A generic three-dimensional static force distribution basis for a medical needle inserted into soft tissue.
Adeline L G Robert, Grégory Chagnon, Ivan Bricault, Philippe Cinquin, Alexandre Moreau-Gaudry

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Abstract: In this paper, a generic loading basis that acts on needles inserted into human tissue is presented. To achieve this purpose, a needle was inserted 62 times into a pig shoulder and a CT scan (Computerized Tomography) was acquired after each insertion to determine the needle’s final trajectory. By using static Beam and B-spline theories, a generic loading basis was then determined from these trajectories. It was first validated on theoretical cases and then on 20 needles inserted into in-vivo human tissue. Such a basis can be worthwhile to highlight what are the loading forces effectively applied on a needle inserted in complex tissue. The knowledge gleaned from this study may be useful to model predictive deformation of the medium that is encountered in a real clinical situation and also to control in real time the motion of the needle during a percutaneous intervention.
Dear Pr. David Taylor,

Please find enclosed the manuscript entitled - A Generic Three - Dimensional Loading Basis Applied to a Surgical Needle Inserted into a Complex Tissue - by Adeline Robert\textsuperscript{1}, Grégory Chagnon, Ivan Bricault, Philippe Cinquin, Alexandre Moreau-Gaudry\textsuperscript{1} to be submitted as a research paper to the Journal of the Mechanical Behavior of Biomedical Materials. All co-authors have seen and agree with contents of the manuscript. We certify that the submission is original work and is not under review at any other publication.

In this manuscript, we report the results of the first study on a generic loading basis that acts on needles inserted into human tissue.

We believe that our findings could be of interest to the readers of the Journal of the Mechanical Behavior of Biomedical Materials because they bring new and strong approaches for the understanding of forces applied on inserted needles. Contrary to the all approaches, we do not consider the forces in a dynamic cases but as a succession of static stages as commonly encountered in percutaneous needle intervention. We free ourselves thus from the measurement of the resultant axial force. Moreover, our work was validated on in vivo human tissue during real clinical procedures.

We hope that the editorial board and the reviewers will agree on the interest of this study.

Sincerely yours,

Adeline Robert on behalf of the authors

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Medical intervention

Model assumption

Image processing and segmentation

Input CT-images

Beam theory
B-spline theory

B-spline theory

Loading basis
9 modes

Forces (N)

Needle length (mm)

Medical intervention

Graphical Abstract (for review)
Highlights:

as far as the authors are aware, this paper is the first to propose a static load distribution approach applied to a needle during a percutaneous procedure. The key contributions are:

- A new approach to achieve load distribution applied on a needle
- The association of two theories (B-spline, beam) to model the forces
- Extensive validation to theoretical cases
- Extensive validation to in vivo tissue (human) in daily clinical situations
A Generic Three-Dimensional Loading Basis Applied to a Surgical Needle Inserted into a Complex Tissue

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Abstract

In this paper, a generic loading basis that acts on needles inserted into human tissue is presented. To achieve this purpose, a needle was inserted 62 times into a pig shoulder and a CT scan (Computerized Tomography) was acquired after each insertion to determine the needle’s final trajectory. By using static Beam and B-spline theories, a generic loading basis was then determined from these trajectories. It was first validated on theoretical cases and then on 20 needles inserted into in-vivo human tissue. Such a basis can be worthwhile to highlight what are the loading forces effectively applied on a needle inserted in complex tissue. The knowledge gleaned from this study may be useful to model predictive deformation of the medium that is encountered in a real clinical situation and also to control in real time the motion of the needle during a percutaneous intervention.

Keywords: Needle deformation, Load distribution, Ex vivo measurements, In vivo forces, B-spline theory, Beam theory

\textsuperscript{∗}The clinical trial, performed in the Grenoble University Hospital, was authorized by the AFSSaPS, the relevant French regulatory authority for biomedical research, and by the Comité de Protection des Personnes Sud-Est V, an institutional French review board (ClinicalTrials.gov identifier: NCT00828893).

