jaumann rate formulation for deviatoric stress rate

\[ \sigma_{ij} = 2 \int_0^t G(t - \tau) D'_{ij}(\tau) d\tau \]

- Simple Maxwell-Kelvin model
  - FO = 0: Maxwell model (fluid like)
  - FO = 1: Kelvin model (solid like)

- Relaxation functional

\[ G(t) = G + (G_0 - G_\infty)e^{-\beta t} \]

- Evolution equation for the stress

\[ \dot{s}_{ij} + \frac{1}{\tau} s_{ij} = (1 + \delta_{ij}) G_0 \dot{e}_{ij} + (1 + \delta_{ij}) \frac{G_\infty}{\tau} \dot{e}_{ij} \]
**MAT_SOFT_TISSUE**

- Element types: Solids and shells (*Belytschko-Tsay* ELTYPE=2)
- Suitable for ligaments, tendons, fascia
  - Isotropic Mooney-Rivlin matrix
  - Collagen fiber reinforcements (transversely isotropic)
  - Simple compression law

\[
W = C_1(\tilde{I}_1 - 3) + C_2(\tilde{I}_2 - 3) + F(\lambda) + \frac{1}{2}K[\ln(J)]^2
\]

\[
\frac{\partial F}{\partial \lambda} = \begin{cases} 
0 & \lambda < 1 \\
\frac{C_3}{\lambda^3} [\exp(C_4(\lambda - 1)) - 1] & \lambda < \lambda^* \\
\frac{1}{\lambda} (C_5\lambda + C_6) & \lambda \geq \lambda^*
\end{cases}
\]

- Exponential behavior of collagen fibers in the intervertebral disc [*Holzapfel et al.* 2005]

- Sample parameters for tendons in *Quapp & Weiss* 1998

passive anisotropic
*MAT_SOFT_TISSUE

C1-C5: Stress parameters

XK: Compression modulus

XLAM: Stretch ratio $\lambda^*$ at which fibers are straightened

XLAM0: Initial fiber stretch (optional)

FANG: Fiber angle in local shell coordinate system (shells only)

FAILSF/M: Failure stretch ratio of fibers and matrix (shells only)

FAILSHR: Shear strain at failure at a material point (shells only)

Remaining parameters: Computation of initial fiber directions

### 3-d Material Models
*MAT_SOFT_TISSUE

Same logic as for composite materials

AOPT<0

AOPT=0

AOPT=2

AOPT=3

3-d Material Models
- General recommendation for anisotropic materials
  - Switch on invariant node numbering
    - The material coordinate system is automatically updated following the rotation of the element coordinate system
    - The response of the orthotropic shell elements can be very sensitive to in-plane shearing deformation and hourglass deformations
  - Invariant node numbering is invoked by *CONTROL_ACCURACY
    - INN=2 (shells)
    - INN=3 (solids)
*MAT_SOFT_TISSUE_VISCO

- Viscoelastic option
  - Convolution integral with six-term Prony series as relaxation function
  - Hyperelastic part represents static case

\[ S(C, t) = S^e(C) + \int_0^t 2G(t - s) \frac{\partial W}{\partial C(s)} ds \quad G(t) = \sum_{i=1}^6 S_i \exp \left( \frac{t}{T_i} \right) \]

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- C1-C5: Factors in the Prony series (stress parameters)
- T1-T6: Characteristic times for Prony series relaxation kernel
**MAT_HEART_TISSUE**

- Heart tissue model described in *Walker et al. 2005*
- Backward compatible with an earlier heart model of *Guccione et al. 1991*
- Hyperelastic material model
  - Strain energy depending on *Green-Lagrangean strain* $\mathbf{E}$
    \[ W = \frac{C}{2}(e^Q - 1) \]
    \[ Q = b_f E_{11}^2 + b_t (E_{22}^2 + E_{33}^2 + E_{23}^2 + E_{32}^2) + b_{fs} (E_{12}^2 + E_{21}^2 + E_{13}^2 + E_{31}^2) \]
  - Stress computation and co-variant push forward
    \[ \mathbf{S} = \frac{\partial W}{\partial \mathbf{E}} - p J \mathbf{C}^{-1} + T_0 \{ t, Ca_0, \ell \} \]
    \[ \mathbf{\sigma} = \frac{1}{J} \mathbf{F S F}^T \]
  - Active fiber stress component is defined by time-varying elastance model
    \[ T_0 = T_{\text{max}} \frac{Ca_0^2}{Ca_0^2 + ECa_{50}^2} C_t \]
  - $T_{\text{max}}$: maximum isometric tension achieved at the longest sarcomere length
  - $ECa_{50}$: Length-dependent calcium sensitivity
  - $C_t$: activation function
*MAT_TISSUE_DISPERSED*

