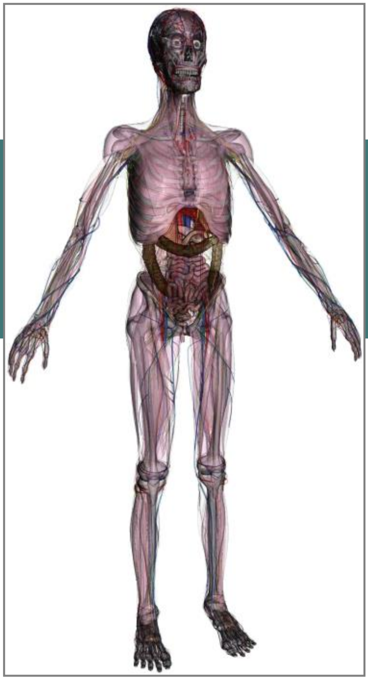


[www.anatomium.com]

DYNAmore GmbH

Biomechanical Material Models in LS-DYNA

N. Karajan, D. Fressmann, T. Erhart & P. Schumacher

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

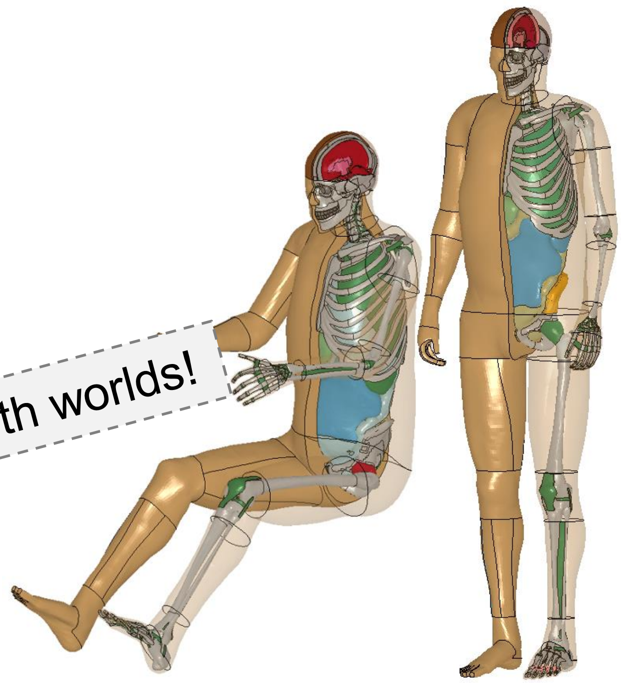
Human Models

- 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]

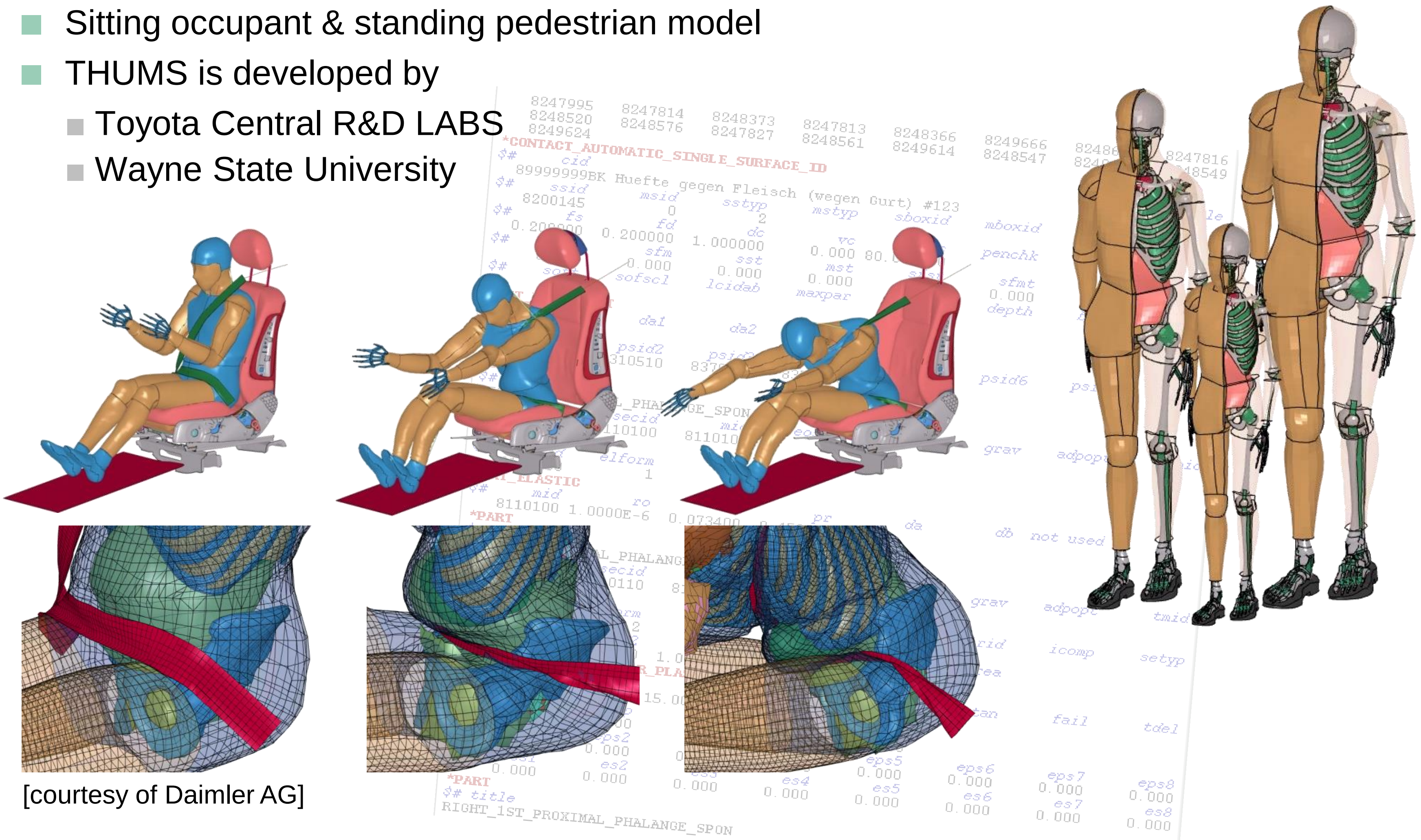
[www.anybody.com]

LS-DYNA unites both worlds!



[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

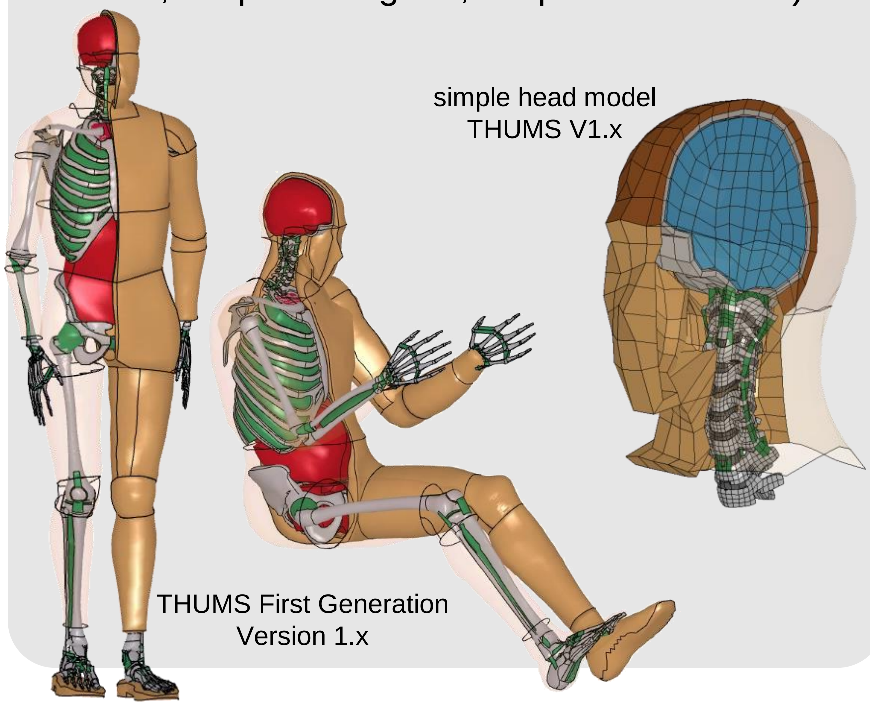


[courtesy of Daimler AG]

- THUMS Model Versions 1.x and 3.0
  - Mostly based on literature data (geometry and material properties)
  - Simple materials (mostly elastic, elastic-plastic, viscoelastic)

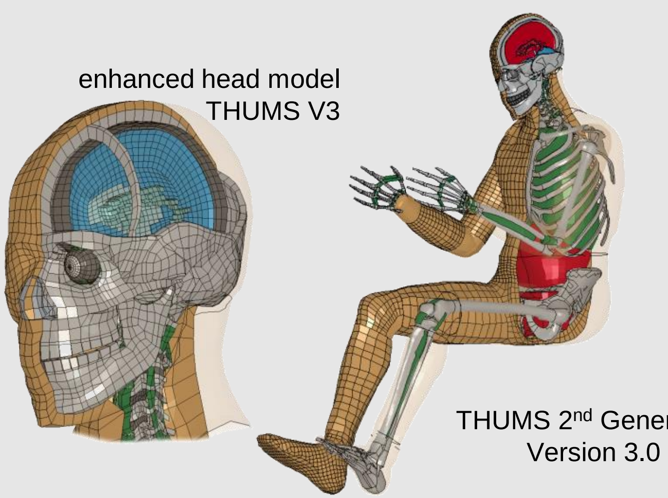
Versions 1.4/1.6 (2004-06)

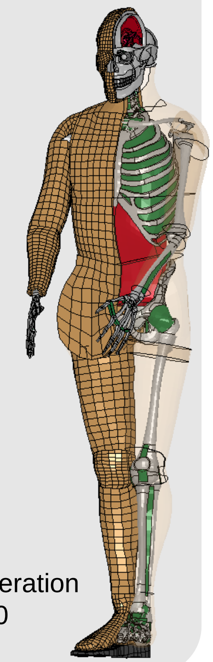
- kinematical model (skeletal structure, joints, flesh, simplified organs, simple head model)



Version 3.0 (beginning of 2008)

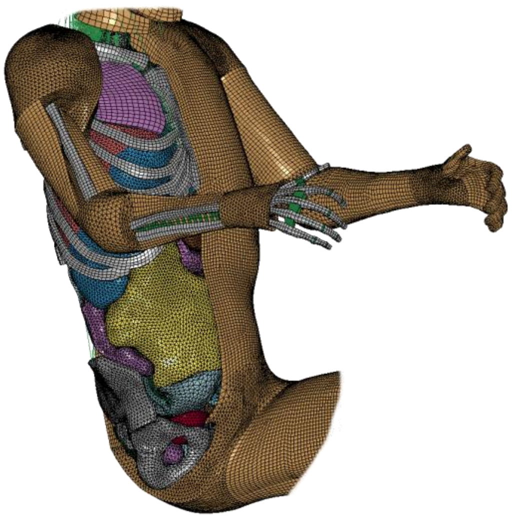
- refined head model (based on CT-scans)
- also: material adaptations, slight geometrical changes
- theoretically head injury simulations possible



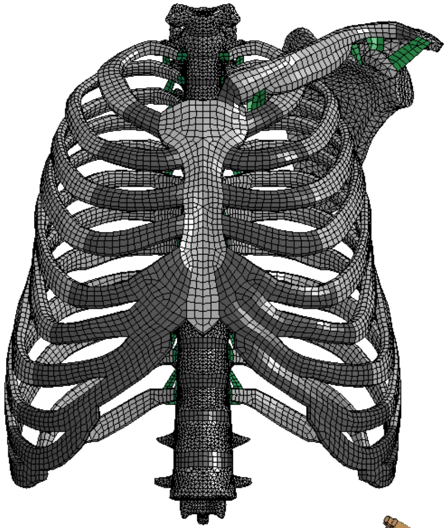


THUMS Model Version 4 (since end 2010)