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1. Introduction

Many medical procedures are based on the insertion of needles to perform a diagnostic (biopsy) or therapeutic (drug injections) act. A medical needle consists of a slender stainless steel tube (cannula) with a plastic hub in the proximal end and a sharp (bevel) tip on the distal end. In all cases, the aim of such procedures is to put the needle tip on a target with accuracy, which is also the purpose of numerous work (Abolhassani and Patel (2006), Alterovitz et al. (2008), Teoh and Chui (2008)). Furthermore, the needle trajectory must respect the surrounding anatomical structures identified beforehand in order to prevent undesired lesions and to maximize the benefit/risk ratio of the act. Different strategies have been developed and are currently being developed to improve the positioning of needles in soft tissue. They may involve the use of robotic (Hungr et al. (2011)), optical (Park et al. (2010), Desjardins et al. (2011)) or electromagnetic (Hushek et al. (2004), Lei et al. (2011)) devices. Regardless of the strategy used, taking into account the applied loading could further optimize their accuracy.

A significant amount of work has been achieved dealing with the modeling of the forces involved in percutaneous needle insertions. As reported in the Van Gerwen et al. (2012) paper, numerous survey of literature exist covering topics from needle insertion force modeling to the planning and control of flexible needles using image guidance (Abolhassani et al. (2007), Misra et al. (2008), Cowan et al. (2010)). However, none focus on experimental data and quantitative force information.

A comprehensive overview of the measurement of all the forces encountered is difficult to give due to the large number of variables involved in needle tissue interactions and the large variety of experimental approaches. Nevertheless, Van Gerwen et al. (2012) provide a survey of experimental insertion-force data available in the literature to highlight the influence of various factors on needle-insertion-force. They summarize data obtained during insertion of a needle or similar tool into any kind of material, either artificial or biological, living or dead. They conclude that for both practical and ethical reasons, measuring forces in vivo is more difficult to come by, but that these measurements are essential for understanding the true needle-tissue interac-
tions in humans provided that the ex vivo results can be extended to in vivo studies.

All the studies identify the various forces applied to the needle during its insertion in a tissue, i.e. in a dynamic case (Kataoka et al. (2002), Simone and Okamura (2002), Podder et al. (2006)). These forces are usually determined from the measurement of the resultant axial force by a load cell put at the outside extremity of the needle in the direction of the motion. All studies show that the axial force is the summation of different forces distributed along the needle shaft as described in Van Gerwen et al. (2012) and as represented in figure 1. The stiffness (or puncture) force (axial) occurs when the tissue boundary is breached by the needle contact, the cutting force (axial) appears to cut the tissue during the insertion and the frictional force (orthogonal and axial) is due to the increasing contact area between the needle and the tissue. The orthogonal component of the last one is the only force which occurs in a static case.

Van Gerwen et al. (2012) report also numerous works (DiMaio and Salcudean (2002), Crouch et al. (2005), Dehghan et al. (2008)) using indirect methods to reconstruct the axial load distribution along the needle, on artificial materials (polyvinyl chloride, silicone gel), based on external force and tissue displacement measurements. They assume that indirect methods are practical due to the facilities in modeling artificial tissue and in measuring markers displacement with camera or ultrasound. Their indirect methods are based on finite element (FE) models and on experimental axial force as well as tissue-displacement measurements.

Contrary to the previous approaches, Park et al. (2010) propose a model in which the forces applied on the needle are distributed along it. This is made through a series of radial impulses, at any orientation in the \((x, y)\) plane and located at intervals of \(L/10\) anywhere along the needle, where \(L\) is the length of the needle. They proposed also a concentrated axial and radial force applied on the needle tip. Although no explanation is given for this distribution.

Very few studies give in vivo measurements of forces in surgical procedures and even less of loading along the needle shaft. All studies consider the forces applied in dynamic cases and not as a succession of static stages.
Nevertheless, such a process for needle insertion is commonly encountered in clinical practice and more precisely in percutaneous procedures, in which the needle is gradually pushed between image acquisitions. Moreover, to our knowledge, whatever the measurement or the model used, nobody proposes a generic loading basis applied along the needle that has been validated on in vivo human tissue. This loading basis is a set of linearly independent load informations that, when associated, can represent every load distribution.

In this paper, we provide a method for determining the loading basis along the needle shaft by using Beam and B-spline theories. Our aim is to focus on a better understanding of the static forces that are applied on a needle introduced in complex tissue and that may explain the deformation of the needle (the axial force components are not considered). Our work can be divided into three parts: in the first part, the forces applied to needles successively inserted into an ex-vivo pig model were determined. In the second part, the results were validated on theoretical cases based on mechanical theory. In the third part, an in vivo validation was performed on a series of real human interventions. Finally, some concluding remarks close the paper.

2. Methods and Materials

First of all, the various methodological points used in this study are presented: we first summarize some elements of mechanical beam theory that are useful for understanding this work. We complete this with a description on data registration. Finally, convenient notions on B-splines are exposed for the modeling of the needle deformations.

