1 Introduction

Numerous surgical cutting tools such as medical needles and biopsy punches (BPs) are frequently used in minimally invasive procedures, aimed both at the extraction of tissue or fluid samples and at the delivery of medical fluids for common operations like regional anesthesia and special treatments. For instance, a skin biopsy is a medical procedure where a cutaneous lesion is extracted from the human body and delivered to a pathologist to perform the diagnosis. In a punch biopsy, in particular, hollow needles, usually termed BPs, are pressed down into the skin to extract a tissue sample. BPs are the primary medical tools adopted for diagnosing skin disorders [1], such as skin cancer, which has the highest incidence with respect to all other cancers combined in the U.S. [2]. If the punch biopsy is performed incorrectly, a pathologist’s interpretation of a skin biopsy can be limited or erroneous. In this context, the cutting efficiency of these medical tools plays a crucial role. Although the previously mentioned devices have been widely adopted, the problems related to effective pain reduction and accurate cutting have so far not been completely addressed. Reducing pain and improving cutting accuracy [3] for the extraction of an optimal quantity and quality of tissue samples are two key objectives in this research area as they are crucial to treatment outcomes and patient care. In the current work, the focus will be placed on the cutting performance of novel bioinspired microserrated biopsy punches.

The concept of improving soft tissue cutting by providing microserrations on the cutting tip of BPs during medical procedures has been inspired by nature and, in particular, by the mosquito’s maxilla. In fact, its cutting edge features numerous microteeth whose function is to enhance the insertion of the maxilla and diminish nerve stimulation during puncture. Atkins and several other researchers [4–6] have already performed some preliminary studies related to the presence of serrated cutting edges that mimic certain animal features, such as the mosquito’s proboscis, stinger of bees, and fish teeth. Izumi et al. [7] have studied the performances of insertions executed with silicon microneedles that mimic the mosquito’s maxilla. Similar needles were also realized by Aoyagi et al. [8] and Oka et al. [9]. The needles were characterized by a serrated shape inspired by the geometry of mosquito’s proboscis and were capable of indenting a layer composed of hard silicon. Kong and Wu [10] performed a study on the prediction of the insertion force of the mosquito’s fascicle from an experimental and computational standpoint. They identified the nanometer tip radius of the mosquito’s fascicle as one of the causes of its extremely low insertion force (18 μN). However, few research works have been performed on the impact of microserrations similar to those seen on the mosquito’s proboscis on insertion forces and their possible implementation in medical devices such as biopsy punches.

Microfeatures on the BPs’ tip hold the potential to positively reduce the penetration force and ultimately diminish the patient’s discomfort. Han et al. [11,12] observed the effect of microtextures realized on several biopsy needles on their insertion force. Further, the authors and Han [13] performed some investigations on tissue cutting by means of microserrated medical devices [14,15].

This paper is aimed at expanding the previously referenced studies and exploring the application of microserrations to biopsy punches. Commercial and microserrated BPs are investigated with the purpose of clarifying the influence of microserrations on tissue cutting forces. First, the experimental setup for the manufacturing of BPs is explained. Second, the performance of commercial and serrated BPs is studied through BP penetrations into phantom tissue. Finally, three-dimensional (3D) and two-dimensional (2D) finite element method (FEM) simulations are performed to provide additional insights into stress generation in the cutting area. The
experimental results will prove that significant reductions in the puncture force of BPs are achievable.

2 Materials and Methods

2.1 Laser Ablation Process. To generate microfeatures on the tip of commercial biopsy punches, a picosecond laser with a wavelength of 532 nm was used. This laser [14] was coupled with a five-axis motion system and a controller with a resolution of 0.01 μm. The system facilitates the rotation and translation of the workpiece during the laser ablation process [16,17].

The setup is displayed in Fig. 1. The main components are as follows: (1) needle fixtures and (2) focusing and vision subsystem. The needle fixture subsystem consists of a microadjuster, an adapter, an ER8 collet, and a shaft. The ER8 collet holds the biopsy punch during the laser machining process, and it is mounted on a shaft, which is connected to the laser’s stage through a microadjuster. The focusing and vision subsystem consists of a polarizing beam splitter, an optical illuminator, a charge-coupled device camera, a camera lens, and a notch filter. The charge-coupled device camera captures magnified images of the workpiece, and its focal position is aligned with the laser focus [12].

