Minimally disruptive needle insertion: a biologically inspired solution

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The mobility of soft tissue can cause inaccurate needle insertions. Particularly in steering applications that employ thin and flexible needles, large deviations can occur between pre-operative images of the patient, from which a procedure is planned, and the intra-operative scene, where a procedure is executed. Although many approaches for reducing tissue motion focus on external constraining or manipulation, little attention has been paid to the way the needle is inserted and actuated within soft tissue. Using our biologically inspired steerable needle, we present a method of reducing the disruptiveness of insertions by mimicking the burrowing mechanism of ovipositing wasps. Internal displacements and strains in three dimensions within a soft tissue phantom are measured at the needle interface, using a scanning laser-based image correlation technique. Compared to a conventional insertion method with an equally sized needle, overall displacements and strains in the needle vicinity are reduced by 30% and 41%, respectively. The results show that, for a given net speed, needle insertion can be made significantly less disruptive with respect to its surroundings by employing our biologically inspired solution. This will have significant impact on both the safety and targeting accuracy of percutaneous interventions along both straight and curved trajectories.

1. Introduction

The use of minimally invasive (MI) interventional tools for soft tissue surgery promises reduced tissue trauma, shorter procedure time and shorter recovery times for the patient. Common procedures involve needle insertions for diagnostic biopsy of cancerous tissue [1], brachytherapy [2] or drug delivery [3]. However, the difficulties and limited access associated with MI procedures warrant smarter modalities, which provide the surgeon with better control and assistance. In the near future, robot-assisted surgery has the potential to improve MI surgery.

In recent years, several robotically controlled steerable needle-based solutions have been proposed, which can accurately control tip deflection to follow planned curved paths within soft tissue [4]. Actively controlling the needle can increase placement accuracy, while simultaneously lowering risk, e.g. by avoiding obstacles or critical regions in tissue that lie between entry point and target. However, while most models can control the needle’s trajectory, tissue motion and deformation during the insertion process remain difficult to quantify and predict, leading to significant targeting inaccuracies, as described by Deurloo et al. [5]. Internal regions of soft tissue around the needle experience the largest shape changes during an insertion, owing to the complex cutting process. These regions are therefore likely to show the largest discrepancy between pre-operative images and surgery. Percutaneous procedures that require multi-targeting (i.e. targeting multiple targets within an organ) or reinsertion are therefore under risk of inaccurate insertions if (i) the planned path is not accurately adjusted to account for the current tissue configuration or (ii) the dexterity of the flexible needle is not sufficient to reach the moving target.

Recent studies have presented systems that use a steerable needle with prior identification of parameters that minimize target motion [6], or that track motion...
A laser-based image correlation technique [19] was developed to measure internal displacements close to the needle interface and at high resolution, where Lagrangian displacements and strains are computed to measure the material response due to the interactions between a needle and its surroundings [20]. Subsequently, FE simulations, supported by single-plane measurements obtained with the same set-up as in [20], were developed to demonstrate reduced target motion due to varied insertion parameters in a previous study [21].

In this study, we present an extended, fully three-dimensional experimental configuration, which enables measurements to be taken across the entire needle width. This results in both displacements and strains being resolved along the interface and at high resolution, where Lagrangian displacements and strains are computed to measure the material response due to the interactions between a needle and its surroundings. The needle’s multi-part design allows biologically inspired motion profiles to be investigated, where each segment is moved reciprocally rather than simultaneously [17]. Reciprocal motion has been shown to be advantageous due to the tensile support offered by segments being retracted to those being inserted—a phenomenon used, for example, in drills designed for interplanetary exploration [18]. Such motion also causes an anchoring effect due to the friction experienced at the needle–tissue interface, which leads to significant changes in the interaction between the needle and the surrounding tissue.

In the following section, the needle’s multi-part design and a definition of the biologically inspired motion profiles for the insertion are presented. The experimental design and the image acquisition around the needle for measuring three-dimensionally resolved displacements and strains close to the needle interface are then described. Subsequently, time- and depth-resolved results for insertions performed with both conventional and biologically inspired motion profiles are presented. The paper ends with a discussion of the results and conclusions.