Second, the tools and materials used during our experiments are described: some for the construction of the loading basis and others for the reconstruction of the needle deflections.
2.1. Method

2.1.1. Application of beam theory to needles

Needles can be considered as monodimensional structures (beam) as the length of the neutral axis is large compared to the dimensions of the cross-sections. The following two assumptions are also true for needles:

- the radius of curvature of the neutral axis is large compared to the cross-sections dimensions,
- the possible variations of the surface of the cross-section are weak and progressive.

Beam theory is therefore an appropriate tool for studying needles and for obtaining the relationship between their deflection and the applied load. A
needle deflection from a medical act is shown in figure 2.

The beam being brought to deform, it will be supposed that 1) the deformations undergone by the beam, as well as the displacements which can be measured, remain small (small deformation assumption) so that the solid remains in the elastic domain, and 2) the points of external load application remain constant throughout. In fact, the tip deflection is relatively small for all the experiments (less than 10% of the needle length) such that small-strain linear beam theory applies.

Based on these hypotheses, the bending moment-curvature beam equation can therefore be written as:

$$\gamma \approx \frac{d^2y(x)}{dx^2} = \frac{M_f(x)}{EI} \quad (1)$$

where $\gamma$ is the curvature, $y$ is the displaced position, $x$ is the coordinate along the beam, $M_f$ is the bending moment, $E$ is the Young’s modulus of
the material and $I$ is the second moment of area. The shear force $V$ and the transverse load $q$ can be deduced from the bending moment:

$$\frac{dM_f(x)}{dx} = -V(x) \quad (2)$$

$$\frac{d^2M_f(x)}{dx^2} = q(x) \quad (3)$$

2.1.2. Registration of data

The needle deflection analyzed in this study was obtained using a specially developed software, which makes possible to segment and record points in space along a needle from a set of CT (Computerized Tomography) images as shown in figure 3.

Figure 3: Needle segmentation from CT images. Visualization of the 2 orthogonal planes $P1$ and $P2$ defined to perform a relevant study on the whole of the needles.

To be able to carry out a coherent study on these deflections, the same reference mark was defined for all the needle deflections. Its origin corresponds to the outside extremity of the needle (plastic hub). Its primary axis
is represented by the least-squared line of the segmented points to the origin. Then by specifying the plane with least squares method, the total reference mark is built. Thus two orthogonal planes were fixed as shown in figure 3. The first, called $P_1$, represents the plane of greater deflection (least-squared plane). The second, $P_2$, represents the secondary curves. The representation of the needle projection in the two planes are presented in figure 4. Furthermore, all the needles used to build the loading basis have the same length and diameter, which makes it possible to standardize the basis and thus apply it to any needle.

![Segmented needles and B-splines](image)

(a)  
(b)  

Figure 4: Registration of the needles deflections when inserted into a pig shoulder. Visualization of the deflection in $P_1$ (a) and $P_2$ (b)

2.1.3. Mathematical model

Sixty-two 3D needle deflections were used to build the 3D generic loading basis. Each 3D deflection was approximated as a uniform cubic B-spline $S_d$ ($d = 1 \ldots 62$), i.e. as a linear combination of basis cubic B-splines. A regular grid of $n + 1$ real values $t_i$ ($i = 0 \ldots n$, with $n = 6$ in this study), called knots with $0 = t_0 < \ldots < t_n = L$, was defined and the expression of $S_d$ was:

$$\forall t \in [t_0 \ t_n], S_d(t) = \sum_{i=-3}^{n-1} P_{d,i} N_{i,k}(t)$$  \hspace{1cm} (4)
where \( k = 3 \) is the degree of the B-spline, \( P_{d,i} \) are the control points for the \( d^{th} \) 3D needle deflection and \( N_{i,k} \) are the B-splines base each determined starting from the preceding ones by recurrence:

\[
\forall t \in [t_0, t_n], N_{i,0}(t) = \begin{cases} 
1 & \text{if } t \in [t_i, t_{i+1}] \\
0 & \text{else}
\end{cases}
\]

For \( k \geq 1 \) and \( \forall t \in [t_0, t_n], \)

\[
N_{i,k}(t) = \frac{t - t_i}{t_{i+k} - t_i} N_{i,k-1}(t) + \frac{t_{i+k+1} - t}{t_{i+k+1} - t_{i+1}} N_{i+1,k-1}(t) \quad (5)
\]