2.2 Microserration Design. Microserrations were generated on a commercial BP with a cannula with a circular cutting edge. This cannula features an outside diameter of 2.4 mm (D), an inside diameter of 2 mm (d), a bevel length of 0.7 mm (l), a tip radius of 15 μm, and an included angle (θ) of 10 deg at the cutting edge (Fig. 2).

To generate microfeatures on the BP cannula, its cutting edge was reshaped by laser ablation. The laser power was set to 0.75 W and the frequency of laser pulses to 100 kHz. The serrated profile is defined by its radius at the tip (r_s), the arc radius (r_c), and the angle between two consecutive serrations (α). Figure 3 shows the laser path used to generate the microfeatures on the BP cannula.

In this work, the laser followed an offset path, shown in red in Fig. 3(a), with respect to the ultimate shape. The laser trajectory is characterized by a pulse overlap of 90%. It is progressively ablating the material, eventually leading to the generation of the desired features along the cutting edge of the biopsy punch.

Several microserration patterns were designed, manufactured, and tested on different BP cutting edges. Each pattern had a circular profile (Fig. 3) with different arc radii (r_c) which ranged from 50 μm to 600 μm (Table 1). The radius at the serration edges (r_s) after laser processing, is approximately equal to 15 μm for all microserrated BPs (Table 1), while the radius at the cutting edge (r) is not affected by the ablation process (Fig. 2).

The minimal arc radius for the BP serrations (r_s) was set to be equal 50 μm. Microserrations with a serration radius (r_s) smaller than 50 μm will not significantly impact the original geometry of the BP’s cutting edge. In fact, the tip radius (r) of the BP cutting edge is equal to 15 μm (Fig. 2), and any serration with a serration radius (r_s) of the same order as the tip radius (r) would not lead to any effective change in the cutting edge geometry. The serration configurations are presented in Table 1.

Following the laser ablation process, the geometry of each BP cutting edge was verified by using a 3D profilometer (Fig. 4).

2.3 Fracture Mechanics Approach. During the insertion of the biopsy punch, the puncture force can be analyzed by studying the fracture mechanics related to the BP insertion [18]. According to the J-integral method [19], a crack will propagate inside the tissue when the energy generated by the insertion is equal or greater than the energy needed to extend the crack (R), i.e.,

\[ J \geq R \]  

where R is defined as the tissue fracture toughness, and it represents the energy required to propagate the crack [18].

<table>
<thead>
<tr>
<th>ID</th>
<th>D (mm)</th>
<th>a</th>
<th>r_s (μm)</th>
<th>r_c (μm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>C1</td>
<td>2.4</td>
<td>20</td>
<td>15</td>
<td>50</td>
</tr>
<tr>
<td>C2</td>
<td>2.4</td>
<td>16</td>
<td>15</td>
<td>150</td>
</tr>
<tr>
<td>C3</td>
<td>2.4</td>
<td>12</td>
<td>15</td>
<td>250</td>
</tr>
<tr>
<td>C4</td>
<td>2.4</td>
<td>10</td>
<td>15</td>
<td>400</td>
</tr>
<tr>
<td>C5</td>
<td>2.4</td>
<td>10</td>
<td>15</td>
<td>600</td>
</tr>
</tbody>
</table>

Each configuration is identified by the BP diameter (D), number of microserrations (a), and microserration radii (r_s and r_c) as shown in Fig. 3.
nonlinear energy release rate \( (J) \) is equal to the derivative with respect to the contact area \( (A) \) of the difference between the strain energy of the tissue \( (U) \) and the external work applied by the punch \( (W) \) \[20,21\] as

\[
J = \frac{d(U - W)}{dA} \tag{2}
\]

When tissue fracture occurs at the end of phase I (as it will be shown later in Fig. 7), the punch is stationary, and its work \( (W) \) during crack propagation is equal to zero, so \( J \) accounts only for the strain energy

\[
J = \frac{dU}{dA} \tag{3}
\]

In this scenario, it is extremely challenging to calculate \( U \), because of the complexity of the crack geometry. Further, the strain energy can be derived from the stress by adopting the constitutive equations of the tissue. According to Mahvash and Dupont \[22\], it can be assumed that the pressure \( (p) \) over the contact area \( (A) \) between the punch and the tissue is constant, and the resulting stresses in the soft tissue are proportional to the contact pressure. In light of these considerations, it is possible to establish a proportional relationship between \( J \) and the contact pressure \( (p) \), which also includes a crack intensification factor \( K \) \[23\] and a coefficient \( m \), which accounts for the nonlinearity of the material as

\[
J \propto Kp^m
\tag{4}
\]

