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Research Paper

A stochastic model for soft tissue failure using acoustic emission data

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ABSTRACT

The strength of soft tissues is due mainly to collagen fibers. In most collagenous tissues, the arrangement of the fibers is random, but has preferred directions. The random arrangement makes it difficult to make deterministic predictions about the starting process of fiber breaking under tension. When subjected to tensile stress the fibers are progressively straighten out and then start to be stretched. At the beginning of fiber breaking, some of the fibers reach their maximum tensile strength and break down while some others remain unstressed (this latter fibers will assume then bigger stress until they eventually arrive to their failure point). In this study, a sample of human esophagi was subjected to a tensile breaking of fibers, up to the complete failure of the specimen. An experimental setup using Acoustic Emission to detect the elastic energy released is used during the test to detect the location of the emissions and the number of micro-failures per time unit. The data were statistically analyzed in order to be compared to a stochastic model which relates the level of stress in the tissue and the probability of breaking given the number of previously broken fibers (*i.e.* the deterioration in the tissue). The probability of a fiber breaking as the stretch increases in the tissue can be represented by a non-homogeneous Markov process which is the basis of the stochastic model proposed. This paper shows that a two-parameter model can account for the fiber breaking and the expected distribution for ultimate stress is a Fréchet distribution.

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1. Introduction

The non-linear elastic behavior of collagenous soft tissue is well understood, and complex models based on fiber arrangement have been developed (Natali et al., 2009; Kroon and Holzapfel, 2008). On the other hand, soft tissue failure shows a much more complex behavior. It is not so well understood and needs additional research (Ionescu et al., 2006). The experimental data needed to develop models of failure are scarce and the adaptation of existing failure formulations for common engineering materials to soft tissues is difficult because of the microstructure existing in soft collagenous tissues.

There are different types of soft tissue failure modes. The geometrical description of the last stages of failure has

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proved to be elusive with high degree of randomness. Reliable failure models is important for some medical procedures such as esophageal dilatation (Fernández-Esparrach et al., 2011) and some other more unusual practical situations involving penetrating trauma injuries. This kind of injuries represents a high socioeconomic cost and represent a significant source of morbidity (Gugala and Lindsey, 2003).

The present work focuses on the understanding of the cumulative effect of internal micro-failures before macroscopic failure. For this purpose a stochastic cumulative damage model based on inhomogeneous renewal-rewarded process is used (an inhomogeneous Markov process measures the number of micro-failures and the "reward" variables control the damage occurred in such a failure). The model presented here differs from other interesting probabilistic models found in the literature (Pradhan and Chakrabarti, 2003; Kabir et al., 2006) and is innovative in that it provides the distribution of the ultimate stress. On the other hand, another important innovation is the use of Acoustic Emission (AE) to detect experimentally almost all of the relevant micro-failures in soft tissues.

In Section 2 the experimental setup is described and the two-parameter stochastic model is explained. Section 3 contains the results of the model applied to a human esophagus sample (the distribution for the ultimate stress is computed and the dissipated energy are given). Section 4 discusses the implications of the results.

2. Methods

A typical specimen of human esophagus from a donor (PMHS) was used for the experimental work. The age of the PMHS was 63 years old, and its decease cause was not associated with any esophageal disease (biometric data: male; Body Mass Index, 25.5 $\mathrm{kg \cdot m^{-2}}$). This single sample produced $N=472$ microfailures, which statistically is an adequate medium-size sample. A tensile test was conducted for the sample in order to obtain the stress–strain curve (the acoustic emission measurements and the strain–stress measures of the tensile test were simultaneous). A conventional servo-hydraulic testing machine (microtest EM2/20) was used for the measurement of force. The strain was computed from the data of a camera using motion tracking (see Fig. 1). An important technical issue was the design of the clamps. The clamps were made out of non-porous polymeric material (Nylon 6) for two main reasons: (1) a porous material would have produced adherence and local dehydration in the sample, (2) in addition, being less rigid the polymeric material allows a better fit to the soft tissue. As it can be observed in Fig. 2, the planar clamps are formed by two sets of twin plates, each set is located at the edge of the rectangular sample of tissue. The clamps used for all tests were specifically designed for the occasion. The thickness and dimensions were adjusted in order to ensure that deformations of the clamps are completely negligible and the application of pressures and forces is fairly uniform. Preventing the creeping of the tissue was specially difficult because of the existence of water and moisture in the tissue; which in some cases acted as lubricant. After some preliminary testing and the addition of some extra drills the pressure was increased. This pressure ensures no significant sliding and, thus, the strain measures are correct (if there had been any sliding, the strain measure would have been distorted). The experimental setting is the same described in Sánchez-Molina et al. (2014). For the esophageal tissue, a stress–strain relation of type

$$\tau = Ae^{-b\varepsilon^2}\varepsilon \tag{1}$$

was obtained, the measured parameters were $A=5385$ kPa and $b=864.2$ (here τ is the [Piola] axial stress, and ε is the [Green-Lagrangian] strain). This relation (1) is deduced from the Yang–Gregersen–Deng constitutive equation (Deng et al., 1994; Yang et al., 2006), and it was found quite accurate ($r^2 = 0,9977$) for the data. For the purposes of this study, the exact constitutive equation used is not very important, as long as the model approximates well the stress–strain curve (all adequate models lead to a distribution function for the ultimate stress of the same type). The relation (1) is needed to calculate the stress when a failure is detected because the camera measures the strain, not the stress.

In addition, the experimental setup included four acoustic emission sensors (see Fig. 1) that detected the occurrence of micro-failures inside the sample (each micro-failure releases a certain amount of elastic energy which can be detected by sensors).

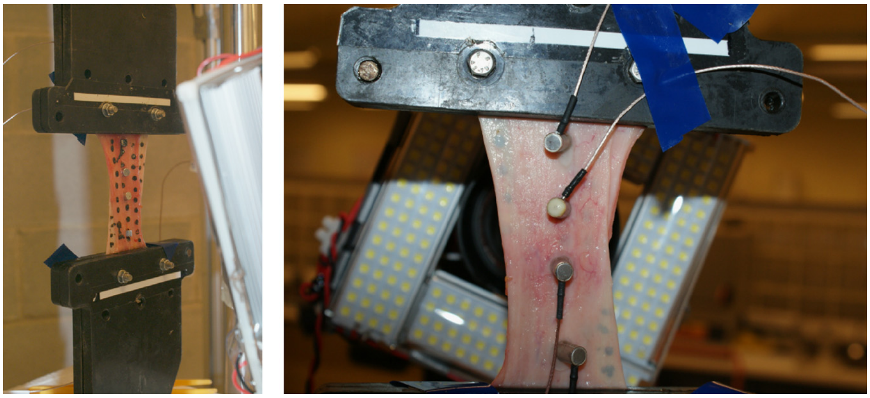

Fig. 1 – Experimental setting for the uniaxial tensile tests. ***Right*****: sample in place for testing, with clamps and acoustic sensors, the upright face is the** ***muscularis externa*** **layer.** ***Left*****: detail of the acoustic sensors in the close-up face is the** ***mucosa*** **layer.**

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