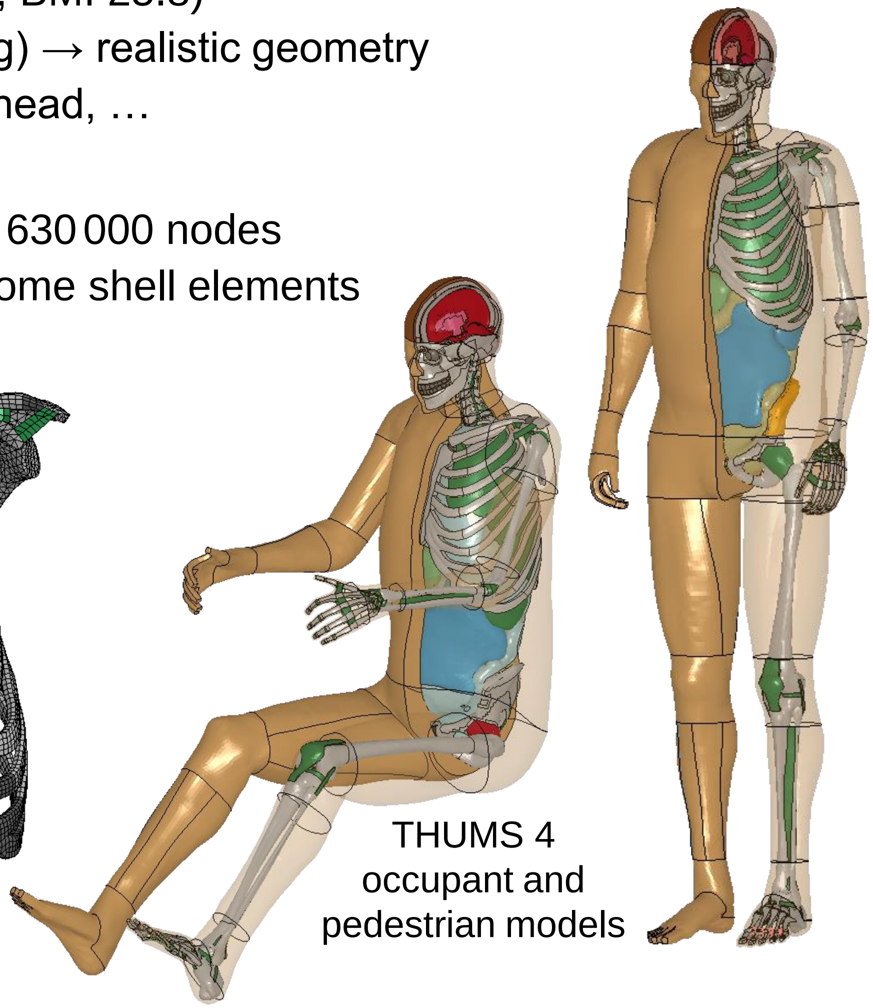
- Geometry obtained from medical CT scans
  - Basis: 39 year-old male (173cm, 77.3kg, BMI 25.8)
  - Scaled to AM50 model (178.6cm, 74.3kg) → realistic geometry
  - High detailing of joints, internal organs, head, …
- Model parameters
  - Element size 3-5mm, 1.8 Mio elements, 630 000 nodes
  - Mainly hexahedrons/tetrahedrons and some shell elements



occupant upper body

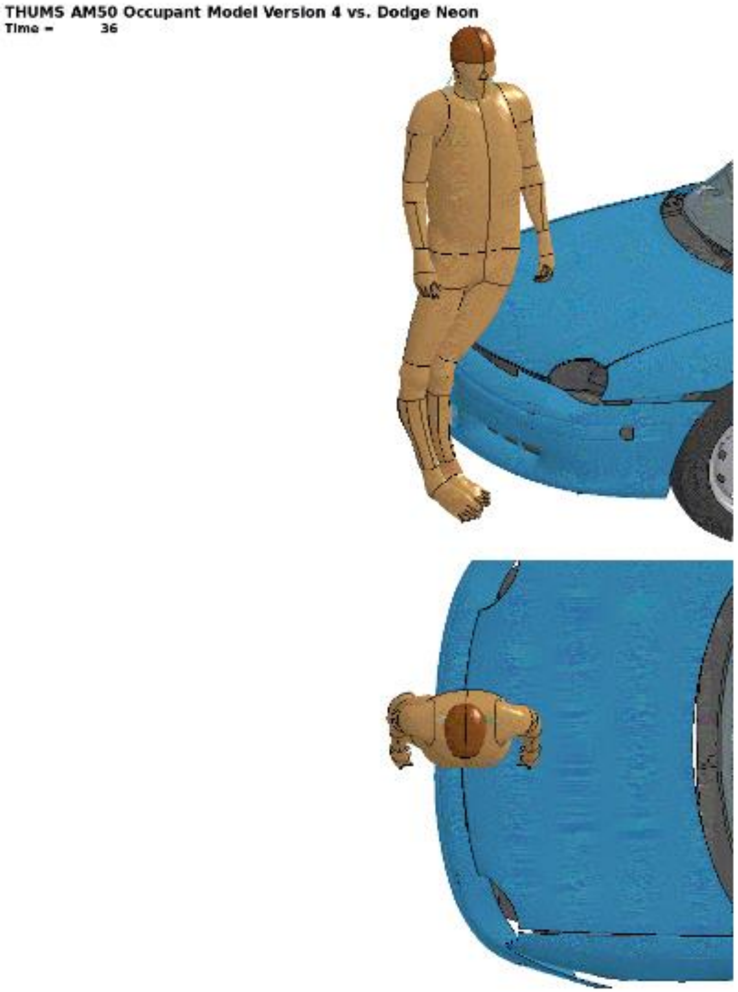


pedestrian thorax

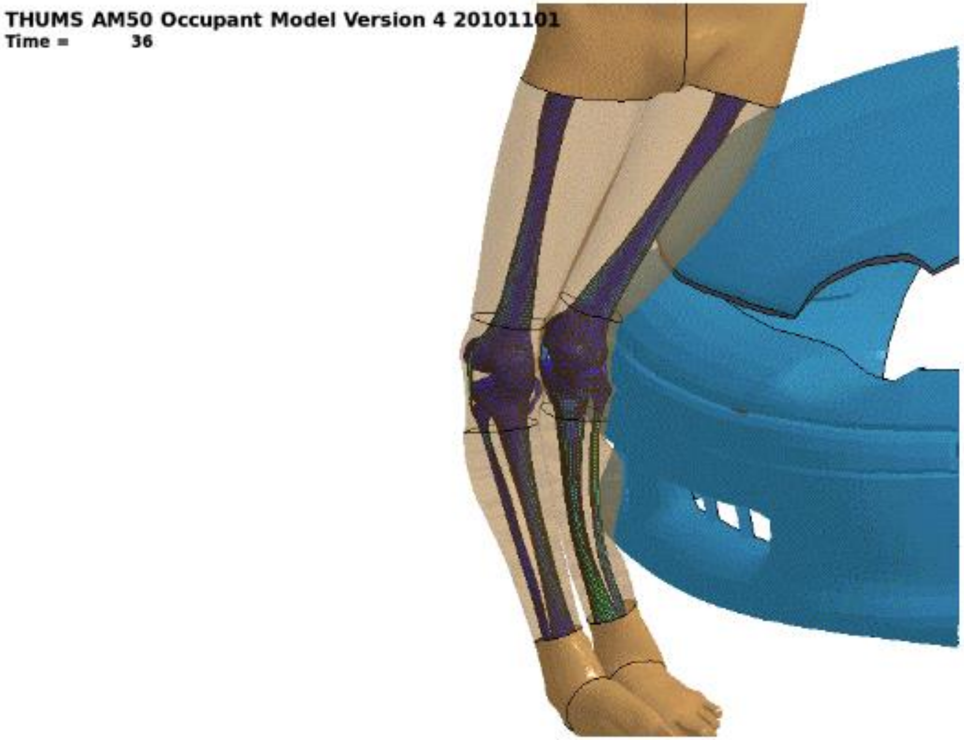


THUMS 4 occupant and pedestrian models

- Pedestrian frontal impact with THUMS V4

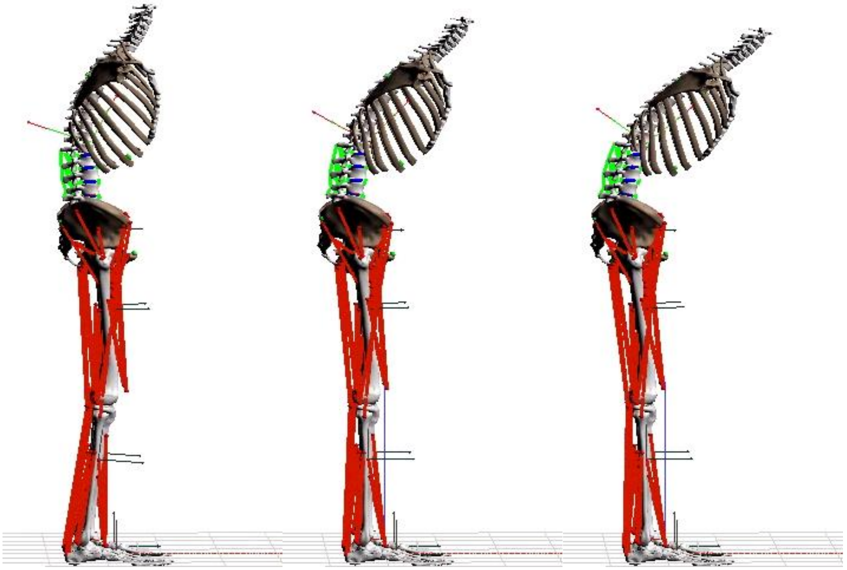


Internal loads during impact



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

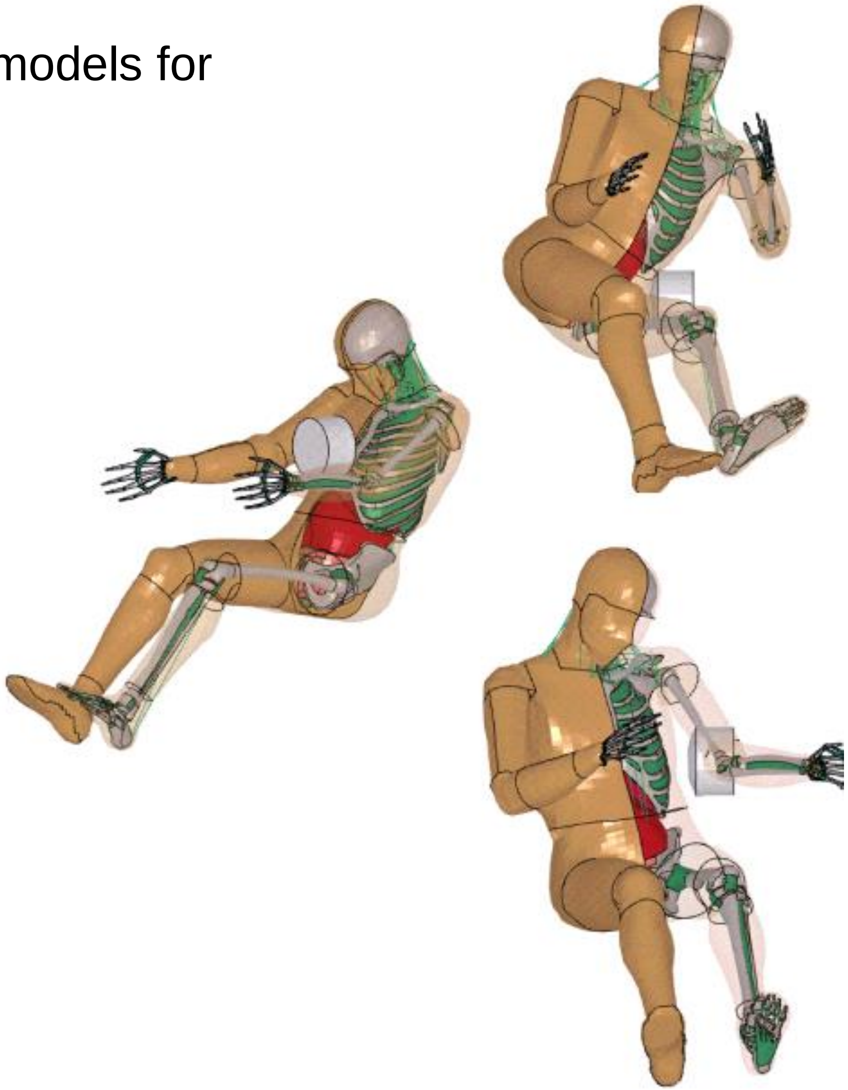
  - Forward kinematics
    - muscle forces are measured and prescribed
    - motion is computed



[courtesy of Prof. Syn Schmitt]