Since the contact pressure \( (p) \) is equal to the ratio between the axial force \( (F) \) that pushes the BP and the contact area \( (A) \), Eq. (4) can be rearranged as \[22\]

\[
F \propto \frac{R}{K} A \tag{5}
\]

Therefore, according to Eq. (5), the axial force at puncture is proportional to the contact area between the BP and soft tissue \[22\]. During insertions performed with microserrated BPs, the contact area \( (A_{\text{SERR}}) \) is lower than the contact area \( (A_{\text{STAND}}) \) during insertions performed with standard BPs. In fact, microserrated edges exert the cutting force over reduced contact areas, i.e.,

\[
A_{\text{SERR}} < A_{\text{STAND}} \tag{6}
\]

\[
\frac{\sqrt{R}}{K} A_{\text{SERR}} < \frac{\sqrt{R}}{K} A_{\text{STAND}} \tag{7}
\]

This effect leads to the decrease in the puncture force during insertions with microserrated BPs and to the increase in localized stresses in the soft tissue to be further discussed in Sec. 4. Because this phenomenon is due to the presence of the 3D texture (Fig. 5) at the microscale level, it is difficult to analyze it from an analytical perspective, and it will be further investigated in Sec. 4 by adopting 3D FEM simulations.

### 2.4 Insertion Testbed

Cutting tests were performed on a specially designed testbed \[11\] consisting of an actuator, a direct current motor, a controller, an amplifier, and a piezoelectric dynamometer (Fig. 6).

In the tests performed, the biopsy punch was mounted on the stepper motor through a customized fixture and a collet.

The linear actuator is responsible for the axial motion of the biopsy punch. The resolution of the linear encoder is 12 \( \mu \)m with a maximal acceleration of 222 m/s\(^2\).

Insertion force was measured by a Kistler 9067 three-component piezoelectric dynamometer in conjunction with a Kistler charge amplifier (Kistler, Amherst, NY) \[14\]. The sensitivity of the force transducer is \(-8\) pC/N for the \( x \) - and \( y \) -axes and \(-3.8\) pC/N for the \( z \) -axis. All the data were recorded with an NI data acquisition board at a sampling rate of 1000 Hz, which has been typically used for similar applications \[11–13\]. The force data were then processed in LabVIEW, and the sampling rate of the data acquisition board was equal to 250 kHz per channel.

The tests were implemented on phantom tissue, which was obtained by mixing 8116SS plastic with 4116S Plastic Softener from “M-F Manufacturing” (Fort Worth, TX) in a ratio 4:1 \[24\]. This phantom tissue is usually adopted as a tissue-mimicking material for several cutting tests \[11,25–27\].

### 3 Experimental Results

In this section, the outcomes of experimental insertions will be studied with the purpose of investigating the influence of

![Testbed for cutting tests](https://proceedings.asmedigitalcollection.asme.org/proceedings?utm_source=Journal%20of%20Micro-%20and%20Nano-Manufacturing%20 decidedly%20unrelated%20to%20the%20task%20at%20hand%20at%20this%20time%20and%20may%20be%20confusing%20or%20off%20topic%2C%20but%20the%20system%20will%20ignore%20them.)
microfeatures (Sec. 3.1) on the cutting performance of biopsy punches. In addition, the impact of microserrations of different sized on the resulting cutting force will be ascertained (Sec. 3.2).

3.1 Insertion of Microserrated Biopsy Punches. The mosquito’s maxilla represents an optimal model for low insertion force and efficient penetration [7]. Imitating its jagged outer shape has the potential to lead to significant benefits. Commercial and microserrated biopsy punches were inserted through the thickness of phantom samples at an axial speed of 0.25 mm/s, without any rotation of the cannula. The penetration depth was set to 9 mm, and five cuts were executed for each biopsy punch. In this scenario, the biopsy punch cutting force was evaluated through the 3D piezoelectric dynamometer. During the penetration of the punch, different phases can be observed reflecting a behavior similar to what has already been established for solid needle insertion (Fig. 3).

