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Towards a Soft-Tissue Cutting Simulator using the Cohesive Zone Approach

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Abstract— **Advancements in computational techniques have provided the ability to utilize detailed models in surgical training systems. The development of more complicated procedures coupled with reduced training time for medical residents has driven the need for accurate reality-based medical simulators. Extensive research has been conducted in the area of modeling general deformation of biological tissue; however few studies have focused on the physical properties of specific tool tissue interactions such as cutting. This paper presents a fracture mechanics based method to model the scalpel cutting of porcine liver by implementing a cohesive zone approach. Using *in vivo* cutting data, the parameters of the cohesive zone model are defined for the scalpel cutting process of soft biological tissues.**

I. INTRODUCTION

Current trends in the medical community have driven the need to create realistic surgical training systems to aid in the educational process of medical residents. The need for reality-based medical simulators has grown out of the limited time available to residents to practice surgical tasks as well as the high fatality rate associated with surgical errors. To this end, the development of a reality-based, real-time surgical training system for common surgical tasks will provide the foundation required for improving upon the surgical skills of medical residents.

To develop a realistic surgical training system, a vital aspect of the simulator is a solid foundation of accurate models to govern the physical behavior of the tool-tissue interaction. A common task during general surgical procedures is the act of cutting or dissecting soft tissue. In studying the simulation of soft-tissue cutting, current approaches can be segmented into two distinct areas. From the simulation side, there is a focus on the element segmentation aspects, whereas modeling-based research has focused on the physics of the cutting process. Element segmentation techniques vary from simple element removal [3] to complex adaptive meshing strategies [8]. These methods, however, focus on the process of splitting the mesh in a finite element code and do not take into account the physical aspects of the cutting process. In a reality-based simulator, studying the physics of the cutting problem will prove to add levels of accuracy in addition to haptic capabilities.

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Studies into the physical aspect of the cutting process have mainly focused on applying fracture mechanics principles to model the cutting process. Work has been conducted in the areas of scissor cutting of soft tissues [9], guillotine loading with razor blades [7], and cutting with a scalpel blade [2][1]. In most of these studies the focus was on the determination of the fracture toughness of the biological material. A slightly different approach was used to determine the local effective modulus for cutting of liver tissue with a scalpel blade [2]. These approaches do study the physics of the cutting process, however, they are limited to defining the onset of the cutting process without additional focus on how the crack propagates after initiation.

In this study, a cohesive zone approach is utilized to model the scalpel cutting process in porcine liver. The cohesive zone method is a means of simplifying the physical aspect of the cutting process into a manageable number of variables capable of modeling the onset of fracture as well as the propagation of the crack in the material. Cohesive zone methods have been used for a large number of fracture problems but have mainly been used in studying debonding processes where the cohesive zone represents the strength of adhesion between two materials [12][11]. The major limitation of the cohesive zone approach is the inability to discern information relating to the direction of the fracture over time. Cohesive zone models assume a straight fracture pattern in front of the loaded region [4]. In the case of surgical cutting this does not present a problem as the fracture process occurs on a trajectory in line with the cutting motion. Details about the cohesive zone approach will be discussed in the subsequent sections.

This paper aims at utilizing the cohesive zone approach to derive models that will be useful in representing the physical aspects of the cutting process in medical training simulators. By using a physics-based approach, reality-based haptic and graphic feedback can be included in the simulation of cutting tasks resulting in more realistic medical training systems. Thus, this paper presents a method for the development of a cutting model based on *in vivo* experimental data for the cutting of porcine liver and cohesive zone theory.

II. METHODS AND RESULTS

A. Cohesive Zone Theory

The cohesive zone approach simplifies the fracture process into a form that allows for modeling of a crack initiation as well as the further propagation into the material [10]. It is a physics-based approach that utilizes three material specific parameters: the cohesive energy, cohesive strength and cohesive separation distance (Figure 1). From these parameters a traction-separation curve is generated which represents the tractions across the cohesive region from the initial loading through the material degradation to ultimate failure. A schematic of a standard traction-separation curve is displayed in Figure 1a. The cohesive strength, τ_{max}, corresponds to the peak of the traction-separation curve. The cohesive separation distance, δ_{sep}, defines the total length of the curve (physically it is the width of the crack front) and the cohesive energy correlates to the total area under the curve.

