Design Choices in Needle Steering—A Review

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***Abstract*—Alignment errors can arise during needle tip placement in deep-seated tissue structures. This can lead to undesired diagnostic and therapeutic outcomes. Path correction by means of needle steering has been investigated in scientific studies for the past decades. Several approaches exist, each of them with their own strengths and weaknesses. This paper describes the various needle-steering techniques, and discusses their applicability in terms of mechanical design choices in order to assess and guide on-going work in this research area. Included steering techniques apply needle base manipulations, bevel tips, precurved stylets, active cannulas, programmable bevels, and tendon actuated tips. Techniques are classified as either passive or active, based on how steering is achieved. Mechanical properties of developed needles differ largely. Flexural rigidity, for instance, was found to vary with a factor 10^6. Mechanical interactions, such as torsion and buckling, are described individually. Different research objectives have led to different steerable needle designs. Design criteria are typically based on these objectives, and not necessarily on clinical needs. However, the effectiveness of steering techniques depends heavily on this design, the navigation medium, and the intended task. In the proposed classification scheme, this dependence is quantified by the flexural rigidity.**

***Index Terms*—Instrument–tissue interactions, mechanical design, medical robotics and systems, motion control, steerable needles.**

I. Introduction

NEEDLES have a crucial role in the modern clinical environment. They are used to extract or inject fluids, perform tissue biopsy, and introduce catheters, ablation electrodes, radioactive seeds, or other instruments to the body. Fig. 1 shows a trocar needle consisting of a base, cannula, stylet, and tip.

The difficulty of needle interventions is getting the tip at the right location. So far, targeting errors have been attributed to human factors, imaging limitations, needle deflections, and dynamic tissue reactions including soft tissue deformations and sliding of multilayered structures [1]. In addition, target movements due to physiological processes, e.g., breathing [2] or heartbeat, may introduce placement difficulties. Consequences of tip misplacement may be an additional damage due to repuncturing [3], misdiagnosis (false negatives), poor dosimetry, and tumor seeding [4].

To further reduce the damage to healthy tissue, there is both the desire to use fine needles, and to expand their use for deeper or more difficult to reach targets. The combination of these factors poses some interesting challenges in terms of the needle–tissue interaction mechanics. After all, there must be a limit to the controlled use of thin and flexible instruments in complex or poorly known environments. For instance, cannula bending and buckling may result from a sudden increase in reaction forces on encountering a tissue membrane. The magnitude of this response relies on mechanical needle properties, such as the cannula's flexural rigidity.

Needle steering describes a research field in which dynamical needle–tissue interactions are used to correct potential placement errors during insertion. Simple translations and rotations of the needle base with respect to the insertion point presumably describe the first needle steering attempts in clinical practice. These base manipulation methods are still in use today, as complete reinsertions of the needle are said to increase the risk of complications [3]. Meanwhile, other steering techniques have been investigated, as will be discussed in this review.

In particular, the modeling and control community picked up needle steering as a practical case to implement a number of fundamental, kinematic, and mechanical navigation models. The elaboration on needle movement restrictions and elastic constraints plays a central role in this field. Several review studies have focussed on this subject [1], [5]–[8]. However, it has been acknowledged that little attention has been paid to steerable needle design [9]. Therefore, this review will focus on actuation mechanisms and mechanical design aspects of developed needles. This is done irrespective of the initial research motivation, whether this was the evaluation of a mechanical system or that of an underlying navigation model. Although the functionality of a mechanical design may be largely influenced by the underlying navigation model, the reverse is also true. The correctness of a navigation model and the relevance of selected parameters are directly related to the design. In fact, design choices may not only affect interaction models quantitatively, but also the framework for which models hold in terms of disturbance rejection and control robustness. The sensitivity to tissue density variations on needle bending is, for instance, influenced by the selected tip type [10]. Alternatively, buckling phenomena are typically not a part of the navigation plan, yet they are often reported in validation experiments [7], [11].

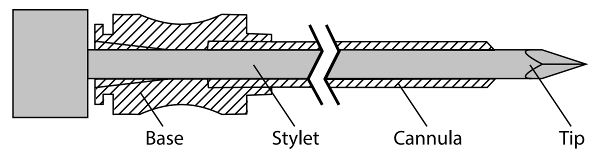

Fig. 1. The trocar needle is composed of a cannula and a stylet. The cannula base is used to handle the needle. The stylet typically contains the tip, used for puncturing tissue.

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II. Review Approach and Focus

A. Methods

A literature search was performed in PubMed and Web of Science (last updated on 24-08-2014) on needle steering, with search queries containing combinations of the words needl*, steer*, robotic*, biop*, interact*, and forc*. Search results were divided in fundamental studies (49), prepuncture alignment techniques (60), and steering techniques (87).

In terms of approach, needle tip placement techniques can be divided in initial aligning and subsequent correction or steering tasks. Prepuncture alignment can be robotically assisted. Insertion after needle alignment is either performed manually [12] or automatically [2] under image guidance. An overview of several setups, typically comprising a robotic arm, has been provided by Cleary *et al.* [13]. The inverse approach, where the target is aligned with a straight needle path, makes use of anatomical knowledge and visual feedback to manipulate the tissue by means of blunt probes. Two examples of such systems are discussed by Reed *et al.* [7]. The scope of this review will, however, not extend to these alignment tasks.

Papers on fundamental studies were read, which contributed to our general understanding of the underlying mechanical interactions. This review, however, primarily extends to those studies involved with steering tasks and descriptions of steerable needle designs.