2. Material and methods

The needle used for all insertions was made out of four axially interlocked segments (figure 1a), with an outer diameter (OD) of 4 mm, produced using rapid prototyping technology (Connex 350, Objet Inc., USA). The segments could slide along one another and were actuated independently by an actuation box able to generate four independent, 200 mm linear motions. All segments possessed a flat bevel tip, inclined by 20° from the needle centre axis, as illustrated in figure 1b, and were rapid prototyped out of the rigid material VeroGrey (Shore Hardness D 83-86, Objet Inc.).
Tissue phantoms were prepared out of gelatin, with a concentration of 3.5% by weight, provided by Sleaford Quality Foods Ltd, UK. Gelatin powder was mixed with heated water for 10 min. At this concentration, gelatin was shown to be an acceptable phantom material, with similar stiffness and frictional properties to ex vivo porcine brain [22]. For the optical measurements, the gelatin was seeded with fluorescent melamine resin beads (10 μm in size, rhodamine B-marked, Sigma Aldrich Co.) before solidification, and then poured into transparent boxes with dimensions of 86 × 28.5 × 62 mm. Each box contained 110 ml of seeded gelatin. The particle number density was approximately 19 particles mm\(^{-3}\). In indentation testing, the added particles were shown to have no influence on the material response of the phantom material. Samples were stored in a domestic refrigerator at 14°C for 12 h and tested on the following day, after the gels had reached room temperature (20 ± 2°C).

2.1. Needle actuation via biologically inspired motion profiles

Insertions with the four-part needle were performed with two different motion profiles, at equivalent net speeds. In the first motion profile, all four segments were aligned and pushed simultaneously at the same speed (figure 1b, top). This type of insertion is referred to as direct push (DP).

For the biologically inspired motion profile, the four parts of the needle were actuated cyclically, at different speeds and directions, throughout the insertion process. Segments were pushed forward sequentially with a stroke length (l) of 4 mm. As one segment was pushed forward, the remaining three segments were pulled backwards (figure 1b, bottom). The amount of pull-back (p) was empirically set to 30%, as it was shown to offer a good compromise between reduced tissue motion and higher individual segment speed [21]. Each cycle began with the leading segment moving forward, followed by the segment lying opposite, and so forth. A cycle was completed when each of the four segments was pushed forward, resulting in all segments being aligned again (figure 1c).

As all four segments were pulled back during one cycle, the distance travelled forward by the needle was not equal to l, but slightly less, with l(1 − p). In order to compare DP and pullback (PB) insertions, the net forward motion was set to be the same, i.e.

\[
s_{\text{net}} = \frac{l(1 - p)}{4t_s},
\]

where \(s_{\text{net}}\) is the speed of the DP insertion and \(t_s\) is the time one segment takes to move forward by l. This ensured that the needles reached the same depth in the same amount of time. \(s_{\text{net}}\) was set to 0.5 mm s\(^{-1}\), a needle insertion speed in the range used in brain [23].

2.2. Experimental set-up for three-dimensional laser-based image correlation

The needle was inserted into transparent boxes containing the fresh, seeded tissue phantoms in the experimental set-up shown in figure 2. Ten PB insertions and eight DP insertions were performed. Full-field displacements were measured in five planes across the needle width. A 4.5 mW, 532 nm diode laser beam was deflected by a high precision, single-axis galvo mirror (GV2S01, Thorlabs Ltd, UK), which scanned the measurement planes sequentially (figure 3). The laser beam was formed into a vertical, thin light sheet by three cylindrical lenses. The laser excited the fluorescent particles (with excitation around 540 nm and emission around 584 nm) in the tissue phantom and the emitted light was captured by charge-coupled device (CCD) cameras (nvBlueFOX 223C, MATRIX VISION GmbH, Germany), which were positioned on two opposite sides of the needle and perpendicular to the light sheets. In front of each camera was a coloured glass long pass filter (cut-off wavelength at 550 nm, OG550, Thorlabs Ltd, UK), which blocked the laser light and let the fluorescent particle light pass. This led to an increased signal-to-noise ratio for the particles with respect to the image background and hence more reliable correlation results, compared to particle images of scattered light. Additionally, unwanted reflections from the needle interface were blocked by the filter, leading to more accurate correlation results close to the needle interface.

The two CCD cameras were triggered synchronously with the galvo scanner. Scanning positions were ordered symmetrically around the centre axis of the needle:

- **Position A**: on the needle centre axis,
- **Position B**: 1 mm offset, and
- **Position C**: 2 mm offset, coinciding with the edge of the needle.