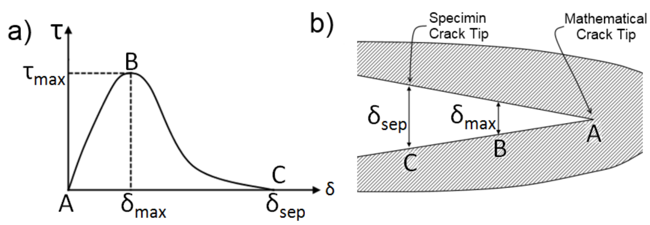

Fig. 1. a) Example traction-separation curve b) Cohesive zone region.

Physically speaking, the cohesive zone approach reduces the complex fracture process into the simplified traction-separation law. When a material is loaded, a cohesive zone develops in front of the crack tip. This zone starts at point A (see Figure 1b) which correlates to the location in the material where loading due to the crack begins to become apparent. This point does not correspond with the location of the physical crack in the specimen but is located ahead of that point where the effect of loading begins, as can be seen in Figure 1b. In the region from the physical crack tip, point C, to the mathematical crack tip, point A, the traction-separation curve defines the loading of the material. Point B is located between the mathematical and physical crack tips and represents the location of maximum stress within the cohesive region. From point A to point B, the sample is being loaded. At point B, the sample reaches it critical stress point and starts to fail. In the region from point B to point C the sample degrades until it reaches the point of complete physical separation at point C [10].

By using this approach it is possible to define the initiation and propagation of the crack front through the study of loading properties in the region directly in front of the crack tip. Once the material is loaded to a certain extent and the damage region completes, the crack physically opens in the material. Implementation through the use of the finite element method is direct as the parameters required for cohesive zone calculations (stress, strain and energy) are directly available in the existing analysis [5]. By comparing the loading conditions in the finite element analysis to those of the defined cohesive zone, it is possible to determine the degree of damage within the material. At the critical level of damage, as defined by the cohesive model, the material is known to be degraded to the extent that separation exists. This is then reflected in the simulation by physically separating the mesh itself, resulting in the formation or propagation of the crack.

B. *In Vivo Experiments*

A test device was designed and constructed for the collection of the *in vivo* experimental data required for derivation of the three cohesive parameters. The experimental setup has also been used for probing tasks on *in vivo* porcine liver and the details pertaining to the device layout can be found in [6]. The device itself was designed to perform a controlled linear cutting motion while recording both the force imparted on the scalpel blade and the surface displacement of the surrounding tissue throughout the cutting process. A close-up of the liver and scalpel setup can be seen in Figure 2. The force sensor connected to the scalpel blade is a six-axis force/torque sensor (JR3 Model number: 20E12A-I25) capable of resolving the cutting force into three individual components with axes aligned along the direction of the cutting motion, perpendicular to the blade surface and in the vertical direction. In addition to the force information, the location of the blade and the location of the surface markers (as seen in Figure 2) can be tracked throughout the experiment via a MicronTracker stereo camera system (Model H40, manufactured by Claron Technology Inc) to within a resolution of 0.25mm. During the experiment, the scalpel is made to cut the tissue at a rate of 1.25mm/s while recording both the force and image data from which the cohesive parameters are derived.

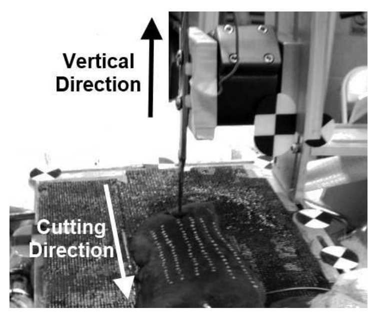

Fig. 2. *In vivo* experimental setup.

A depiction of the force profile obtained during the experiments is shown in Figure 3. This image displays the force acting on the scalpel blade in the direction parallel to the cutting motion and the vertical direction. The forces acting to push the blade from side to side were negligible due to the symmetric nature of the blade. As can be seen in the figure, a well pronounced sawtooth pattern appears during the cutting process which represents the individual fracture processes which occur in the tissue through the extent of the cut. It was this level of detail that was desired for the cohesive zone modeling. Therefore, for a single cutting experiment, the data was split into 18 individual cutting segments which were analyzed individually to obtain a list of cohesive parameters that could be used for simulation purposes.