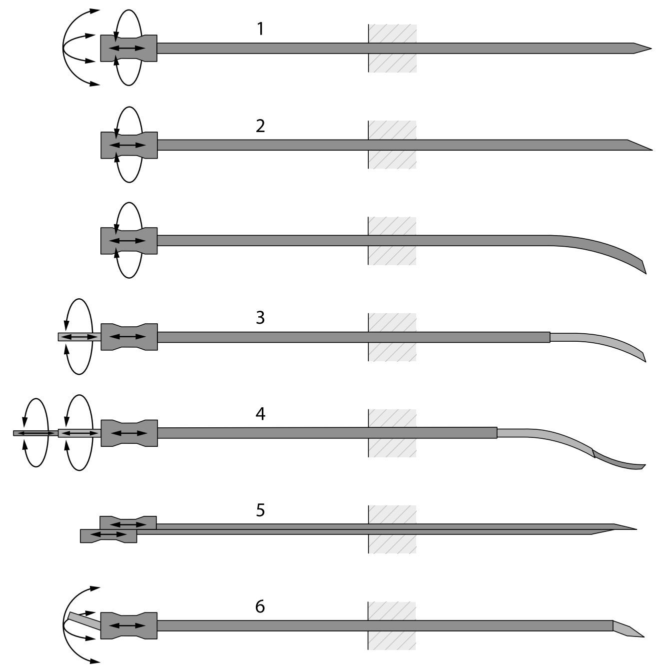

Fig. 2. Illustration of the discussed steerable needles and their degrees of freedom in actuation. The depicted techniques are, respectively, (1) base manipulation, (2) bevel tip (with and without precurve), (3) precurved stylet, (4) Active cannula, (5) programmable bevel, and 6) tendon actuated tip steering. The programmable bevel is here presented with two segments, versions with four segments are available as well.

B. Results

It was found that 68% ($n = 59$) of the needle steering articles and proceedings appeared in robot, control, and automation journals, 23% ($n = 20$) appeared in (biomechanical) engineering journals, and 9% ($n = 8$) appeared in imaging journals. Furthermore, based on the introductory examples, programmed environments, or tissue models adopted, it was found that 59% ($n = 51$) of the articles did not sketch a clinical setting for needle steering, 14% ($n = 12$) focused on the prostate, 10% on the liver ($n = 9$), 8% on the brain ($n = 7$), 6% on the lungs ($n = 5$), and 1% on the breast, heart, and kidney ($n = 1$ each). Of the experiments, 39% ($n = 34$) was validated in a phantom environment, 32% ($n = 28$) in a virtual environment, 16% ($n = 14$) to some extent in *ex vivo* biological tissue, 10% ($n = 9$) in air, 1% ($n = 1$) in water, and 1% ($n = 1$) in *in vivo* biological tissue. Of the biological tissue studies, three studies performed placement accuracy experiments and reported an average error; ~2 mm [14], [15], and ~3 mm [16]. One study reported a single measurement for a path tracking error ~0.5 mm [17].

III. Classification of the Steering Techniques

In vehicle dynamics, active steering is primarily used to describe a speed-dependent adaptation of the overall steering ratio between steering wheel input and front wheel output. For needle steering, input–output relations with respect to control parameters, such as speed, and the environment are not that intuitive (a topic that will be discussed in more detail in Section VI). Therefore, we apply a different definition of the term active steering in this paper, active steering denotes the possibility for the "driver" to adapt the steering output by means of actively modifying the mechanical design and, thereby, the device-environment interactions. For cars, this would resemble the possibility to alter the orientation of the front wheels. This excludes the possibility of lifting the car in its entirety and changing its orientation.

By this definition, needle steering can be divided into passive and active techniques. In both cases, the needle cannula will eventually bend. During passive needle insertion, bending forces are a direct result of the needle–tissue interactions. For active needles, either the tip or the cannula shape can be modified without tissue contact. During insertion in tissue, this modification typically also translates to a cannula deformation. In this paper, six steering techniques are distinguished, as shown in Fig. 2. This figure also presents the degrees of freedom used in needle control. These steering techniques are, respectively, base manipulation, bevel tip steering, precurved stylet steering, active cannula (AC) steering, programmable bevel steering, and tendon actuated tip steering. These techniques will be discussed in this approximate chronological order, although the timeline of the developments may overlap. The first two techniques are considered passive, the latter four active.

A. Needle–Tissue Interaction Models

For a better understanding of needle-steering techniques, a short introduction of the relevant mechanical interactions is in order. Axial forces acting on a needle during insertion are typically divided into puncture forces, cutting forces, and friction. Studies seem to suggest that cutting results in an approximately

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constant force, and that friction increases approximately linearly with insertion depth. Force measurements by experiment have been summarized by van Gerwen *et al.* [18]. Lateral forces that result in bending are less easily measured. A theoretical understanding of these interactions is given by navigation models, such as discussed in [1] and [5]–[8]. This provides a good starting point for discussing the selected steering techniques.

Kinematic models are used to analyze and predict the range of motions of a given mechanism without relating this to the cause of motion. For needle steering, this allowed us to adopt several kinematic descriptions from the field of vehicle dynamics. Examples are the unicycle model [19], the bicycle model [20], the Dubins car model with binary left/right steering [21], and the underwater vehicle model with nonholonomic constraints [22]. These can also be described using differential geometry [23]. Nonholonomic constraints are present in practically all needle steering models and describe the limitations of possible endpoints of the needle tip with respect to the previous path taken and the velocity-dependent constraints of the system. Typically, there is a zero velocity constraint implemented for lateral displacements in tissue and needle movements are described by a constant radius of curvature. To navigate under these conditions, several path optimization algorithms (e.g., Markov decision processes, artificial potential field, and penalty-based methods, or rapidly exploring random trees) have been adopted and tested (for a review see [1]). The practical uncertainties of needle–tissue interactions should be incorporated in the development of motion planners. They may have a substantial impact on both the probability of success and the optimal path computation time [21].

Making these kinematic models accurate for a large range of conditions (model robustness) requires substantial experimental data. Alternatively, descriptive relations between steering in- and outputs can be implemented, for example, between the "steering offset"—a control input—and the curvature of a multisegment programmable bevel [24]. Note that this is already similar to finding symbolic expressions that explain the underlying mechanics.

Mechanical models for needle steering have included both numerical, e.g., FE models and symbolic approaches. Needle bending is often predicted by considering a cantilever beam loaded at the tip [25], and supported along the needle length by virtual springs [26] or a distributed load [27]. These models often presume quasi-static motion [26], and neglect friction along the shaft. A typical mechanical model is shown in Fig. 3. The advantage of this approach over purely kinematic models may be evident when departing from initial conditions, e.g., by considering both a single and a double bend [27].

