**Graphical Abstract**

**Needle Design**

- Needle 1: Bevel (1) - 18°, Inclination (1) - 72°
- Needle 2: Bevel (2) - 25°, Inclination (1) - 63°

**Testing and Experiment setup**

- Static Load/Compressive Test
- Load Chart

**Modelling of needle design system (FEA)**

- 3D model of needle

**Analysis of needle design**

- Comparison of Experiment vs. Simulation for Needle 1 and Needle 2
- Graph showing deflection vs. insertion depth for Needle 1 and Needle 2

**Simulation Results (i.e. Deflection and Contact Stress)**

- Graphs showing deflection and contact stress for Needle 1 and Needle 2
Dynamic analysis of hollow needle insertion into soft gel - Coupled Eulerian Lagrangian based three-dimensional finite element modelling

Mohamed Gouse Jushiddi¹-³, John Mulvihill²-³*, Drahomir Chovan¹, Aladin Mani¹, Camelia Shanahan¹,², Christophe Silien²,³, Syed Ansar Md. Tofail¹-³, Peter Tiernan²,³,*

¹ Modelling, Simulation and Innovative Characterisation (MOSAIC), Bernal Institute and Department of Physics, University of Limerick, Limerick, Ireland.
² School of Engineering, Faculty of Science and Engineering, University of Limerick, Limerick, Ireland.
³ Bernal Institute, University of Limerick, Limerick, Ireland.

Mohamed Gouse Jushiddi¹-³ (E-mail: Mohamed.Jushiddi@ul.ie)
Dr. John Mulvihill²-³ (E-mail: John.Mulvihill@ul.ie)
Dr. Drahomir Chovan¹ (E-mail: Drahomir.Chovan@ul.ie)
Dr. Aladin Mani¹ (E-mail: Aladin.Mani@ul.ie)
Dr. Camelia Shanahan¹,² (E-mail: Camelia.Shanahan@ul.ie)
Dr. Christophe Silien²,³ (E-mail: Christophe.Silien@ul.ie)
Dr. Syed Ansar Md. Tofail¹ (E-mail: Tofail.Syed@ul.ie)
Dr. Peter Tiernan² (E-mail: Peter.Tiernan@ul.ie)

*Corresponding Authors (Postal address):
Dr Peter Tiernan
School of Engineering, Faculty of Science and Engineering, University of Limerick, Limerick, Ireland.
Tel #: +353-61213503;
E-mail: Peter.Tiernan@ul.ie

Dr John Mulvihill
School of Engineering, Department of Biomedical Engineering, University of Limerick, Limerick, Ireland.
Tel #: +353-61217719;
E-mail: John.Mulvihill@ul.ie
Abstract
Planning and practice of surgical procedures can be improved through the use of modelling. This study provides an insight into the hollow needle (i.e. cannula) and needle-tissue interactions using a modelling approach, thus enabling the optimisation of needle-tip designs not only for training but also for the planning of surgical procedures. Simulations of needle insertion into agar gel were performed using a Coupled Eulerian Lagrangian based finite element (FE) analysis, adapted for large deformation and tissue fracture. The experimental work covers needle insertion into 3% agar gel using a needle with a bevel tip of various angles, to assess the accuracy of the simulations. The simulated needle deflection and insertion force for two needles (i.e. Needle 1 with 18° bevel angle and Needle 2 with 27° bevel angle) were compared with corresponding experimental results. The contact stress (i.e. contact pressure) on the needles from the agar gel during the insertion of the needles were also studied. Observations indicate that varying the needle bevel angle from 27° to 18° results in a decrease of the peak force (i.e. puncture force) and an increase in needle deflection. Quantitatively, the percentage errors between the experimental data and the FE model for the total insertion force along the z-direction (i.e. Z Force) for Needle 1 and 2 were 4% and 4.8% (p > 0.05), respectively. Similarly, needle deflection percentage errors along the x-z plane were 5.7% and 10% respectively. Therefore, the forces and needle deflection values predicted by the simulation are a close approximation of the experimental model, therefore validating the Coupled Eulerian Lagrangian based FE model.

