A simple damage and fracture model of brain parenchyma for haptic brain surgery simulations

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Abstract

Dissection and removal of lesion areas are fundamental operations in brain surgeries. Therefore, damage and fracture models are needed to simulate dissection and removal operations. Generally, brain tissues show strong ductility; however, conventional fracture or damage models cannot reproduce ductile fractures well. In this paper, a simple damage and fracture model of brain parenchyma is proposed for real-time haptic surgery simulations. Although the proposed model does not require iterative calculation, it can reproduce ductile fracture while maintaining sufficient accuracy. The finite element method (FEM) is used to perform numerical simulations. In the proposed damage model, it is assumed that micro-damage begins when von Mises stress exceeds a certain threshold in an element, and the micro-damage grows with increased von Mises stress. The stiffness decreases as the micro-damage grows. When the integrity of an element becomes smaller than a certain threshold, the element is removed to express the occurrence of a fracture. These steps were formulated algorithmically. In order to verify the proposed damage and fracture model, tensile tests were conducted using porcine brain parenchyma. Parameters for the proposed damage model were identified using the results of the tensile tests. Tensile test simulations were performed using the identified parameters. The simulations effectively reproduced the stress-strain curves obtained in the tensile tests using porcine brain tissues.

Key words: Brain surgery simulation, Damage mechanics, Tensile tests, Mechanical properties, FEM

1. Introduction

In brain surgery, the surgeon must be very careful to remove lesion areas without damaging surrounding areas. If a surrounding area is damaged, important life functions may be lost. Therefore, surgeons must accumulate and update abundant knowledge and skills in order to provide patients with effective medical services. Animals, cadavers, plastic models, and apprenticeship have been used as traditional training methods. Novice surgeons have developed their skills using these traditional training methods. However, it is difficult to reproduce patients’ individual cases using animals, cadavers, and plastic models, and hence training has been limited to fundamental operations.

In recent years, surgery simulation has become increasingly popular as a surgery training method. Surgery simulators have the following advantages:

- It is easy to repeat a particular surgical operation under the same environment. Iterative training is an efficient way to improve basic surgery skills for novice surgeons.
- Brain models of individual patients can be made from preoperative medical images (e.g., CT or MRI images). Therefore, a surgery simulator can be used to make a surgery plan using a patient’s realistic brain model.
- Rare cases can be reproduced as models. Training for such rare cases would be highly effective, even for experienced surgeons.
A number of simulators have been developed, such as the LapVR™ Surgical Simulator (CAE Healthcare Inc.), the dV-Trainer® (Mimic Technologies Inc.), and so on. They have been used in various surgical specialties such as laparoscopic surgery. The validity of such surgery simulators has been confirmed and reported (Iwata et al., 2011, Perrenot et al., 2012).

Compared with laparoscopic surgery simulators, the number of developments in brain surgery simulators has been limited. A few attempts have been made so far. Alaraj et al. and Fenz et al. presented simulations of cerebral aneurysm clipping (Alaraj et al., 2015, Fenz et al., 2015). Delorme et al. presented a brain surgery simulator NeuroTouch and accomplished a tumor-debulking task (Delorme et al., 2012). All of these efforts dealt with single-scenario in brain surgery, and only a limited area around the lesion area was modeled. In order to realize a surgery simulation that can deal with a series of surgical operations using a whole brain, we have developed a brain surgery simulator (Fukuhara et al., 2014, Sase et al., 2015). Figure 1 shows the developed brain simulator.

In brain surgeries, the operative procedure is generally constructed using basic skills such as peeling of the blood vessels and the brain parenchyma to gain access to the surgical target area, retracting parts of the brain with spatulas to ensure an operative field, and maintaining appropriate tension with aspiration to allow ease of cutting with scissors. A common point of difficulty in these skills is the possibility of damaging brain tissue when an operator applies excessive force to the brain. Damage to the brain may cause serious aftereffects. Therefore, an accurate damage and fracture model should be implemented in brain surgery simulators.