Steering techniques that consist of multiple interacting parts, may require additional models to deal with internal mechanics. In depth descriptions of the combined curvature of precurved concentric tubes by means of Euler–Bernoulli beam expressions have, for instance, been presented in parallel by two research teams [28], [29]. Combining internal and external mechanical models is, however, a challenging task for which the cannula preshapes may limit the achievable shapes within tissue [30]. Currently, concentric tubes are often either described as catheter-like or minimally invasive tools (simplifying tissue contact) [30], or as systems with stiff outer cannulas (simplifying internal mechanics) [16].

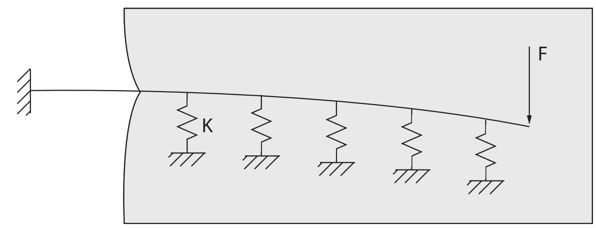

Fig. 3. Simple mechanical representation of a needle, modeled as a cantilever beam inserted in tissue. Tissue support is, here, expressed by a series of springs, and the load F describes the lateral component of the asymmetric tip–tissue interaction.

Finally, to enhance our understanding of puncture and cutting forces, needle–tissue interactions at a microstructural level can be considered. In order to predict crack propagation, the integral of the energy release rate during cutting can be compared to the materials fracture toughness [31]. In this study, material deformations are predicted by a modified Kelvin model. Initial boundary displacements, and subsequent tip insertion, crack growth, and tissue wedging effects can lead to various levels of stored strain energy and crack propagation phenomena, called modes of interaction. These modes may also affect needle–tissue contact conditions and, thereby, the viability of modeling assumptions. The analysis of cutting speeds on stable and unstable crack propagation was, for instance, helpful in understanding needle insertion force [32].

B. Bending

The flexural rigidity of a structure describes the resistance to bending under orthogonal force components. Orthogonal forces during insertion result from asymmetric needle–tissue interactions. Although the needle design can clearly play a part in this [10], asymmetric interactions largely come from tissue factors, such as inhomogeneities and relative movements. Arguments for this have readily been provided in the introduction on placement difficulties of nonsteered needles. The flexural rigidity of a needle can be considered as a measure of the needle's "sensitivity" to medium changes. Dependent on the steering technique, this sensitivity is either a prerequisite for steering or a conflicting mechanism working against it. Table I presents the flexural rigidity E·I, with E be the material's Young's modulus, and I be the needle's second moment of area, of evaluated needles. As this metric requires a linear elastic material, rubber-like programmable bevels are left out. Flexural rigidity is determined based on the needle size, shape, and material, as visualized in Fig. 4. Whenever multiple tubes are present, the principle of superposition is used. For convenience, the following values are used: E_{ss} (stainless steel) = 200 GPa, E_{NiTi} (nitinol) = 50 GPa, and E_{PEEK} (PEEK polymer) = 3.4 GPa. Exact values may differ somewhat, dependent on heat treatments, crystal structures, etc. Clinically used needles are also shown as a reference. They are divided between fine 20–25-G needles and large 14–19-G needles [33].

TABLE I
THEORETICAL FLEXURAL RIGIDITY [N $\cdot$M^2] OF EVALUATED DESIGNS

Steering technique	min.	max.	refs.
Fine needles	6×10^{-4}	7×10^{-3}	–
Large needles	7×10^{-3}	2×10^{-1}	–
1) Base manipulation	3×10^{-3}	7×10^{-3}	[4], [40]
2) Bevel	4×10^{-5}	1×10^{-3}	[41, 42]
Bevel + precurve	1×10^{-6}	7×10^{-4}	[43], [44]
3) Precurved stylet	–	7×10^{-3}	[45]
4) AC (tip)	7×10^{-4}	2×10^{-1}	[16], [29]
(base)	7×10^{-4}	1	[16], [29]
5) Programmable bevel	n/a	n/a	[46]
6) Tendon actuated tip	–	8×10^{-3}	–

As a reference, clinically used fine (20–25 G), and large (14–19 G) needles are shown. Due to protruding tubes, the flexural rigidity of ACs at the base and tip can differ.

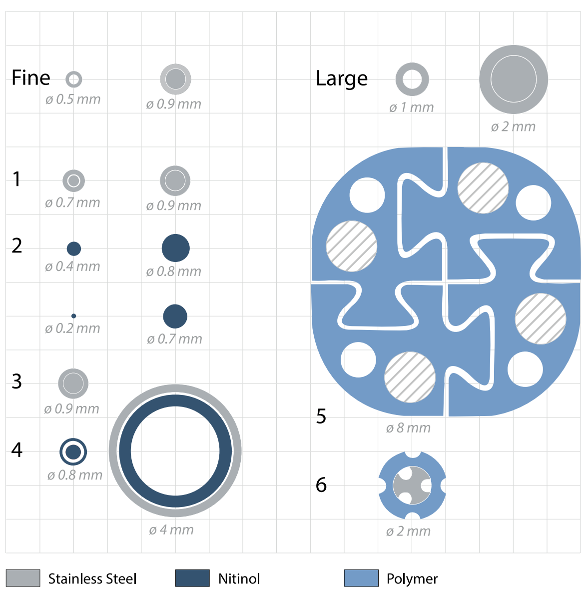

Fig. 4. Cross sections of the needles evaluated in Table I. The second row at 2) presents precurved bevel tips. Hatching denotes the material was not specified.

C. Torsion

Besides flexural rigidity, torsional stiffness is of importance in the mechanical design of steerable needles. In particular, for those that require rotations to steer. Initially, torsion was frequently neglected for the sake of presenting simpler and more intuitive interaction models. Torsional considerations are, however, receiving more and more interest, e.g., [34], [35]. This is underlined by the large share of acknowledged control issues that deal with torsion effects and out-of-plane motions [36]–[39]. Nevertheless, with the given information, torsional stiffness could not be assessed in a similar way to flexural rigidity for all needle designs, where possible torsion effects are discussed individually.