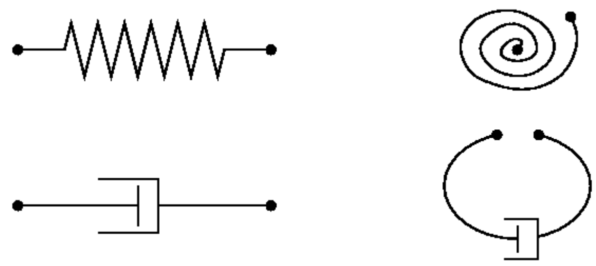
Material Models

- 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
  - *ELEMENT_DISCRETE & *SECTION_DISCRETE
    - Springs
    - Dampers



- Available material models

*MAT_S01:	*MAT_SPRING_ELASTIC
*MAT_S02:	*MAT_DAMPER_VISCOUS
*MAT_S03:	*MAT_SPRING_ELASTOPLASTIC
*MAT_S04:	*MAT_SPRING_NONLINEAR_ELASTIC
*MAT_S05:	*MAT_DAMPER_NONLINEAR_VISCOUS
*MAT_S06:	*MAT_SPRING_GENERAL_NONLINEAR
*MAT_S07:	*MAT_SPRING_MAXWELL
*MAT_S08:	*MAT_SPRING_INELASTIC
*MAT_S13:	*MAT_SPRING_TRILINEAR_DEGRADING
*MAT_S14:	*MAT_SPRING_SQUAT_SHEARWALL
*MAT_S15:	*MAT_SPRING_MUSCLE

tendons, fixators, bracelets, other passive structures

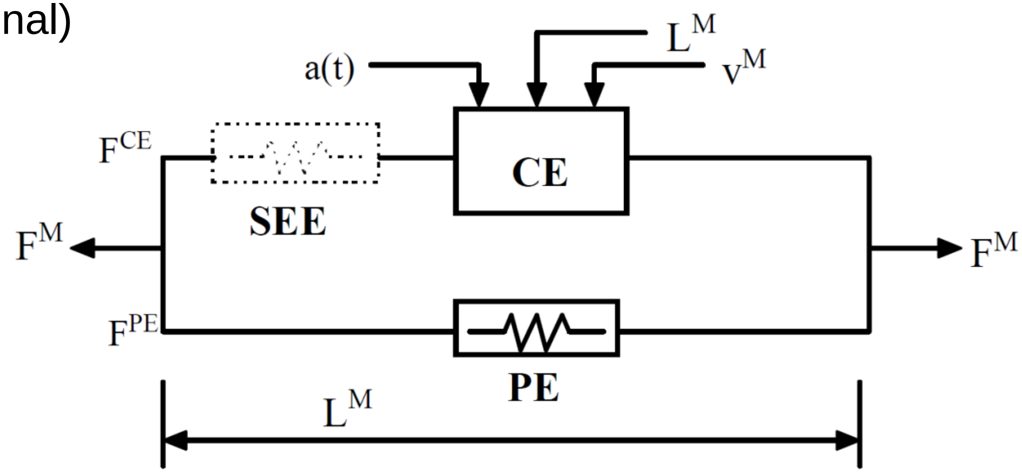
muscles (active & passive)

*MAT_SPRING_MUSCLE (*MAT_S15)

- Hill-type muscle model
- Based on *Hill* 1938, *Zajak* 1989, *Winters* 1990
- Rheological model
  - CE: contractile element
  - PE: parallel elastic element
  - SEE: serial elastic element (optional)
- Muscle force computation

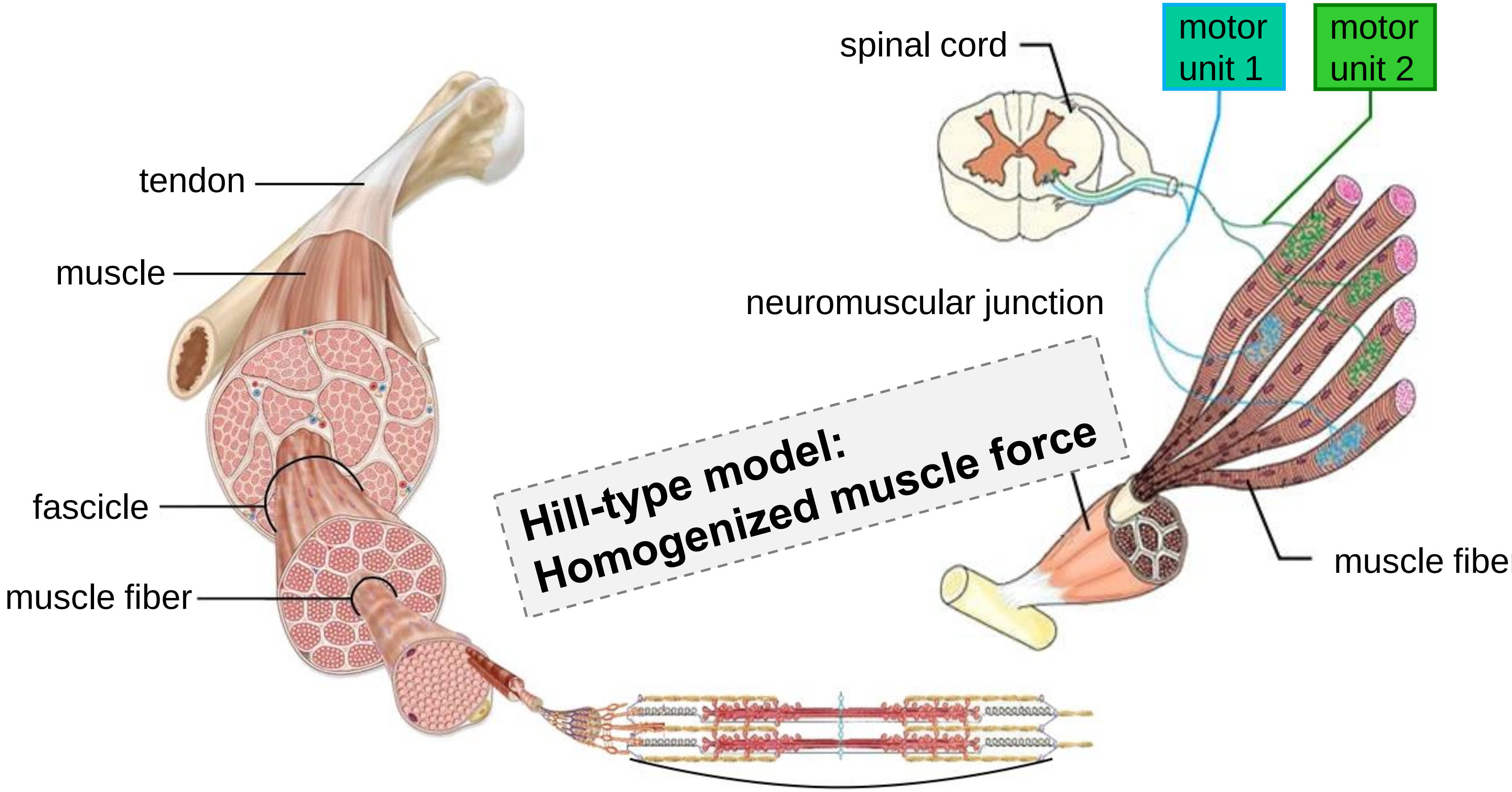
$$F^{M} = F^{PE} + F^{CE}$$

$$F^{SEE} = F^{CE}$$




```
*MAT_SPRING_MUSCLE
$---+----1----+----2----+----3----+----4----+----5----+----6----+----7----+----8
$#      mid         lo      vmax        sv         a      fmax        tl        tv
          1      0.000     0.000  1.000000     0.000     0.000  1.000000  1.000000
$#      fpe       lmax       ksh
      0.000      0.000     0.000
```

- 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]

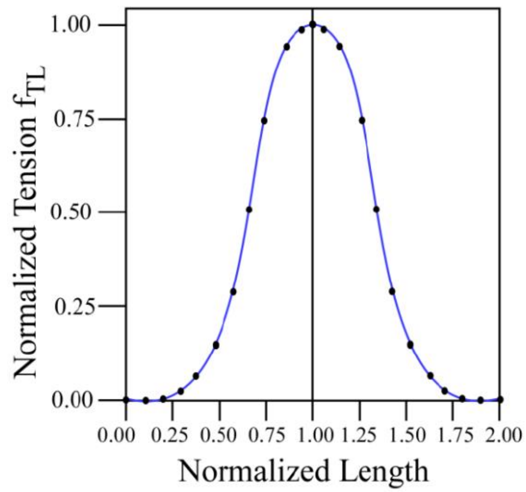
*MAT_SPRING_MUSCLE (*MAT_S15)

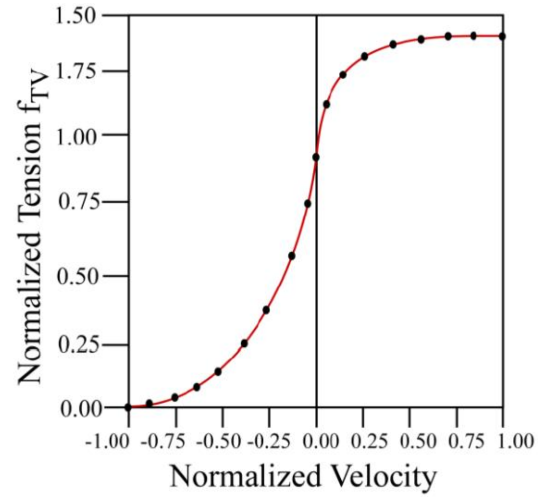
Contractile element CE

$$F^{CE} = a(t)\, F_{\max}\, f_{TL}(L)\, f_{TV}(V)$$

$$L = \frac{L^M}{L_0}$$

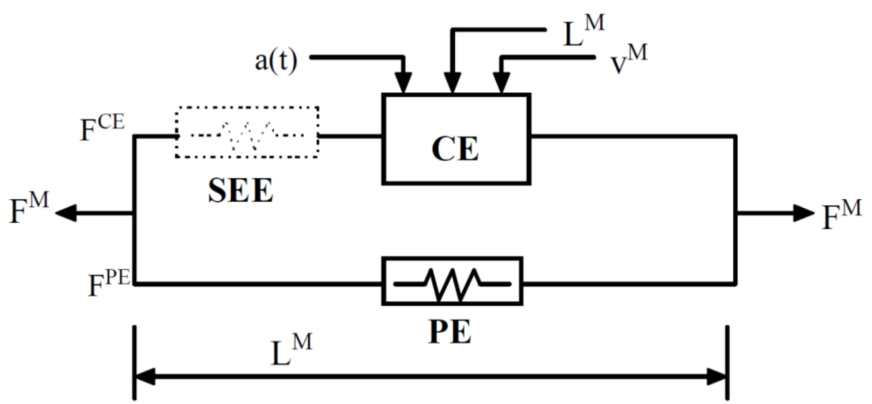
$$V = \frac{V^M}{V_{\max}\, S_V(a(t))}$$





- 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

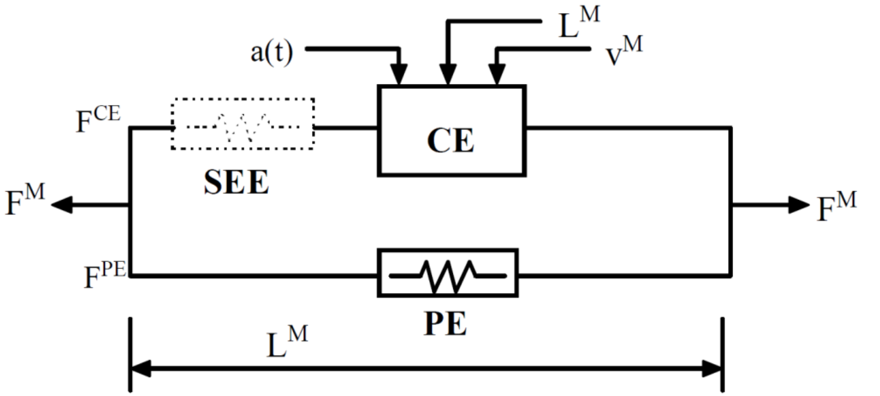


*MAT_SPRING_MUSCLE (*MAT_S15)

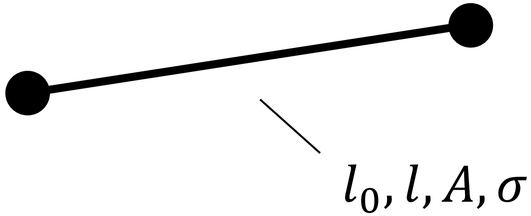
- Passive Element PE
  - FPE Force vs. length function for PEE
    - <0: absolute value gives load curve ID
    - >0: constant value of 0.0 is used
    - =0: exponential function is used

