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The significance of rate dependency in blade insertions into a gelatin soft tissue phantom

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A B S T R A C T

An improved understanding of the detailed tool–tissue interactions in soft tissue phantoms is necessary to optimise the design and actuation of novel surgical devices. Incorporating rate dependency of these interactions provides general, rather than specific, working conditions. Blade insertion experiments and accompanying finite element simulations examine contact interactions, strain energy release rate and deformation in a gelatin phantom with a modulus of approximately 10 kPa. Insertion rates of 0.25–2 mm/s show frictional response and strain energy release rate more than doubling. This demonstrates the importance of testing phantoms over the full range of operating conditions. The role of the coefficient of friction between the tool and the tissue and the related frictional energy expenditure are also studied and discussed.

1. Introduction

Steering a needle tip to a target deep within soft tissue is a challenge that has attracted significant recent attention. The potential applications and benefits of such a steerable device are outlined well in [1]. Other devices have been presented that also seek to attain a similar goal [2–5]. Amongst these research efforts, bevel-tipped needles use asymmetric forces at the needle tip to direct the device ahead of flexible actuating cable. A different type of bevel-tipped device constructed using four interlocking segments – each segment flexible in its own right and capable of independent actuation – is presented in Ko et al. [6]. It is this four-part device that forms the basis of the experimental and numerical investigation presented here.

In the development and testing of biomedical devices for use in soft tissue it is normal to use tissue phantoms. These offer the benefit of controllable, consistent mechanical properties as well as being readily available at relatively low cost. Gelatin has been used as a viable soft tissue phantom for its low elastic modulus of about 10 kPa and similar density to biological tissues. Understanding interactions between the gelatin tissue phantom and the four-part probe is therefore critical to the development of the device itself. This development can take the form of design improvements [7] or revised actuation strategies.

Manufacturing prototype devices, particularly those that are both novel and complex, is time consuming and costly. Finite element models are a useful way to test ideas and gain a greater understanding of experimentally observed behaviour as a result of the extra diagnostic insight that they offer. It is therefore vitally important to create finite element models with accurate parameters that truly reflect all of the significant physical phenomena that occur during needle insertions. A modelling process that captures the cutting of gelatin and contact interactions was implemented in [8]. The cohesive zone model proposed by [7] was implemented to capture material failure during needle insertions.

The significance of friction in needle insertions into soft tissue has been demonstrated when applied to the cycling of a bevel-tipped device [9]. Katoaka et al. [10] also sought to separate the effects of friction from cutting forces using a bespoke device for needle insertions. Cutting by a needle in soft tissue is too important to be neglected in detailed numerical models. The strain energy release rate is used as a convenient means of encapsulating a variety of phenomena that lead to the failure of brittle materials [7,11,12]. An advantage of the strain energy release rate is its compatibility with a cohesive approach to material failure in finite element simulations.

It is also clear that the significance of friction to the needle insertion process merits further investigation into the model used. In particular, the coefficient of friction can be expected to show some degree of velocity dependence. Studies on gelatin have also revealed that the strain energy release rate is dependent on the rate at which loading is applied [13]. Consequently, opportunities may arise where these more complex, rate dependent behaviours can inform optimal actuation strategies for needle insertions.

Generally it appears that interactions are rarely studied from the perspective of varying substrate properties. More normally

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modifications to needle geometries and actuation strategies are presented during the development process [14,15]. However, without a more complete understanding of the phantom behaviour and the intrinsic characteristics of the tool–tissue interactions, it is difficult to achieve optimisation in a fully integrated design strategy. In particular, once the rate dependence of the tissue is considered, the range of some input parameters such as insertion speed can be usefully bounded and exploited.

In [8], finite element modelling with cohesive failure was used to simulate needle insertion into a gelatin tissue phantom. Computational requirements for the finite element algorithms dictated that 2D plane-strain models were used to calibrate material parameters. The plane-strain assumption, in a 3D process with a cylindrical needle and a planar crack of finite width, led to the calibration of strain energy release rate in the cohesive elements being unduly sensitive to the measured crack width. Here, by implementing a revised experimental method using a wide blade instead of a cylindrical needle, a plane strain condition is largely enforced and the influence of any uncertainties in the planar crack width is made insignificant.

Many studies [1,3,5,10] perform experimental needle insertions to quantify the performance of a particular configuration of steerable needle. This is not the aim of the work presented here. Instead, parameters are sought that enable predictive numerical simulations to be built, based on quantified interaction models between a tool and the surrounding tissue phantom. Although the studies presented in [7–9,15] also had similar aims, here the experimental observations are used to extract information which enable characterisation of both the cutting process and the frictional behaviour of the tool–tissue interaction. These are analysed in a single tissue phantom, used in the work of [6,14], over a range of realistic insertion rates. The benefit of such an approach is the contribution towards versatile, high resolution and predictive finite element models of tool–tissue interactions. These finite element models can then be used in the development of optimal tool geometries and actuation strategies.