IV. PASSIVE NEEDLE STEERING

First, the classic base manipulation technique will be described. Subsequently, bevel tip needles will be discussed. The vast majority of steerable needle research deals with passive needle steering, and of this group, most work is performed on bevel tip steering. Besides nonholonomic constraints, passive tip steered needles are generally subjected to unilateral constraints as these needles only curve when pushed in tissue [47]. As long as the needles are constrained by tissue and remain in their linear elastic range, they are likely to follow the same path during retraction.

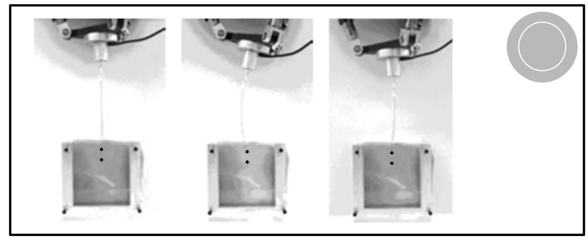

Fig. 5. Obstacle avoidance by changing the needle base orientation during insertion (adapted from [26]). As a reference, the right-top corner shows a cross-sectional view of the needle, similar to that in Fig. 4.

A. Base Manipulation

One of the first models to automate needle steering was based on the manual base manipulation techniques [4]. Subsequently, robotically controlled open-loop base motions were used to regulate the needle tip position and orientation with respect to a target location, as shown in Fig. 5 [26]. A planar finite-element model including potential fields with regions of repulsion and attraction was defined. Subsequent iterative path planning resulted from minimizing the total path potential. Validation studies in phantom material were conducted, but feasible needle paths were not always found due to needle movement constraints.

In order to reduce computational complexity, a planar beam model subjected to friction and supported by virtual springs was developed [26]. Minimizing the total spring energy during path planning would resemble finding a navigation solution with minimal lateral tissue pressure. This was used to optimize performance. In an inverse kinematic approach, this yielded a set of input base movements. A test trial in a closed loop (x-ray), *ex vivo* setup resulted in a tracking error of about 0.5 mm for a 40-mm deep insertion [17]. Peak steering moments at the needle base of around 25 N $\cdot$ mm were reported. Under ultrasound guidance in phantom tissue, a 1-mm error was found for 35- to 40-mm deep insertions [40]. Interestingly, the latter study used an adaptable virtual spring model. Local tissue motions were assessed with speckle tracking ultrasound to subsequently update the tissue stiffness coefficients. Two phantoms with different stiffness values were used and recognized by their displacement relative to the needle tip.

Steering by means of base manipulation is subjected to depth dependence. As the moment arm decreases and the tissue resistance increases with depth [7], accurate tip placement will be hampered [26], and the required steering moments around the base can be expected to increase rapidly. Therefore, base manipulation studies often focus on superficial targets.

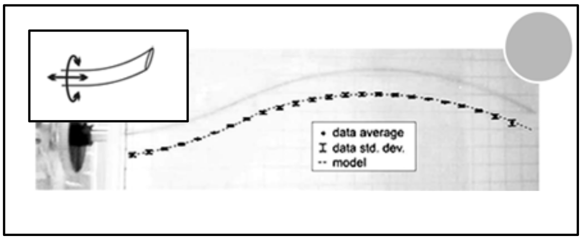

Fig. 6. Forward propagation and rotation of a bevel tip needle. The imbalance in tip-tissue reaction forces will cause the cannula to bend (adapted from [6]).

In terms of the mechanical design, base manipulation studies use unmodified fine (20–22 G) stainless steel needles with various tips, e.g., Franseen [4] or bevel [40]. The effect of tip type or bevel orientation has not been investigated or discussed.

B. Bevel Tip

It was demonstrated in a silicone model that needles with a bevel tip bend more than those with a cone or triangular pyramid tip [48]. Several research groups have therefore focused on steering possibilities of thin and flexible bevel tip needles. As shown by the arrows in Fig. 6, these needles are typically controlled by translating or rotating the needle along its longitudinal axis [42], [49]. It is often assumed that asymmetric interaction forces at the tip will cause a bending effect with constant curvature [20], although this assumption does not necessarily hold for more compliant soft tissues [49].

The insight in steering in controlled environments has reached advanced levels. Both phenomenological and mechanics-based models have been introduced to control steering. Some studies have readily accounted for target movements [50]. However, in biological tissue, bevel tip steering is more difficult. In *ex vivo* tissue, no significant effects in curvature were found by varying bevel angle [51]. *In vivo*, the curvature of bevel tip needles was even described as negligible [43].

Mechanically, interaction forces are related to the insertion speed [18], [42], and the tissue's rupture toughness [52]. The curvature of inserted bevel tip needles is not necessarily affected by insertion speed [42], but can be increased by

1) decreasing the needle diameter [51],
2) decreasing the bevel tip angle [42],
3) increasing the bevel surface [44],
4) introducing a precurve near the tip,
5) introducing a flexure near the tip [9], [53],
6) a combination of the above.

In addition, research was done to control not only steering direction, but also the radius of curvature. This was achieved by investigating duty cycled needle rotations [44] and actively variable bevels [24]. Duty cycling is defined as the ratio of the rotation period T_{rot} to the cycle period T. Here, $T = T_{\mathrm{rot}} + T_{\mathrm{ins}}$, with periods T_{ins} of pure insertion, and periods T_{rot} of rotated insertion [50]. Theoretically, during a 100% duty cycle, straight needle paths would be possible with bevel tip needles. It was found that needle rotations may reduce target displacements, but also cause helical marks in tissue phantoms (with diamond shaped tips), a possible indicator for additional iatrogenic tissue damage [54]. To account for patient movements, intraoperative 2-D (camera) motion replanning with duty cycling was assessed with movable obstacles [50]. With respect to the clinical use of rotating bevel tip needles, it should be noted that some procedures require a certain bevel orientation during needle insertion [55]. Furthermore, it was suggested that bevel direction could make the difference between parting or cutting of fibrous structures [56]. To the author's knowledge, conflicts in technical and clinical demands for bevel steering have not been investigated, so far.