Each camera was exposed when the galvo scanner was held in locations A, B and C on its respective side. Position A, the centre plane, was recorded by both cameras simultaneously for validation purposes. Table 1 shows the trigger configuration of both cameras in the experimental design. The scanning speed was approximately 4.8 mm s\(^{-1}\) for a needle width of 4 mm, with an exposure time of 100 ms for each image acquisition.

The light sheet position was calibrated using the fixed needle in the experimental set-up. Using two apertures, the scanner was positioned in line with the needle axis so that the light sheet coincided with the needle centre. The rotational axis of the scanner was positioned at a distance of 500 mm from the centre of the camera’s field of view. For the edge of the needle (2 mm), this resulted in a mirror angle of approximately 0.23°. Owing to the long distance between the rotating galvo scanner and the measurement location, measurement planes were assumed to be parallel at the location of the field of view.

For measurements during needle insertion, two field-programmable gate arrays (FPGAs) were used, configured via LasVIEW (CRI0 9014, National Instruments Inc.). A flowchart of the experimental set-up is shown in figure 4. The first FPGA controlled the motion profile of the needle, as defined in the previous section. The second FPGA module was employed to manage the timing and positioning of the two cameras and the galvo scanner. The camera exposures were triggered via a transistor–transistor logic (TTL) signal and the galvo position was controlled via an analogue output. This allowed the synchronized exposure of the cameras at each scanning position.

2.3. Image registration and displacement computation

To ensure consistent data analysis across the whole needle width, the field of view of both cameras was transformed into
the same coordinate space by registering the recorded frames of the second camera to the first one. Frames of the measurement plane that lay on the needle centre axis (Position A) were recorded by both cameras simultaneously. Images recorded simultaneously in the centre measurement plane were paired and the rotation and scale factor of camera 2 with respect to camera 1 were found using Matlab (Mathworks Inc., USA). Matched features of the needle shape between both images were detected using the speeded-up robust features algorithm (SURF) [24]. Using corresponding point pairs, the transformation matrix (scale and rotation) was found using the M-estimator sample consensus algorithm (MSAC) [25]. By applying the transformation matrix to the frames recorded by the second camera, the fields of view of both cameras were aligned, resulting in consistent full field measurements in five measurement planes across the needle.

For each scanning position, images of the fluorescent particles in the specific light plane inside the tissue phantom were recorded. Over the course of the insertion, particles enter and exit the measurement planes, hence making conventional particle tracking techniques unfeasible. Instead, frame-to-frame velocities in a spatially fixed grid were obtained via image correlation and interpolated to obtain Lagrangian displacements (as in Oldfield et al. [20]). Similar set-ups for scanning particle image velocimetry in flow applications, such as by Hori et al. [26], have been used to measure complex instantaneous vortex structures in a volume. In our work, a stereoscopic approach was not pursued, as the goal was to measure Lagrangian deformations over the entire insertion. While stereoscopic methods can provide instantaneous out-of-plane displacement components, particles cannot be tracked after leaving the plane. Thus, two-dimensional displacements were computed in the five measurement planes across the needle.

Subsequent frames in each measurement plane were cross-correlated to obtain instantaneous displacements. The needle shape was segmented from the particle images so that correlation results were not biased close to the interface. Recorded frames were 1360 px × 1024 px in size, which resulted in a field of view of approximately 25 × 20 mm. Frames were correlated in two passes, with square subsets of 128 px and 64 px, respectively, and 50% overlap, using a particle image velocimetry algorithm (PIVlab [27]). Rigid body experiments showed no significant effects on measurements due to image distortion. Magnification, due to the different refractive indices of the gelatin and the container’s walls to air, was calibrated by using the needle diameter in the field of view as a reference. The resulting spatial resolution of the correlation grid was approximately 0.6 mm. Starting from the first recorded frame with the needle outside of the tissue phantom, virtual material points were defined by the initial correlation grid. Over subsequent frames, Lagrangian material locations \( x(t) \) were spatially interpolated. Points in the undeformed state \( X \) at time \( t = 0 \) were mapped into the deformed state at time \( t \):

\[
x(t) = \chi(X, t).
\]