\( S_d \) was also broken up into two planar cubic B-splines \( (S_d^1, P_{d,i}^1, i = -3 \ldots n-1) \) and \( (S_d^2, P_{d,i}^2, i = -3 \ldots n-1) \) in \( P1 \) and \( P2 \), respectively. The deflections \( S_j^j (j = 1, 2) \) in the \( j^{th} \) plan were:

\[
\forall t \in [t_0, t_n], S_d^j(t) = \sum_{i=-3}^{n-1} P_{d,i}^j N_{i,k}(t) \quad (6)
\]

The \( P_{d,i}^j \) coefficients were totally defined by minimizing the following problem:

\[
u_0(S_d^j(0) - y_0^j)^2 + u_1 \left( \frac{dS_d^j}{dt}(0) - dy_0^j \right)^2 + \sum_{k=1}^{m} w_k (S_d^j(x_k^j) - y_k^j)^2 + \tau \int_0^L \left( \frac{d^2S_d^j}{dt^2}(t) \right)^2 dt \quad (7)
\]

where \((x_k^j, y_k^j)_d, k = 0 \ldots m\) are the projections of the \((m+1)\) segmented points of the \( d^{th} \) needle in the plan \( P_j \) \((j = 1, 2)\), \( y_0^j \) and \( dy_0^j \) are the informations on the position and the orientation of the proximal extremity, \( u_0, u_1, w_i \) and \( \tau \) are the weights chosen to acquire a good compromise between the smoothness of the result and the interpolated cubic B-spline.

Because of the B-splines model, the different derivatives could be computed for each new shape, thus giving access to the bending moment and the shear forces according to the paragraph 2.1.1.

2.1.4. Principal Component Analysis (PCA)

In this section, our use of PCA (Cootes (2000), Cootes et al. (2001)) is described, in order to build a model of medical needles by analyzing the
shape of a set of such needles. This model makes it possible to extract relevant information related to this set of shapes and may mimic shapes similar to those of the training set. It gives a compact representation of allowable variation, but it is specific enough not to allow arbitrary variation different from that seen in the training set.

Our approach requires that the topology of the object cannot change and that the object is not so amorphous that no distinct landmarks can be applied. According to paragraph 2.1.1, needles are comparable to a beam in view of the various mechanical assumptions required for beam theory: they do not exhibit large deflections. According to paragraph 2.1.2, needles are aligned into a common coordinate frame: the shapes of the needles are considered to be independent of the position, orientation and scale of the needle and are hence comparable. Moreover, each needle is modelled by a uniform cubic B-spline according to the same methodology. This methodology guarantees the coherence of landmarks, that are defined as the control points.

Given a set of needle shapes \( S_d \) \( (d = 1 \ldots 62) \), and more precisely, given a set of associated control points \( P_d = P_{d,i}, i = -3 \ldots n - 1 \), a mean shape \( \bar{S} \) may be defined from its mean control points \( \bar{P} = \bar{P}_{d,i}, i = -3 \ldots n - 1 \). A new set of control points \( P \), associated to a new shape \( S \) may be approximated by:

\[
P \approx \bar{P} + Db
\]

where \( D = (D_1|D_2|\ldots|D_p) \) contains \( p \) eigenvectors of the covariance matrix, with

\[
\sum_{i=1}^{p} \lambda_i \geq 0.95 \sum_{i=0}^{\text{Tot}} \lambda_i
\]

where \( \lambda_i \) is the eigenvalue associated to the eigenvector \( D_i \) and \( b \) is a \( p \) dimensional vector defining a set of parameters of the deformable model given by

\[
b = D^T(P - \bar{P})
\]

2.1.5. Synthesis of the study’s methodology

To build the loading basis, known needles are used and their deflections are recovered. First, all needle deflections are adjusted in order to make them comparable. Second, all needle deflections are model by B-splines. Then, by
associating the B-spline theory and beam theory, the bending moment and shear forces can be found. Finally, a statistical approach is employed to extract the relevant information from the set of deflection as well as bending moment and shear force, of which the basis will be made up. This generic 3D loading basis can then be applied to reconstruct any needle deflection with a linear combination of the linearly independant load vectors.

2.2. Material

2.2.1. Pig and needle

A pig shoulder was selected to carry out the tests with a known needle. This choice was justified by the similarity of the mechanical properties of pig tissue compared to human tissue, as well as the presence of tendons, muscles and bones, allowing an insertion into a complex medium. A fresh shoulder (300x200x100mm) was prepared by laying it on a plastic case in the CT scan. The outside wall of the muscle was free.