The kinematic constraints and control difficulties of bevel tip needles meet each other in a tradeoff. The path of thin needles is easily affected by tissue irregularities [7]. This path was found more sensitive to tissue density fluctuations than that caused by a triangular or diamond tip [10]. These kinds of effects should be considered in the placement plan [57]. In addition, thin needles are easily susceptible to buckling near the entry point [42] or even within tissue [7]. A validation study with a 0.86-mm nitinol needle (which is thick for bevel tip steering) in *ex vivo* liver tissue acknowledged that several trials had to be terminated because the needle buckled before penetrating an encountered structure [11]. Another recent study, using thinner needles, reported similar findings [15].

During bevel steering, torsional damping and friction can become substantial [35], [37], [58], [59]. Needle rotations of 180° (often used in 2-D control models to steer left or right) can lead to phase differences between the tip and base of up to 45° after a 10-cm insertion [36]. This potentially leads to large out of plane steering errors. Camera tracking of the needle in phantom tissue has been used as a dead-beat observer to estimate the actual bevel orientation [47]. However, the quality of this estimate relies on the amount of sample points, i.e., the insertion depth. In addition, torsional dynamics can be implemented in control models [35]. Physical experiments in plastisol with time-varying torsion dynamics in a closed-loop control scheme showed improvement over purely kinematic control [34], [60]. Alternatively, it was found in phantom tissues that rotating or "wiggling" the needle may release the built-up strain energy [58], a finding that may be coherent with the observation that post placement needle revolutions diminish target displacements [54].

Prebends and precurves have been introduced to enhance the steering behavior of bevel tip needles [44], [51], [61], [62]. As there's a thin, undefined line between the two, both having a radius of curvature, only one is "small" and the other "large," we will adopt the term precurved needles. In terms of control, precurved needles show a velocity dependence in contrast to normal bevel needles [62]. The investigation of duty cycling of precurved needles showed that needle tips may not immediately follow the base movement due to tissue resistance. In addition, care should be taken to prevent sudden snapping motions to the unstrained needle state, which can lead to tip position discontinuities of up to 4 mm [63]. Besides snapping, corkscrew insertion motions should be carefully assessed. Replacing the precurve with a flexure near the tip potentially reduces tissue damage during these procedures [9].

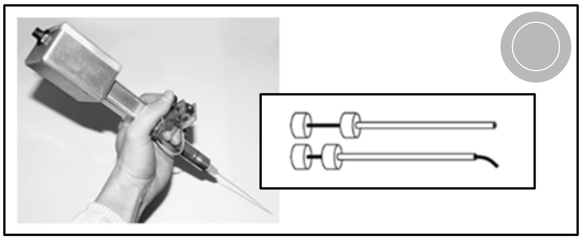

Fig. 7. As the precurved stylet and outer cannula are translated with respect to each other, the length of the exposed precurved tip is altered (adapted from [45]).

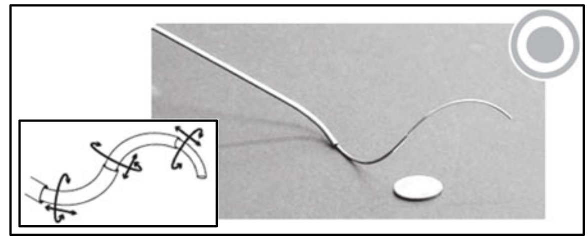

Fig. 8. Example of an AC consisting of three tubes (adapted from [38]).

Most often, bevel steering experiments are performed with thin, solid, and superelastic nitinol stylets, ranging in diameter between 0.36 and 0.83 mm [41], [42]. This would resemble 21–28-G needles. Precurved needles may be thinner (down to 0.15 mm, $<$34 G) [44]. Eventually these stylets could be replaced by cannulas, such as the 0.86-mm nitinol tube [11], to be of use in a clinical procedure. This would, however, decrease the needle flexibility and steering range. Alternatively, these stylets could be used as guidewire, although this may affect the tip placement and should be investigated.

V. Active Needle Steering

During active steering methods, either the tip or cannula shape is actively adapted. Adapted tip shapes affect the needle–tissue interactions, which will indirectly result in cannula shape alterations. In this section, the main concepts of active needle steering will be discussed. Mechanically, these needles may be more complex than the earlier discussed passive needles. In terms of steering constraints, they are often expected to be more indulgent to variations in the navigation medium. Most techniques make use of a combination of protruding stylets or cannulas, some of them being precurved. In case of multiple interacting tubes, the internal ratio is of importance in flexural rigidity values. This ratio is either in balance (close to 1) or dominated. In the latter case, the stiffer (typically outer) cannula determines the combined shape.

As bending of actively steered needles is at least partly the result of internal mechanisms, the need for bilateral path planning should be assessed. Similar to stiffness interactions among tubes, the interaction with the environment can be described as either balanced or dominated. Dominating systems rely on internal actuation mechanisms for steering. Balanced systems rely on tissue interactions and may to some extent be guided by tissue on their way out.

A. *Precurved Stylet*

Fig. 7 illustrates a protruding precurved stylet used as a guidewire to steer the cannula [45]. On following the stylet path with the cannula, tissue reaction forces will cause the cannula to bend in the same direction. The degree of steering can be controlled by varying the exposed length of the stylet. A path planning algorithm was developed to determine the required stylet exposure length and cannula insertion depth for use in a 2-D ultrasound feedback system. Once in place, the actuation mechanism and stylet can be retracted, leaving only the cannula in place. From a control perspective, a reorientation of the steering direction can be achieved in its retracted state, minimizing tissue interference and torsion. An initial prototype of this method was fabricated and tested in *ex vivo* porcine tissue.

For this technique, a modified 20-G Chiba needle was used. The stylet was curved opposite to the bevel direction over a tip length of approximately 20 mm. In order to compensate for the bevel and travel along a straight line, the stylet was kept slightly exposed in its base configuration. Mechanically, a stiffness balance was sought. When completely retracted, the stylet curve should be straightened out by the cannula. Tissue interaction forces should, however, still be able to affect the cannula shape.