In surgical procedures, dissection can be classified into two types: sharp dissection and blunt dissection. Sharp dissection (also referred to as cutting) is separation of tissues with sharp instruments, such as scalpels or scissors. Blunt dissection is separation of tissues along natural lines of cleavage with blunt instruments or fingers. Several studies have been conducted on cutting of biotissues. Element subdivision (Courtecuisse et al., 2014, Paulus et al., 2015) and node snapping (Steinemann et al., 2006, Lim et al., 2007) are common and effective methods to express cut wounds. However, the number of elements increases rapidly in element subdivision methods, requiring recalculation of the total stiffness matrix, and ill-shaped elements are easily generated in the node-snapping method. Therefore, both element subdivision and node snapping are not suitable for real-time haptic simulation. Moreover, a cut wound was defined as a cut line, and the fracture mechanics was not mentioned in these studies. A few studies of blunt dissection of biotissues have been conducted so far (Mora et al., 2009, Delorme et al., 2012). Mora et al. and Delorme et al. used a volume-sculpting technique to present tissue separation. Although the stiffness of tissues (Delorme et al., 2012) or the contact force (Mora et al., 2009) was presented, fracture mechanics and reaction forces when fractures occur were not discussed.

In order to characterize a material’s resistance to fracture, a field of research, fracture mechanics, can be cited. Fracture mechanics focuses on the propagation of cracks in materials, and brittle fractures can be reproduced accurately. However, conventional fracture mechanics cannot reproduce ductile fractures well. In order to express ductile fractures, damage models such as the Lemaitre model (Lemaitre et al., 1985) have been proposed. The Lemaitre model is a damage model for metals, and requires a yield function that specifies elastic and plastic regions. In the Lemaitre model, damage progresses only in the plastic region. However, in biological tissues such as brain tissue, it is difficult to define a yield function that specifies elastic and plastic regions by referring to results of tensile tests.

In order to represent damage to brain tissue in a simulation, the damage mechanics of brain tissue should be clarified. There have been many earlier studies of the mechanics of brain tissue using tensile tests (Rashid et al., 2014), compression...
Motion direction

Fig. 2 Overview of the tensile test system. This system consists of a motion table for applying tension to a specimen and a load cell for force measurement. Specimens are fixed to the plates with instant glue.

Fig. 3 Sequences of a tensile test at a velocity of 0.1 mm/s.
(a) initial state, (b) micro-damage, (c) macro-damage growth, and (d) complete separation.

tests (Pervin et al., 2009), and shear tests (Rashid et al., 2013). However, most of these focused on deformation before failure. Fracture mechanics of other biotissues such as skin (Annaidh et al., 2012), liver (Kemper et al., 2010), and muscle (Taylor et al., 1993) have also been investigated. However, fracture and damage mechanics of brain tissue have not been investigated thoroughly. Therefore, this study focuses on damage mechanics of brain parenchyma.

This paper is organized as follows. Section 2 presents the results of tensile tests of porcine brain parenchyma. In Section 3, a damage and fracture model suitable for real-time computation is proposed. In Section 4, mechanical parameters of the damage model are identified with the results of tensile tests by solving an optimization problem, and simulation results using the proposed damage model are compared with the experimental results of tensile tests to validate the model.

2. Tensile tests of porcine brain parenchyma

In order to investigate fracture mechanics of brain tissue, tensile tests were performed. In the tensile tests, porcine brains were used because the mechanical properties of porcine brain parenchyma are similar to those of the human brain (Nicolle et al., 2004). Uniaxial tension and fracture tests were performed to determine the mechanical characteristics of brain tissue.

2.1. Specimen preparation and equipment

Three fresh porcine heads were collected from a local slaughter house on the day of the experiment. Swines were roughly six months old. We removed the brains from the heads just before the experiments, and tested within 12 hours postmortem. Thirteen specimens were taken from different areas of the brain cerebrum. Firstly, specimens were cut from the brain parenchyma using a cork borer; thus, specimens were cylindrical. A constriction was then formed at the middle of each specimen so that the specimen was fractured at the constriction in the tensile test. As a result of the forming, each specimen was shaped as a hyperboloid. The lengths of hyperboloid-shaped specimens were 15±1 mm, the diameters of both ends were 6±1 mm, and the diameters of the constrictions were 4±1 mm. There were individual differences among specimens because of a variation in size and different composition. Brain parenchyma is composed of gray matter, which contains numerous cell bodies, and white matter, which contains mainly myelinated axons. Some researchers have found no difference between the elastic properties of the white matter and gray matter, while others have found differences (Kyriacou et al., 2002). In this work, we assumed that white matter and gray matter have the same elastic properties, and did not separate them in preparing the specimens. Moreover, previous researches reported that the anisotropy of brain tissue was relatively low (Prange et al., 2002, Nicolle et al., 2005). In this study, the brain tissue is assumed as an isotropic material. Specimens were preserved in a physiological saline solution before the experiments.

