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CHARACTERIZATION OF AORTIC TISSUE CUTTING PROCESS: EXPERIMENTAL INVESTIGATION USING PORCINE ASCENDING AORTA

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Abstract

Understanding biomechanical responses during soft tissue cutting is important for developing surgical simulators and robot-assisted surgery with haptic feedback. The biomechanics involved in the aortic tissue cutting process is largely unknown. In this study, porcine ascending aorta was selected as a representative aortic tissue, and tissue cutting experiments were performed using a novel tissue cutting apparatus. The tissue cutting responses under various cutting conditions were investigated, including differing initial tissue lateral holding force and distance, cutting speed, cutter inclination angle, tissue anatomical orientation and thickness. The results from this study suggest that a “break-in” cutting force of about 4 – 12 N, a cutter “break-in” distance of 5 – 15 mm, and a continuous cutting force of 2 – 4 N were needed to cut through the porcine ascending aorta tissue. For all testing conditions investigated in this study, the cutting force vs. the cutter displacement curves exhibited similar characteristics. More importantly, this study demonstrated that tissue cutting involving one or more of the following conditions: a larger lateral holding force, a smaller lateral hold distance, a higher cutting speed or a larger inclination angle, could result in a smaller “break in” cutting force and a smaller “break-in” distance. In addition, it was found that the cutting force in the vessel longitudinal direction was larger than that in the circumferential direction. There was a strong correlation between the tissue thickness and the cutting force. The experimental results reported in this study could provide a basis for understanding the characteristic response of aortic tissue to scalpel cutting, and offer insight into the development of surgical simulators.

Keywords

Soft tissue; tissue cutting; ascending aorta; cutting variables; minimally invasive surgery

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INTRODUCTION

A quantitative understanding of tissue-tool interactions is a critical and enabling step towards realistic haptic (force and tactile) modeling in surgical simulators and robot-assisted surgery [1, 2]. Surgical simulators offer an efficient and economical means to train novice surgeons on various surgical procedures. Additionally, surgical simulators can also be used to develop new surgical techniques and serve as a pre-and intra-operative planning tool. However, in order to develop a realistic model of tissue-tool interactions, the soft tissue mechanical properties and failure characteristics while interacting with surgical tools are needed to be modeled accurately[1]. Recent advances in surgery have also focused on minimizing the invasiveness of surgical procedures. A significant paradigm shift has occurred for some procedures, particularly in the robot-assisted minimally invasive surgery (RMIS), in which surgeons no longer directly touch or see the structures on which they operate[3]. The RMIS holds great promise for improving the accuracy and dexterity of a surgeon and minimizing trauma to the patient. However, clinical success with RMIS has been marginal. It is hypothesized that the lack of haptic feedback presented to the surgeon is a limiting factor[4].

The core task of surgery is “cutting and sewing.” During the process of cutting or dissecting, the tissues undergo a physically “damaging” process. Understanding of the damage thresholds such as the amount of forces required to cut through the tissue is critical for assessing interaction mechanisms. Aside from the damaging forces, the tissue cutting process also involves many variables such as the nonlinear anisotropic tissue properties, cutting speed, tool geometry, relative motion and inclination angle of the tool with respect to the tissue. The impact of these variables on the tissue cutting process is significant and remains to be quantified.

Considerable efforts in previous studies of tissue-tool interaction have been focused on needle insertion forces [5–9] and experimental and simulated responses of various soft tissues during needle insertion [10–19] due to its minimally invasive nature. However, experimental and theoretical studies to characterize the soft tissue cutting process have been limited [1]. McCarthy *et al* [20, 21] studied the sharpness of edge blade and its influence on the forces generated during the cutting process, but the materials chosen were not soft tissue (i.e. polyurethane elastomers). For realistic soft tissue studies, fracture toughness of human skin was studied by Pereira *et al.* [22] via cutting with scissors and of porcine skin by Comley and Fleck [23] via a trouser tear test. Mahvash *et al.* [24, 25] modeled the forces applied to scissors during cutting chicken skin, as well as various other materials including paper, plastic and cloth. They also developed an analytical model based on the concepts of contact mechanics and fracture mechanics. Picinbono *et al.* [26, 27] implemented the large-displacement linear elasticity model of liver tissues for laparoscopic surgery. Their focus was to optimize the tissue deformation model rather than modeling tissue fractures and cutting forces, and thus no validation or comparing experimental and simulated data was reported. Experimental study of porcine liver cutting mechanics was performed by Chanthasopeephan *et al.* [28, 29] who investigated the impact of cutting speed and blade inclination angle. They showed that the liver tissue deformation resistance decreased as the cutting angle varied from 90 to 45 degrees and as cutting speed increased. Recently, Valdastrì *et al.* [30] developed a more advanced scalable experimental platform with a triaxial force-sensitive scalpel, by which cutting forces in 3-directions were measured for porcine lung, liver and kidney.