The needle, commonly used for interventional radiology procedures, was a 22-gauge needle with a length of 200mm and a Young’s modulus estimated at 200GPa (stainless steel). The needle tip was bevelled at an angle of 40 degrees. The thin diameter of such medical needles make them minimally invasive and also allows them to be deformed to avoid or reach a target. The needle was fully inserted into the pig shoulder for a total of 62 insertions. By means of this set of experiments, various factors on needle-insertion-force as the influence of insertion method and tissue characteristics, enumerated by Van Gerwen et al. (2012), are highlighted and thus the various possibilities of needle deflection which could take place in percutaneous interventional radiology are obtained. The different conditions used during needle insertion (rotation of the needle, forces applied to the needle hub, number of deflections) to create a certain level variability in the resulting needle deflection, are detailed in table 1.
<table>
<thead>
<tr>
<th>Needle number</th>
<th>Condition of insertion</th>
<th>Application of efforts</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 - 4</td>
<td>1 rotation of 180°</td>
<td>without</td>
</tr>
<tr>
<td>5 - 8</td>
<td>1 rotation of 180°</td>
<td>in the direction of bevel</td>
</tr>
<tr>
<td>9 - 12</td>
<td>1 rotation of 180°</td>
<td>in the orthogonal direction to bevel</td>
</tr>
<tr>
<td>13 - 16</td>
<td>1 rotation of 90°</td>
<td>without</td>
</tr>
<tr>
<td>17 - 20</td>
<td>1 rotation of 90°</td>
<td>in the direction of bevel</td>
</tr>
<tr>
<td>21 - 24</td>
<td>1 rotation of 90°</td>
<td>in the orthogonal direction to bevel</td>
</tr>
<tr>
<td>25 - 28</td>
<td>2 rotation of 180°</td>
<td>without</td>
</tr>
<tr>
<td>29 - 32</td>
<td>2 rotation of 180°</td>
<td>in the direction of bevel</td>
</tr>
<tr>
<td>33 - 36</td>
<td>2 rotation of 180°</td>
<td>in the orthogonal direction to bevel</td>
</tr>
<tr>
<td>37 - 40</td>
<td>2 rotation of 120°</td>
<td>without</td>
</tr>
<tr>
<td>41 - 44</td>
<td>2 rotation of 120°</td>
<td>in the direction of bevel</td>
</tr>
<tr>
<td>45 - 48</td>
<td>2 rotation of 120°</td>
<td>in the orthogonal direction to bevel</td>
</tr>
<tr>
<td>49</td>
<td>1 rotation of 180°</td>
<td>significant in the direction of bevel</td>
</tr>
<tr>
<td>50</td>
<td>1 rotation of 90°</td>
<td>significant in the direction of bevel</td>
</tr>
<tr>
<td>51</td>
<td>2 rotation of 180°</td>
<td>significant in the direction of bevel</td>
</tr>
<tr>
<td>52</td>
<td>2 rotation of 120°</td>
<td>significant in the direction of bevel</td>
</tr>
<tr>
<td>53 / 54</td>
<td>5 steps of insertion</td>
<td>without</td>
</tr>
<tr>
<td>55 / 56</td>
<td>5 steps of insertion</td>
<td>in the direction of bevel</td>
</tr>
<tr>
<td>57 / 58</td>
<td>5 steps of insertion</td>
<td>without</td>
</tr>
<tr>
<td>59 / 60</td>
<td>5 steps of insertion</td>
<td>in the direction of bevel</td>
</tr>
<tr>
<td>61 / 62</td>
<td>5 steps of insertion</td>
<td>in the orthogonal direction to bevel</td>
</tr>
</tbody>
</table>

Table 1: Description of the condition of needle insertion.

2.2.2. Theoretical cases and needles

Theoretical needles are defined with an aim of validating the loading basis. The most classical static loading case is chosen (a single force applied at the distal extremity of the needle) as corresponding to the model often used in literature. In such a model, it can be assumed that the trajectory of the needle through the tissue is primarily dependent on forces on the bevel tip (Alterovitz et al. (2005), Webster et al. (2006), Duindam et al. (2010)). Moreover, different boundary conditions can be added to generate any other
curvatures and to reveal inflection points. At last, a distributed loading is proposed to change the mathematical degree of the needle deflection in beam theory. These theoretical cases are determined according to the equation in section 2.1.1. They are then segmented by the same method than the pig needle to include a segmentation error and put in the same reference mark. The three types of loading considered are presented in figure 5.