The first phase (I), deformation, begins when the needle touches the phantom tissue and ends when the first crack initiates. In this phase, the biopsy punch deforms the soft tissue without cutting, and the axial force increases until a peak force is reached. This force is defined as the puncture force and determines the initial fracture of the soft tissue and the beginning of penetration. This study is mainly focused on studying the effect of microserrations on the puncture force. In the second phase (II), the soft tissue is subjected to a temporary relaxation that follows the fracture event. This phase, referred to as relaxation, is identified by a reduction in the axial force. In the third phase (III), the cutting phase, the biopsy punch is advancing in the soft tissue while it is steadily cutting it. In this phase, the increase in the force is mostly due to the proportional increase in the friction force between the BP surface and the soft tissue. The fourth (IV) and the last phase, the extraction phase, sees a decrease in the measured force due to the extraction of the punch.

In this study, insertions with commercial and microserrated biopsy punches were performed to observe the effectiveness of the technique and highlight the advantages of microserrations (Fig. 8). By comparing the force profiles for biopsy punches with and without microserrations, it is evident that the forces at puncture are substantially different.

Figure 8 shows the axial force for a commercial and a microserrated BP with a “C4” texture (Table 1). The presence of microserrations on the BPs cutting tip leads to a reduction in the puncture force from 1.19 N to 0.75 N. An unpaired T-test, which assumes unequal variances between samples, was performed by using the force data measured during commercial and microserrated BP insertions. For microserrated BP with a C4 texture (Fig. 8), the axial force ($\mu = 0.87$, $\sigma = 0.1$) proved to be 22.5% lower ($p$-value = 0.016) than the axial force measured during insertions with commercial BPs ($\mu = 1.12$, $\sigma = 0.04$). This phenomenon can be explained by the concentration of higher localized stresses (Sec. 2.3) in the contact area between the microserrations and soft tissue, which provokes its earlier fracture. In fact, the application of microserrations also leads to a reduction in the tissue’s initial displacement (Fig. 8) at fracture from 6.6 mm ($d_{\text{STAND}}$) to 6.05 mm ($d_{\text{sERR}}$).

3.2 Texture Depth Effect in Microserrated Biopsy Punches Insertion. To better investigate the influence of microserrations, similar experiments were conducted using biopsy punches with different texture sizes (Table 1). The measured cutting forces were approximately between 0.8 N and 1.15 N for all tested punches. Microserrations with larger texture depths (Fig. 9) lead to smaller cutting forces and consequently to higher variations in the axial force ($\Delta F = (F_{\text{sERR}} - F_{\text{STAND}})/F_{\text{STAND}} \times 100$). Unpaired T-tests were run by comparing the forces measured during insertions of commercial and microserrated BPs (Table 2). The results show that microserrated biopsy punches lead to a reduction in the BP axial force between 7% and 30%.

The fact that deeper textures lead to lower cutting forces is mainly due to the manner in which microserrations engage with soft tissue during BP penetration. For instance, insertions performed with BPs with a texture radius of 400 $\mu$m present a smaller contact area between the BP and tissue than insertions performed with BPs with texture radius of 150 $\mu$m. The presence of larger microserrations on the BP likely determines higher localized stresses in soft tissue, since the contact area is smaller. This will also be observed in the following computational study (Sec. 4).

The reduction in the cutting force is not consistent ($\Delta F = 2%,$
Fig. 9) only for microserrated punches with a texture radius of 50 \( \mu \text{m} \), since its value is comparable to the tip radius at the BP’s cutting edge \( r \), which is equal to 15 \( \mu \text{m} \) and does not dramatically affect the geometry of the cutting edge.

4 Numerical Modeling of Serrated Biopsy Punches Penetration

In this section, computational studies related to the insertion of microserrated biopsy punches will be performed. In Sec. 4.1, a comparison between a standard and a microserrated BP will be presented by adopting 3D FEM simulations aimed at studying the stresses induced in the contact area by the microserrated profiles. In Sec. 4.2, the impact of the microserration radius on the cutting performance of BPs will be evaluated through 2D FEM simulations. This choice is dictated by the high computational cost of 3D FEM simulations for modeling serrated cutting edges and the possibility of adopting a plane stress condition for this case.

4.1 Computational Study of Biopsy Punches Insertion. To provide additional insights into the efficacy of microserrations, a finite element model was formulated (Fig. 10) with ABAQUS, while the mesh for the biopsy punch and the soft tissue was realized with HyperMesh. In this computational study, two different cases were analyzed where the same tissue block was indented by a commercial BP in the first scenario and by a microserrated BP, with a texture depth of 600 \( \mu \text{m} \), in the second scenario.