The surgical blade is one of the main surgical tools for dissecting cardiovascular tissues, and to our knowledge, the biomechanics involved in the surgical blade cutting process of cardiovascular tissue is largely unknown. In this study, porcine ascending aorta was selected

as a representative aortic tissue due to its relatively large size and easy availability. Tissue cutting experiments were performed using a novel tissue cutting apparatus. The tissue cutting responses under various cutting conditions including: differing initial tissue lateral holding force and distance, cutting speed, cutter inclination angle, tissue anatomical orientation, and thickness, were investigated.

MATERIALS AND METHODS

Tissue Preparation

Fresh porcine hearts from 6–9 months old pigs were obtained from the Animal Technologies Inc, Erwin Tyler, TX, and transported on ice by overnight courier shipment to our laboratory. The tissues were either tested immediately upon arrival or stored in a -20°C freezer until testing. All tests were completed within 7 days of tissue harvest. Stemper *et al.* 2007 [31] showed that there is no significant change in failure mechanics between fresh and frozen porcine aorta specimens stored at -20°C for up to 3 months. The $\sim 25 \times 30$ mm samples were excised from the ascending part of the aorta, and the circumferential and longitudinal directions were noted. All adherent fatty tissue was carefully removed. The thickness of the specimen was measured by using a Mitutoyo (model 7301, resolution: 0.01mm) non-rotating thickness gauge [32]. Thickness was measured and averaged from five different locations on the specimen.

Tissue cutting device

A novel tissue cutting apparatus was designed and fabricated, as shown in Fig. 1. The testing apparatus consisted of three linear actuators, two load cells (SMT S-type, Interface Inc, Scottsdale, Arizona) and a tissue holding fixture. The stepper motor unit (Parker-Hannifin, Rohnert Park, CA) has a maximum speed of 50 rev/s, and a resolution of 50,800 steps/rev, connected to a carriage using 1 thread/25.4 mm lead screws, resulting in a net movement of $1 \mu\text{m}/\text{step}$ [33]. The load cell has a maximum error of $\pm 0.05\%$ in nonlinearity, $\pm 0.03\%$ in hysteresis, and $\pm 0.02\%$ non-repeatability. A test specimen was held laterally by the tissue holding fixture in the direction perpendicular to the cutting direction. The holding force was adjustable by controlling the displacement of the two holding linear actuators. The two holding linear actuators were coupled such that a synchronized motor motion was enforced. Tissue clamping was realized by two holding plates on both sides of the tissue. To ensure uniform and secure tissue clamping, the holding plates were lined with coarse sandpaper. The lateral holding force was recorded by load cell 2 as shown in Fig. 1. The cutting motion was realized by the third linear actuator. The cutting force was recorded by load cell 1. The entire system allows for adjustment of the cutting speed, blade inclination angle and the blade being used. Displacement of the blade tip, cutting force, cutting speed, and holding force were measured simultaneously throughout the experiments. The blade was cleaned with alcohol after each cut to remove any tissue debris from the blade. The total time for tissue clamping and cutting was within 2–3 minutes.

Tissue cutting variables

To gain a quantitative understanding of the vessel tissue cutting process, in this study, we explored the impact of the tissue cutting variables listed in Table 1. These variables included the tissue initial lateral holding force and lateral holding distance, the cutting speed, the cutter inclination angle, the tissue anatomical orientation and thickness. The inclination angle is defined as the orientation of the cutting edge of the scalpel blade with respect to the tissue sample. At 0° the blade edge and tissue edge are parallel. A size #11 Bard-Parker scalpel (BD Medical, Franklin Lakes, NJ) was employed in the tissue cutting experiments. The blade has a thickness of 0.4 mm, a wedge angle of 23° and an edge radius of $2 \mu\text{m}$. For each testing condition, 3–5 specimens were tested. For the study of the tissue thickness