C. *Cohesive Parameter Derivation*

After completion of the *in vivo* cutting tests, the experimental data was analysed to derive the parameters required for the definition of the cohesive zone model. Only two

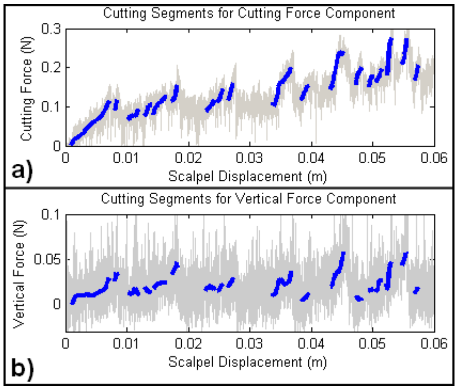

Fig. 3. Forces in the a) cutting and b) vertical directions.

of the three values defining the traction-separation curve are required for the complete statement of the model as the three variables are intrinsically linked. From the data collected, it was apparent that the most accurate parameters to determine were the cohesive strength and the cohesive energy. The cohesive separation distance is defined as the critical width of the crack at the instant of failure. Despite the effort to measure this parameter with the smaller markers, the tissue exhibited a large amount of deformation due to the scalpel loading procedure, composed mainly of elastic deformations, which limited the ability to accurately measure the cohesive separation distance. Both the cohesive strength and the cohesive energy, however, could be derived from the measured force profile.

Using the raw force values obtained during the experiment, the force profile was split into individual loading segments (Figure 3) through the use of a low-pass Butterworth filter aimed at reducing the noise while maintaining the previously mentioned sawtooth pattern. The location of the fracture was recorded as the point of maximum force for each of the loading segments. Thus, the data was split into the 18 segments as represented by the blue lines in Figure 3. The methodology as presented henceforth will be for the first cut segment, however a similar process was applied to analyze all individual loading segments.

The cohesive strength is a representation of the maximum normal and tangential tractions that occur along the scalpel blade during the cutting process. Thus the first step in determining the cohesive strength parameters was to find the forces in the cutting and vertical directions at the time of fracture; these values were directly derived from the filtered force segments previously mentioned. Next, to derive the tractions, the measured force values were distributed along the length of the blade to determine the local loading. To do this, the shape of the scalpel blade as well as the blade depth must be known at the instant of fracture. To determine the profile, imaging techniques were used to record both the profile and the cross sectional dimensions of the blade. To determine the depth of the blade the imaging data from the experiment was used. A combination of markers placed on the plate holding the liver, on the scalpel, and the surface was used to create a methodology for the determination of the depth of the blade at the point of fracture.

The goal of the analysis was to find the local tractions acting on the scalpel blade; the maximum of which, at the time of fracture, constitutes the cohesive strength. As the measurement of the force profile along the blade was not feasible, the projected length of the blade in both perpendicular directions was used to distribute the measured load. First, a pre-defined length was established to split the scalpel-tissue contact length into discrete sections. In this study, lengths of 0.2, 0,8 and 2.0mm sections were used to determine the effect of the section length. During the sectioning process, the normal to the blade at each section was also recorded for later use. Using the projected length of each section along the cutting and vertical directions, the measured force values in the cutting, $F_{c_{crit}}$, and vertical, $F_{v_{crit}}$, directions were transformed to local loads in the cutting, F_c, and vertical, F_v, directions acting on the center of each section as follows:

$$F_c = (ProjectedSectionDepth) \times \frac{F_{c_{crit}}}{L_z} \quad (1)$$

and

$$F_v = (ProjectedSectionWidth) \times \frac{F_{v_{crit}}}{L_x} \quad (2)$$

where L_z and L_x correspond to the maximum depth of the blade and width of the blade in contact with the tissue at the location of fracture.

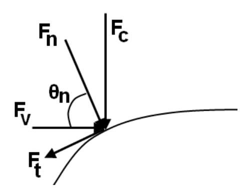

Fig. 4. Free body diagram of force distribution on scalpel blade.