The BP was described as a discrete rigid body. Its geometry was set to be equal to the geometry of the BPs adopted during the experiments (Fig. 2). A translational motion was applied to the BP that moves parallel to the thickness of the tissue, and it is constrained with respect to the remaining translational and rotational motions. The BPs were meshed with three-dimensional triangular (R3D3) and quadrilateral (R3D4) elements. Specifically, the finite element mesh of the commercial BP mesh was built with 37,255 nodes, 37,105 R3D4 elements, and 24 R3D3 elements. More elements were used to mesh the microserrated BP to have a better definition of the serrated edges. The mesh of the microserrated BP was built with 148,637 nodes, 146,689 R3D4 elements, and 3896 R3D3 elements.

The soft tissue block was considered as a deformable body, and it was represented by a cylinder with a radius of 7.5 mm and a thickness of 3 mm. Since the model is symmetric with respect to the \( xz \) and \( yz \) planes, a quarter of the cylinder with symmetric boundary conditions was used in the simulations (Fig. 10) as the soft tissue. The base of the tissue was completely constrained, while the external cylindrical surface was assumed to be free. The mesh was built by adopting three-dimensional eight-node hexahedral elements (C3D8) and three-dimensional six-node wedge elements (C3D6). The finite element mesh consists of 4,833,310 nodes, 4,669,704 C3D8 elements, and 31,536 C3D6 elements. The mesh consists of 441 elements along the radius, 72 elements around the circular edge, and 35 elements along the thickness of the block. However, in the cutting area, which interacts with the BP serrations, the number of elements has been consistently increased. In fact, to define the stress gradient in the tissue being cut by a cannula with a radius at the tip of 15 \( \mu \text{m} \), the tissue was meshed with 1 \( \mu \text{m} \) elements in the tissue–cutting edge interaction zone (Fig. 10(d)).

The tissue was modeled by adopting hyperelastic constitutive models, since they provide a suitable framework to analyze large strains (>1–2%). In this study, the tissue was modeled by using the Arruda–Boyce model, since it better describes the large stretch behavior of soft tissues. Its strain energy density function is defined as \( \psi \) [28]

![Fig. 10 3D finite element model of BP insertion. The mesh of the BP and of the soft tissue is shown in the top view (a), 3D view (b), and side view (c). The mesh is refined in the proximity of the microserrations (d).](https://proceedings.asmedigitalcollection.asme.org/micronanomanufacturing/article-pdf/5/4/041004/5950772/jmnm_005_04_041004.pdf)
\[ \psi = c_1 \left[ \frac{1}{2} \left( I_1^p - 3 \right) + \frac{1}{20 \mu_m^2} \left( I_1^p \right)^2 - 9 \right] + \frac{11}{1050 \mu_m^4} \left( I_1^p \right)^3 - 27 + \frac{19}{7000 \mu_m^6} \left( I_1^p \right)^4 - 81 \right] + \frac{519}{673.750 \mu_m^8} \left( I_1^p \right)^5 - 243 \right] + \frac{1}{D_1} \left( \frac{F^2}{2} - \ln(J) \right) \]

(8)

where \( c_1, \mu_m, \) and \( D_1 \) are material constants, which are equal to 0.002, 1.600, and 0.73, respectively. These material constants were already adopted in previous computational studies for tissue-mimicking material for cutting tests [13].

During the simulation, the cutting edge of the BP was pushed toward the tissue. The stresses generated in the contact zone during cutting with the commercial BP and with microserrated BP were compared. All the model parameters were kept identical in the two simulations, including the mesh, boundary conditions, and constitutive model of the soft tissue. Since soft solids can fail under critical tensile stresses [29], the first principal stresses in the tissue contact zone were investigated.

Figure 11 shows the contour plot of the maximum principal stress when the commercial BP reaches a penetration depth of 500 \( \mu m \). The stresses are highly localized in the cutting edge–tissue contact zone. The maximum tensile stress is found to be distributed at the external side of the cutting edge–tissue contact zone, and it is approximately equal to 1 MPa (Fig. 11(b)). The tensile stress results from the stretching of the tissue surface, while the internal side of the cutting edge–tissue contact zone is characterized by negative values of the principal stress.

For insertions performed with microserrated BPs (Fig. 11), the simulations show that for the same BP penetration depth (500 \( \mu m \)), the stresses are exclusively localized at the corner of the microserrations, and they are approximately equal to 5 MPa (Fig. 11(d)), which is five times higher than the maximum principal stresses recorded for indentation with commercial BPs.