parameter, we selected a group of 16 specimens with varied thicknesses of 1.1 mm – 3.3 mm. In the following sections, unless noted otherwise, the following testing conditions were applied: the initial holding force was 2 N, the cutting speed was 30 mm/min, the holding distance was 10 mm, the inclination angle was 25 degrees, the averaged tissue thickness was 2.1 mm, and the cutting orientation was along the circumferential direction of the ascending aorta. One way ANOVA analysis with the Tukey test was performed for all pair wise multiple comparisons. Significance was determined by a p-value of 0.05 or less. All variables were reported as mean and standard error.

RESULTS

General characteristics of the aortic tissue cutting response

A representative tissue cutting response was shown in Fig. 2a for cutting and holding forces vs. tool displacement curves. The solid (blue) curve represented the tissue cutting force and the dashed (pink) curve represents the tissue lateral holding force. To quantify the characteristics of the tissue cutting process, we identified the characteristic points labeled as “a”, “b”, “d”, and “e” on the curves. All experimental data we obtained followed a similar pattern as that shown in Fig. 2a. It can be seen that on the solid (blue) line, for this particular specimen, the cutting force increased from 0 N to 8 N before the blade cut into the tissue at point a. Once the blade broke the surface of the tissue, there was a sudden drop of the cutting force to 2 N at point b. The cutting force then remained low and fluctuated about 3 N. The force fluctuation might be due to the cutting energy accumulation and release processes during the tissue cutting. The tissue lateral holding force, the dashed (pink) curve in Fig. 2a, was one of the cutting parameters that affected the cutting force. As shown in the Fig. 2a, the holding force similarly increased from the initial 4 N to about 6 N before the blade cut into the tissue at point d. After point d, there was a sudden drop of the holding force back to the initial 4 N at point e. The magnitude of the holding force continuously decreased as the cutting process progresses. This decrease of the holding force might be due to the fact that there was less tissue left to be held as the cutting progresses.

In Fig. 2a from *b* and *e* points onwards, the cutting force and holding force both decreased. However, when Fig. 2a was converted into Fig. 2b by changing the holding force to the hold tension (holding force/uncut tissue length), it can be seen that even though the holding force was smaller, the holding tension was actually increased due to the smaller uncut tissue length. Note that uncut tissue length was obtained by using the initial length of the tissue minus the blade displacement. Due to the fact that the tissue may wrinkle in front of the blade during the cutting, the uncut length calculated was an approximated value. Therefore, the hold tension was approximated as well.

Impact of initial holding force

A total of 25 specimens were tested, with 5 specimens in each group and 5 initial holding forces of 0, 2, 4, 6 and 8 N. The averaged tissue thickness was about 2.1 mm. The cutting forces vs. cutter displacement curves were shown in Fig. 3a from four specimens with 0, 2, 4, and 6 N initial holding forces. It can be seen that the initial holding force had a significant impact on the tissue cutting force. The cutting force increased more quickly for a larger initial holding force prior to the “break-in” point. However, the “break-in” cutting force and “break-in” cutter displacement decreased with the application of a larger initial holding force.

As shown in Fig. 2b, as the initial holding force incrementally increased from 0 to 6 N, the “break-in” cutting force F_a decreased from 9 to 5 N, the mean cutting force F_c and the “break-in” distance L_a also decreased. One way ANOVA analysis indicated that a

significant difference between “break-in” cutting forces occurred for initial hold forces of 0 and 6 N ($p<0.001$), 2 and 6 N ($p=0.01$), and 4 and 6 N ($p=0.038$). For the mean cutting force F_c , a significant drop also occurred at the 6 N initial holding force ($p<0.05$). Significant differences were observed between 0 and 4 N ($p=0.027$), 2 and 6 N ($p<0.01$), and 4 and 6 N ($p=0.031$) for the L_a as well.