In the tensile tests, both ends of the specimens were glued to plastic plates by instant glue (cyanoacrylate). One plate was set to the top platen attached to a load cell, and the other plate was set to a lower platen attached to a motion table. As shown in Fig. 2, the motion table pulled the bottom of the specimens downward (i.e., in the negative direction of the Y-axis), and the tensile forces of the specimens were measured by a load cell (TEAC Corp. TU-UJ1.5N).
2.2. Results of tensile tests

In the tensile tests, specimens were pulled at two different velocities of 0.1 mm/s and 1.0 mm/s. Figure 3 shows the progress of how the brain tissue deformed from the initial state to complete separation at a tensile velocity of 0.1 mm/s. Eight successful datasets were obtained: four datasets at 0.1 mm/s and four datasets at 1.0 mm/s.

The stress-strain curves at each velocity are plotted in Fig. 4. In this paper, the stress was calculated from the force and the original end area of the specimens. In other words, it represents nominal stress. Since the data obtained from the load cell includes high-frequency noise, a Savitzky-Golay filter (frame size = 100) was used for noise removal. Fitting of a quintic function (a fifth-degree polynomial) to the datasets at each velocity was performed. The fitted curves are plotted together with the original data in Fig. 4. The fitted curves for the tensile tests at the fast (1.0 mm/s) and slow (0.1 mm/s) tension velocities are compared in Fig. 5. As shown in Fig. 5, both curves are nearly the same when the strain is low. Thereafter, the curve at the high tension velocity reaches a higher ultimate tensile strength (UTS) and larger strain. This shows that the stress-strain curve for brain tissue depends upon tensile velocity. A similar biomechanical phenomenon in anterior cruciate ligament has been reported (Noyes et al., 1974).

2.3. Differences between brain and other soft biological tissues

Stress-strain curves of other biotissues such as skin (Annaidh et al., 2012), liver (Kemper et al., 2010), and muscle (Taylor et al., 1993) have been reported. Skin, liver, and muscle showed brittle fracture characteristics as illustrated in Fig. 6, while brain tissues showed ductile fracture characteristics as presented in Figs. 4 and 5.

Although ductile fracture characteristics have been studied in damage mechanics, most works have been done for metals; thus, such works have assumed elasto-plasticity. As shown in Figs. 4 and 5, explicit elasto-plasticity characteristics are not evident in the stress-strain curves of brain tissue. Therefore, we propose a damage and fracture model for brain parenchyma that can reproduce ductile fracture characteristics.
3. A damage and fracture model for brain parenchyma

3.1. Conventional fracture models for biotissues

Most surgery simulators developed so far use the finite element method (FEM) to compute deformations of organs and biotissues. The simplest method to simulate surgical cuts and dissections is to remove an element (Forest et al., 2002, Nakayama et al., 2011, Chen et al., 2014). In the removing element method, maximum principal stress, shear strain energy, or maximum shear stress of an element has been generally used as an index to evaluate whether an element should be removed or not. The blue solid line in Fig. 7 shows the stress-strain curve given by a tensile test simulation of a hyperboloid-shaped soft tissue model using the removing element method. The removing element method reproduces brittle fractures.

A considerable number of studies have been conducted on fracture mechanics for metals (e.g., Anderson, 2005). In the field of computational fracture mechanics, the extended finite element method (XFEM) (Moës et al., 1999) has been used widely to analyze crack growth. Some researchers have applied XFEM to surgery simulation in order to simulate surgical cuts and dissections (Gutiérrez et al., 2010, Vignon et al., 2004). A tensile test simulation of a hyperboloid-shaped soft tissue was performed using XFEM. The obtained stress-strain curve is plotted as a green solid line in Fig. 7. ABAQUS/CAE Student Edition 6.14 was used for the XFEM simulation. The maximum principal stress criterion was used to predict damage initiation in the XFEM-enriched region. Displacement and linear were chosen for the type of damage evolution and the softening method, respectively.