Keywords: Hollow-Needle insertion; finite element analysis; coupled eulerian lagrangian method; dynamic analysis; bevel-tipped needle; three-dimensional.
1. Introduction

Needles are ubiquitous medical tools widely used in minimally invasive surgical and percutaneous procedures such as injection, regional anesthesia, blood sampling, biopsy and brachytherapy [1-3]. They come in a wide variety of needle tip configurations, such as bevel-tip, lancet tip, back-bevel tip, trocar, franseen, conical etc. depending on the medical procedure in which they are used [1-4]. The minimally invasive surgical procedure is a common method that involves the insertion of a needle to a target inside the body for either tissue sample removal (i.e. biopsy) or drug delivery (i.e. brachytherapy). During this procedure, the accuracy of needle placement is of utmost importance, as tip misplacement in drug delivery or biopsy of an unintended tissue region may reduce the effectiveness and conclusiveness of the diagnosis and treatment, respectively. The inaccuracy of needle placement or needle deviation from its intended path is due to several factors such as tissue deformation (i.e. inhomogeneity and anisotropy), physiological processes (i.e. fluid flow and respiration) and anatomical obstructions [5-7]. However, the success of this procedure involves other factors such as deflection of the needle from the target with the deformed tissue during insertion caused by needle tip force and friction, resulting in target placement error [8]. These inaccuracies and target placement errors can be reduced by using a robotic system for needle insertion that predicts needle deflection and also steers the needle towards the target [8-14]. Such use of a robotic system requires knowledge of needle-tissue interaction dynamics that can be gained through experimentation or simulation-based methods such as finite element (FE) analysis [2, 3, 5, 15-21].

In recent years, substantial research has been conducted on needle-tissue interaction through experimentation of needle insertion into soft gel [16-19]. These studies investigate the effect of parameters such as diameter, bevel angle, needle insertion speed, different tip needle and gel elasticity, which influence the interactions between the needle and soft tissue. In an experimental based study by Jiang et al. it was shown during needle insertion that varying parameters (i.e. diameter, bevel angle, insertion speed, and tip shape) result in alterations of the needle forces and gel rupture [16]. Gerwen et al. carried out a review of the experimental studies of needle force interactions with tissue and found that typical peak forces are in the order of 0.1 to 10 N. The study also highlighted that blunt tipped needles produce higher peak axial force than the conical, diamond and bevel needles, despite smaller diameters [17]. Extensive research has been carried out in the development of the surgical needle-tissue
interaction model to simulate real-time behaviour using Finite element (FE) method [5, 6, 20, 21]. FE analysis of the needle-tissue insertion mechanism in surgical simulation needs to take several factors into account, such as material mechanical behaviour (tissue/soft gel), material and geometry non-linearity (large deformation), material rupture (puncture) and boundary conditions (contacts) [5-7].

Misra et al. presented a comprehensive review of the literature related to the modelling of needle-tissue interaction using the FE method and its application in modelling simulations for the invasive and non-invasive method in medical field [19]. FE analysis with an element deletion method can be used to simulate puncture or rupture at the needle tip [5, 20]. However, this modelling method requires small elements in the region of needle-tissue interaction, which results in high computational time and may result in unreliable results due to deletion of elements in the vicinity of the needle when large deformation occurs (i.e. needle penetration). Subsequently, Oldfield et al. proposed an FE method with a cohesive zone element model or crack propagation using a cohesive element to simulate tissue rupture and needle-tissue interaction [20]. In this technique, the cutting and subsequent needle penetration path need to be defined prior to needle insertion, and the initiation of crack needs to be defined with the crack (i.e. notch) on the surface of the bulk material. However, the drawback of this method is that the needle path is unknown during insertion and may lead to a distribution of cohesive elements over a region with very fine elements, as needle deflects during insertion; therefore, making it difficult to define the crack propagation path and leading to higher computational time. In the literature [2, 18, 20], most of the linear elasticity models are based on indentation and element deletion or element splitting (i.e. adaptive remeshing technique).

Recently, another FE method was proposed by Yamaguchi et al. using ALE ( Arbitrary Lagrangian Eulerian) based technique [21]. This method is used in fluid based analysis, which involves large deformation and fracture in continuum models. The advantages of this method include its ability to resolve arbitrary enclosing boundaries, to have changeable zoning for obtaining an optimum resolution, with improved accuracy in a problem where fully Lagrangian calculations are not possible, and to function with time steps much larger than possible with explicit methods [22]. The study showed that the FE method enables dynamic analyses of needle insertion and can be used to study needle-interaction forces and needle deflection using a soft tissue FE model and different tip needles. Coupled Eulerian-
Lagrangian (CEL) method in Abaqus is a similar technique which provides to capture the advantages of both Lagrangian and Eulerian method with the ability to simulate large deformation and fracture problems using fluid-structure analysis [23].