Impact of tissue holding distance

A total of 25 specimens were tested with 5 specimens in each group for 5 groups with holding distances of 2, 5, 10, 15 and 20 mm. The averaged tissue thickness was 2.2 mm. It can be seen in Fig. 4a that a larger tissue lateral holding distance resulted in a larger “break-in” cutting force. The “break-in” cutting force almost tripled from 4.5 N to 12 N with the increased lateral holding distance from 2 mm to 20 mm. A significant increase in the “break-in” cutting force was found between the 15 mm and 20 mm holding distances ($p=0.048$). However, there was no statistically significant difference between the mean cutting forces after the “break-in” point, with each varying between 2 – 4 N. The magnitude of force oscillation was smaller with a small holding distance. For the tissue holding distance of 20 mm, we observed a large degree of tissue buckling at the tool tip. Significant differences in the mean L_a were observed between initial holding distances of 2 mm and 10 mm ($p=0.011$), as well as 5 mm and 15 mm ($p=0.007$).

Impact of cutting speed

A total of 25 specimens were tested with 5 specimens in each group for 5 groups with cutting speeds of 7.5, 15, 30, 75 and 150 mm/min. As shown in Fig. 3a, with other cutting parameters kept unchanged, a higher cutting speed resulted in a lower “break-in” cutting force and a shorter “break-in” distance L_a . The “break-in” cutting force doubled from 4 to 8 N as the cutting speed decreased from 150 to 7.5 mm/min. A significant difference between the “break-in” cutting forces was found between the 15 and 75 mm/min cutting speed ($p=0.039$). After the “break-in” point a, the mean continuous cutting forces were similar regardless of the cutting speed. There was a significant difference in the mean cutting forces between speeds of 7.5 mm/min and 150 mm/min ($p=0.01$). For all other conditions there was no difference between the mean cutting forces. There was also a significant difference observed for the mean L_a values between cutting speeds of 7.5 and 75 mm/min ($p<0.05$), and 15 and 150 mm/min ($p<0.05$).

Impact of tool inclination angle and tissue orientation

A total of 15 specimens were tested with 5 specimens in each group for 3 groups with tool inclination angles of 0, 25 and 45 degrees. It can be seen from Fig. 6a that a larger tool inclination angle resulted in a smaller “break-in” cutting force. With the 45 degree inclination angle, the “break-in” cutting force was less than 4 N, whereas with the 0 degree inclination angle, the “break-in” cutting force was over 10 N. Significant differences in the “break-in” cutting forces were observed for all pair-wise comparisons between cutting inclination angles of 0, 25 and 45 degrees. However, there were no significant differences among the mean cutting forces for all three inclination angles. A significant difference was observed in the “break-in” distance L_a for pair-wise comparisons between inclination angles of 25 and 45 degrees ($p=0.004$), as well as 0 and 45 degrees ($p<0.001$).

To study the impact of the tissue vessel orientation, 4 specimens were cut in each orientation (circumferential and longitudinal). It can be seen from Fig. 6b that a larger “break-in” cutting force was required for cutting along the longitudinal orientation than the circumferential orientation. We also observed that the amplitude of the cutting force after the “break-in” point was greater when cutting along the longitudinal orientation. It is shown in Fig. 6b that the mean cutting force and the average “break-in” distance were also larger

when cutting along the longitudinal orientation than along the circumferential orientation. The “break-in” cutting forces were significantly different with mean values of 5.99 and 8.23 N ($p=0.029$) along the circumferential and longitudinal orientations respectively. The mean cutting forces were 2.83 and 3.30 N. The average “break-in” distances were 8.24 and 9.13 mm. However; there were no statistically significant differences in the F_c or L_a between the two orientations.

Impact of tissue thickness

A group of 16 tissue specimens were selected with continuous thickness variation from 1.45 mm to 3.3 mm. It can be seen from Fig. 7a that thicker tissue required larger peak and mean cutting forces, as one would expect. In addition, the “break-in” distance also increased with the tissue thickness. There were strong correlations between the thickness and the three cutting parameters, i.e., the peak cutting force, the mean cutting force and the “break-in” distance ($p<0.001$) (Fig. 7b). Furthermore, when the peak cutting force, the “break-in” distance and the mean cutting force were normalized with respect to the tissue thickness, the trend lines of the normalized cutting forces stayed flat, indicating that the normalized cutting forces were nearly independent of the tissue thickness.