In the removing element method, several elements tend to reach the fracture stress simultaneously. In XFEM, a crack grows quickly after fracture initiation. Both the removing element method and XFEM effectively simulate brittle fractures, as shown in Fig. 7. However, it is difficult for both methods to simulate ductile fractures, which are seen in tensile tests of brain tissues (Figs. 4 and 5).
3.2. Conventional damage models

Continuum damage mechanics has been studied enthusiastically so far (Murakami, 2012). It is generally considered that a fracture process consists of the nucleation of microscopic cracks and voids, growth and coalescence of microscopic cavities, and resultant macroscopic cracks, as illustrated in Fig. 8.

When dealing with microscopic discontinuity effects in continuum mechanics, discontinuous structures should be represented as macroscopic continuous fields by averaging mechanical effects. The minimum volume that satisfies this condition is called a representative volume element (RVE).

Gurson represented the damage state by the void volume fraction \( f \) (Gurson, 1977), which is given by

\[
f = \frac{V_{\text{void}}}{V^e},
\]

where \( V^e \) and \( V_{\text{void}} \) are the volume of an RVE and the total volume of voids, respectively. Gurson proposed a yield function for a damaged material as follows (Gurson, 1977).

\[
F(\sigma_{\text{Mises}}, \sigma_H, \sigma_y, f) = \left( \frac{\sigma_{\text{Mises}}}{\sigma_y} \right)^2 + 2f \cosh \left( \frac{3\sigma_H}{2\sigma_y} \right) - f^2 - 1 = 0,
\]

where \( \sigma_{\text{Mises}}, \sigma_H, \) and \( \sigma_y \) are von Mises stress, mean stress, and yield stress, respectively. When \( f = 0 \), Eq. (2) corresponds to the von Mises yield criterion (Murakami, 2012).

In the Gurson-Tvergaard (GT) model (Chu and Needleman, 1980), the evolution equation for the void volume fraction is given by

\[
f' = (1 - f)e_{kk}^p + \Lambda e_{eq}^d,
\]

where \( e_{kk}^p \) is the plastic volumetric strain rate and \( e_{eq}^d \) is the equivalent plastic strain rate. \( \Lambda \) is a coefficient.

Most continuum damage mechanics models (e.g., Rice et al., 1969, Gurson, 1977, and Lemaître, 1985) are based on elastic-plastic theory. Therefore, nonlinear solutions (e.g., using the Newton-Raphson scheme) are needed by invoking iterative calculations. However, such iterative calculation methods are not suitable for haptic surgery simulations because reaction force, deformation, and dissection must be calculated in real time. Furthermore, iterative calculation cannot guarantee the convergence of a solution at a critical time.

Therefore, only the evolution equation for the volume fraction is implemented in a linear dynamic FE model. In order to evaluate the damage model given by Eq. (3), tensile test simulations were performed using the same specimen model presented in Section 3.1. The coefficient \( \Lambda \) in Eq. (3) is a function of \( e_{eq}^d \) in the GT model; however, a constant value is set for \( \Lambda \) in order to simplify the simulation.

Figure 9 (a) shows the stress-strain curve given by tensile test simulation with a tensile velocity of 0.1 mm/s. \( \Lambda \) and other parameters were given manually so that the resultant curve reproduced the fitted curve obtained from tensile tests with 0.1 mm/s presented in Section 2.

Figure 9 (b) shows the stress-strain curve given by tensile simulation with a tensile velocity of 1.0 mm/s. The same parameters were used as in the case of the velocity of 0.1 mm/s. Figure 9 (b) clearly shows the disadvantage of the damage model given by Eq. (3). If the tensile velocity is changed, the damage model does not reproduce the experimental results well.
3.3. Proposed damage model

Although the fracture behavior of brain tissue is different from that of metals, the fracture process can also be expressed by three steps: (1) micro-damage initiation, (2) micro-damage growth, and (3) coalescence of micro-damage, as illustrated in Fig. 8.