(a) ![Concentrated radial force](image)
(b) ![Double concentrated radial forces](image)
(c) ![Constant loading](image)

Figure 5: Theoretical cases. a) Concentrated radial force, b) Double concentrated radial forces, c) Constant loading.

2.2.3. Patients and needle

A clinical trial was performed in the Grenoble University Hospital involving CT-guided interventional radiological procedures such as biopsies, injections. This clinical trial was authorized by the AFSSaPS, the relevant French regulatory authority for biomedical research, and by the Comité de Protection des Personnes Sud-Est V, an institutional French review board (ClinicalTrials.gov identifier: NCT00828893). Twenty CT-images of patients with needles inserted were acquired. Contrary to the needle used on the pig shoulder, neither geometrical data (radius, length) nor materials characteristics (Young modulus) of the needles involved in this clinical trial were known. This data was acquired to evaluate the basis for any needle even if no information on mechanical properties is available.
3. Results

In this section, the construction of the loading basis is presented. The results of needle reconstructions, with this basis, are exposed for some needles (chosen to show the different curvatures and errors) as well as the all reconstruction errors.

3.1. Construction of the loading basis

As shown in figure 6, for the deflection needle model, 95% of the variability in the training set could be explained using the first nine among 18 basis modes ($p = 9$).

Figure 7 illustrates the distribution of the $b_i$ parameters of the deformable model for the $i^{th}$ principal deformation mode of the model for each needle. By applying the upper limits, it is ensured that the shape/moment/efforts generated is similar to those in the original training set. At these upper limits, the shape/moment/efforts of the principle modes of variation are shown in figure 8 (modes 1 to 3), figure 9 (mode 4 to 6) and figure 10 (modes 7 to 9). Each mode present a complex form with different curvatures in the two planes $P_1$ and $P_2$. Furthermore, none of them corresponds to a simple force applied at the needle extremity.

3.2. Validation of the deflection

The performance of the loading basis was checked on all the needle cases (pig needles (cf. section 2.2.1), theoretical needles (cf. section 2.2.2) and patient needles (cf. section 2.2.3)) presented in the study. Each shape was rebuilt using the loading basis in the two planes $P_1$ and $P_2$ (cf. section 2.1.2). As previously described in the paragraph 2.1.4, in order to interpret a new needle shape, the parameters which best match the model instance to the image data must be found. The basis linear combination coefficients are those which minimized the euclidean distance between the model instance and the image data. All following figures (12 to 19) show some of the reconstructed needles. The points represent the segmentation of the needle as indicated in the paragraph 2.1.2. The reconstruction of the needle, i.e. the
optimal linear combination of loading base is represented by the dashed lines.

3.2.1. Validation criteria

For each of the following cases, reconstruction errors ([Q2, Q3], Maxtip)\textsubscript{P1,P2} (with Q2 the median value, Q3 the upper quartile and Max\textsubscript{tip} the maximum value of the tip errors) are calculated in (mm) and in (‰) by considering that all needles were normalized to 200mm in the two planes. These errors consider the euclidean distance between the reconstructed shape and the needle segmented points for all needles. Max\textsubscript{tip} represent the maximum error on the needle tip, whose the placement is very important in percutaneous procedures. In figure 11, the first bar represents the median value (Q2) and the second one the upper quartile (Q3). The maximum values of the tip deflection errors for each cases are show in the table of the figure 11.

3.2.2. Validation of the pig cases

Figure 12 shows the reconstruction of 2 pig needles, which were used to determine the loading basis. In figure 12.(a) and figure 12.(b), the importance of the initial modeling of the needle, i.e of the determination of the P1 and P2 planes, is emphasized. These planes cannot be inverted. Indeed, the reconstruction in figure 12.(a) is carried out in the planes, P1 and P2, defined for data registration (cf. section 2.1.2) whereas in figure 12.(b), these planes are inverted resulting in an incorrect reconstruction. This shows that the two planes do not have the same meaning and that the deformation modes are quite different in the two planes. In the figure 12.(c), another pig needle is shown with a reconstruction in the correct planes. The reconstruction errors (‰) for all pig needles are shown in figure 11.(c) and figure 11.(d) for P1 and P2 respectively: ([0.07, 0.22], 4.51)\textsubscript{P1}, ([0.01, 0.03], 7.8)\textsubscript{P2}.

