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Estimation of fracture toughness of liver tissue: Experiments and validation

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ABSTRACT

The mechanical interaction between the surgical tools and the target soft tissue is mainly dictated by the fracture toughness of the tissue in several medical procedures, such as catheter insertion, robotic-guided needle placement, suturing, cutting or tearing, and biopsy. Despite the numerous experimental works on the fracture toughness of hard biomaterials, such as bone and dentin, only a very limited number of studies have focused on soft tissues, where the results do not show any consistency mainly due to the negligence of the puncturing/cutting tool geometry. In order to address this issue, we performed needle insertion experiments on 3 bovine livers with 4 custom-made needles having different diameters. A unique value for fracture toughness ($J = 164 \pm 6\,\mathrm{J/m^2}$) was obtained for the bovine liver by fitting a line to the toughness values estimated from the set of insertion experiments. In order to validate the experimental results, a finite element model of the bovine liver was developed and its hyper-viscoelastic material properties were estimated through an inverse solution based on static indentation and ramp-and-hold experiments. Then, needle insertion into the model was simulated utilizing an energy-based fracture mechanics approach. The insertion forces estimated from the FE simulations show an excellent agreement with those acquired from the physical experiments for all needle geometries.

1. Introduction

The last two decades have witnessed significant advances in the fields of medical robotics, image-guided surgery, and computer-aided surgical planning and simulation. In all of these fields, accurate modeling of the interaction forces between surgical instruments and soft organ tissues has proven crucial for both the realistic simulation and the proper execution of the medical procedures. However, the precise estimation of these forces through a model requires the knowledge of material properties of the soft organ tissues targeted by the surgical instruments. While many of these properties have already been extensively examined, some are left unnoticed, such as fracture toughness, the resistance of a material to fracture. Only in a few exceptional studies, the emphasis was placed on estimating the fracture toughness of a soft biological material [1–6], investigating the geometrical effects of the instruments in tissue penetration through models based on fracture toughness [7], and measurement of the interaction forces via needle insertion experiments [8–10].

The fracture toughness of the soft tissue targeted by the surgical tools plays a critical role in medical procedures, such as catheter insertion, robotic-guided needle placement, suturing, cutting or tearing, and biopsy. Specifically, all these procedures involve tissue damage to a certain extent, which should be kept to a minimum in order to avoid any medical complications [11–13]. Thus, the knowledge of fracture related material properties, especially the fracture toughness, is of utmost importance.

Despite the significant amount of work carried out to determine the fracture toughness of hard biomaterials, such as bone [14] and dentin [15], estimation of fracture toughness of the soft tissues has come under the focus of only a limited number of studies that mostly rely on a fracture mechanics approach based on energy balance [16,17]. For instance, Azar and Hayvard [1] inserted suture, syringe and biopsy needles with diameters ranging from 0.71 mm to 2.1 mm into porcine liver to calculate the crack size and the fracture toughness of the liver. In particular, two consecutive insertions were made into the same spot on the liver; the first one creating the crack and the second one being a free-pass. Then, the fracture toughness was calculated by taking the difference between the fracture and the viscoelastic works first, and then dividing the difference by the crack area. The calculated fracture toughness of the porcine liver varied between 75.8 and 185.6 $\mathrm{J/m^2}$. Chanthasopeephan et al. [2], on the other hand, employed a scalpel as the cutting tool, and the fracture toughness of pig liver was estimated to vary between 186.98 and 224.83 $\mathrm{J/m^2}$, with a standard deviation reaching to 142 $\mathrm{J/m^2}$ in some experiments. Cutting with scissors was considered by Pereira et al. [3] to estimate the fracture toughness of the human skin, where samples were obtained from the hands of two cadavers. The fracture toughness of the dorsal skin was estimated as $1777 \pm 376\,\mathrm{J/m^2}$ along the longitudinal

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direction and as $1719 \pm 674\,\mathrm{J/m^2}$ along the circumferential direction, while the palmar skin had an estimated fracture toughness of $2365 \pm 234\,\mathrm{J/m^2}$ along the skin creases and $2616 \pm 395\,\mathrm{J/m^2}$ across the skin creases. Comley and Fleck [4] estimated the toughness of porcine dermal and adipose skin tissue (soft connective tissue under the dermal layer) via a trouser tear test as $17{,}000\,\mathrm{J/m^2}$ and $4100\,\mathrm{J/m^2}$, respectively. Misra et al. [5] used an experimental set-up to robotically steer Nitinol needles having different diameters and bevel tips into 3 plastisol gels in different stiffnesses and a porcine gel. Using a single insertion, the rupture toughness of the plastisol gels, ordered in increasing stiffness, was estimated as 115.40, 218.19 and $221.04\,\mathrm{J/m^2}$. The rupture toughness of the porcine gel, on the other hand, was estimated as $82.28\,\mathrm{J/m^2}$. Utilizing a trouser tear test, Shergold [6] measured the fracture toughness of the silicone rubbers of grades Sil8800 and B452 as $3100\,\mathrm{J/m^2}$ and $3800\,\mathrm{J/m^2}$, respectively. In an attempt to corroborate their penetration models for sharp-tipped and flat-bottomed punches [18], Shergold and Fleck [7] carried out penetration experiments on skin and skin-like silicone rubber, where they investigated the effect of the punch-tip geometry on the mechanics of penetration based on the experimental data obtained in [6].