Surgical procedures, such as biopsy and drug delivery based diagnostic, use a hollow needle (i.e. cannula) and a stylet, both of them jointly advancing to the site of biopsy (i.e. tissue removal) or drug delivery (i.e. seed implantation). The hollow needle (i.e. cannula) advances further deeper into the tissue to cut a tissue sample or drug delivery, whilst the stylet is held fixed in place [24]. In this study, the aim is to develop a three-dimensional (3D) FE model that resembles the surgical needle (i.e. cannula) insertion mechanism into phantom tissue (i.e. soft gel) and to study needle-tissue interaction forces for improved needle design and material-tissue interaction. Phantoms are usually tissue mimicking materials that are generally used to investigate the needle-tissue interaction and for the training of clinician in image-guided needle interventions [25]. In general, the use of biological tissues are not always a feasible option, due to practical and ethical issues. Alternatively, gels can exhibits viscous, non-linear elastic behaviour; mimicking biological tissue, therefore models are quite complex [16]. In general, gels are more durable and can be customised to the needs of specific research activities or training.[25]. As a first approximation, the present study focused on developing an FE analysis linear elastic model that predicts the needle-gel interaction dynamics as a substitute for the soft biological tissue. In the analysis, a Coupled Eulerian Lagrangian (CEL) based FE method was used to perform dynamic analysis of needle insertion into a soft gel. A review of the literature shows that the force and deflection are the functions of solid needle geometry [17]. However, the outcome of these studies cannot be applied to hollow needles as the needle geometry and the contact area between the needle cutting edge and tissue are significantly smaller. Furthermore, this results in lack of understanding and insight of the hollow needles and its variation of cutting forces along the cutting edges of the hollow needle [24]. To the best of the authors’ knowledge, this study investigates for the first time the modeling of a 3D hollow needle insertion into the gel using FE based CEL method and presents a comparison of simulation results with experimental data acquired through the use of a high-speed camera. The model also predicts and could help us to understand, how a hollow needle, particularly the cutting edge and needle tip design, vary needle forces and deflection during the needle insertion process. This model can form an indispensable tool not only for planning, training, and practice of surgical procedures in
handheld or robot-assisted surgery but also in the design of more effective hollow needle tip geometries.

2. Material and Method

2.1 Specimen preparation and test conditions
Agar gel is commonly used for the cell culture of cartilage, soft tissue as an alternative biomaterial and also been used as a phantom material for material property characterisation using image techniques [21, 26]. Gel with Agar (Agar-05038, Sigma-Aldrich, USA) concentration (weight/volume, w/v) of 3% was prepared by dissolving powdered agar in distilled water. The solution was sealed and heated for 15 mins at 90-95 °C and magnetically stirred until the agar powder was completely dissolved. Finally, the solution was left in the beaker to cool down to 60°C before it was poured into a cylindrical mould and left overnight at room temperature to cure. To ensure consistent test conditions across the experiments, transparent acrylic Petri dishes (cell culture dish), 85 mm in diameter and 15 mm in height were used as a mould and covered with a lid to maintain humidity and to prevent losing its moisture content. Storage time before testing was kept constant throughout the study, within allowable limits (18 h and 30 ± 5 min), to ensure comparable gel properties and reproducible data. The preparation of all specimens and the testing were performed at controlled laboratory conditions; 20 °C with 50% relative humidity. Agar gel samples were punched out in a cylindrical shape, nominally 50 mm in diameter and 10 mm in height. Ten cylindrical test samples were prepared for the test (compression test: Ø50 x 10 mm cross-head speed: 2.5 mm/s).

2.2 Characterization of mechanical properties of agar gel
A uniaxial compression test on agar samples as shown in Fig.1 (a) was performed to obtain mechanical properties such as gel elastic modulus, yield strength and strain to failure, which are used as input in FE analysis to create a material model to simulate gel. The nominal stress-strain curve was generated from the results of the uniaxial compression test. During the experiment, loading in compression mode was conducted on the cylindrical gel sample positioned between the loading plates of a Tinius testing machine with a 1000 N load cell (Tinius Olsen, H25KS Tinius Olsen Ltd. - United Kingdom). To achieve frictionless contact between the load introduction plate and the test sample during testing and ensure pure compression of the sample, paraffin oil was applied on the outer surface of the gel samples.
The test was repeated for ten samples at a compression speed of 2.5 mm/s and the resulting force and displacement were recorded. Subsequently, this information is used to calculate nominal stress ($\sigma_n$) and nominal strain ($\varepsilon_n$), which are given by:

