The tongue and muscle surfaces were defined by a geometric model that describes the shapes of the entire tongue and of individual muscles within the tongue. Individual muscle surfaces were based on MRI data, the work of Takemoto (2001), and segmentation of images from the Visible Human Project. Surfaces were created in the software package Rhinoscript (McNeel, Inc.).

The finite element model is derived by meshing the entire volume of the tongue model with 7000 hexagonal finite elements (left). Elements within muscle surfaces define the muscles within the solid model for computation (right).

The basis of the model is derived from the work of Reiner Wilhelms-Tricarico describing the tongue fiber directions as woven vector fields. The foundation of the model is compared to the active portion for the fiber or to that of the passive matrix. Stress generated within the tongue is ultimately constrained by the passive isotropic component (matrix) and is in addition to the anisotropic hyperelastic strain energy density function is added and is applied at each muscle.

\[
\Psi = \frac{1}{2} \sum_{l=1}^{12} (\lambda_l - 1)^2
\]

In the kinematic description, the deformational changes are given by a deformation gradient, \( F \), which is multiplicative split into deviatoric, \( F_d \), and a volumetric component, \( F_v \), that are dependent on the Jacobian, \( J \).

\[
J = \det F = \prod_{i=1}^{n} (1 + \varepsilon_i^F)
\]

The finite element model is derived by meshing the entire volume of the tongue model with 7000 hexagonal finite elements (left). Elements within muscle surfaces define the muscles within the solid model for computation (right).

The constitutive model is based on an anisotropic hyperelastic strain energy density function. The passive part, \( \Psi_{\text{pass}} \), is based on isotropic Ogden parameters. For the active component, an anisotropic hyperelastic strain energy function is added and is applied at each muscle fiber element at each contact. The active function, defined with a plastic passive component, depends on fiber direction and externally defined levels of activation. Both passive and active strain energy components depend only on the deviatoric parts of the deformation while a pressure model depending on volume change is used to characterize the compression behavior of the model. Computations are based on a three-field variational principle, in which the pressure, \( p \), and an activation field \( a \) are coupled for the appropriate description of the kinematics. This new constitutive model was based on the work of Blemker, Simo, and others which was incorporated as FORTRAN code into MSC Nastic for computations.

The deformation of the model is essentially a mapping function \( \psi(x,t) \) by a displacement field \( u(x,t) \) in which it is used to index changes and is not explicitly time.

In the kinematic description, the deformational changes are given by a deformation gradient, \( F \), which is multiplicative split into deviatoric, \( F_d \), and a volumetric component, \( F_v \), that are dependent on the Jacobian, \( J \).

\[
F = F_d F_v
\]

A modified Cauchy stress tensor is used in the strain computations from which the principal stretches \( \lambda_i \) are derived.

\[
\lambda_i = \sqrt[3]{J_i}
\]

The active Cauchy stress in the fiber direction, written as a tensile stress tensor, is:

\[
\sigma = \lambda_i a \max (0, \lambda_i - 1) \frac{1}{A} \frac{\partial}{\partial \lambda_i} \frac{1}{A} a \lambda_i m @ m
\]

where \( A \) is a normalization constant of 1.435.

The nearly incompressible behavior of the tissue is modeled as an additional degree of freedom representing pressure, which is constant for an element, and which is derived from a pressure energy density function that depends on the Jacobian, \( J \), of the deformation and the bulk modulus \( K \).

\[
\delta J - \frac{1}{2} \varepsilon \cdot \nabla \phi \nabla \phi = -p \nabla \phi
\]