The presence of microfeatures along the tip of the BP cannula leads to a rapid rise in stresses in localized areas, which favors the initiation of fracture and the consequent cutting of the soft tissue. Figure 12 shows the comparison between the experimental insertion force profiles and the ones computed by finite element simulations for commercial and serrated BPs.

The behavior of the axial force is similar to the one observed in the experiments (Fig. 8) although the force values are considerably lower. In fact, Fig. 12 corresponds to a BP penetration of 500 \( \mu m \) in the effort to limit the computational cost of the simulation.

Despite the fact that these simulations do not analyze the crack initiation in soft tissue, they provide a framework to understand the reasons why microserrations on devices are more effective in soft tissue cutting. The serrated profile causes an earlier rise in contact forces, which leads to an earlier fracture of the tissue and lower cutting forces in comparison to plain edge profiles.
4.2 Computational Study of Different Serrations. To further investigate the impact of the microserration size on cutting efficiency, a finite element analysis was performed (Fig. 13) with ABAQUS standard. Two scenarios were considered where the same tissue block was indented by microserrated BPs with a texture radius of 50 μm in the first scenario and by a texture radius of 400 μm in the second scenario. A 2D finite element model, rather than a 3D one, was built for each scenario. In fact, in the first model, the BP is characterized by a fine microtexture radius (50 μm) that would require an excessive computational cost if implemented in an FEM 3D model.

The microserrated biopsy punch was defined as a rigid body. The BP geometry was approximated by a rectangular blade 2.75 mm wide and 2 mm thick. In the simulations, the BP indented the tissue for 0.5 mm, and it was constrained in the other directions.

The size of the tissue slab cross section is equal to 10 mm in width and 4 mm in thickness. A plane stress assumption was made in the model to describe micron-scale deformations without excessively impacting the computational costs of the simulation. The base of the tissue was completely constrained, while the other sides were set free of loads. The tissue was meshed with 106,133 linear quadrilateral elements of type CPS4R. The minimum size of the elements is equal to 10 μm in the cutting zone. As in the previous FEM simulations, the tissue was modeled by the Arruda–Boyce model.

One simulation for each microserrated BP insertion was performed. By comparing the contour plots related to the maximum penetration depth reached by the BP (Fig. 14), it could be noticed that the contact area was smaller for microserrated BPs with a larger textured depth (r = 400 μm), as it was described in Sec. 3.2. This leads to higher localized stresses in soft tissue. For a cutting depth of 0.5 mm, BP cutting edges with 400 μm serrations induce in the soft tissue a maximum principal stress ($S_{\text{maxp}}$), which is two times higher than the maximum principal stress that is induced by BP with 50 μm serrations. Even though the values of $S_{\text{maxp}}$ (Fig. 14) are much lower than the values found during 3D simulations (Sec. 4.1), they explain how serrations with different sizes have a different impact on soft tissue cutting. Microserrations with a larger texture depth lead to an earlier rise in tensile stresses and cause an earlier fracture of the tissue and a decrease in the cutting force, as it was indicated from the experimental results (Fig. 9).

5 Conclusions

This paper investigates the effects of serrated biopsy punches on soft tissue cutting. A setup for laser ablation of BP cutting edges was implemented and microfeatures were manufactured on their cutting edges. Insertion tests were performed with the aim to investigate the influence of BP microserrations on the cutting...
force. Three-dimensional finite element simulations have been performed regarding the deformation of hyperelastic soft tissue undergoing large deformations. This computational study was able to capture the micro- and macro-scale aspects of the cutting phenomena before the fracture of the hyperelastic soft tissue.

The results show that microserrations on the cutting tip of the BP cannula lead to a significant reduction in the cutting forces and potentially to less pain during soft tissue cutting. The 2D finite element simulations performed provide insights related to the comparison between the performances of BPs, characterized by microserrations of different sizes.

In the future, additional computational studies related to micro-scale modeling and soft tissue cutting should be performed. This end, it is crucial to develop FEM models to predict the cutting of the tissue and the evolution of the insertion forces over a wider range of displacements. The rise in stresses in localized tissue areas due to the presence of BP microserrations should be further investigated to analyze possible implications in terms of tissue damage and consequences on surface integrity.

The manufacturing of microserrations on BPs characterized by different geometries and dimensions should be considered. According to this study, the presence of microserrations could increase the BP cutting efficiency, but could also make the BP cutting edge more fragile and prone to breakage. Therefore, in the future, an optimum balance should be identified to obtain efficient and resilient microserration geometries.

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