3.2.3. Validation of the theoretical cases

Reconstructions of the three theoretical cases are presented in figure 13. The reconstruction errors (‰) for all theoretical needles are shown in figure 11.(e) and figure 11.(f) for P1 and P2 respectively: ([0.39, 0.59], 3.93)\textsubscript{P1},
3.2.4. Validation of the in vivo cases (human)

While the material properties of the pig needles were known, this was not the case for the in vivo human shape estimate. It was possible to reconstruct the needle deflection for needles with different lengths and different diameters. Two reconstructions are shown in figure 14. The reconstruction errors ($\%_e$) for patient needles are shown in figure 11.(a) and figure 11.(b) for $P1$ and $P2$ respectively: $([0.07,0.11],7.41)_{P1}$, $([0.01,0.03],5.02)_{P2}$.

4. Discussion

Model parameters

Because of the B-splines model, it is possible to build a synthetic and representative loading basis of the reality and to represent relevant information. Any shape/moment/efforts can be broken up into this basis while associating the different derivatives of the B-spline theory to the beam theory.

The weights $u_0$, $u_1$, $w_i$ of the B-spline model are chosen equal to 1. $\tau$ has a value of 100 as a good compromise between the interpolation and the smoothness qualities of the cubic B-spline. These weights were set up at these values at the beginning of this study after some registration on known cases.

Accuracy of reconstruction

Needle misplacement errors are revealed by clinical studies to be due to several causes such as human errors, imaging limitations, target uncertainty, tissue deformation and needle deflection. The desired tolerance for the accuracy of needle insertion in clinical practice depends on the application and thus is not a well-defined general information. It could be assumed that the precision of diagnosis and the effectiveness of the treatments would increase for a large numbers of insertions if the misplacement error was kept
low. In fact, a millimetric accuracy is required in procedures such as biopsies (prostate, liver, breast and kidney) while sub-millimetric accuracy is desired in brain or foetus procedures (Van Gerwen et al. (2012)). This loading basis performs well in all cases and is compatible with the clinical practice and the current navigation systems that require a sub-millimetric accuracy.

The use of PCA allows to identify the principal variations modes, which are the most representative of the plausible variations of the needles. It gives thus a compact representation of allowable variation. Indeed, from an 18 vectors basis, it simplifies down to a basis of 9 meaningful vectors.

To show that the base of loading is generic, reconstructions of theoretical cases were presented. The reconstruction errors obtained on degenerated theoretical cases confirm the robustness of the model. Indeed, 75% of the errors are under 0.59‰, 0.43‰ for $P_1$ and $P_2$ respectively.

In this paper, it is also shown that the loading basis can be used to reconstruct needles deflection during medical interventions in a relevant way. The results show that 75% of the errors are under 0.11‰, 0.03‰ with a maximum error tip at 7.41‰ and 5.02‰ for $P_1$ and $P_2$ respectively. Moreover, it is noted that the errors obtained on the human cases are lower than those obtained with the experimental cases, suggesting that the different methods of insertion carried out on the pig were exhaustive enough to represent a true environment.

**Static approaches**

During the insertion of a needle in percutaneous needle intervention, the correct trajectory of the needle is verified by iterative CT images in order to prevent the complication of the medical act if anatomic structures are not respected. Thus, such procedures are a succession of needle insertions and image acquisitions. This results in a succession of static images of the needle. A static loading assumption could therefore provide a better model for understanding and apprehending real clinical needle behavior. Indeed, the model of this work may be particularly well adapted to this case in order to optimize the reconstruction procedure as presented in Park et al. (2010) or in Yang et al. (2010) for the positioning of strain gauges and sensors.
As an illustration of these remarks are referring to the results presented in table 1, needles 53 to 62 were inserted with five steps of insertion, like in a clinical procedure under CT guidance i.e. by several static stages. For these 10 needles, the intermediate shapes were reconstructed in figure 15. What was highlighted is the fact that the part of the already inserted needle does not change in a consistent way during successive stages of insertion. This confirm a significant amount of work, which assume that the trajectory of the needle through the tissue is primarily dependent on the bevel tip (Alterovitz et al. (2005), Webster et al. (2006)).