\[
\varepsilon_n = \frac{l}{l_0} \quad \text{and} \quad \sigma_n = \frac{F}{A_0}
\]

where $l_0$ and $l$ are the initial and compressed length of the gel sample respectively, while $A_0$ is the initial cross-sectional area of the gel sample and $F$ is the compression force. Similarly, true stress ($\sigma_t$) and true strain ($\varepsilon_t$) are calculated from the nominal stress-strain curve as follows:

\[
\varepsilon_t = \ln \frac{l}{l_0}
\]

\[
\sigma_t = \frac{F}{A_0} (1 + \varepsilon_n)
\]

In general, tissues are inhomogeneous and exhibit non-linear, anisotropic and viscoelastic behaviour which makes them more complex to model. As a first approximation, this study focuses on linear elastic large deformation dynamic model that predicts needle-interaction forces in three-dimensions. Fig. 1(c) shows the compression stress-strain curve which is close to linear up to strains of 10-15%. The relationship was non-linear for strains greater than 15%, due to a combination of material inhomogeneity and viscoelasticity [27]. The region after the strain of 15% compression was neglected as most of the probing and insertion experiments operate with strains below 15% [17, 18]. From the initial slope of the curve (i.e. linear region) as shown in Fig. 1(c), Elastic Modulus ($E$) was approximately found to be 0.297 MPa. As soft gel is considered as nearly incompressible, the Poisson’s ratio ($\nu$) was set as 0.49 (i.e. 0.47 to 0.5 is commonly used) [18-21]. In linear elastic modelling, $E$ and $\nu$ are the two material parameters used to define the soft tissue properties [18]. The FE simulation was performed using the parameters presented in Table1.
Fig. 1. Compression test of Agar Gel at a compression speed of 2.5mm/sec: (a) Cylindrical gel sample between loading plates (b) Gel sample completely damaged after compression test and (c) Compression stress-strain response. (i) Red continuous line indicates the nominal stress-strain curve and (ii) Blue dotted line indicates the true stress-strain curve. (i.e. 0-15% strain is considered to be linear region and 15-40% strain – non-linear region).

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Elastic Modulus ((E)) (MPa)</th>
<th>Poisson’s ratio ((\nu))</th>
<th>Density ((\rho)) (g/cm(^3))</th>
</tr>
</thead>
<tbody>
<tr>
<td>Values</td>
<td>0.297</td>
<td>0.49</td>
<td>1.172</td>
</tr>
</tbody>
</table>

2.3 Needle Insertion Experiment

To evaluate the needle insertion approach, an experimental procedure was developed to measure the needle insertion forces and deflection with a uniaxial insertion device. In the experiments, the needle was inserted into the agar gel at a constant velocity of 2.5mm/s to a penetration depth of 50 mm with a uniaxial tensile tester (insertion device) as shown in Fig. 2. A similar test is performed in the industry with the custom made insertion devices to measure the needle deflection and reaction forces [20, 21]. In this test, the penetration forces at the needle tip and friction forces at the needle shaft are measured using a 100 N load cell (S-Type, Tinius Olsen DBBMTOL-100N) placed above the needle holder (i.e. chuck and collet). The size of the needle was Ø 0.9 x 125 mm based on a 20G surgical needle (stubs wire-gauge standard, ISO 9626) while the agar block dimensions were 100 x 40 x 150 (mm). The angles of the bevel needle tip were 18° (i.e. Needle 1) and 27° (i.e. Needle 2) as shown in Fig.4. Elastic modulus \((E)\) and Poisson's ratio \((\nu)\) of the 20G bevel tip steel needles (custom-made, Cook Medical, Ireland) used were 200 GPa and 0.29 GPa, respectively [28]. To
estimate the needle deflection, a rectangular perspex box mould (100 x 40 x 150 mm) for a gel with one open end (Fig.3) and a graph sheet printed on acetate paper glued to the front of Perspex box were prepared. The deflection of the needle was captured on videos and photos taken from the lateral side with a digital single-lens reflex camera (Nikon D5300, Nikon Corp, Japan) during insertion. Two 20 G one-plane symmetric needles with bevel angles of 18° and 27° were each inserted into agar gel ten times, leading to a total of 20 trials. The insertion force at the final depth of the needle at 50 mm was recorded along with the initial puncture force to validate the needle insertion simulation model (Fig. 4). In each case, the force along the z-direction (i.e. Z force) against the needle insertion depth was measured.