Using this information, it was possible to determine the normal and tangential forces acting on each of section of the scalpel blade. Figure 4 displays a free body diagram of the force distribution of an individual section. Thus by using, θ_n, F_c, and F_v, we can determine the magnitude of the critical normal and tangential forces as:

$$F_n = F_c cos\left(90 - \theta_n\right) + F_v cos\left(\theta_n\right) \quad (3)$$

and

$$F_t = F_c cos\left(\theta_n\right) - F_v cos\left(90 - \theta_n\right) \quad (4)$$

Finally, using the defined section length and the information recorded earlier for the thickness of the scalpel blade (0.365mm), the forces can be transformed into the normal and tangential tractions, τ_n and τ_t, acting along the length of the blade as:

$$\tau_n = \frac{F_n}{SectionLength \times BladeThickness} \quad (5)$$

and

$$\tau_t = \frac{F_t}{SectionLength \times BladeThickness} \quad (6)$$

A depiction of the traction distribution using the 0.8mm sectioning can be seen in Figure 5. The cohesive strength parameters correlate to the maximum traction value in the normal and tangential direction. The exact values for the tractions can be found in Table I. As the magnitude of the section size decreased, the maximum tractions converged to a unique value. Therefore, values presented in Table I are for the smallest 0.2mm length sections.

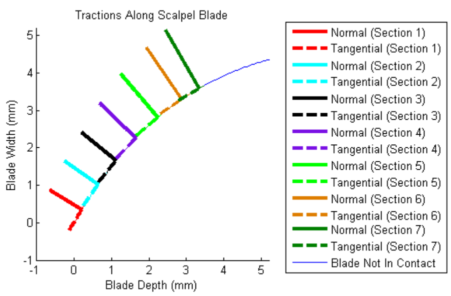

Fig. 5. Depiction of the tractions distributed along the scalpel blade using 0.8mm sections.

The next step was to determine the value of the cohesive energy for each cut segment. The cohesive energy is the total energy required to complete the fracture process; thus it encompasses a combination of the forces measured in the cutting and vertical directions. Using the force profiles from Figure 3, the total energy was found for each segment by integrating the forces in the cutting and vertical directions individually. The combination of these parameters constitutes the total cohesive energy of each cut segment. The cohesive energy values can be found in Table I for each cut segment.

TABLE I

COHESIVE PARAMETERS.

Cut Segment	τ_n (kPa)	τ_t (kpa)	Energy (mJ)
1	75.64	35.22	0.404
2	75.36	34.00	0.056
3	57.69	32.31	0.159
4	64.34	34.54	0.122
5	76.01	36.96	0.286
6	95.93	42.05	0.175
7	73.39	36.39	0.180
8	85.76	42.73	0.134
9	97.96	46.17	0.146
10	113.71	51.43	0.536
11	84.20	44.42	0.181
12	129.81	52.08	0.528
13	103.49	56.14	0.194
14	92.61	45.86	0.144
15	102.49	47.01	0.192
16	143.48	62.20	0.229
17	139.69	59.03	0.262
18	101.34	49.67	0.137
Average	95.16	44.90	0.225

III. CONCLUSIONS AND FUTURE WORKS

Through this process, fully defined cohesive zone models have been developed for the scalpel cutting of porcine liver. As can be noted in Table I, there is a large amount of variation present in the magnitude of the cohesive parameters. This is due to the highly variable nature of the soft-tissue itself. Although an average data set has been found, the soft-tissue variability is an important characteristic that should be incorporated into any simulation system to increase the realism of the training. Therefore, when including the cohesive zone models into a finite element simulation, a selection of individual cohesive parameters for each cut could be used to increase the realism by changing the cutting characteristics slightly for each cut segment.

The next step in the project is to use a commercially available finite element program to validate the models derived from the *in vivo* scalpel cutting tests. To do this, the cohesive elements will be included in the simulation along the cutting path with the cohesive parameters presented in Table I. The resultant tissue behavior and reaction forces on the scalpel blade will then be compared to the experimentally determined values to check the validity of the model.

Upon validation, the cohesive zone model will be combined with the models developed for general soft-tissue deformations to create a more comprehensive, reality-based surgical simulator. By including a physics-based cutting model, such as the proposed cohesive zone approach, realistic haptic feedback will be possible not only for general tool-tissue interactions, but for more detailed aspects of the medical procedure such as cutting or dissecting of soft-tissue.

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