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

- 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

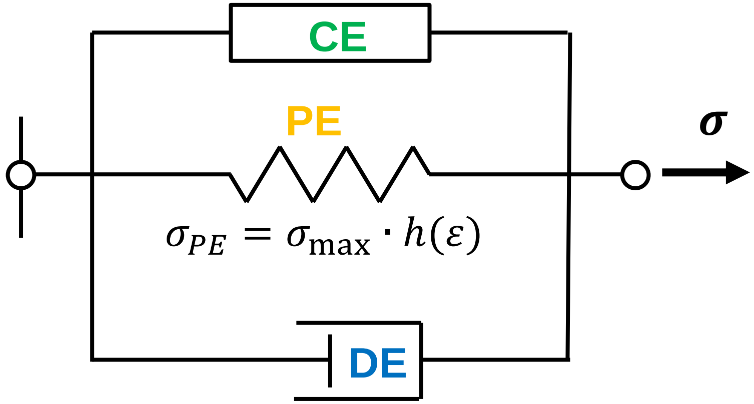


  - Beam type 3: Truss element with 6 material models
    - *MAT_001: Elastic
    - *MAT_003: Elastic-plastic
    - *MAT_004: Elastic-plastic thermal
    - *MAT_027: Mooney-Rivlin rubber
    - *MAT_098: Simplified Johnson-Cook
    - *MAT_156: Hill's muscle model
  - Beam type 6: Discrete beam/cable with 1 material model
    - *MAT_071: Non-linear elastic

*MAT_MUSCLE (*MAT_156)

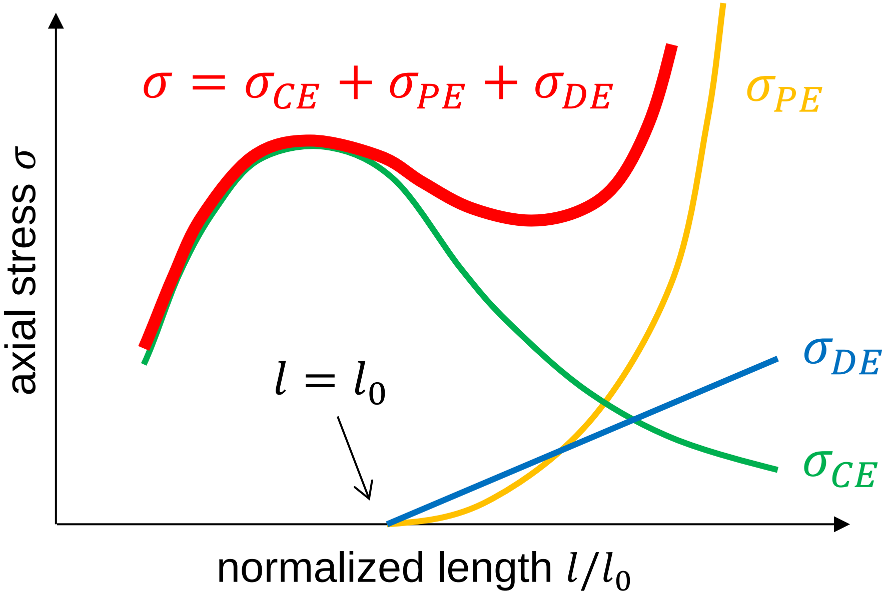
- Based on Hill-type muscle model *MAT_SPRING_MUSCLE
- Available for discrete and truss beam elements
- Extended by damper element (DE)
- Rheological model
  - CE: force generation by the muscle
  - PE: energy storage from muscle elasticity
  - DE: muscular viscosity
- Muscle force computation

$$\sigma_{CE} = \sigma_{\max} \cdot a(t) \cdot f(\varepsilon) \cdot g(\dot{\varepsilon})$$



$$\sigma_{PE} = \sigma_{\max} \cdot h(\varepsilon)$$

$$\sigma_{DE} = D \cdot \varepsilon \cdot \dot{\varepsilon}$$



*MAT_MUSCLE (*MAT_156)

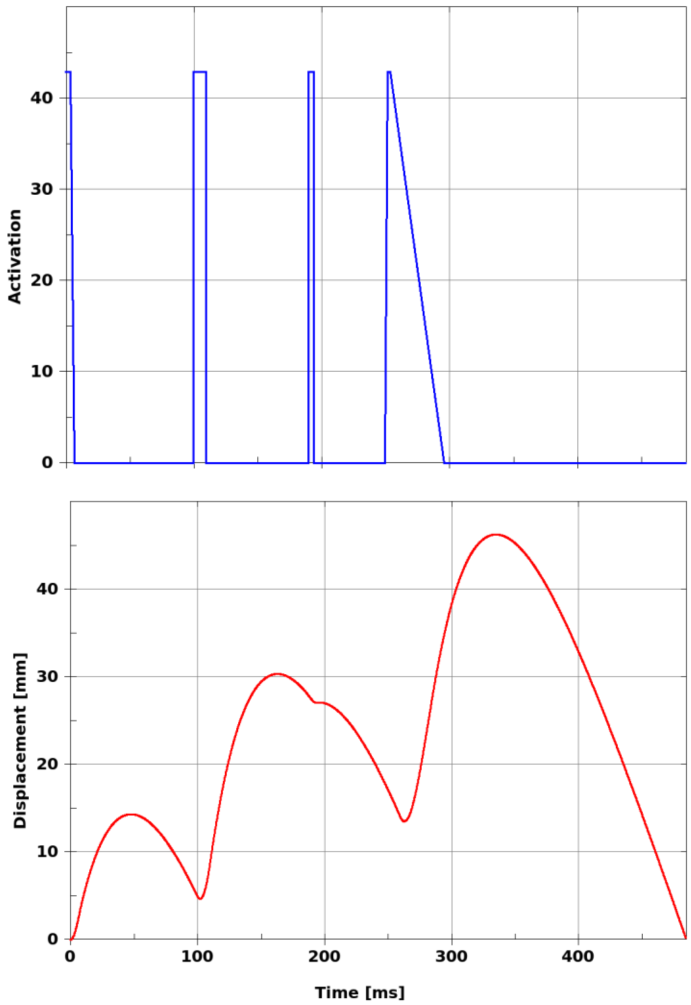
```
*MAT_MUSCLE
$---+----1----+----2----+----3----+----4----+----5----+----6----+----7----+----8
$#      MID        RO       SNO       SRM       PIS       SSM       CER       DMP
          1   1.05e-6       1.0       2.0     0.003       1.0       2.0       1.0
$#      ALM       SFR       SVS       SVR       SSP
      -43.0       1.0     -46.0     -47.0     -48.0
```

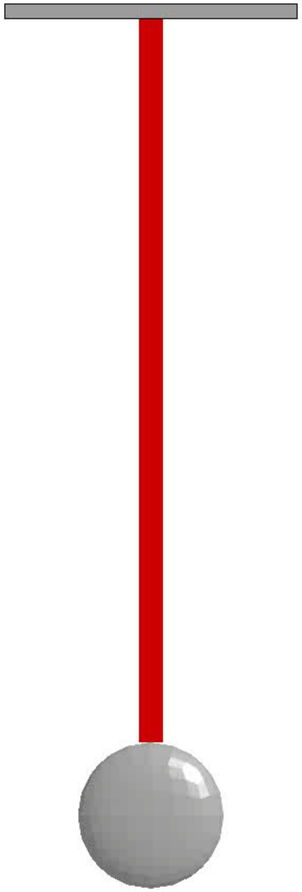
- RO: Density
- SNO: Initial stretch ratio l_0/l_{orig} (nodal distance / original length)
- SRM: Maximum strain rate
- PIS: Peak isometric stress
- SSM: Strain when the stress in SSP is maximal
- CER: Exponential rise of SSP (if SSP=0)
- DMP: Damping constant
- ALM*: Activation level vs. time
- SFR*: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

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

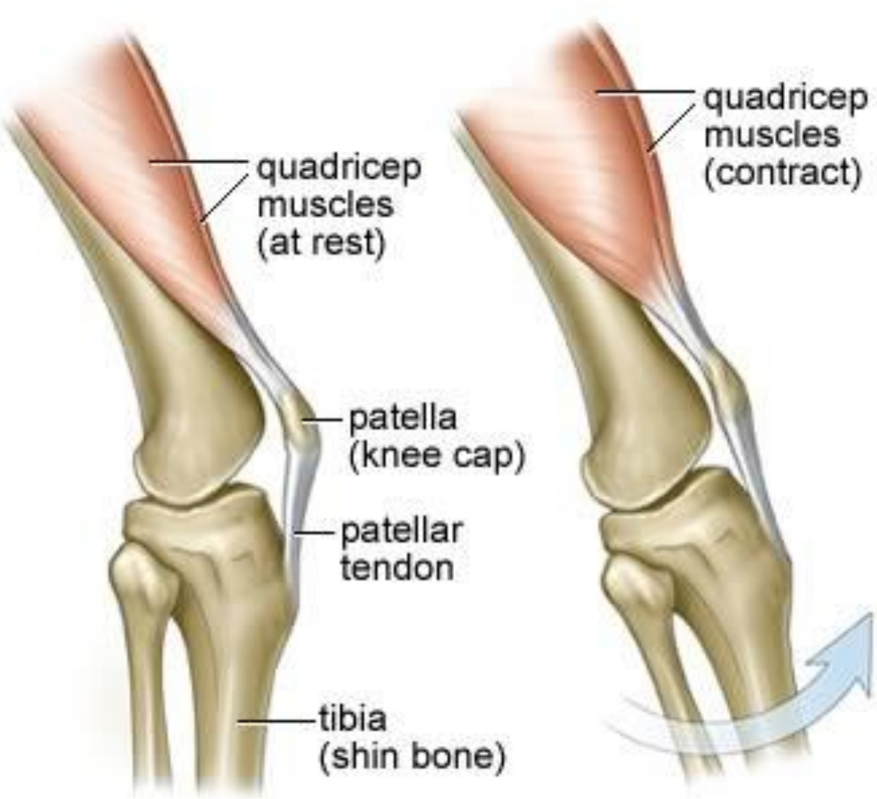
*MAT_MUSCLE (*MAT_156)

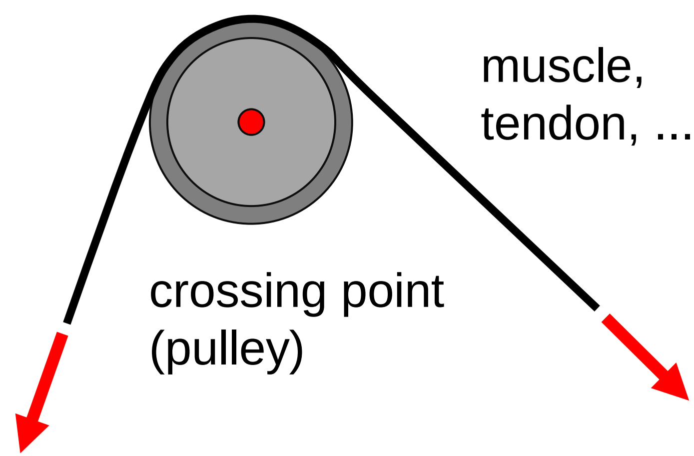
- Example of an activation (Prof. Syn Schmitt & Julian Blaschke, INSPO, Uni-Stuttgart)





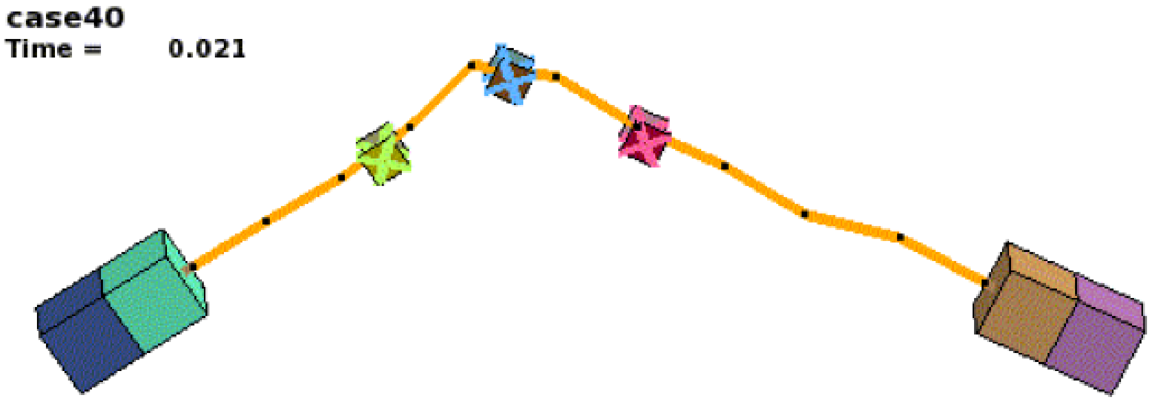
- Problem of deflecting forces
  - FEM model of bent muscles or tendons which are guided by bones around joints
  - Examples: Ankle, elbow, knee,