Fig. 2.Uniaxial tensile tester. (a) Tinius Dual Column tester fitted with a 100N load cell and custom made collet-chuck fixture to hold a needle, (b) Agar mould inside the perspex glass encasing, and (c) Digital single-lens reflex camera to capture deflection of the needle.
2.4 CEL-based finite element analysis

A CEL based linear FE model was developed as shown in Fig. 4 [23]. The FE model consists of the 3D models of the needle and of the agar gel block. The needle is constrained along the x and y axis at the proximal rigid part (red part of Fig. 4a) and a displacement of 50 mm was applied along the z-axis to this rigid part of the needle. A friction coefficient of 0.02 was set to define tangential interaction between the needle and the agar gel after some trial runs [25]. The geometries for the size of needles and agar gel were adopted corresponding to the experimental setup (Fig. 2) and the angles of the bevel needle tip (Fig. 3). The needle geometry was discretized using linear solid hexahedral elements with reduced integration and hourglass control, while the agar gel meshed with linear eulerian brick, reduced integration, hourglass control (Fig. 4b, c). The number of meshes totaled 340,800 elements. The smallest element size adopted was 0.5 mm in all models. A dynamic explicit CEL based FEA was performed using ABAQUS V14 [23]. The material properties of agar gel determined from the experiment were used in creating the agar gel material model (Table 1). The solid hollow needle was moved downward at the constant speed of 2.5 mm/s as the in needle insertion experiment. The total simulation time was set at 20 s for the 50 mm insertion, which matched the experiments.
2.5 Statistical analysis

The experimental data are represented as a mean ± standard error about the mean; for deflection, peak force and insertion force. The normality of these datasets was examined using the D’Agostino and Pearson test, all of which were found to be normally distributed. Significant differences were identified between experimental groups of continuous variables using Student t-test for the normally distributed data. For comparison of experimental to simulation data for both needles, a two-way analysis of variance (ANOVA) test was used to compare the mean values of the different groups. A p-value (alpha value) < 0.05 was considered statistically significant for all tests.

3. Results

3.1 Needle Deflection

Comparison of needle deflection for the two bevel tips between the experimental and simulation results is shown in Fig. 5. The difference in mean deflection for the 18° and 27° bevel angle needles, compared to experiments, were 0.25 mm and 0.4 mm, respectively. The
deflection of the needle during the experiment is captured from the forward direction for the lateral side of the 18° bevel tip needle at 50 mm approx. is shown in Fig. 6. The photos of needle insertion in agar gel at 10, 20, 30, 40, and 50 mm during needle the insertion experiment is shown in appendix A (Fig. A1).

![Graph showing needle deflection comparison](image1)

*Fig. 5. Comparison of needle deflection at 50mm in-depth between results of experiment and simulation for Needle 1 (18 deg) and Needle 2 (27deg).*

![Graph showing needle deflection](image2)

*Fig. 6. Needle deflection at 50mm in-depth: (A) Simulation result for Needle 1 (18deg) (i.e. Deflection (D) = 3.91 + 0.45 outer diameter (O.D) of needle) and (B) Experimental result for Needle 1 (18deg) (i.e. D = 2mm/dot).*
3.2 Needle Forces

A typical needle insertion process involves three basic phases of interaction as shown in Fig. 7, phases which are distinguished as follows:

(a) Deformation (Phase 1): The first phase (Fig. 7) starts when the needle tip comes in contact with the tissue/gel surface or tissue/gel boundary and ends when the tissue/gel surface or boundary is ruptured and the insertion force reaches a peak value known as peak force ($F_N$). The actual event of rupturing of the boundary is referred to as “puncture”.

(b) Tip Insertion (Phase 2): The second phase commences when the tissue/gel surface or boundary is ruptured and it ends when the tissue or gel (i.e. phantom soft tissue) slides over from the tip onto the shaft.

(c) Tip and Shaft Insertion (Phase 3): The third phase starts just after the tip insertion, which is a transition from tip to the shaft and ends when the needle encounters a new tissue boundary (internal), or it is stopped. During this insertion phase, the needle tip is subjected to cutting force and needle shaft to a varying frictional force due to the increase in contact area between the shaft and tissue as the needle is advanced.