In the finite element method (FEM) for linear elasticity problems in infinitesimal strain theory, the element stiffness matrix $K^e$ is given by:

$$K^e = \int_V (B^e)^T C^e B^e dv,$$

where $V^e$ is the volume of the element, $B^e$ is a displacement in the strain matrix, and $C^e$ is a strain in the stress matrix. In this study, models in the simulation were divided into linear tetrahedral elements. Therefore, Eq. (4) is rewritten into

$$K^e = V^e (B^e)^T C^e B^e,$$  \hspace{2cm} (5)

The stress $\sigma^e$ of an element is calculated by:

$$\sigma^e = C^e B^e u^e,$$  \hspace{2cm} (6)

where $u^e$ is a displacement vector of an element.

In this study, an element is considered as an RVE. The integrity $\omega$ of the RVE is defined as

$$\omega = \begin{cases} 1 & \text{corresponds to the undamaged state}, \\ \omega_t < \omega < 1 & \text{characterizes the damaged state}, \\ \omega \leq \omega_t & \text{corresponds to a completely damaged state}, \end{cases}$$  \hspace{2cm} (7)

where $\omega_t$ is the threshold for fracture.

The three steps of a fracture process in an element are summarized as follows:

1. **Micro-damage initiation:** The von Mises yield criterion is used for micro-damage initiation. When the von Mises stress of an element exceeds a previously specified threshold, it is assumed that micro-damage is initiated in the element. This is expressed as

$$\omega = \begin{cases} 1 & (\sigma_{\text{Mises}} < \sigma_t), \\ \omega \in [0, 1) & (\sigma_{\text{Mises}} \geq \sigma_t), \end{cases}$$  \hspace{2cm} (8)

where $\sigma_{\text{Mises}}$ is the von Mises stress of an element and $\sigma_t$ is the stress threshold for micro-damage initiation.

2. **Micro-damage growth:** In the proposed damage model, it is assumed that the volume of micro-damage $V_{\text{void}}$ grows owing to the increase in von Mises stress. The void volume fraction $f$ is formulated by Eq. (1). The evolution equation for the void volume fraction is given by Eq. (3) in the GT model. In the proposed damage model, the evolution equation is formulated as

$$f = \begin{cases} \gamma(1 - f)(\sigma_{\text{Mises}} - \sigma_t) & (\sigma_{\text{Mises}} \geq \sigma_t), \\ 0 & (\sigma_{\text{Mises}} < \sigma_t), \end{cases}$$  \hspace{2cm} (9)

where $\gamma$ is a coefficient. In this step, the integrity $\omega$ of the RVE is defined as

$$\omega = 1 - f.$$  \hspace{2cm} (10)

The stiffness loss increases with $V_{\text{void}}$ growth. The damaged stiffness matrix $K^e(\omega)$ of an element after stiffness loss is given by

$$K^e(\omega) = (V^e - V_{\text{void}})(B^e)^T C^e B^e = (1 - f)V^e(B^e)^T C^e B^e = \omega K^e.$$  \hspace{2cm} (11)

3. **Coalescence of micro-damages:** When the void volume becomes sufficiently large, the voids in the RVE coalesce with voids in adjacent RVEs. Therefore, it is assumed that if the integrity $\omega$ of an RVE becomes smaller than a threshold $\omega_c$, coalescence of microscopic voids evolves rapidly into fracture. Hence, the element is removed to express fracture. In this paper, $\omega_c$ was set to 0.005.

4. Tensile test simulation

In order to verify the proposed damage model, tensile simulations were performed. The simulation results are compared with the stress-strain curves plotted in Fig. 4.

Numerical calculation is performed on a workstation KRONOS S810R (CIARA Inc.). It is possible to overclock the multi-core CPU (Intel Core i7 3960X 3.3 [GHz], 6 cores) to 4.8 [GHz].
4.1. Numerical Analysis

A linear, isotropic, viscoelastic brain tissue model was used in the simulation. Corotational FEM (Müller et al., 2002) was used in the tensile test simulation to deal with geometrical nonlinearity.

