In-Silico User Study Case: Wearable Feedback Haptic Device for Rehabilitation

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Complete List of Authors:	André, António Diogo; INEGI; University of Porto Faculty of Engineering, Parente, Marco; INEGI, IDMEC; University of Porto Faculty of Engineering Martins, Pedro; INEGI; Universidad de Zaragoza,

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António Diogo André[1,2] and Marco Parente[1,2] and Pedro Martins[1,3]

Abstract

Soft smart materials have useful properties for addressing everyday problems affecting human health and well-being, having a positive societal impact. For instance, these materials can serve as sensors for breath monitoring or as soft actuators to stimulate muscles impaired by injury or illness. A notable example of their versatility lies in piezoelectric materials, which can function both as passive elements (utilising the direct piezoelectric effect) and as active elements (employing the converse piezoelectric effect). This dual functionality showcases the broad potential of smart materials in various applications. The present study is an *in silico* simulation of a wearable piezoelectric material (polyvinylidene fluoride - PVDF), using finite element analysis (FEA) to evaluate the effectiveness of the touch sensation provided by the haptic device on human skin, using different actuators geometries and voltage input intensities. Moreover, the main active element, a PVDF-based soft actuator, was fully characterised in terms of the piezoelectric matrix, using an inverse finite element approach. In conclusion, the findings point to promising results when using this haptic technology for re-educational therapies.

Keywords

In-silico study, FEM analysis, PVDF-based materials, Wearable haptic device, Rehabilitation

Introduction

According to the World Health Organisation (WHO) **?**, in 2022, approximately 1300 million people experience some kind of physical disability. A significant part of these mobility impairments require or will need in the future some sort of therapeutic rehabilitation, motion training or motion assistance during a period of time after the onset of disorders **?**.

During the last decades, the medical and research community have been trying to mitigate this problem and accelerate the recovery of those who are suffering from lack of mobility **?**. As a result, society's efforts have led to a trend of developing rehabilitative wearable devices, with the possibility to assist without tethering the patient to a specific location. However, when based on soft materials, those devices evidence limited ability to perform motion tasks **?**, being essentially developed for haptic stimuli solutions on human skin.

Skin is the largest organ of the body with different functions associated. It is the first barrier against germs, helps to regulate the body temperature and, most importantly, provides the main and the first sensory channel of the body to perceive external stimuli, based on touch.

Through the skin, people can interact with the surrounding objects, feeling their properties, such as weight, temperature, textures and motion **?**. The interaction between skin and objects results in nerve stimuli, which are primarily received by different receptors in the skin and later sent to the brain. Pacinian corpuscles, Meissner corpuscles, Merkel complexes, Ruffini corpuscles, and C-fiber LTM (low threshold mechanoreceptors) are the mechano-receptors that can detect even innocuous stimuli **?**, for example, responding to displacements of the skin on the order of micrometres (μm). Different stimuli, such as mechanical, electrical and thermal can feedback those receptors and transmit different sensations and information through the nerves **?**.

To take advantage of the high sensitivity of skin, different technologies have been developed, using different powered-up soft materials, for rehabilitation and re-educational therapies through the use of external devices. As reviewed by André et al **?**, different soft actuators can be used to provide mechanical, electrical or thermal feedback on the skin, such as electroactive materials (piezoelectric), magnetic responsive materials, thermally responsive materials or even photo-responsive materials. All these smart materials, in particular piezo polymers due to their versatility, can play an important role in actuation for tactile sensation since they are light, flexible and biomimetic.

Piezoelectric materials, in particular polymers such as polyvinylidene fluoride (PVDF), have been largely studied in rehabilitation solutions since they can be used as sensors **??** or actuators **??**. For example, Pan *et al.* **?** developed a mechanomyography (MMG) sensor using

[1]Associated Laboratory of Energy, Transports and Aeronautics (LAETA), Biomechanic and Health Unity (UBS), Institute of Science and Innovation in Mechanical and Industrial Engineering (INEGI). Campus da FEUP, Rua Dr. Roberto Frias, 400, 4200-465, Porto, Portugal
[2]Faculty of Engineering of University of Porto (FEUP). Rua Dr. Roberto Frias, s/n, 4200-465, Porto, Portugal
[3]i3A, Universidad de Zaragoza, Zaragaoza. C. de Mariano Esquillor Gómez, s/n, 50018, Zaragoza, Spain

Corresponding author:
Pedro Martins
Email: palsm@fe.up.pt

PVDF's piezoelectric properties for lower limb rehabilitation exoskeleton. They concluded that the approach used improved the sensitivity in driving the device. Gariya *et al.* **?**, developed a pneumatic soft actuator with a PVDF membrane for sensing the bending deformation of the actuator, which was tested for the medical application of human finger rehabilitation.