It is evident from the typical graph for needle insertion in soft tissue/gel as shown in Fig. 7, that the varying frictional force leads to an increase of total insertion force (i.e. Cutting and Friction). Hence, the total insertion force ($F_I$) comprises of puncture force (i.e. cutting) and friction force (i.e. penetration). Fig. 8 demonstrates the comparison of force-needle insertion curve for the experimental and simulation results in 18° bevel tip needle (Needle 1). For the oscillations (i.e. noise) in the simulated reaction force curve (i.e. Z Force) from FE analysis, the dataset was smoothened and normalized using data analysis software (Origin Pro8). The measured initial peak force (i.e. maximum, mean and minimum) from all the 10 repeated experiments for Needle 1 and Needle 2, along with the simulated peak force from the FE analysis are plotted in Fig. 9. Similarly, Fig. 9 also represents the result of measured total insertion force for both the Needle 1 and Needle 2 along with the simulated insertion force from the FE analysis, respectively. The percentage of error between the measured (mean) from experiment and simulated peak force ($F_N$) for needles 1 and 2 are 15.2% and 15.6% (Table. 2). However, the percentage of error between the measured and simulated total insertion force ($F_I$) for both the needles is 4% and 4.2% respectively.
Table 2: Comparison of needle insertion forces at 50mm in-depth between results of experiments and simulation for Needle 1 and Needle 2

<table>
<thead>
<tr>
<th>Needles</th>
<th>Angle</th>
<th>Bevel Length (mm)</th>
<th>Initial Peak Insertion Force, $F_N$</th>
<th>Total Insertion Force, $F_T$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Measured (Mean) (N)</td>
<td>Simulated (N)</td>
</tr>
<tr>
<td>Needle 1</td>
<td>$\xi = 18^\circ$</td>
<td>2.95</td>
<td>0.106</td>
<td>0.125</td>
</tr>
<tr>
<td>Needle 2</td>
<td>$\xi = 27^\circ$</td>
<td>2.01</td>
<td>0.108</td>
<td>0.128</td>
</tr>
</tbody>
</table>

Fig. 7. A typical needle insertion process (i.e. Phase 1 – Deformation ends with Peak force ($F_N$); Phase 2 – Tip insertion; and Phase 3 – Tip and shaft insertion ends with total insertion force ($F_T$).
The contact stresses on the needle-tissue interaction can be divided into two components: normal stress (i.e. axial) and shear stress (i.e. frictional). However, these stresses result from the contact pressure (CP) between the needle and tissue/gel interaction during the needle insertion process. Comparison of needle contact stress for needles 1 and 2, based on simulations, is shown in Fig. 10. The contact stress distribution for Needle 1 along the cross-

*Fig. 8. Comparison of reaction force (i.e. Z-Force) between results of one representative experiment and simulation for Needle 1(18deg).*

*Fig. 9. Comparison between all representative experimental results (mean ± standard error about the mean) of needle insertion forces (i.e. peak and insertion force), to model, predicted results (i.e. simulation).*

### 3.3 Needle Stresses

The contact stresses on the needle-tissue interaction can be divided into two components: normal stress (i.e. axial) and shear stress (i.e. frictional). However, these stresses result from the contact pressure (CP) between the needle and tissue/gel interaction during the needle insertion process. Comparison of needle contact stress for needles 1 and 2, based on simulations, is shown in Fig. 10. The contact stress distribution for Needle 1 along the cross-
sectional surface in the $x$-$z$ plane for initial puncture and for each 10 mm insertion, is shown in Fig.11. Subsequently, after a puncture, the maximum contact stress appeared to be moving along the shaft from needle tip throughout the needle insertion for both of the models. The initial peak contact stresses at the event of puncture during needle insertion for Needle 1 and Needle 2 were 125 MPa and 110 MPa, respectively.

Fig. 10. Comparison between Needle 1 (18deg) and Needle 2 (27deg) contact stresses on the needle (i.e. z dir-stress) from needle insertion simulation.

Fig. 11. Contact stress distributions around the needle for Needle 1 (18°) (MPa): (a) Puncture, (b) step10, (c) step20, (d) step30, (e) step40, and (f) step50.
4. Discussion

In this study, a 3D FE analysis model was developed for hollow needle insertion into soft agar gel through CEL based FE method using dynamic fracture analysis. Applying a CEL method to structure analysis (i.e. FE analysis) makes it possible to do fluid-structure analysis with large deformation and fracture. The aim of this study was to create a numerical FE approach to enable the better understanding of the needle-tissue interaction \textit{in silico} and ultimately optimise the performance of hollow needle tip design to overcome the existing shortcomings of current hollow needle design. Moreover, the needle insertion experimental test setup model into soft gel was developed and quantitatively analysed to evaluate and validate the simulation model.