B. *Active Cannula*

An extensively studied cannula deformation method, shown in Fig. 8, uses multiple precurved concentric tubes, ACs. Initial studies on balanced tube pairs were presented in parallel by two independent research teams [28], [29]. By means of rotating and extending the tubes with respect to one another, cannula curvature, tip position, and tip orientation can be adapted. Since the actuation forces are generated internally, these instruments do not need environmental contact to steer. In terms of trajectory control, balanced tube pairs are, in fact, not very tolerant to variable environments [11]. Therefore, these balanced ACs are often described as compliant, catheter-like mechanisms, useful for applications in open space or fluid-filled cavities [29].

Piecewise circular approximations have been used to model the overall shape of balanced AC systems. These assumptions hold while the overlap of curved sections is sufficiently short and the radius of curvature sufficiently small. The introduction of torsion in balanced AC models has shown that, otherwise, noncircular overall shapes are also possible [30]. Internal interactions, but also the contact between an extended tip and tissue may disturb this torsion balance. When disregarded, torsional wind up can lead to both excessive placement errors and sudden snapping motions between stable cannula configurations [38], [39]. This risk is particularly large when balanced tube pairs are rotated to approximately 180°, i.e., when trying to straighten out a balanced cannula pair. A bifurcation in the energy landscape will appear, and on sufficient rotation, one of the cannulas will snap from a local minimum in potential energy to the global minimum. A proper design of the stiffness pairs or otherwise a

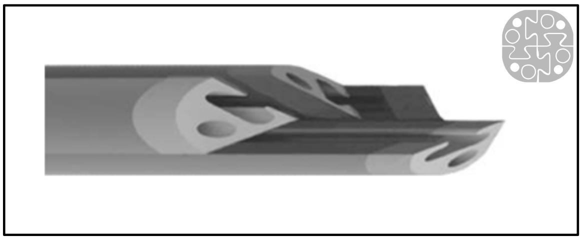

Fig. 9. Programmable bevel consisting of four interlocked segments (adapted from [22]).

Fig. 10. Example of a tendon actuated active tip-steered needle with a ball joint and a conical tip.

limit to the allowed relative tube rotations would be required to prevent this [38].

To the author's knowledge, the AC steering method is the only steering technique that has been combined with a therapeutic modality (ultrasonic ablation) in an *ex vivo* tissue study [64]. This prototype consisted of two tubes: one precurved inner tube and one straight dominating outer tube. Note that this nonbalanced tube configuration is remarkably similar to the just discussed precurved stylets. In the past, similar needles have been presented and marketed [65]–[67]. To date, this is, however, considered the only viable AC design for which the shaft can follow the tip during advancement in tissue [16]. Placement accuracy measurements yielded tip errors on randomly picked targets in *ex vivo* liver tissue of 3.3 ± 2.7 mm (mean $\pm$ std), under 2-D ultrasound tracking [16]. Information on the needle path or target depth was not presented.

As the description of ACs is very generic, the used needle designs vary considerably. Some fundamental experiments were performed with nitinol precurved tubes and straight stylets with outer diameters down to 0.8 mm [38]. In contrast, 4.2-mm needles were used in the *ex vivo* liver study [16], meant to carry a small elastic interventional tool within it.

C. Programmable Bevel Tip

A quick developing steerable needle design, based on biomimetic concepts, makes the use of a programmable bevel tip [24]. The latest version of this device consists of four interlocked segments that can slide along one another, as shown in Fig. 9. Each segment is actuated by a linear motor at the base and allows the respective part of the tip configuration to change. Therefore, we will consider this needle an active mechanism. In comparison to passive bevels, cannula bending largely results from asymmetric tip–tissue interactions.

Tip control and steering occurs on behalf of the relative offsets of the interlocked segments. These offsets have been related to a particular radius of curvature in a particular test environment (6 wt% gelatin). Trocars are used at the gelatin surface to prevent buckling. One of the lumens contains an electromagnetic position sensor, which leads to the tip, where it is used for closed-loop control.

The virtues of rapid prototyping have contributed to a quick evolution of this steering technique, for which (more or less) every successive article presents a somewhat altered needle design. Considerable improvements include a decrease in needle diameter from 12 to 4 mm in a prototype meant for 2-D navigation tasks (using a two-segment needle) [68], and down to 8 mm in a prototype that allows 3-D steering (using a four-segment needle) [46]. A 6-mm four-segment design was used to assess target migration during "reciprocal motions" (actuating segments in turn), while navigating on a straight path [69]. This study produced insightful qualitative assessments of the relations between actuation approach and insertion rate on phantom strain, displacement, and relaxation phenomena.

Steering results in phantom material were obtained by evaluating the eight primary bending directions (every 45°). This was achieved by either single-segment actuation or by actuating two-adjacent segments simultaneously. The found variation in steering response for different directions was attributed to different boundary conditions acting on the gelatin phantom [46]. As we are convinced by the significance of boundary conditions on needle steering, we advise the randomization of insertion location during motion assessment studies. Overall, the found path curvatures ranged up to $\sim$0.017 mm^{-1} for the single-segment actuation directions, and up to $\sim$0.010 mm^{-1} for the double-segment actuation directions. This difference was attributed to the difference in flexural rigidity of the two configurations [46], illustrating the importance of this factor for both navigation constraints and modeling.

D. Tendon Actuated Active Tip

Tendon-actuated tip-steering techniques have been integrated in a variety of medical instrument designs, such as laparoscopic tools [70] or endovascular guidewires [71]. In needle steering, this technique has received little attention. During active tip steering, the needle tip shape can be actively modified. Tip actuation occurs outside the patient and the mechanical propagation of actuation signals can, for instance, be achieved by cable or wire. The needle tip is placed on the top of a flexible cannula. In between, any type of compliant structure or joint mechanism can be used. Cannula bending results from asymmetric tip–tissue interaction forces, comparable to the operation of passive bevel tips.