Nevertheless, the use of PVDF, as a haptic feedback element, has been presented in the literature with distinct purposes. Ege and Balikci **?** introduced a haptic interface using transparent thin films of PVDF actuators to feedback touch displays. Maeda *et al.* **?** developed a wearable haptic augmentation system using a skin vibration sensor made from PVDF. In their studies, they pointed to an increase in haptic sensation of, approximately, 5% when compared to no feedback condition.

The present study aims to perform an *in-silico* user study, as proof of concept using the finite element method (FEM), of a wearable device for re-educational therapies, using PVDF/ionic liquid (IL) as haptic actuators. These feedback elements, with distinct tip geometries (rectangular, circular and triangular), will be fed with different voltage intensities, to study the influence of input voltage on the touch perception. With this approach, the authors hope to understand if the developed methodology can be effectively applied in rehabilitation scenarios and if the skin can be stimulated through gentile touch sensations provided by PVDF/IL's actuators. This work uses the index finger as a reference model, due to its high sensibility **?**.

Inverse Finite Element Analysis

André et al **??** characterised PVDF/IL haptic actuators experimentally, in terms of porosity and microscopic defects (high resolution scans were obtained); mechanical properties (i.e. Young modulus (E) and yield stress (σ_{Yield})); chemical properties (i.e. degree of crystallinity (χ) and electroactive phases (β phase)); and electrical, piezo and conductive properties (i.e. dielectric permittivity (ϵ'), AC conductivity (σ_{AC}) and piezoelectric constant (d_{33})). However, those properties are not sufficient to define a material model able to reproduce the behaviour observed experimentally, namely the electromechanical performance **?**.

To better mimic the experimental phenomenon observed, an inverse finite element analysis approach was used to evaluate the piezoelectric matrix.

Piezoelectric Constitutive Equations

The standard form of the piezoelectric constitutive equations can be presented in four different forms by taking either two of the four field variables as independent. Considering a tensorial representation of the strain-electric displacement form and the components of stress and electric fields as the independent variables **???**, it comes

$$S_{ij} = s^E_{ijkl} \cdot T_{kl} + d_{kij} \cdot E_k \tag{1}$$

$$D_i = d_{ikl} \cdot T_{kl} + \epsilon^T_{ik} \cdot E_k \tag{2}$$

where the index i, j, k, l represent the different components or directions in the tensor equations, S_{ij} are the strain components, D_i are the electric displacements, s^E_{ijkl} are the elastic compliance constants, T_{kl} are the stress components, d_{kij} are the piezoelectric constants of the material, ϵ^T_{ik} are the permittivity constants and finally E_k is the electric field component.

In a matrix form, the equations (1) and (2) are given as **?**,

$$\begin{bmatrix} S \\ D \end{bmatrix} = \begin{bmatrix} s^E & d^t \\ d & \epsilon^T \end{bmatrix} \cdot \begin{bmatrix} T \\ E \end{bmatrix} \tag{3}$$

where the superscripts E and T denote that the respective constants are evaluated at constant electric field and constant stress, respectively, and t represents the transpose.

Assuming the study of a piezoelectric thin film, such as the case under analysis, matrix 3 can be further simplified. If the thin structure is assumed as a thin beam, based on the Euler-Bernoulli beam theory or Rayleigh beam theory **?**, the stress components T_{22}, T_{33}, T_{23}, T_{13} and T_{12} are negligible ($T_{\neq 11} = 0$), since only the one-dimensional bending stress (T_{11}) has a non negligible value. Moreover, if the electrodes are placed perpendicular to the 3-direction, equation (3) becomes

$$\begin{bmatrix} S_{11} \\ D_{33} \end{bmatrix} = \begin{bmatrix} s^E_{11} & d_{31} \\ d_{31} & \epsilon^T_{33} \end{bmatrix} \cdot \begin{bmatrix} T_{11} \\ E_{33} \end{bmatrix} \tag{4}$$

Matrix (4) shows that the piezoelectric constant, d_{31}, is crucial and also needed to characterise properly the piezo polymer PVDF/IL thin film.