Displacement gradients were then calculated from bilinear FEIs in which displacement fields \( u \) are interpolated over the element, using a digital image correlation toolbox [28]. From displacement gradients, Green–Lagrangian finite strain tensors were obtained, with the components:

\[
e_{ij} = \frac{1}{2} \left( \frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i} + \frac{\partial u_k}{\partial x_i} \frac{\partial u_k}{\partial x_j} \right).
\]

The two normal and the shear strain components, \( e_{11}, e_{22} \), and \( e_{12} \), respectively, were combined into a single quantity, in order to express the resulting state of strain around the needle. It is
formed by the principal strains, $e_1$ and $e_2$:

$$e_{1,2} = \frac{e_{11} + e_{22}}{2} \pm \sqrt{\left(\frac{e_{11} - e_{22}}{2}\right)^2 + \left(e_{12}\right)^2}.$$ 

Here, the effective strain $e_{\text{eff}}$ was chosen, as this measure was used in other studies to quantify strain within tissue [29]:

$$e_{\text{eff}} = \sqrt{\frac{1}{2}(e_1^2 + e_2^2)}.$$ 

As strains are calculated from displacement gradients of the bilinear elements, results are sensitive to noise in heterogeneous regions with large gradients, as discussed by Lava et al. [30]. Tool–tissue interactions cause large deformations very close to the needle. Therefore, spatially smoothing displacements prior to building the gradients is required in order to obtain continuous strain fields. Here, a local regression filter, which fitted a data point and surroundings that fell within a radius of seven neighbouring points to a linear plane, was applied to the interpolated displacement fields. Control rigid body experiments showed this smoothing led to more consistent strain results.

### 3. Results

During needle insertions with DP and PB motion profiles, frames were recorded in five measurement planes across the needle width. Displacement and strain fields were computed for each plane, resulting in sliced volume measurements across the needle width while the needle travelled through the tissue phantom (figure 5).

For comparison, a point at the same insertion depth of approximately 31.4 mm was chosen for DP and PB experiments. At this point, all four parts of the needle were aligned during the PB motion profile and the needle had approximately travelled to the centre of the field of view. Displacements and strains in the vicinity of the needle were averaged and compared across the needle width. Displacements that fell within a neighbourhood of 1–3 mm distance from the needle interface were averaged. The evaluation points of the grid for the strain computation formed by bilinear FEs lay within the displacement grid, which led to losses of information at the grid edges. Therefore, strain evaluation points that fell within a neighbourhood of 1.5–3.5 mm from the needle interface were considered.

#### 3.1. Depth-resolved displacement and strain measurements for different motion profiles

Computed displacements on the needle centre plane (Positions A1 and B1) measured with both cameras should match closely after image registration and correlation. By registering images during a needle insertion at the same point of time, the field of view of both cameras was aligned (figure 6a). This enabled the extraction of measurements from all measurement planes at the same location relative to the needle. Displacements at a position that was 4 mm behind the aligned needle tip were extracted for all experiments. At this location, the substrate has been fully parted by the needle, leaving the deformed material surrounding the needle shaft in focus. Figure 6b shows box plots of measured displacements on the needle centre axis for all experiments using DP and PB motion profiles, split into recordings A1 and A2 of the respective cameras. Means and standard deviations of displacements measured by both cameras match very closely for both experiments (table 2). An unpaired $t$-test was performed for both experiments and showed non-significant means of samples of both cameras in DP and PB experiments, confirming the
accurate alignment of the fields of view and the accuracy of the image correlation. Hence, measurements by both cameras in the centre plane were treated as one sample group.

The depth-resolved distributions of displacements and strains for DP and PB motion profiles are shown in figure 7. Displacements and strains are lowest on the needle edges and tend to be higher towards the needle centre plane, with DP mean strains increasing from 4.9% on the edge to 7.2% in the centre. In general, displacement and strain magnitudes are significantly reduced in PB experiments in comparison to DP. Whereas the highest displacement and strain of DP lie on the centre plane ($z = 0$ mm), the peak for PB lies on the offset plane at $z = -1$ mm. For statistical analysis, centre plane measurements have been analysed via an unpaired $t$-test, showing statistically significant differences between DP and PB for both displacement magnitude and effective strain ($p$-values of $4 \times 10^{-12}$ and $1 \times 10^{-13}$, respectively). Displacements and strains of PB compared to DP were reduced by 30% and 41%, respectively (table 3).