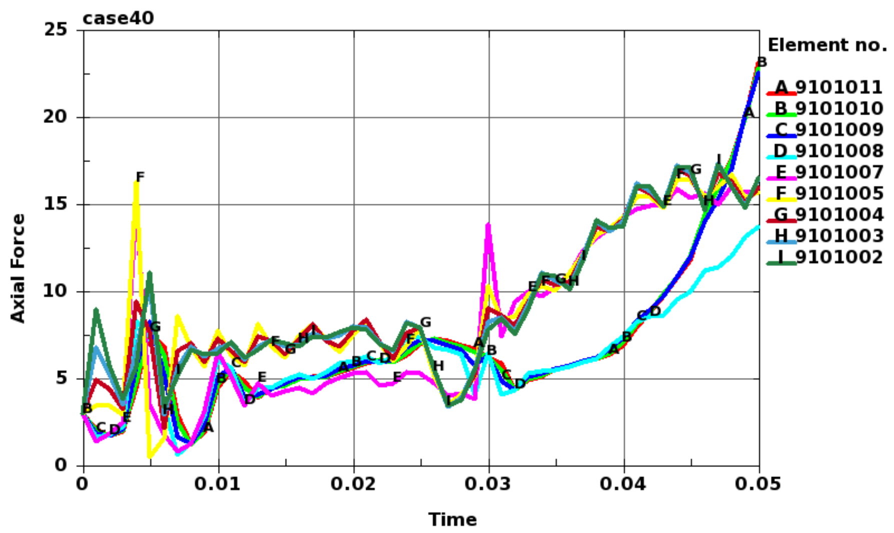


➡ Quantities of interest: friction, angle, forces, …

- Classical modeling technique
  - Truss elements with *MAT_MUSCLE and *CONTACT_GUIDED_CABLE



- Problems
  - Non-smooth contact
  - Non-uniform axial forces

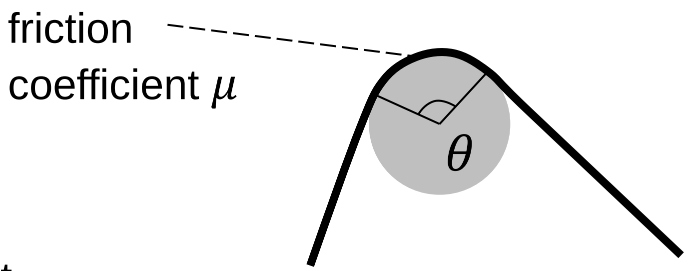


- Keyword *ELEMENT_BEAM_PULLEY [*Erhart* 2012]
  - Pulleys allow continuous sliding of truss elements through a sharp change of angle
  - Available for *MAT_ELASTIC, *MAT_CABLE_DISCRETE_BEAM , *MAT_MUSCLE

```
*ELEMENT_BEAM_PULLEY
$---+----1----+----2----+----3----+----4----+----5----+----6----+----7----+----8
$#     PUID      BID1      BID2      PNID        FD        FS      LMIN        DC
         42         1         2         1       0.2       0.3       0.1       1.0
```

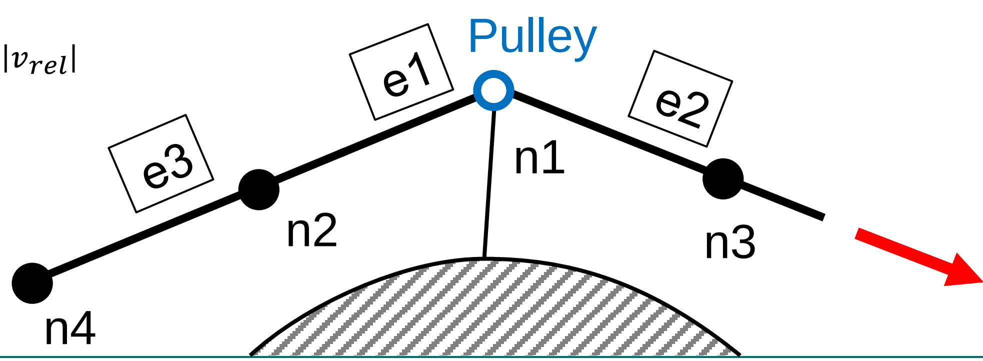
PUID	Pulley ID.
BID1	Truss beam element 1 ID.
BID2	Truss beam element 2 ID.
PNID	Pulley node, NID.
FD	Coulomb dynamic friction coefficient.
FS	Optional Coulomb static friction coefficient.
LMIN	Minimum length.
DC	Decay constant

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

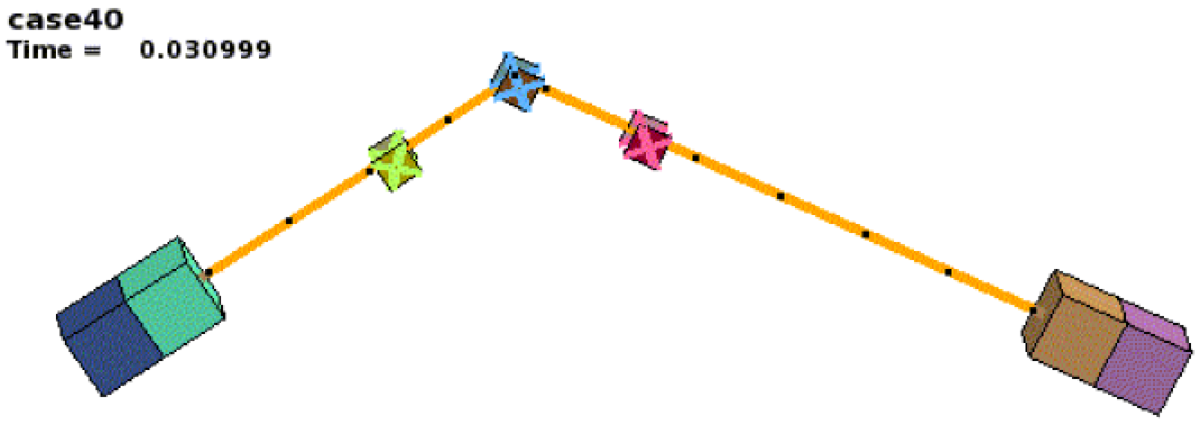


slip condition: $T_2 \leq T_1 e^{\mu\theta}$

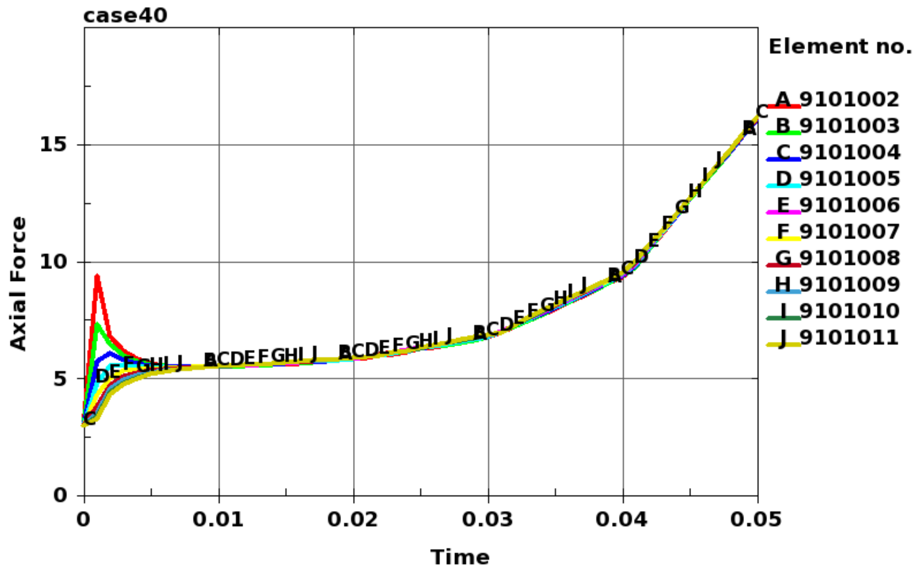
Euler-Eytelwein Equation



- 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

passive isotropic

*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

$$\sigma_\varepsilon(\varepsilon) = \sum_{i=1}^{k} C_i \varepsilon^i \qquad \sigma_V(t) = \int_0^t G(t-\tau)\frac{\partial \sigma_\varepsilon[\varepsilon(\tau)]}{\partial \varepsilon}\frac{\partial \varepsilon}{\partial \tau}d\tau \qquad G(t) = \sum_{i=1}^{n} G_i e^{-\beta t}$$

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

$$\sigma(\varepsilon,t) = \sigma_{SR}(\varepsilon) + \sigma_V(t)$$

$$\sigma_V(t) = \int_0^t G(t-\tau)\frac{\partial \varepsilon}{\partial \tau}d\tau$$

passive isotropic

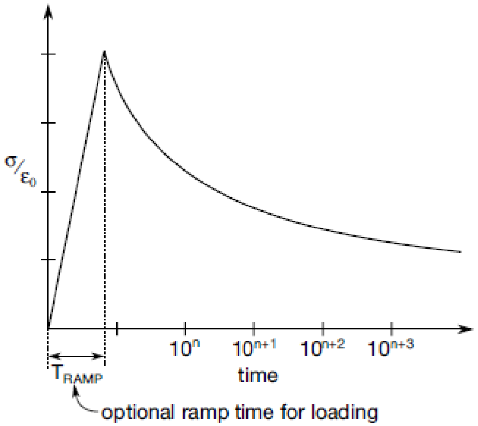
- *MAT_LUNG_TISSUE
  - Hyperelastic model for heart tissue *Vawter* 1980
    - Isochoric and volumetric strain-energy function

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

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

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

$$\sigma_{ij} = \int_0^t g_{ijkl}(t-\tau) \frac{\partial \varepsilon_{kl}}{\partial \tau} d\tau \qquad g(t) = \sum_{i=1}^{n} G_i e^{-\beta_i t}$$



    - Optionally prescribed relaxation of *MAT_GENERAL_VISCOELASTIC via *DEFINE_CURVE

passive isotropic

*MAT_BRAIN_LINEAR_VISCOELASTIC

- Simple material model for solid elements only

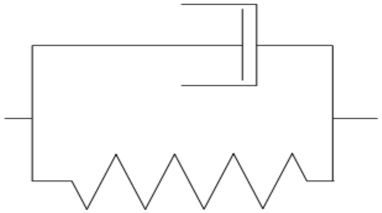
```
*MAT_BRAIN_LINEAR_VISCOELASTIC
$---+----1----+----2----+----3----+----4----+----5----+----6----+----7----+----8
$#      MID        RO      BULK        G0        GI        DC        FO        SO
         42      1.05      23.0       2.1       0.2       1.3       1.0
```

- *Jaumann* rate formulation for deviatoric stress rate

$$\overset{\nabla}{\sigma}_{ij} = 2\int_0^t G(t-\tau)D'_{ij}(\tau)dt$$

- 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

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

passive anisotropic

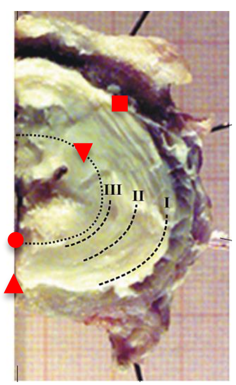
*MAT_SOFT_TISSUE

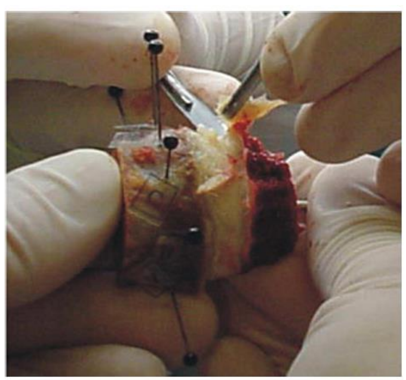
- Element types: Solids and shells (*Belytschko-Tsay* ELTYPE=2)
- Suitable for ligaments, tendons, fascia
- Hyperelastic material law [*Weiss, Maker & Govindjee* 1996, *Puso & Weiss* 1998]
  - Isotropic Mooney-Rivlin matrix
  - Collagen fiber reinforcements (transversely isotropic)
  - Simple compression law