**Comparison with other approaches**

DiMaio and Salcudean (2002), Crouch et al. (2005) and Dehghan et al. (2008) used indirect methods to reconstruct the axial load distribution along the needle on artificial materials, based on external force and tissue displacement measurements. An indirect means of estimating the applied force distribution involves the measurement of the resulting tissue deformation using a grid of black dots that can be tracked by the camera as they move. If the relationship between applied load forces or stresses and tissue displacement is known, then the distribution of force applied along the needle shaft can be computed. In this study, an indirect method is also used, based on the easier to implement beam theory instead of finite element analysis. Furthermore, the load distribution along the entire needle length can be found and not only the axial load.

Kataoka et al. (2002), Simone and Okamura (2002), performed experiments for measuring the resultant force, during needle insertion into biological materials using a load cell mounted on the outside extremity of the needle in the direction of motion. However, this force depends on the insertion tissue and is not validated on other cases. In this work, the loading basis comes from pig experiments, but are generic enough to be validated on human and theoretical cases.

**Comparison with a similar approach**

Contrary to most other approaches, Park et al. (2010) propose a static randomized force profile with concentrated radial and a axial tip forces. The radial impulses have an amplitude of $0 - 0.07$ N in the $(x, y)$ plane. The con-
centrated axial and radial forces applied on the needle tip have a maximum amplitude of 0.1 N. However, to our knowledge, there is no explanation for these amplitudes. For such loading profiles, the needle undergoes at most one curvature in one plane. In the present study, the amplitude of forces is from 0 – 4 N and the loading basis can consider more than one curve in 3D.

5. Conclusion

In this paper, a generic loading basis was presented that acts on needles during a medical procedure using needle shape data collected from a total of 62 needle insertions into a pig shoulder. The needle shape data was collected by segmenting and recording 3D points along a needle from CT-images. Furthermore, using static Beam and B-spline theories, this generic loading basis was determined and then validated on theoretical cases as well as on 20 needles inserted into human tissue during real clinical procedures.

References


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Figure 6: Eigenvalues and inertia of the eigenvalues determined by PCA
Figure 7: Visualisation, for all needles, of the coefficients distribution for the first nine principal deformation modes. Theoretical needles (stars), pig needles (diamonds), patient needles (circles)
Figure 8: Effect of varying each of first three needle model shape(mm) / moment(Nmm) / efforts(N) parameters in turn between given limits. Each pad shows the effect of varying one of the shape parameters (or moment and efforts) keeping the others at zero for the two plan along the needle (0-200mm). The median (points) are the mean shape / moment / efforts.
Figure 9: Effect of varying from fourth to sixth needle model shape (mm) / moment (Nmm) / efforts (N) parameters in turn between given limits. Each pad shows the effect of varying one of the shape parameters (or moment and efforts) keeping the others at zero for the two plan along the needle (0-200mm). The median (points) are the mean shape / moment / efforts.
Figure 10: Effect of varying the last three needle model shape (mm) / moment (Nmm) / efforts (N) parameters in turn between given limits. Each pad shows the effect of varying one of the shape parameters (or moment and efforts) keeping the others at zero for the two plan along the needle (0-200mm). The median (points) are the mean shape / moment / efforts.
Figure 11: Reconstruction errors with standard deviation in (‰) and (mm) in the two planes P1 and P2: first bar (median value (Q2)), second bar (upper quartile (Q3)). The table represent the maximum value of the tip errors. (a) Patient needles errors: P1, (b) Patient needles errors: P2, (c) Pig needles errors: P1, (d) Pig needles errors: P2, (e) Theorical needles errors: P1, (f) Theorical needles errors: P2.
Figure 12: Deflection reconstruction of experimental cases (pig): a) Needle 1 in the correct planes P1 and P2, b) Needle 1 if the planes P1 and P2 are reversed, c) Needle 21 in the correct planes P1 and P2. The dots represent the segmented needle and the dashed lines represent the reconstructed needle with the loading basis.
Figure 13: Deflection reconstruction of theoretical cases.  
(a) Reconstruction of the concentrated radial force case in the planes P1 and P2.  
(b) Reconstruction of the double concentrated radial forces case in the planes P1 and P2.  
(c) Reconstruction of the constant loading case in the planes P1 and P2.  
The dots represent the segmented needle and the dashed lines represent the reconstructed needle with the loading basis.
Figure 14: Deflection reconstruction of patient cases. a) Needle 027 ARS-NAV in the planes P1 and P2, b) Needle 020 BOM-NAV in the planes P1 and P2. The dots represent the segmented needle and the dashed lines represent the reconstructed needle with the loading basis.

Figure 15: The successive representation of a needle trajectory in various steps of its insertion (five insertion steps). Inside the tissue, the needle keep the same trajectory.