- General hyperelastic invariant formulation for dispersed orthotropy in soft tissues
- Suitable for heart valves, arterial walls or other tissues with one or two collagen fibers
- Stress computation

\[
S = \kappa J(J - 1)C^{-1} + \mu J^{-2/3} \text{DEV} \left[ \frac{1}{4} (I - \bar{C}^{-2}) \right] + J^{-2/3} \sum_{i=1}^{n} \left[ \sigma_i(\lambda_i) + \varepsilon_i(\lambda_i) \right] \text{DEV}[K_i]
\]

Neo-Hooke model

Fiber contribution

- Deviatoric part of a tensor

\[
\text{DEV}[\cdot] = (\cdot) - \frac{1}{3} \text{tr}[(\cdot)C]C^{-1}
\]

- Passive fiber contribution

  - Crimped model by *Freed et al. 2005*
    \[
    \lambda < \Lambda \quad \sigma = \xi E_s(\lambda - 1)
    \]
    \[
    \lambda > \Lambda \quad \sigma = E_s(\lambda - 1) + E_f(\lambda - \Lambda)
    \]

  - Exponential model
    \[
    \sigma = C_1 \left[ e^{\frac{C_2}{2}(\lambda^2 - 1)} - 1 \right]
    \]

3-d Material Models
*MAT_TISSUE_DISPERSED

- Example of an arm with passive muscle material
  - Geometry provided by Prof. O. Röhrle, Uni-Stuttgart, Fraunhofer IPA
  - Bones modeled as rigid bodies
  - Prescribed motion of the forearm
  - Here: Uniform fiber direction

Note: Location dependent fiber orientation via diffusion tensor MRI of the muscle
Recall the background information

- Skeletal muscles and their activation
  - Sarcomeres are the contractile or functional unit of the muscle

[Graphs are courtesy of Benjamin Cummings]
■ Background information

■ Skeletal muscles and their activation
  ■ Muscle force depends on the sarcomere length

\[ S_{\text{muscle}} = S_{\text{iso}} + S_{\text{aniso}} + S_{\text{tension}} \]

\( S_{\text{muscle}} \) — normalized muscle force

\( S_{\text{iso}} \) — passive part
\( S_{\text{aniso}} \) — active part

Courtesy of Prof. O. Röhrle, Uni-Stuttgart, Fraunhofer IPA
**MAT_TISSUE_DISPERSED**

- Active fiber contribution by Guccione et al. 1993
- Stress in the muscle fiber

\[ \sigma = T_{\text{max}} \frac{Ca_0^2}{Ca_0^2 + ECa_{50}^2} C(t) \]

\[ ECa_{50} = \frac{(Ca_0)_{\text{max}}}{\sqrt{e^{B(l_r \sqrt{2(\lambda-1)+1-l_0})-1}}} \]

- Activation function

\[ C(t) = \frac{1}{2} \left( 1 - \cos \omega(t) \right) \]

\[ \omega = \begin{cases} \frac{\pi t}{t_0} & 0 \leq t < t_0 \\ \frac{t - t_0 + t_r}{t_r} & t_0 \leq t < t_0 + t_r \\ 0 & t_0 + t_r \leq t \end{cases} \]
- *MAT_TISSUE_DISPERSED*
  - Example of an activated fusiform muscle
  - Definition of the fiber alignment
    - Local coordinates (AOPT = 0)

[Graphs are courtesy of Benjamin Cummings]
- **MAT_TISSUE_DISPERSED**
  - Example of an activated fusiform muscle
  - Simulation results using LS-DYNA
    - Muscle force

3-d Material Models
- **MAT_TISSUE_DISPERSED**
  - Example of an arm with active muscle material
    - Geometry provided by Prof. O. Röhrle, Uni-Stuttgart, Fraunhofer IPA
    - Bones modeled as rigid bodies
    - Motion by activation of bizeps
Summary

Out of more than 280 available material models for

1-d discrete elements and finite element
- *MAT_SPRING_*
- *MAT_SPRING_MUSCLE
- *MAT_CABLE_DISCRETE_BEAM
- *MAT_MUSCLE

3-d finite elements
- *MAT_OGDEN
- *MAT_MOONEY_RIVLIN
- *MAT_QUASILINEAR_VISCOELASTIC
- *MAT_LUNG_TISSUE
- *MAT_BRAIN_LINEAR_VISCOELASTIC
- *MAT_SOFT_TISSUE(_VISCO)
- *MAT_HEART_TISSUE
- *MAT_TISSUE_DISPERSED

- no longer being developed and extended
- more versatile
- passive isotropic
- passive transverse isotropic
- active anisotropic
Thank you for your attention!

Your LS-DYNA distributor and more