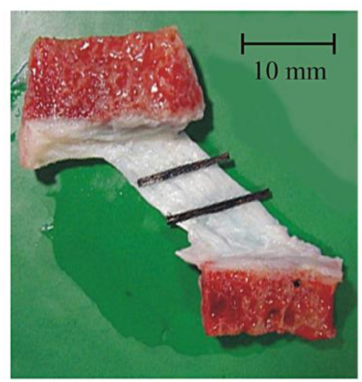
$$W = C_1(\tilde{I}_1 - 3) + C_2(\tilde{I}_2 - 3) + F(\lambda) + \frac{1}{2}K[\ln(J)]^2$$

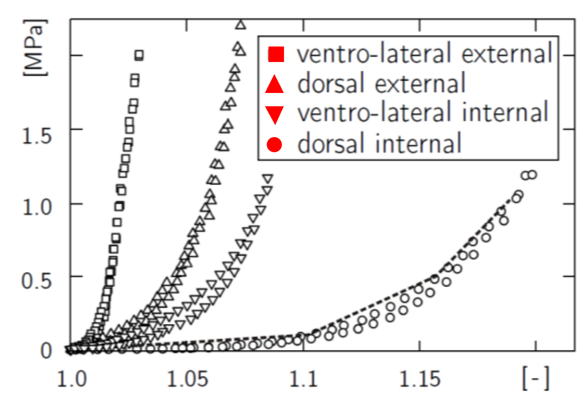
$$\frac{\partial F}{\partial \lambda} = \begin{cases} 0 & \lambda < 1 \\ \frac{C_3}{\lambda}\left[\exp\left(C_4(\lambda - 1)\right) - 1\right] & \lambda < \lambda^* \\ \frac{1}{\lambda}(C_5\lambda + C_6) & \lambda \geq \lambda^* \end{cases}$$

- 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

```
*MAT_SOFT_TISSUE
$#     mid        ro        c1        c2        c3        c4        c5
        42      1.05      23.0       2.1       0.2       1.3       1.0
$#      xk      xlam      fang     xlam0    failsf    failsm   failshr
     0.000     0.000     0.000     0.000     0.000     0.000     0.000
$#    aopt        ax        ay        az        bx        by        bz
     0.000     0.000     0.000     0.000     0.000     0.000     0.000
$#     la1       la2       la3      macf
     0.000     0.000     0.000         1
```

- C1-C5: Stress parameters
- XK: Compression modulus
- XLAM: Stretch ratio λ^* 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

passive anisotropic

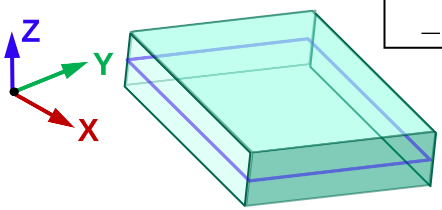
*MAT_SOFT_TISSUE

```
*MAT_SOFT_TISSUE
$#     mid        ro        c1        c2        c3        c4        c5
        42      1.05      23.0       2.1       0.2       1.3       1.0
$#      xk      xlam      fang    xlam0    failsf    failsm   failshr
     0.000     0.000     0.000     0.000     0.000     0.000     0.000
$#    aopt        ax        ay        az        bx        by        bz
     0.000     0.000     0.000     0.000     0.000     0.000     0.000
$#     la1       la2       la3      macf
     0.000     0.000     0.000         1
```

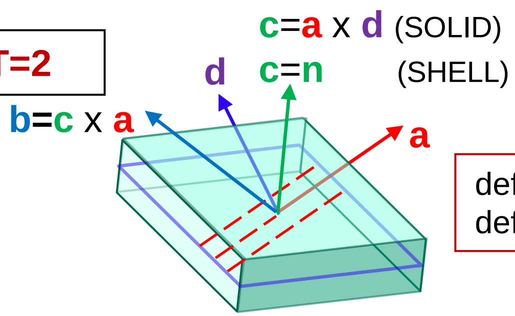
- Same logic as for composite materials

AOPT<0

*DEFINE_COORDINATE…
_NODES/_SYSTEM/_VECTOR



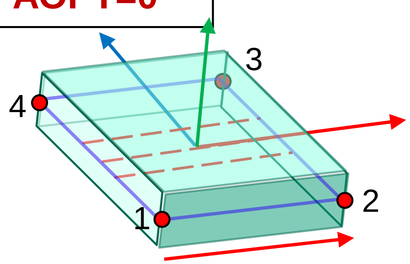
AOPT=2



define a (SHELL)
define a & d (SOLID)

AOPT=0

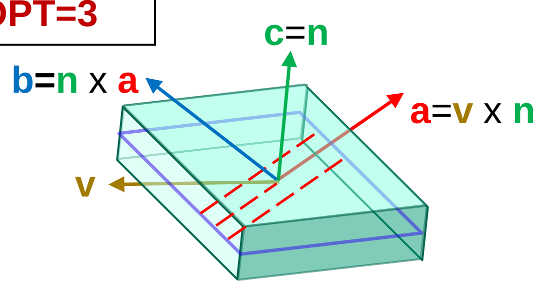
a-direction defined based on element coordinate system



can be changed with:

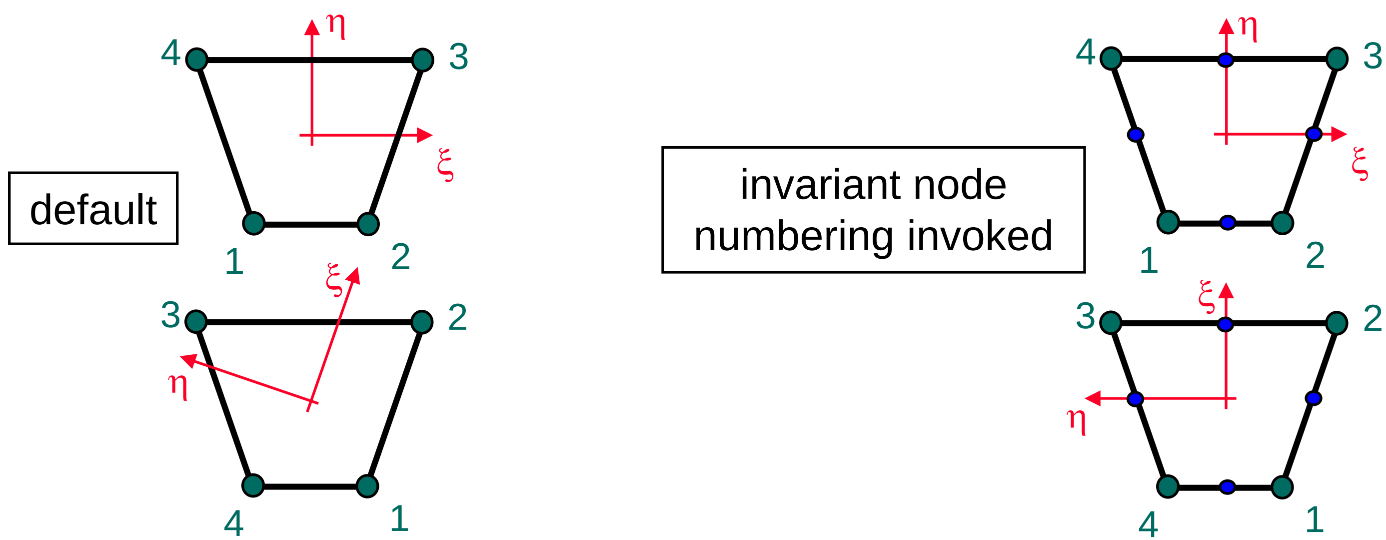
*ELEMENT_SHELL_BETA
*ELEMENT_SOLID_ORTHO

AOPT=3



define v

- 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)



passive anisotropic

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

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

```
*MAT_SOFT_TISSUE_VISCO
$#     mid        ro        c1        c2        c3        c4        c5
        42      1.05      23.0       2.1       0.2       1.3       1.0
$#      xk      xlam      fang     xlam0
     0.000     0.000     0.000     0.000
$#    aopt        ax        ay        az        bx        by        bz
     0.000     0.000     0.000     0.000     0.000     0.000     0.000
$#     la1       la2       la3      macf
     0.000     0.000     0.000         1
$#      s1        s2        s3        s4        s5        s6
     0.000     0.000     0.000     0.000     0.000     0.000
$#      t1        t2        t3        t4        t5        t6
     0.000     0.000     0.000     0.000     0.000     0.000
```

- 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-Lagrange*an 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

$$\mathbf{S} = \frac{\partial W}{\partial \mathbf{E}} - p\,J\,\mathbf{C}^{-1} + T_0\{t,\, Ca_0,\, l\} \qquad \boldsymbol{\sigma} = \frac{1}{J}\,\mathbf{F}\,\mathbf{S}\,\mathbf{F}^T$$

  - Active fiber stress component is defined by time-varying elastance model

$$T_0 = T_{\max} \frac{Ca_0^2}{Ca_0^2 + ECa_{50}^2} C_t$$

    - $T_{\max}$: maximum isometric tension achieved at the longest sarcomere length
    - ECa_{50}: Length-dependent calcium sensitivity
    - C_t: activation function

active anisotropic

active anisotropic

*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

$$\mathbf{S} = \underbrace{\kappa J(J-1)\mathbf{C}^{-1} + \mu J^{-2/3}\mathbf{DEV}\left[\frac{1}{4}(\mathbf{I}-\bar{\mathbf{C}}^{-2})\right]}_{\text{Neo-Hooke model}} + \underbrace{J^{-2/3}\sum_{i=1}^{n}[\sigma_i(\lambda_i)+\varepsilon_i(\lambda_i)]\mathbf{DEV}[\mathbf{K}_i]}_{\text{Fiber contribution}}$$

Neo-*Hooke* model — Fiber contribution

  - Deviatoric part of a tensor $\mathbf{DEV}[\bullet] = (\bullet) - \frac{1}{3}\mathrm{tr}[(\bullet)\mathbf{C}]\mathbf{C}^{-1}$

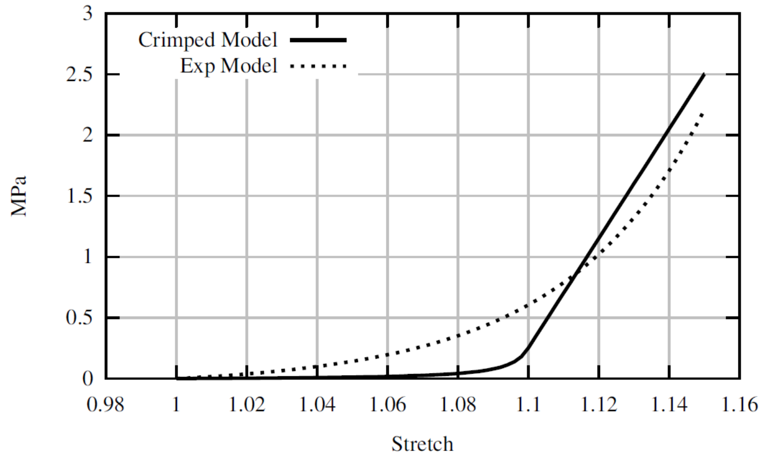
- Passive fiber contribution
  - Crimped model by *Freed et al.* 2005

$$\lambda < \Lambda \qquad \sigma = \xi E_s(\lambda - 1)$$

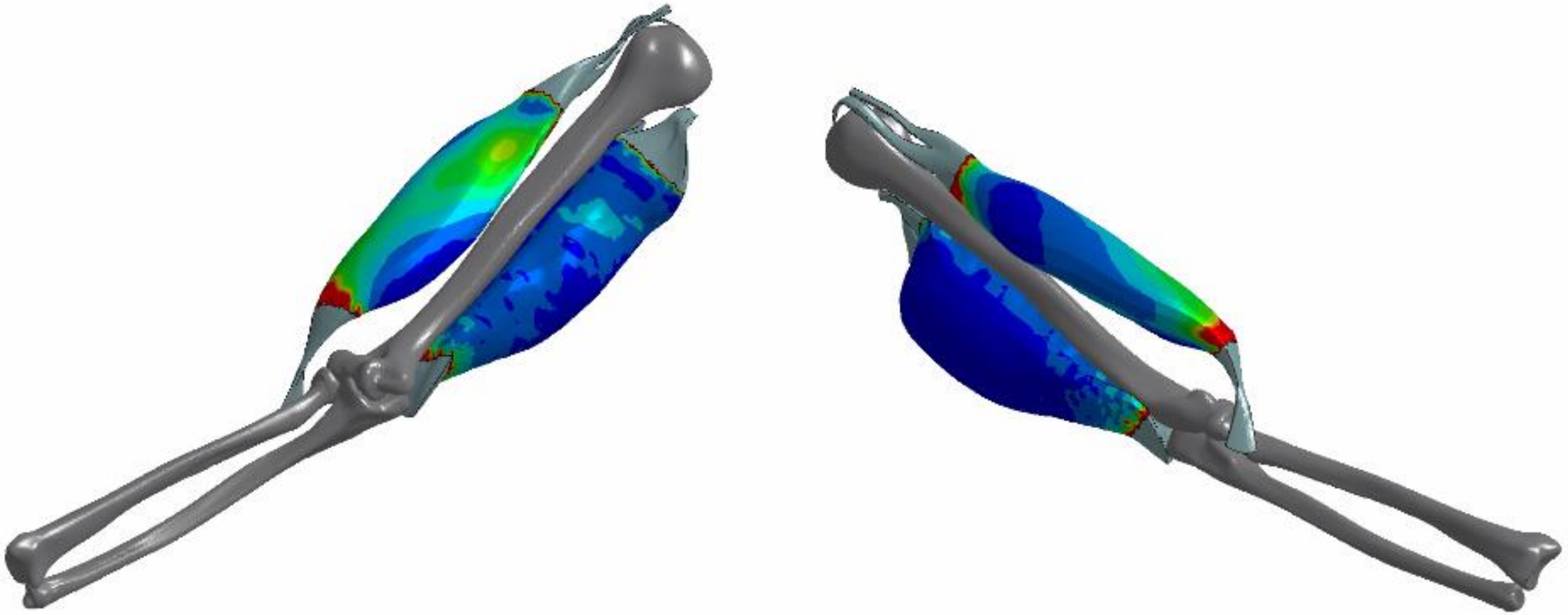
$$\lambda > \Lambda \qquad \sigma = E_s(\lambda - 1) + E_f(\lambda - \Lambda)$$