The FE study performed for assessing the needle-tissue interaction mechanism involved creating FE models and simulating needle insertion and interaction with a phantom tissue (i.e. soft gel). The frictional coefficient of 0.02 was set analytically as the contact condition between the needle and gel because it was found to be in good agreement between the experimental and simulation result. In addition to this, T.L. de Jong \textit{et al.} carried out a research to study the phantom tissue (PVA) as a liver tissue mimicking material and found PVA 4 m\% with 2FT (freeze-thaw) cycles is only specimen comparable to ex vivo human liver. Moreover to these result, it was found that the estimated median friction slope for the insertion into phantom tissue is in the range between 0.01 N/mm to 0.02 N/mm \cite{25}. A mesh sensitivity study with varying mass density was also performed to investigate the effect of reducing global mesh size and increasing the step time ($\Delta t$) on the results of FE simulations. However, since large noise and oscillation in the curve occurred for simulated FE forces, for the larger density and mesh size, a finer mesh, and smaller $\Delta t$ was still recommended. In order to reduce the computational time and accuracy of the forces, a trade-off between the larger density and optimum mesh size for the needle was considered. The simulation took approximately 70 hr on a 12 core CPU (i.e. Intel dual-core Xeon 2.6 GHz, Memory: 16Gbyte), with mass scaling the density of needle to $10^5$ times as large as the true value to finish analysis. Accordingly, Kataoka \textit{et al.} proposed a needle insertion simulation model using Eulerian hydrocode FE method and highlighted the influence of density scaling to the simulated force has very less or no significant difference \cite{29}. In regards to the computational time, as of one solution for this issue, the general purpose graphics processing unit (GPGPU) has been developed which has accelerated the use of various commercial software in recent
years [23]. In other words, with the application of GPGPU in the dynamic analysis (i.e. explicit analysis), real-time computational based surgical simulation will appear soon.

In the literature [2, 5, 8, 15, 16, 30], it is clear that the shape and size of the needle play a substantial role in determining the forces of needle insertion. In general, a needle with a smaller diameters leads to less insertion force but more needle bending [30]. Bevel tipped needles leads to more bending and higher peak force compared to the cone and triangular tipped needle [16]. Although, despite this drawback, bevel tip are prevalent because of maneuverability to the clinician to rotate the needle to deliver therapy in different direction and ease of manufacture. In most of the surgical procedures, such as biopsy or drug delivery uses a needle cannula (hollow needle) for tissue removal or seed implantation. Needle insertion is typically a tissue cutting process, greatly depends on its cutting edge geometry [3]. In this study, for the first time modeling of a 3D hollow needle insertion into the gel, needle tip geometry was defined only by the bevel angle which does not directly reveal any specific information about the cutting edges (Fig. 3).

This study presents both experimental and simulation results of needle-gel interactions. The experiments were conducted under constant velocity, constant needle diameter, needle tip shape, gel elasticity, and varying bevel angle. The uniaxial compression test was suitable for gel specimens because it is simple and eliminates the problem of premature failure arising from clamping as in tensile tests [5, 26, 27]. To define the elastic modulus of agar gel, the initial slope of true stress-strain (i.e. linear region) was considered. It was observed from previous studies that most of the needle or probe insertion method undergo high deformation which is followed by rupture event causing sudden crack propagation inside the gel/tissue and operate within the range of 15% to 20% strain.[17, 18, 31]. To imitate the intraoperative handheld insertion process, or continuous robotic insertion, a constant velocity of 2.5 mm/s opted in needle insertion experiment. It is evident from previous studies that the increase in insertion velocity from 0.5 mm/s to 20 mm/s has exponential growth in total insertion force initially, but starts to remain constant when the speed is up to 5 mm/s or higher [16]. During the experimental process, the interaction between the needle and the agar gel was observed. The results from experimental curves were found to be in good agreement with the typical graph for needle insertion in soft tissue/gel as shown in Fig. 7.

The results from FE simulation were compared with those of experiment that include force and deflection data with agar gel. For needle deflection, the percentage error along the x-z
plane for Needle 1 and Needle 2 are 5.7% and 10%, respectively (Fig. 5). Fig. 5 also demonstrates that a needle with lower bevel angle tends to have more deflection compared to one with higher bevel angle and similar trends were observed when compared to a result obtained by Yamaguchi et al. [21]. Moreover, in clinical practice, there is no defined tolerance for the accuracy of needle insertion, and in general, insertions with limited needle misplacement or target placement errors result in more effective treatment [8]. These suggest that our experimental results are the allowable range for clinical application such as pre-operative surgical planning. In addition, the variation of the bevel angle does not lead to a significant variation between simulation estimates and experimental results in needle deflection (Fig. 5 & 6) (p > 0.05).