Table 2. Measured mean displacements in needle surroundings on centre planes A1 and A2, respectively. The measurements of both cameras match for DP and PB experiments.

<table>
<thead>
<tr>
<th>camera</th>
<th>$N$</th>
<th>mean (mm)</th>
<th>s.d. (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>DP</td>
<td>1</td>
<td>8</td>
<td>1.27</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>8</td>
<td>1.24</td>
</tr>
<tr>
<td>PB</td>
<td>1</td>
<td>10</td>
<td>0.87</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>10</td>
<td>0.88</td>
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Axial strains are compressive in front of the needle, as the tip is approaching, and decrease along the shaft, as material is dragged with the needle during the insertion (figure 8a). Radial strains change from tension ahead of the tip, as the crack begins to open, to compression around the shaft, as the needle occupies the volume inside the sample box (figure 8b). Shear strains are close to zero in front of the needle and increase around the needle shaft, when the crack is formed.

3.2. Evolution of internal deformations

Figure 9 shows the evolution of displacements and strains on the centre plane, averaged across all experiments, at the depth of 31.4 mm within the tissue phantom, as the needle is traveling through it. At the beginning of the insertion, at a 8 mm insertion depth, a local peak is observable, caused by initial puncture of the needle in the sample. Over time, displacements and strains are showing a cyclical increase and decrease for the PB motion profile, as the needle segments are pushed forward and pulled back regularly during the insertion. Consistently, over the course of the insertion, displacements and strains for PB are below those for DP. After the needle tip has passed and the full crack has been developed, displacements and strains for DP increase continuously, but at a slower rate. PB displacements remain stable and strains decrease slightly.

4. Discussion

Deformations resulting from interactions of a needle inserted within a substrate were analysed for experiments performed with conventional DP and biologically inspired PB motion profiles, for a 4 mm, four-part needle. Optical measurements via laser-based image correlation allowed us to obtain a comprehensive set of full-field displacements and strains inside a soft tissue phantom, resolved across five measurement planes on the needle interface over the course of the entire insertion. By extending the image correlation measurements to multiple planes across the whole needle width, the full material response due to friction, cutting and crack forming was captured. This allows us to compare the material response due to
different insertion parameters. Here, we inserted the needle with two different motion profiles.

The image registration and cross correlation of subsequent frames was shown to be reliable by comparing simultaneous measurements of both cameras on the measurement plane located on the central needle insertion axis. Results of both cameras in the vicinity of the needle showed very good agreement (table 2).

After the needle tip has passed, material is deformed the most where the needle has cut the material. Displacements and strains were the lowest at the needle edges and higher towards the needle centre plane (figure 7). While for DP the segments are fully aligned and the crack is formed by its symmetric tip, in PB the material is cut by the bevel tip of the leading segment (§2.1). This results in a different distribution of displacements and strains, with the highest values measured on the measurement plane lying on the bevel tip of the leading segment (position B1).

The PB motion profile causes changing contact interactions between the needle and the surrounding substrate. Overall, displacements and strains during insertion were significantly reduced compared with DP across the needle width (table 3). Strains (normal, radial and shear) in the needle surroundings are also reduced for PB at the same insertion depth (figure 8). This shows that the continuous increase of deformation, owing to an increasing frictional contact between needle and substrate, is less during a PB insertion than during DP. As a result, axial strains are less compressive and shear strains in the substrate are reduced. This reduction of deformations inside the tissue phantom was achieved solely by changing the way the needle was actuated, at the same net insertion speed (as to reach a targeted location in the same amount of time). Whilst the needle burrows itself deeper into the substrate, the relatively small pullback motion leads to sticking contact in the opposite direction, impeding material from being displaced. Conversely, material is continuously dragged with the full needle during the DP insertion. With the material being anchored and remaining closer to its undeformed configuration with PB, effective strains are reduced.

The cyclic nature of the interaction between the needle with the PB motion and the substrate manifests itself in the evolution of internal deformations at a given insertion depth. Although displacements and strains in DP increase continuously, PB shows increases and decreases with every cycle. After the needle tip has passed a point in the substrate, the substrate has been parted and increasing frictional drag in DP leads to an increase in axial strain around the needle interface. In PB, displacements and strains remain fairly stable, as overall material motion is reduced by the back and forth motion of the segments. Strains even decrease slightly, indicating that material is releasing strain energy, while it is kept at the same position over time.