DISCUSSION

Reducing the “break-in” cutting force is desirable during surgery as an elevated “break-in” cutting force may also cause damage to surrounding tissues. Furthermore, as shown in this study, a larger “break-in” force is often associated with a larger “break-in” distance. In many surgical operations, particularly in minimally invasive procedures, the operating space is very limited. Therefore, the quantification of tissue cutting characteristics and knowledge of tissue cutting mechanisms is essential for minimizing the “break-in” force and distance while maintaining a low mean cutting force in a surgical operation. The results from this study suggest that about 4 – 12 N of “break-in” force, 5 – 15 mm “break-in” distance, and 2 – 4 N continuous cutting force were needed to cut through the porcine ascending aorta tissue under various cutting conditions investigated herein.

The aortic vessel wall has complex tissue structures with layers of crimped collagen and elastin fiber bundles at the un-loaded state[34]. The tissue fibers are un-crimped and tensioned during the cutting process due to the interaction between the blade and the tissue. It has been shown that a higher stiffness and/or a larger stored strain energy of a material would facilitate crack initiation and propagation [35]. Similarly, with the application of an initial holding force (Fig. 3), the tissue is pre-tensioned, which increases the instantaneous tissue stiffness and stored strain energy, and thus, facilitates cutting. Consequently, the “break-in” cutting force and “break-in” cutter displacement decrease. In our experiments, as the initial holding force increased from 0 to 8 N, the “break-in” cutting force significantly decreased from 9 to 5 N.

This study demonstrated that the tissue lateral holding distance also has a significant impact on the tissue cutting process. In surgical operations, the lateral holding of the tissue resembles the use of forceps or fingers to hold the surrounding tissues to assist tissue dissection. Indeed, in many surgical procedures, various tools are designed to hold tissues to facilitate tissue dissection. Our experimental results showed that the tissue cutting force and the cutter displacement at the “break-in” point decreased with smaller tissue holding distances. We also observed that with a larger holding distance, the tissue would wrinkle in the cutting direction, causing the tissue to fold and accumulate in the front of the blade, instead of being cut. Similar phenomenon was also observed by Lake and Yeoh [35] and McCarthy *et al.* [21] when cutting elastomeric materials using blunt blades. With a smaller lateral holding distance, the tissue in between would be tensioned relatively easily, thus

resulting in a lower “break-in” force and distance. Our results indicated that a lateral holding distance less than 15 mm was desirable for a surgical dissection of porcine ascending aorta.

Chanthasopeephan et al. [28, 29] performed tissue cutting experiments on porcine livers using #10 Bard-Parker stainless steel surgical blades with seven disparate speeds from 60 mm/min to 1524 mm/min. Their results indicated that a larger cutting speed led to less resistance to local tissue deformation during the cutting process. The impact of cutting speed on the cutting process of soft materials was also investigated using polyurethane materials [20, 21]. Scalpel cutting with different speeds (10, 250 and 500 mm/min) was investigated. It was found that the cutting force at a speed of 250 mm/min was much less than that at a speed of 10 mm/min, but very similar to that at 500 mm/min. Similarly, as illustrated in Fig. 5b in our study, the “break-in” force and “break-in” distance decreased with an increased cutting speed, but the mean cutting force after the “break-in” point was relatively similar in all cases. There was a significant difference between the 15 and 75 mm/min cutting speeds, however, no significant difference between the 75 and 150 mm/min was found. Due to the limitation of our testing apparatus, we did not investigate cutting speeds above 150 mm/min, which may warrant a future study.

Mechanical properties along the circumferential and longitudinal directions of porcine ascending aorta have been quantified [36]. It was found that the tissue has higher stiffness in the circumferential direction than in the longitudinal direction [36]. Histological and morphological studies also demonstrated that collagen and smooth muscle fibers are more aligned along the circumferential direction [37]. In this study, the results indicated that the “break-in” cutting force, the “break-in” distance and the mean cutting force when cutting along the longitudinal orientation were larger than those along circumferential orientation. The higher cutting force in the longitudinal direction could be explained by the fact that more tissue fibers need to be cut in the longitudinal direction. We also observed that the cut surface of the tissue along the circumferential orientation is rougher than that along the longitudinal orientation.