Fig. 10 presents an actively steered needle consisting of a conical steel tip on top of a PEEK cannula, as is currently being developed at our department. In between, a ball joint mechanism is used. The needle stylet has grooves for glass fibers with optical strain sensors (fiber Bragg gratings or FBGs) and the cannula has grooves for steering cables. The cables are connected to the tip and controlled by four-rotary servo motors. In size, the used needle was comparable to 14–15-G needles. Due to material choices, the flexural rigidity (8.2×10^{-3} N $\cdot$ m^2), on the other hand, was comparable to that of a fine 20-G needle.

Closed-loop steering experiments with this needle were performed in phantom tissue. For this, a PI controller was implemented to control the tip location on behalf of an FBG-based needle shape reconstruction. This reconstruction process is presented in [72]. Nine different target locations were defined, and an average targeting accuracy of 6.2 ± 1.4 mm (mean $\pm$ std) was achieved. The majority of this error was attributed to systematic steering differences in the mechanical design. A steering precision of 2.6 ± 1.1 mm, around the average position reached, was found. It was implicated that a comparable targeting accuracy can be obtained with the aid of both mechanical design optimization and the implementation of a fitting mechanical navigation model to replace the current PI controller. More investigation on this behalf is in order.

E. Other Active Steering Methods

A recent addition to the class of automated cannula deformation techniques describes a robotic controller for a hand held tendon driven steerable biopsy needle [3]. A little information was available on this particular needle design. The hand held device has a joystick handle that pulls on actuation cables. As a result, the needle cannula deforms with a constant curvature (in contrast to the tip deformation of the tendon actuated tip). For this study, this joystick was robotically actuated by a couple of DC motors. The planar tip position in air was tracked with a camera. The average number of iterative control steps (8) required to reach the target with sufficient accuracy ($<$1 mm) was reported. This should relate to the number of image captures required for navigation in the lung environment.

Another recent development in active tip steering uses a shape memory alloy wire to generate mechanical strain and cause deformations near the tip [73]. The wire is heated by means of laser light delivered through optical fibers. A unidirectional active steering mechanism was accomplished within a 1.37-mm-diameter nitinol tube. These needles were aimed to reach tip angles up to 10° and lateral deflections in the order of 20 mm. Future development efforts will initially focus on the reduction of power losses and waiting times inherent to the used heating process, and will later focus on phantom studies.

Simulation studies on compliant hinges near symmetric magnetized needle tips were performed to study the feasibility of magnetic tip actuation [74], [75]. Advantages of such a system would be the initial straight tip orientation, and the ability to steer in any direction by changing the location of the surrounding magnets. As the actuators can be kept at a distance, this steering method only requires relatively simple instruments to be introduced into the body. On the other hand, the interactions between a compliant mechanism and soft tissue will be crucial to investigate. Aside from modeling buckling effects and stressing the need for proper hinge dimensioning [74], no test setups or prototypes have, to the author's knowledge, been presented.

Theoretically, cannula deformation can also be achieved by piezoelectric actuators, straining the needle surface on one end and compressing it on the other [76]. Some FEM simulations were performed to show the relations between actuator design and potential tip deflection. It was concluded that a set of thin (0.05 mm) and long (200 mm) actuators would be needed to produce planar tip deflections in the order of 0.15 mm, using a potential difference of 30 V. Suggestions to improve the tip deflection included the application of longer slimmer actuators and the increase of the potential difference. The authors, however, acknowledge that this may go at the cost of actuator robustness and procedural safety.

VI. Discussion

Manual control of flexible needles is said to be difficult and unintuitive [77]. This complicates accurate tip placement and stimulates the development of needle-steering techniques. Needle steering can be divided into passive and active solutions. During passive steering, cannula deformations result from interactions with surrounding tissue. During active steering, either the tip or cannula can, to some extent, be deformed without making tissue contact. Needle steering is constrained by both the navigation medium and the needle itself, including both mechanical design and operational factors, like flexural rigidity and insertion speed. In practice, the navigation medium is highly variable or unknown. In this regard, it has been regularly acknowledged that the insertion success rate can be low with current steerable needle designs [11], [15]. Perhaps the stiffness of clinically applied needles could serve as a guide for future designs. However, it was noted that, of the analyzed needle steering studies, over 50% refrained from touching this subject and did not specify a clinical task or organ to steer to.

The effect of insertion rate during needle steering depends on the navigation medium. Findings in artificial phantom material seem to agree that axial force (regularly related to target movement [78]), but moreover friction, increases with increasing velocity [18]. Findings in biological tissue, on the other hand, suggest that puncture force remains either constant or decreases with increasing velocity [18], [79]. Note that, for steering tasks, insertion rates should not just depend on task optimization in terms of forces or target movements, but also on the control approach, the limitations of human motor control (when applicable), the reliability of available feedback, and the clinical situation at hand, e.g., the desired curvature and the nearby presence of delicate structures. Axial rotations can potentially reduce friction and total axial force [18], even when the rotations are performed after tip placement [54]. However, the risk on tissue damage should be carefully addressed.

Although this review assesses steerable needle designs, it should be noted that some of the discussed steering concepts were not developed with the purpose of validating a particular design. Instead, they were used to validate a mathematical steering model. There is a fundamental difference between the two, as one optimizes to a single robust working solution, whereas the other optimizes to a generic underlying control method. Any confinement to such a control method on behalf of applicability

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TABLE II
FREQUENTLY USED TEST CONDITIONS AND OBJECTIVES IN NEEDLE-STEERING STUDIES

Steering technique	Stylet or cannula	Tip type	Material	Clinical use	Examples of research objectives
1) Base manipulation	Both	Various	SS[2]	Shallow tissue	Min. needle curvature/tissue load [26]
2) Bevel	Stylet	Bevel	NiTi[3]	Deep tissue/organs	Identify parameter-curvature relations [42]
+ precurve	Stylet	Precurved bevel	NiTi	Deep tissue/organs	Max. needle curvature/steering range [44]
3) Precurved stylet	Both	Precurved bevel	SS	Deep tissue/organs	Min. physical dimensions [45]
4) AC (balanced)	Cannula	–[1]	NiTi	Vessels/body cavities	Max. forward kinematic plan accuracy [30]
(dominated)	Both	–	SS/NiTi	Deep tissue/organs	Max. forward kinematic plan accuracy [16]
5) Programmable bevel	Cannula	Adapted bevel	Rubberlike[4]	Deep tissue/organs	Min. closed-loop trajectory error [80]
6) Tendon actuated tip	Both	Conical	PEEK/SS	Deep tissue/organs	Min. closed-loop tip placement error

[1]The tip would be part of the internal stylet or tool.
[2]SS = Stainless steel.
[3]NiTi = Nitinol.
[4]Although various rapid prototyping materials have been used, the cannula is typically described as rubberlike.

is undesired. Although different research goals have led to different test setups, eventually a combination of these goals—a needle and a control model—is desired. As many well-written reviews on control aspects already exist [1], [5]–[8], the critical analysis of needle prototypes and their practical limitations is justified. However, this does not mean that the developed steering models should receive the same criticism or that they are equally limited in practice.