Casing reduced tissue motion and deformation during an insertion can increase the accuracy and hence the success for a medical procedure. If the tissue configuration has changed less due to the needle being inserted, surgeons can rely with higher confidence on internal scans acquired preoperatively. By combining this needle insertion method with tissue constraining techniques, e.g. via suction or compression, tissue mobility can potentially be reduced even further. Additionally, reduced strains mean that less force is acting by the tissue on the needle. This can potentially reduce the risk of buckling for thin and flexible needles that can be actively steered within soft tissue during insertion.

The OD of the needle prototype in this study is larger than that of typical MI surgical instruments, such as those used for biopsy or brachytherapy. However, we have been using scaled up prototypes in testing for several years, as they are easier and cheaper to manufacture and handle within a laboratory set-up. In order to produce results for easy interpretation in the context of previous and parallel work on needle steering [31] and control [15], we chose to keep the OD as 4 mm, which is the same size as was used in these experiments. Over the years, we have continued to reduce the OD of successive prototypes by means of design optimization (e.g. [32]) and more
motion profiles to improve substrate deformation depending on its particular and evolving material characteristics (e.g. moving from grey matter to white matter in the brain) and we have preliminary, unpublished tests that indicate the needle presented here is also able to penetrate stiffer and tougher materials, such as liver.

By carefully optimizing parameters, such as stroke length or pullback, the PB motion profile can be tuned to further reduce disruptiveness. Replicating the experiments with FE models may present an efficient way of optimizing motion profile parameters. Three-dimensional findings presented here can be used to further develop FE models (as presented in [21]) that realistically capture the interactions associated with fracture, as well as friction. The anchoring effect could also be enhanced by increasing directional frictional resistance of the outer needle surface, e.g. with surface textures [35]. However, this may lead to increased tissue trauma when the needle is retracted.

The biologically inspired motion profiles, with lower displacements and strains in the surrounding material, could lead to a more tissue sparing approach to percutaneous intervention, as internal strains can be linked to functional damage [36]. The gelatin soft tissue phantom, however,
shows less viscoelastic behaviour than real soft tissue [22]. As the material around the needle is undergoing repeated cycles of being dragged and pulled back, it is more likely for temporal relaxation to occur, which would lead to a reduction in stress. Investigations using a more viscous material could reveal more insights about time-dependent changes in interactions, such as decreasing strains and stresses inside the substrate (figure 9).

Although the material response of a tissue phantom can only give an indication, histology studies could be used for an assessment of the tissue damage caused by both insertion profiles. New methods to make tissue transparent, such as CLARITY [37], are able to reveal intact brain structure and could be used for detailed damage studies.

5. Conclusion
In this work, we present the most complete, high fidelity tool–tissue interaction study of a biologically inspired needle insertion profile, ‘pull back’, which is compared to conventional ‘DP’ for a 4 mm, four-segment plastic needle. By means of optical measurements inside a soft tissue phantom, we reveal how the material at the needle interface is dragged less during the insertion process, leading to cyclically decreasing displacements and strains inside the substrate for our biologically inspired approach. At the same insertion depth, displacement magnitudes (−30%) and effective strains (−41%) in the needle surroundings were significantly less than after a DP-based insertion, where the full needle is inserted at a matched, constant insertion speed.

By exploiting the benefits of our biologically inspired motion profile, needles can be inserted more accurately into soft tissue and flexible needles can be steered with smaller risk of buckling, enabling safer insertions, with the potential for reduced tissue damage. Tests in ex vivo tissue and advanced numerical models will be required in order to optimize insertion parameters of the needle motion profile for better accuracy and safety.

Additionally, our immediate goal will be to marry this unique insertion strategy with ongoing work on surgical needle steering, for which we are close to a clinically viable prototype for application to neurosurgical drug delivery (www.eden2020.eu).

Data accessibility. See the electronic supplementary material for Excel file with DP and PB displacements and strains, and recording during performed needle insertion with scanning laser-based set-up.

Authors’ contributions. A.L. designed the optical set-up, carried out the experiments, post-processed the data and drafted the manuscript. M.O. defined the pullback motion concept and helped draft the manuscript. F.R. conceived the study, coordinated the work and helped draft the manuscript. All authors gave final approval for publication.

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