The theory behind this approach is outlined in the following section. Subsequently, a refined experimental method and associated numerical model are described including the range of insertion rates typically seen in the development of percutaneous needles and probes. Results and their discussion are combined and include rate dependent coefficients of friction and strain energy release rates. A summary of the main conclusions and areas requiring further research in the future close the paper.

2. Theory

Strain energy release rate, G_c, is commonly used as an all-encompassing term for the processes that lead to cracking. Consequently G_c is now used to quantify failure between bonded layers of material [16] and in complex crack propagation investigations [17,18] as well as in simulating needle insertions [19]. The theoretical basis for establishing G_c for use in a soft tissue phantom is reported in [11,12] and is also described in some detail in Oldfield et al. [8].

Strain energy release rate is immediately compatible with the 'energy dissipated due to failure' in the traction–separation relationship defining cohesive elements in ABAQUS finite element software [20]. As such, it is an effective way to model the damage induced by a cutting tool on a soft tissue phantom—described numerically by a finite element discretisation of its continuum representation.

During needle insertions it is assumed that the damage or crack propagation can be identified through the following energy balance

$$W_{ext} = W_\epsilon + W_f + W_{G_c} \tag{1}$$

where W_{ext} is simply the work done by the externally applied force on the insertion device; W_ϵ is the deformation energy in the gelatin tissue phantom; W_f is the work done to overcome frictional resistance and W_{G_c} is the work done in creating a crack surface. The terms on the right hand side of Eq. (1) consequently need more careful consideration.

Fig. 3 shows that there are two characteristic peaks in the force-displacement responses when inserting a blade into a gelatin tissue phantom. The first peak force is associated with the displacement at which a crack is initiated, δ_{ic}, while the final peak, δ_{tm}, is associated with the displacement at which the blade passes out of the tissue phantom. One notable evolution of the theory in [8] is that W_f is assumed to be negligible until δ_{ic} is reached

$$W_f = \int_0^L F_f(x)\,\mathrm{d}x \tag{2}$$

where $F_f(x)$ is the depth-dependent resistance due to friction and L is the total displacement. The relationships defining $F_f(x)$ are as follows:

$$F_f(x) = \begin{cases} 0 & \text{if } 0 \le x \le \delta_{ic} \\ \dfrac{(x-\delta_{ic})F_{ss}}{(\delta_{tm}-\delta_{ic})} & \text{if } \delta_{ic} < x \le \delta_{tm} \\ F_{ss} & \text{if } \delta_{tm} < x \end{cases} \tag{3}$$

with F_{ss} the 'steady state' resistance to insertion following relaxation from the peak associated with full penetration of the gelatin tissue phantom. A further departure from theory presented in [8] is that the phantom is assumed to be viscoelastic and W_ϵ includes creep dissipation, W_{CD}, viscous dissipation, W_{VD}, and elastic strain energy, W_E

$$W_\epsilon = W_E + W_{CD} + W_{VD} \tag{4}$$

All of the parameters in Eq. (4) are established numerically using finite element simulations.

The remaining term (W_{G_c}) in Eq. (1) leads to G_c for an out of plane crack depth, D, as follows:

$$G_c = \begin{cases} \dfrac{W_{G_c}}{LD} & \text{if } \delta_{ic} \le x \le \delta_{tm} \\ \dfrac{W_{G_c}}{\delta_{tm} D} & \text{if } \delta_{tm} < x \end{cases} \tag{5}$$

with the assumption that W_{G_c} is negligible until the point δ_{ic}, as the transition from elastic deformation to crack formation is rapid—with little energy lost in unrecoverable deformation.

Once established, G_c is input into the traction–separation laws described in [8] and implemented in the finite element software ABAQUS [20]. The remaining parameters of the bilinear traction–separation law are: K_i, the initial elastic stiffness; σ_y, the stress at the elastic limit; δ_y, the separation at the elastic limit; δ_0, the separation at element failure. After fixing G_c, the area under the traction–separation curve, all of the remaining parameters in the traction–separation relationship can be manipulated to ease numerical problems and alter the overall force–displacement response. One essential requirement is that δ_0 is smaller than the width of the insertion device. Without this constraint, the cohesive element will possess some residual strength after the insertion device has passed between its surfaces. This would imply that the crack has not fully formed (contrary to what experimental evidence shows).