In this review, a classification of steering techniques is proposed on behalf of the mechanical functionality of developed needles. Design criteria for these needles can vary and do not necessarily reflect clinical needs. However, the effectiveness of needle steering depends heavily on the navigation medium and, thus, on the intended use. This dependence is quantified by the needle's flexural rigidity. Steering with extremely flexible needles is often validated in well-known phantom materials. Therefore, scaling of needle–tissue interaction mechanics and their uncertainties may be valuable to investigate, not only to ensure realistic movement constraints, but also to assess safety factors, like steering robustness, cannula buckling, tissue slicing, and needle snapping effects.

As a result of differences in experimental designs, test conditions, and performance metrics, the direct comparison of needle-steering techniques is difficult. Steering potential in air or tissue will differ largely. Steering potential also differs among organs [43]. In terms of task performance, some studies aim to reach superficial targets with a minimized needle curvature, whereas other studies focus on steering extremes for deep targets. Radii of curvature and placement accuracies for several combinations of steering techniques, control approaches, and test environments are summarized in [80]. Such experimental results may in absolute sense be indicative, but should always be regarded in their full context. Nevertheless, the collected information on instrument and/or experimental design gives insight in suitable use. Table II presents some of these factors. The discussion that should be provoked by addressing these differences is why we actually need steering. It is often suggested and assumed that needle steering provides new minimally invasive approaches and allows targets to be reached that were formerly inaccessible. Perhaps, the largest benefit of needle steering actually lies in performing current needle interventions more effectively and with less complications.

In terms of steering mechanics, design choices, and previous research focus, some statements can be made regarding the clinical use of steering techniques. Base manipulation may, due to its decreased ability to steer at greater depths, be used for shallow targets, e.g., during breast biopsy or local anesthesia. However, for these applications, alternative methods like tissue fixation, target or needle alignment, and preloading [81] should also be considered. Other passive steering methods, such as bevel and precurved tips need some tissue depth for steering. In terms of suitable environments, these techniques deal with an inherent design conflict. Maximal steering is achieved with high-tip–tissue interaction forces within highly dense tissue. Fluctuations in density are, however, undesired in terms of steering robustness, as the used needles can be extremely flexible. The feasibility of use of precurved tip needles for brain interventions is currently being investigated [14]. Precurved stylets and simple AC designs may be used in deeper interventions, e.g., liver biopsy or brachytherapy. ACs with balanced tube pairs require limited tissue contact and should probably look at catheter interventions or minimally invasive therapies. For programmable bevels, this same division is thinkable. Scaling down of current designs is an ongoing challenge. Possibly, thinner versions will require harder materials, which may decrease friction issues. Increase of friction, or adding fixations, on the other hand, may be investigated for open air operation. This may lead to multichannel designs suitable for the minimally invasive setting, where larger instruments diameters are accepted. Tendon actuated tips are relatively new and aim for deeper therapies such as liver biopsy, but this requires more investigation.

The importance of a human operator to guide, interpret, and safeguard the intervention should not be underestimated. Reed *et al*. stated that the control of insertion speed and the selection of a needle path should be human operated for a needle robot to be clinically viable [7]. It was said that the physician relies on visual feedback from imaging techniques and kinaesthetic feedback from the instrument in combination with a mental image of anatomical structures [26]. Other sensory information, such as tactile feedback from intraprocedural force measurements [1] or needle shape reconstruction by FBGs [72] may also aid in task performance. These additional feedback methods are interesting research topics to investigate, both to increase our understanding of needle–tissue interactions and to enhance the operator's awareness of ongoing events.

To conclude, needle steering can be achieved by either passive or active methods to adjust the needle shape. The magnitude of these adjustments relies on the needle design and the tissue environment, i.e., the mechanical interactions. In practice, steering variability results from uncertainties in these interactions. Although needle–tissue interaction mechanics have been studied in fundamental work, it has been found that validation studies to assess needle designs or navigation methods rarely use mechanical interaction parameters as dependent variable in the experimental design. Typically, validation experiments assess the accuracy of navigation models under known (constant) mechanical conditions, a fixed needle and a fixed phantom. Since, the selection of accurate tissue parameters is important for realistic needle insertion modeling [20], an important step of bringing the current models to practice would include either evaluating this selection sensitivity or making these parameters more adaptable. A noteworthy contribution of the latter is found in a recent assessment of needle placement robustness in a double-layered phantom [50]. Steering constraints in both layers were determined *a priori* and used to adjust the navigation plan. In the end, a robust navigation model may require constantly varying model parameters, the selection of which relies either on *a prior* environmental knowledge, or on actual on-site measurements. Tissue stiffness coefficients may, for instance, be amended on behalf of ultrasound footage [40]. Fundamental studies in needle-tissue mechanics with actually steered needles may help refine estimates of interaction parameters, parametric relations, and their constraints. This type of work may be beneficial both for mechanical design optimization, as well as the realization of generic navigation models. This review stresses the importance of considering mechanical parameters of the tissue and the needle, e.g., flexural rigidity and torsion stiffness, irrespective of the research objective. The variance of these metrics for currently developed needles is illustrated in this paper. Currently, this variation is largely attributed to technical and/or modeling limitations. Eventually, mechanical design should be based on clinical needs.

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