  - Exponential model

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

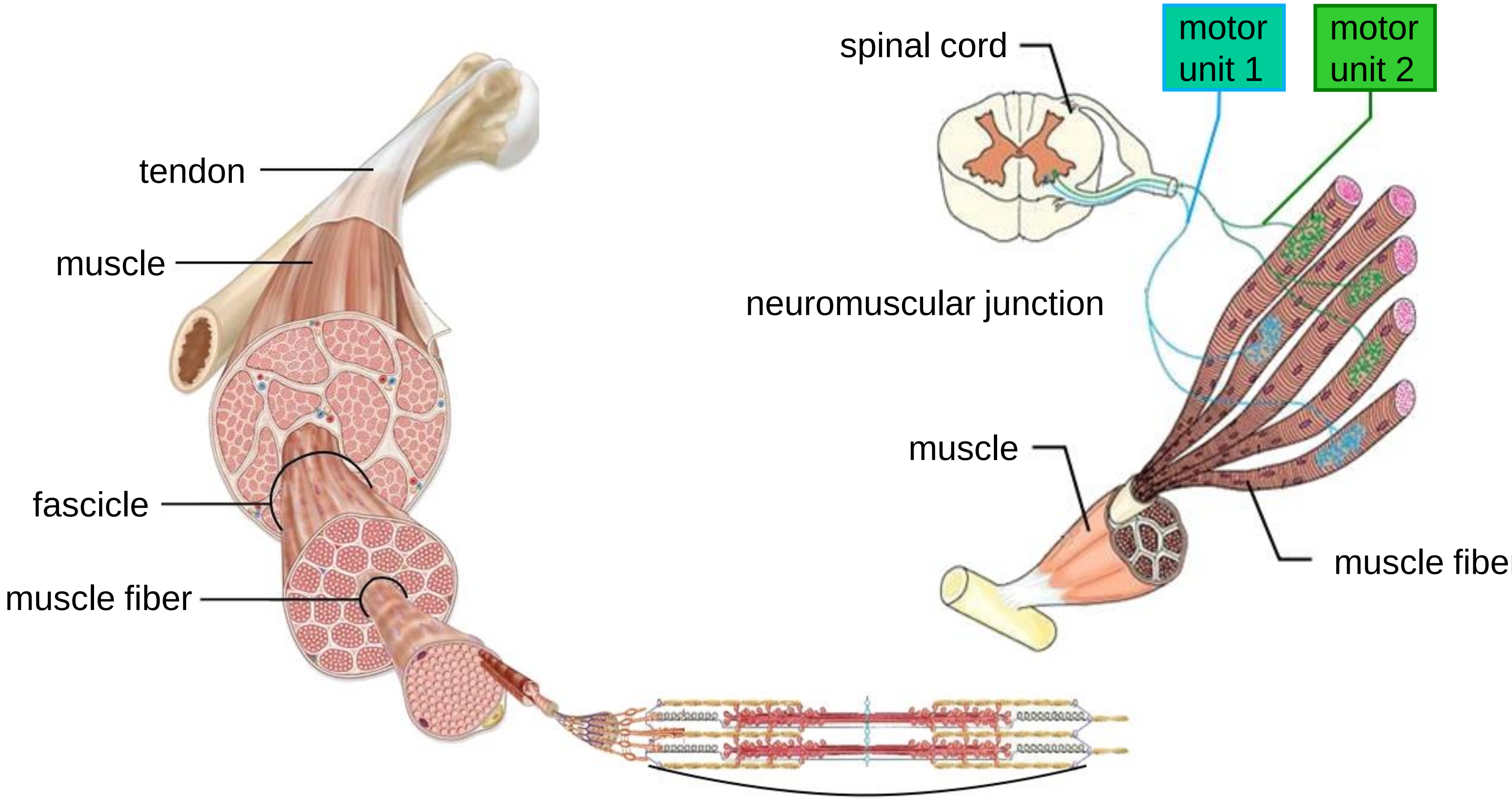


- *MAT_TISSUE_DISPERSED
  - Example of an arm with passive muscle material
    - Geometry provided by Prof. O. Röhrle, Uni-Stuttgart, Fraunhofer IPA
    - Bones modeled as rigid bodies
    - Prescribed motion of the forearm
    - Here: Uniform fiber direction



  - Note: Location dependent fiber orientation via diffusion tensor MRI of the muscle

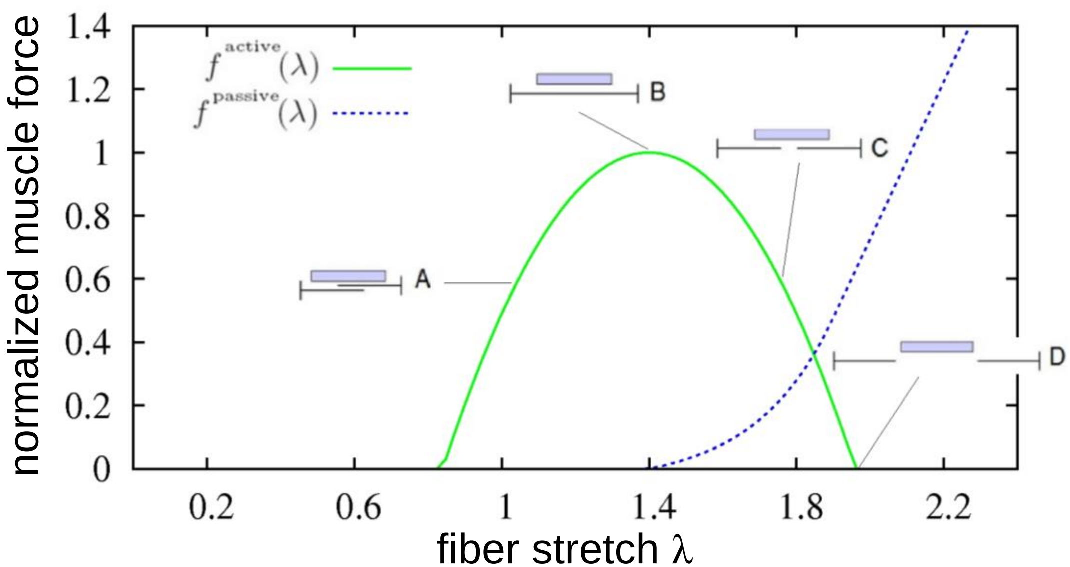
- Recall the background information
  - Skeletal muscles and their activation
    - Sarcomeres are the contractile or functional unit of the muscle

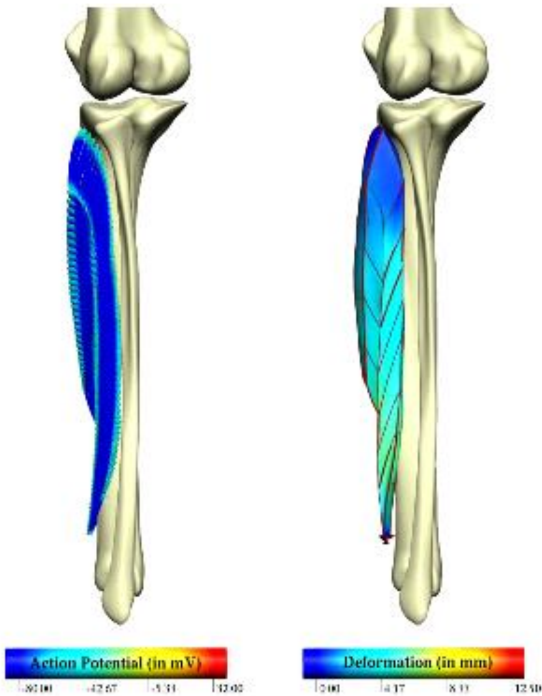


[Graphs are courtesy of Benjamin Cummings]

- Background information
  - Skeletal muscles and their activation
    - Muscle force depends on the sarcomere length

$$\mathbf{S}_{\text{muscle}} = \underbrace{\mathbf{S}_{\text{iso}} + \mathbf{S}_{\text{aniso}}}_{\text{passive part}} + \underbrace{\mathbf{S}_{\text{tension}}}_{\text{active part}}$$





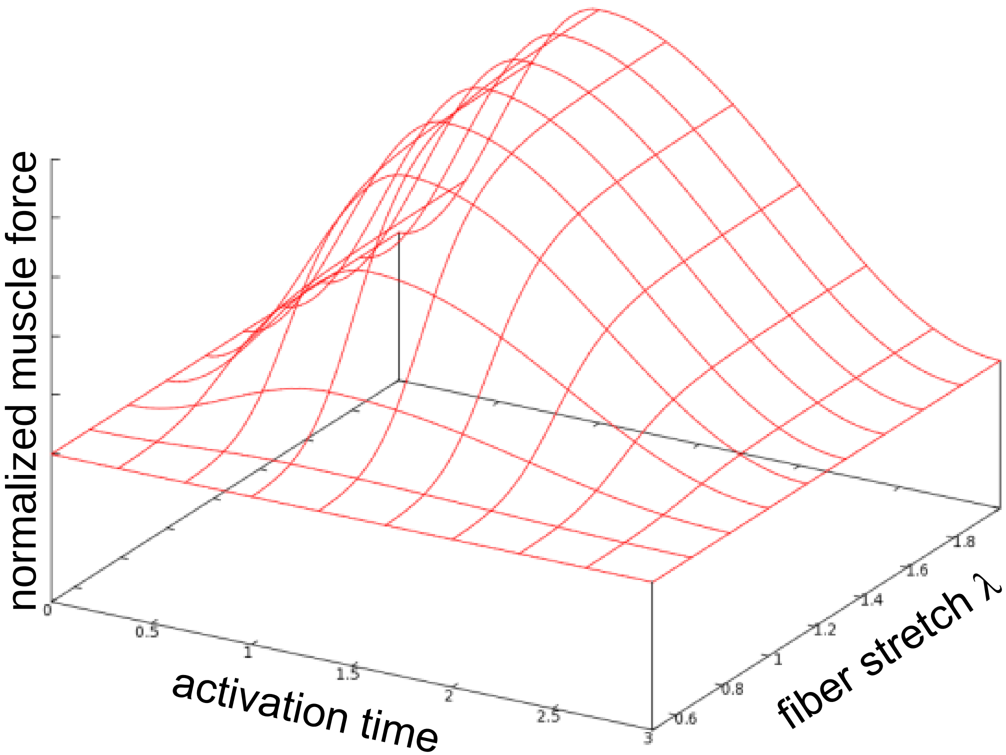
Courtesy of Prof. O. Röhrle, Uni-Stuttgart, Fraunhofer IPA

active anisotropic

*MAT_TISSUE_DISPERSED

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

$$\sigma = T_{\max} \frac{Ca_0^2}{Ca_0^2 + ECa_{50}^2} C(t) \qquad ECa_{50} = \frac{(Ca_0)_{\max}}{\sqrt{e^{B\left(l_r\sqrt{2(\lambda-1)+1}-l_0\right)-1}}}$$