Quantitatively, the percentage errors between the experimental data and the FE simulation for the total insertion force along the z-direction (i.e. Z Force) for Needle 1 and Needle 2 were 4% and 4.8%, respectively (Table 2). Fig. 8, clearly demonstrate that total insertion force in the simulation followed those in the experiment until the needle tip and shaft portion of the needles were completely inside the agar gel. For the insertion force (Fig. 8), there was a peak force during needle puncture as compared to a result obtained by Okamura et al. [30]. It is evident from the results, that needle with lower bevel angle (i.e. Needle 1) tends to have less peak force but higher insertion force compared to higher bevel angle needle (see Table 2). A larger bevel angle (ζ) reduces inclination angle (λ), thereby creating a higher peak force during needle insertion as observed by Han et al. [32]. Moreover, the simulated reaction force from FE analysis and the measured insertion force, obtained from experiments, showed no significant difference for the total insertion force throughout the needle insertion depth, for both of the tested bevel needle tips (Table 2) (p > 0.05). It is evident from the simulation and experimental result, the increase of insertion force after a puncture, is due to the influence of friction between the agar gel and the needle (Fig. 8). In the literature [2, 16, 18, 20, 21], the force at the needle tip were almost flat after the needle tip penetrated the surface, while the friction force proportionally increased. These trend were achieved in our result, because the force at the needle tip and the friction on the needle shaft were separately measured, and it was observed that the cutting force (F_C) was almost flat after the needle tip insertion into the surface (i.e. puncture), while the friction force (F_F) proportionally increased along the needle insertion depth (see Appendix A, Fig. A3).

Finally, the contact stress (i.e. contact pressure) along the needle surface due to the stick-slip effect of during friction on the needle shaft could be the key parameter to study the needle-
tissue interaction for different biological tissues and different material coating for the needles. The results from the simulation may help to predict the needle deflection, insertion force and contact stress distributions during needle insertion as pre-operative surgical planning \textit{in silico} that require accurate needle targeting e.g. biopsy and brachytherapy. These results suggest that the FE model can be used to simulate the 3D gel fracture due to large deformation and tip-gel interaction forces and needle stresses. However, the FE model presented here predicts the forces and deflection of the needle based on a linear elastic material model. Within the limitation of FE based CEL technique in Abaqus, the FE model is limited to needle deformation or deflection and cannot predict deformation for the gel. For more clinical situation, it was proposed to adopt the current setup to investigate the needle-tissue interaction for the porcine liver as a biological substitute to gel and applicability of our approach.

\textbf{5. Conclusion}

The accuracy of the model suggests that it could be used as a tool to study clinical applications such as pre-operative surgical planning. Our results help to understand and predict the needle deflection, needle forces, and contact stress distribution during needle insertion as pre-operative surgical planning \textit{in silico}. This research study provides an insight into how needle-tissue interactions for hollow needles (i.e. cannula) using FE modelling approach can lead towards optimising the needle-tip designs and although modelling real-time surgical simulation not only for training but also in planning and practice of surgical procedures. Furthermore, enabling in modeling a new needle design system on mimetic gels of tissue. This model can form an indispensable tool not only for needle tip design but also help us to understand alternative material or coating process that can enhance the cutting process during needle insertion, thus assisting in optimizing the current needles. For more clinical situation, the plan is to extend the outcome of this study for biological application by incorporating CEL based FE analysis for needle and soft biological tissue interaction.

\textbf{Conflict of interest statement}

The authors have declared that no conflict of interests exists.
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Appendix

*Fig. A1. Needle deflection during needle insertion experiment (18°) (mm): (A) step10, (B) step20, (C) step30, (D) step40, and (E) step50.*
Fig. A2. Needle deflection during needle insertion simulation (18°) (mm): (A) step10, (B) step20, (C) step30, (D) step40, and (E) step50.

Fig. A3. Needle Reaction force (i.e. $R_z$ force) (N) during needle insertion simulation process.
Fig. A4. Needle Reaction forces (i.e. $R_x$, $R_y$, and $R_z$ force) (N1) during needle insertion simulation process.

References