There are several limitations in this study. We treated the ascending aorta as a homogenous material. The arterial wall consists of three layers: the intima, media and adventitia layers. Although heterogeneous across the wall thickness, from the biomechanical perspective, the aortic wall is often treated as a single layer, homogenous tissue [38–40]. Layer-specific tissue mechanical properties have not been well characterized. The sample size for each cutting condition was relatively small. A larger sample size would provide more conclusive results. This paper only includes the experimental investigation of the porcine aortic tissue cutting response. Modeling of the experimental data to study the tissue fracture mechanism is currently ongoing and will be reported in the future. Finally, considering the effects that age and disease state have on the mechanical properties of human aortic tissues, the results presented here can only offer insight in the cutting characteristics of the porcine ascending aorta under various cutting conditions. Further studies are warranted for comparison between human and animal aortic tissues.

CONCLUSION

In this study, the characteristic responses of the porcine ascending aorta to blade cutting were investigated. The effects of major cutting parameters including the initial holding force, lateral holding distance, cutting speed, tool inclination angle, tissue orientation and thickness were investigated. The results demonstrated that the measured cutting force vs. cutter displacement curves for all testing conditions exhibited similar characteristics: a J-shaped curve was obtained before the blade cut into the tissue, and once the blade broke the surface of the tissue, the cutting force drastically reduced as a crack was created in the tissue

and propagated quickly. Consequently the cutting force remained low, but fluctuated. About 4 – 12 N of “break-in” forces, 5 – 15 mm of cutter “break-in” distance, and 2 – 4 N of continuous cutting force were needed to cut through the porcine ascending aorta tissue. The cutting conditions involving a larger initial holding force, a smaller lateral hold distance, a larger cutting speed or a larger inclination angle could result in a smaller “break in” cutting force and a smaller “break-in” distance. The cutting force in the longitudinal orientation was larger than that in circumferential orientation. There was a strong correlation between the tissue thickness and the cutting force. The experimental results reported in this study could provide a basis for understanding the characteristic responses of aortic tissue to scalpel cutting, and offer insight in developing surgical simulators and designing robot-assisted surgical tools.

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Figure 1.

A novel tissue cutting apparatus consisting of three linear actuators, two load cells and a tissue holding fixture. a) The actual cutting device setup and b) a close-up view of the blade and the tissue during cutting. Two synchronized linear actuators are used to apply an identical lateral displacement of the two holding plates (in the direction of the yellow arrows) and the lateral holding force is measured by load cell 2. The third linear actuator is used to move the cutting blade in the direction of cutting (the direction of the red arrow) and the cutting force is measured by load cell 1.

Figure 2.

a) A representative tissue cutting response, marked with characteristic points of a, b, d, e and the corresponding forces of F_a , F_b , F_d and F_e . F_c and F_f are the mean cutting force and mean holding force after the blade cuts into the tissue, calculated by averaging all the force measurements after F_b and F_e , respectively; b) the impact of the initial holding force on the tissue cutting force, indicating that the higher the holding tension, the less the cutting force.

Figure 3.

a) Representative tissue cutting responses at the initial holding forces of 0, 2, 4 and 6 N. b) Impact of initial holding force on the peak tissue “break-in” force F_a , the mean cutting force F_c and the “break-in” cutter displacement L_a .

Figure 4.

a) Representative tissue cutting responses at holding distances of 2, 5, 10, 15, and 20 mm. b) Impact of holding distance on the tissue “break-in” cutting force F_a , the mean cutting force F_c and the “break-in” cutter displacement L_a .

Figure 5.

a) Representative tissue cutting responses at cutting speeds of 7.5, 30, 75, 150 mm/min. b) Impact of cutting speed on the tissue “break-in” cutting force F_a , the mean cutting force F_c and the “break-in” cutter displacement L_a .

Figure 6. Representative tissue cutting responses at a) the inclination angles of 0, 25, and 45 degrees, and b) tissue cutting orientations along the circumferential and longitudinal directions.

Figure 7.

a) Representative tissue cutting responses for different tissue thicknesses. b) Impact of tissue thickness on the tissue “break-in” cutting force F_a , the mean cutting force F_c and the “break-in” cutter displacement L_a . The scattered dots are experimental F_a , F_c and L_a , and the lines are fitted linear trend lines.

Table 1

Tissue cutting parameters

Parameters	Variable values						
Initial holding force, N	0	2	4	6	8		
Cutting speed, mm/min	7.5	15	30	75	150		
Inclination angle, degree	0°			25°		45°	
Holding distance, mm	2	5	10	15	20		
Anatomic orientation	longitudinal and circumferential directions						
Thickness, mm	1.1 ~ 3.3						