The equation of motion in corotational FEM is given by:

$$\mathbf{M}\ddot{\mathbf{x}} + \mathbf{D}(\omega)\dot{\mathbf{x}} + \mathbf{K}'(\omega)\mathbf{x} + \mathbf{F}'_0 = \mathbf{F}_e. \quad (12)$$

where $\mathbf{F}_e$ and $\mathbf{F}'_0$ are the external force vector and external force offset, respectively. $\mathbf{x}$, $\dot{\mathbf{x}}$, and $\ddot{\mathbf{x}}$ are the position, velocity, and acceleration vectors of the nodes, respectively. $\mathbf{D}(\omega)$ and $\mathbf{M}$ are the damping and mass matrices. $\mathbf{K}'(\omega)$ is the modified total stiffness matrix constructed from the updated element stiffness matrix $\mathbf{K}''(\omega) = \mathbf{R}'\mathbf{K}'(\omega)\mathbf{R}^T$, where $\mathbf{K}'(\omega)$ is the damaged element stiffness matrix given by Eq. (11) and $\mathbf{R}^e$ is the rotation matrix of the element. See (Müller et al., 2002) for more details regarding corotational (or stiffness-warping) FEM.

For simplicity and efficiency of computation, the lumped mass approximation is adopted:

$$m_i = \sum_{j=1}^{N_i} \rho V_j / 4, \quad (13)$$

where $m_i$ is the lumped mass of node $i$, $\rho$ is the density, $V_j$ is the volume of element $j$, and $N_i$ is the number of elements that share node $i$. Therefore, the $i$th diagonal block matrix $\mathbf{M}_i$ is $m_i\mathbf{I}_3$, where $\mathbf{I}_3 (\mathbf{I}_3 \in \mathbb{R}^{3\times3})$ is an identity matrix.

In this study, Rayleigh damping is assumed, in which $\mathbf{D}(\omega)$ is given by:

$$\mathbf{D}(\omega) = \alpha\mathbf{K}'(\omega) + \beta\mathbf{M}, \quad (14)$$

where $\alpha$, $\beta$ are scalars. In order to avoid numerical instability, the implicit Euler method is adopted for time integration.

4.2. Mechanical properties

Mechanical properties must be determined. Past works reported that the range of the density and Poisson’s ratio of porcine brain tissue were approximately 1,000–1,140 kg/m$^3$ and 0.45–0.5, respectively (Ayache, 2004). Chen et al. determined the density and Poisson’s ratio of porcine brain from tensile experiments as 1,036 kg/m$^3$ and 0.4925, respectively (Chen et al., 2014). These values were used in the simulation.

On the other hand, the Young’s modulus values determined so far varied a great deal from three to four digits (Kaster et al., 2011, and Rashid et al., 2012). It seems that the tremendously wide range of results arise from strong individual variation and different specimen conditions. For example, it is well-known that the Young’s modulus of brain tissue depends strongly upon the postmortem time (Garo et al., 2007).

Rayleigh damping has been popularly used in finite element dynamic analysis. However, the scalars used in Rayleigh damping do not have any physical meaning. Moreover, the stress threshold for micro-damage initiation and the void volume growth rate coefficient used in the proposed damage model must be determined. Therefore, the Young’s modulus, Rayleigh damping scalars, stress threshold for micro-damage initiation, and void volume growth rate coefficient must be identified so that simulation results fit experimental results.

4.3. Parameter identification

The Young’s modulus $E$, stress threshold $\sigma_t$ for micro-damage initiation, Rayleigh damping scalars $\alpha$, $\beta$, and void volume growth rate coefficient $\gamma$ were identified by solving an optimization problem. The identification process is illustrated in Fig. 10. Tensile test simulations were performed in the optimization loop with a brain parenchyma model illustrated in Fig. 11 at a tensile velocity of 0.1 mm/s. The parameter vector $\mathbf{q}$ was given from an optimization problem solver.
solver to the dynamic simulator. The dynamic simulator performs a tensile test with the given parameter vector and returns the objective function $\mathcal{F}(\mathbf{q})$ to the optimization problem solver. This process is iterated until $\mathcal{F}(\mathbf{q})$ is minimized.

The objective function $\mathcal{F}(\mathbf{q})$ is defined as follows:

$$\mathcal{F}(\mathbf{q}) = \sum_{\varepsilon=0.0}^{0.9} \left( \sigma_{\text{fit}}(\varepsilon) - \sigma_{\text{sim}}(\varepsilon) \right)^2,$$

$$\mathbf{q} = [E \quad \sigma_t \quad \alpha \quad \beta \quad \gamma],$$

where $\mathbf{q}$ is a parameter vector. $\sigma_{\text{fit}}(\varepsilon)$ is the stress given by the fitted curve at the tensile velocity 0.1 mm/s (the red solid line in Fig. 4(a)). $\sigma_{\text{sim}}(\varepsilon)$ is the stress obtained from tensile test simulations.