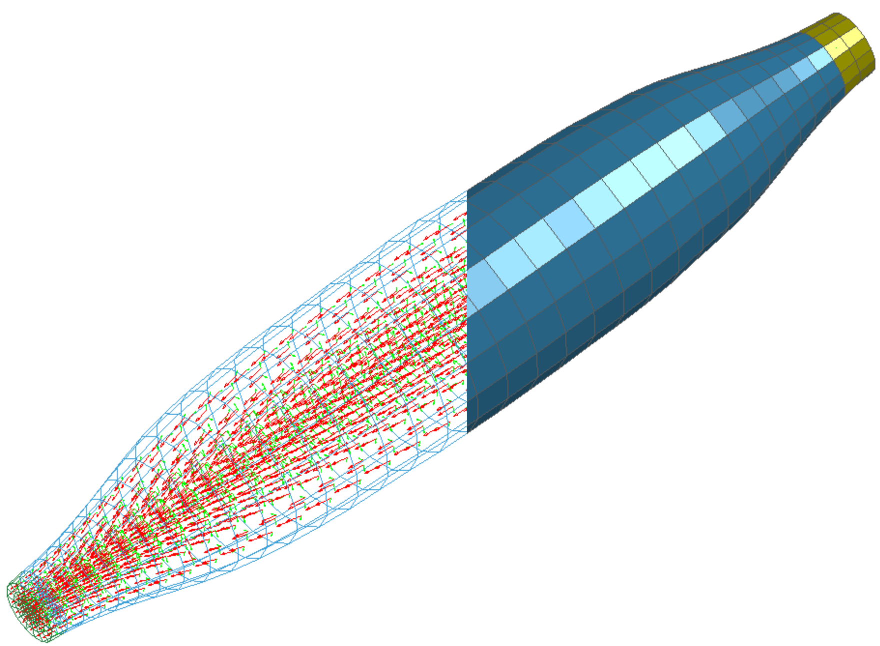
- Activation function

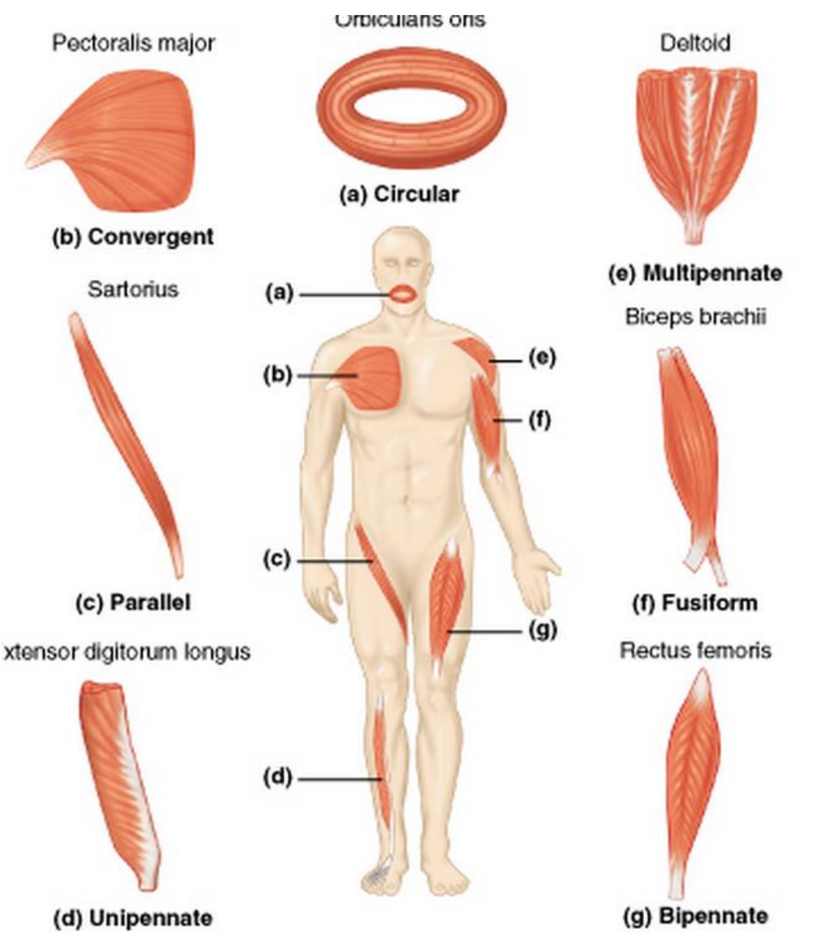
$$C(t) = \frac{1}{2}\left(1 - \cos\omega(t)\right)$$

$$\omega = \begin{cases} \pi \dfrac{t}{t_0} & 0 \le t < t_0 \\ \pi \dfrac{t - t_0 + t_r}{t_r} & t_0 \le t < t_0 + t_r \\ 0 & t_0 + t_r \le t \end{cases}$$

active anisotropic

- *MAT_TISSUE_DISPERSED
  - Example of an activated fusiform muscle
  - Definition of the fiber alignment
    - Local coordinates (AOPT = 0)

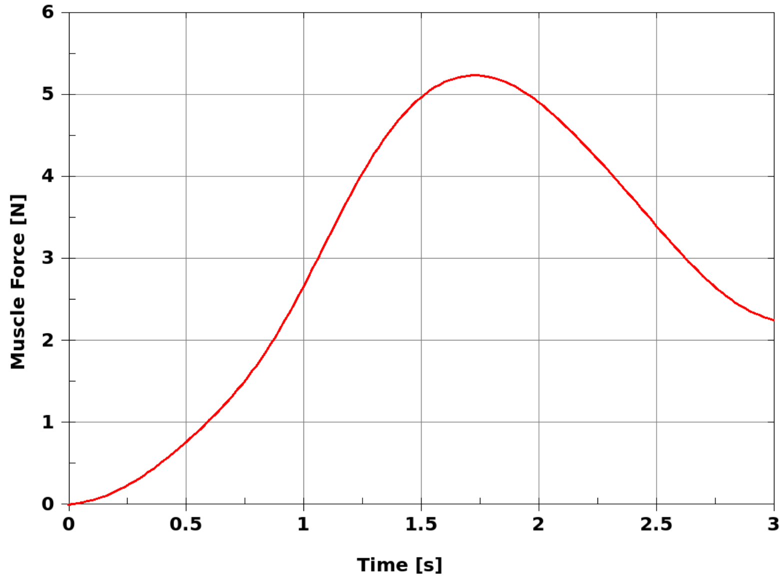




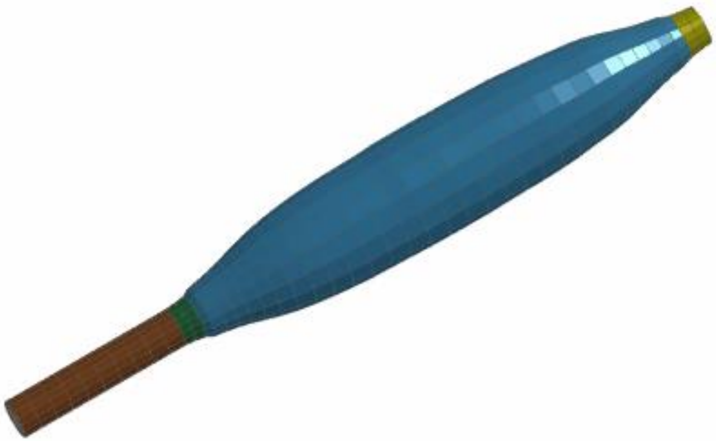
[Graphs are courtesy of Benjamin Cummings]

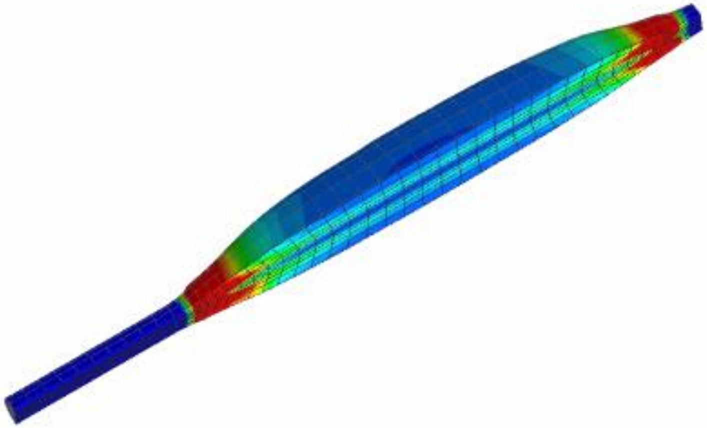
*MAT_TISSUE_DISPERSED

- Example of an activated fusiform muscle
- Simulation results using LS-DYNA
  - Muscle force



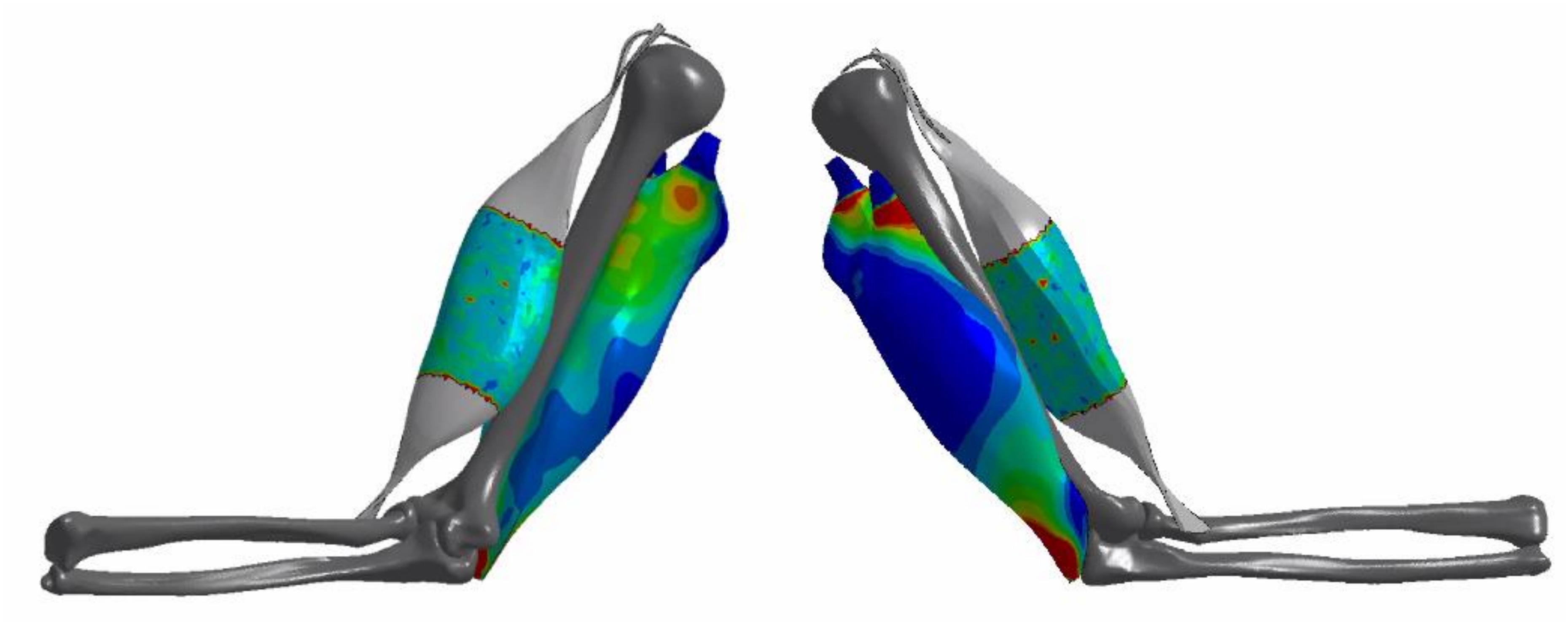
active anisotropic





active anisotropic

- *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

    no longer being developed and extended

    - *MAT_CABLE_DISCRETE_BEAM
    - *MAT_MUSCLE

    more versatile

  - 3-d finite elements
    - *MAT_OGDEN
    - *MAT_MOONEY_RIVLIN
    - *MAT_QUASILINEAR_VISCOELASTIC
    - *MAT_LUNG_TISSUE
    - *MAT_BRAIN_LINEAR_VISCOELASTIC

    passive isotropic

    - *MAT_SOFT_TISSUE(_VISCO)

    passive transverse isotropic

    - *MAT_HEART_TISSUE
    - *MAT_TISSUE_DISPERSED

    active anisotropic

DYNA MORE

Thank you for your attention!

Your LS-DYNA distributor and more