As reported in the literature review, the toughness values estimated in the earlier studies make up a wide range, partially due to the differences in the material properties of the subjects, and partially owing to the methods chosen for testing and evaluation. Although it is not uncommon for the material properties of samples extracted from different animals to show variation due to the individual differences, we hypothesize that part of the large variation in the estimated values is due to the neglected effect of the puncturing/cutting tool geometry on the measurements. In particular, we point out that, even though the energy method has been used to evaluate the fracture toughness of soft tissues in the earlier needle insertion studies, no attention has been paid to the role of needle geometry in these evaluations. To prove this hypothesis, needle insertion experiments were performed in the current study on 3 bovine livers with 4 custom-made needles having different diameters, and the relationship between the fracture toughness and the needle diameter was investigated in detail. In order to validate the experimental results, FE simulations of the needle insertion process were carried out in ANSYS. The good agreement between the computed and experimentally measured interaction forces indicated that the current approach could be successfully adopted in medical procedures that require precise control of tissue cutting forces.

2. Materials and methods

2.1. Theory

The insertion of a needle into a soft tissue can be investigated by dividing the process into multiple distinct phases [8,16,17,19,20]. The process starts with the deformation of the soft tissue under the force exerted by the needle (Fig. 1). Due to the viscoelastic nature of the soft tissue, this deformation continues until a certain threshold is reached in the relation between the viscoelastic work, W_v, and the fracture work, W_f. In particular, during the deformation phase, the value of W_v remains larger than the W_f. As the needle penetrates deeper into the soft tissue, the W_f starts to increase, and eventually becomes equal to the W_v. When the value of the W_f surpasses the value of the W_v, the needle punctures the tissue and rupture occurs. This marks a very brief change of state in the process of insertion; with the occurrence of rupture, the stage of pure deformation ends and a mixed stage of penetration and deformation starts (Fig. 1). At this stage, as the needle continues its movement through the soft tissue, the forces tend to increase until the needle comes to a full stop. A phase of relaxation follows, as the motion of the needle comes to an end and the soft tissue remains in this phase until the needle is extracted from the tissue.

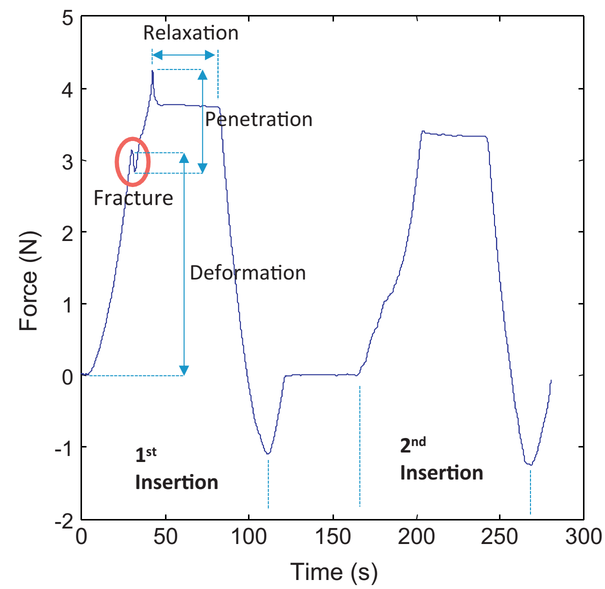

Fig. 1. The phases of needle insertion into soft tissue.

In order to determine the fracture toughness, J, via the energy balance equation, a method involving two subsequent insertions of a needle into the same spot was suggested in [8,17]. In the first insertion, all stages of insertion, namely deformation, rupture, and penetration, are present (Fig. 1). As a result, the energy balance equation for the first insertion is:

$$F_1\,du = J\,d\Delta + d\Delta + P\,du \tag{1}$$

where F_1 is the force acting on the needle during the 1st insertion and du is the change in the needle displacement. Hence $F_1\,du$ is the total work done by the needle during the first insertion, P is the friction force and $P\,du$ is the work done by friction. In Eq. (1), dA is the change in crack area (circumference of the needle times the incremental needle displacement, where the total crack area is the circumference times the current depth). Assuming that friction in the system is accounted for following the rupture, the sum of $J\,dA$ and the friction work becomes equal to the fracture work, W_f; whereas, $d\Delta$, the change in the strain energy, is equal to the viscoelastic work, W_v. As a result, the total work becomes equal to the sum of W_f and W_v.

During the second insertion to the same spot (Fig. 1), which is a free pass, only the penetration stage exists, such that the governing equation becomes:

$$F_2\,du = d\Delta + P\,du \tag{2}$$

where F_2 is the force acting on the needle during the second insertion and has to be smaller than F_1. Since no rupture occurs during the second insertion, the value of fracture work is equal to zero. Since the change in strain energy, $d\Delta$, and the work done by friction, $P\,du$, are exactly the same for both insertions, the subtraction of Eq. (2) from Eq. (1) results in:

$$(F_1 - F_2)du = J\,dA \tag{3}$$

If the left and right hand sides of the above equation are integrated with respect to u, and the lower and upper limits of the integral are