The MATLAB Optimization Toolbox™ was used as the optimization problem solver. A sequential quadratic programming (SQP) solver (the $fmincon()$ function in the MATLAB Optimization Toolbox) was used to solve the optimization problem with nonlinear constraints. The optimization problem was defined as follows:

minimize $\mathcal{F}(\mathbf{q})$,

subject to $E \in [1.0 \times 10^3, 5.0 \times 10^3]$, $\sigma_t \in [1.0 \times 10^2, 2.5 \times 10^3]$, $\alpha, \beta \in [1.0 \times 10^{-4}, 10.0]$, $\gamma \in [1.0 \times 10^{-7}, 5.0 \times 10^{-6}]$.

The initial guess was set to $\mathbf{q}_0 = [2500 \quad 800 \quad 0.6 \quad 0.6 \quad 3.0 \times 10^{-6}]$.

The dynamic simulator was programmed in C++. The $fmincon()$ function of MATLAB called the dynamic simulator as a child process. The optimized parameters were $\mathbf{q}_{\text{opt}} = [2823.8 \quad 1183.6 \quad 1.174 \quad 0.958 \quad 3.316 \times 10^{-6}]$, which are also summarized in Table 1. Sequences of tensile test simulations with $\mathbf{q}_{\text{opt}}$ are illustrated in Fig. 12. Stress-strain curves obtained by dynamic simulation with $\mathbf{q}_{\text{opt}}$ and curves obtained by fitting experimental results with a tensile velocity of 0.1 mm/s are plotted in Fig. 13.
As shown in Fig. 13, the error between the simulation result and the experiment result (fitted curve of the experimental data with tensile velocity of 0.1 mm/s) is small. It proves the usefulness of the parameter identification by solving the optimization problem. Animation pictures of the simulation (Fig. 12) show the fracture process of the brain parenchyma. The fracture develops gradually with the deformation of longitudinal direction. However, as shown in Fig. 12(e), even the applied tensile force is uniform, the fracture surface is knurling. It is because that the fracture is expressed by removing tetrahedral elements in this study. Therefore, the fracture surface depends on the tetrahedral element position. This problem will be improved by minifying the size of elements.

4.4. Verification simulation at different tensile velocities

In Section 4.3, parameters of the proposed damage model were identified so that the simulation results at a tensile velocity of 0.1 mm/s fit the corresponding experimental results. In order to verify the proposed damage model, tensile simulations were performed using the \( q_{opt} \) determined in Section 4.3 at different tensile velocities. A tensile velocity of 1.0 mm/s was used in the verification simulation in order to compare simulation results with experimental results at the same tensile velocity, as plotted in Fig. 4 (b).

Figures 14 and 15 show comparisons between simulation results and experimental results at tensile velocities of 0.1 mm/s and 1.0 mm/s, respectively. As shown in Fig. 15, although parameters were identified using experimental results at a tensile velocity of 0.1 mm/s, the damage and fracture model accurately reproduces the experimental results at a tensile velocity of 1.0 mm/s. This result validates the effectiveness of the proposed damage and fracture model.
5. Conclusions and future works

When a specimen of brain tissue is stretched, the specimen generally shows ductile fractures. However, conventional fracture or damage models cannot accurately reproduce ductile fracture characteristics. In order to simulate dissection and removal operations in brain surgery, a simple damage and fracture model for brain parenchyma was proposed.

The novelty of this work consists of the following two points: (1) fracture characteristics of brain parenchyma were verified experimentally by performing tensile tests using porcine brain parenchyma at two different tensile velocities (0.1 mm/s and 1.0 mm/s), and (2) a simple damage and fracture model was proposed and verified by comparing experimental and simulation results.

In order to achieve real-time haptic simulation, a simple damage and fracture model with a linear FEM was studied in this paper. In the future, a more accurate model for off-line simulation such as a Maxwell non-linear model with a Lemaitre damage model will be studied.

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