Piezoelectric Matrix

Overall, the piezoelectric matrix of PVDF is given by equation (5), since the material evidences anisotropy **??**. That fact implies that the piezoelectric properties change with direction **?**.

$$d_{ij} = \begin{bmatrix} 0 & 0 & 0 & 0 & d_{15} & 0 \\ 0 & 0 & 0 & d_{24} & 0 & 0 \\ d_{31} & d_{32} & d_{33} & 0 & 0 & 0 \end{bmatrix} \tag{5}$$

where the subscript i refers to the direction of plane polarisation, while j is the direction of the induced strain.

However, in the literature, the piezoelectric properties of PVDF more commonly used to characterise the polymeric material are d_{31} and d_{33} **??**. The d_{31} constant is the transverse coefficient, which defines the mechanical strain created in the perpendicular direction to the applied electric input; while d_{33} is the longitudinal coefficient, which defines mechanical strain in the same direction as the applied stress **?**. With a good approximation, the two coefficients are enough to define the phenomenological behaviour observed experimentally. Considering that fact, for this particular study case, matrix (5) could be simplified into matrix (6), ignoring d_{32} coefficient and the rotational and shear components (d_{24} and d_{15}).

$$d_{ij} = \begin{bmatrix} 0 & 0 & 0 \\ 0 & 0 & 0 \\ d_{31} & 0 & d_{33} \end{bmatrix} \tag{6}$$

Inverse-FEM Algorithm

In recent years, the identification and characterisation of the piezoelectric matrix using inverse FEM algorithms have grown and gained popularity ?. Figure (1a) summarises the steps adopted to approximate numerically the piezoelectric constant d_{31}.

In general, the optimisation algorithm runs until achieving a d_{31} value that grants similar numerical behaviour when compared to the experimental behaviour. The experimental displacement (Displ_{exp}) measured at the tip of the PVDF/IL sample, when electrically stimulated with a square pulse of 10 V for 10 seconds, was 1.86 mm ?. The optimisation algorithm optimised the initial d_{31} parameter guess (15x10^{-12} pm/V ?) until the exit optimisation criteria is satisfied: numerical displacement (Displ_{num}) is equal to the Displ_{exp} more or less the error (Err = 15%) (Figure 1b).

Applying an iterative refinement method ?, and after 8 iterations, the optimised d_{31} was 3.75x10^{-07} pm/V, with a Displ_{num} of 1.87 mm (error of 0.8%).

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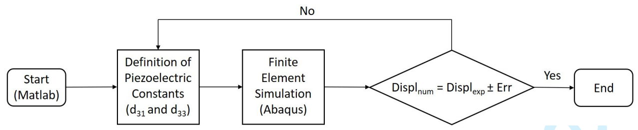
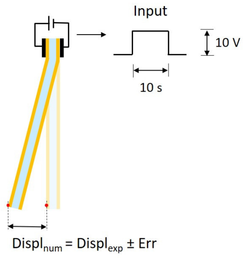


(a)



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(b)

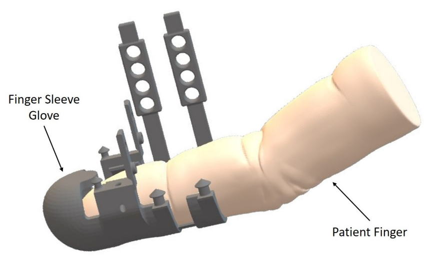
Figure 1. (a) Flow chart for d_{31} optimisation; (b) Exit optimisation criteria when Displ_{num} = $\mathrm{Displ}_{exp} \pm$ Err.

In-Silico User Study Case

A numerical model of the wearable haptic device was developed to simulate the stimuli contact and interaction between an index finger and the device, through gentile and soft haptic touch sensations.

The finger sleeve prototype used was first designed by André et al ?. Briefly, the finger sleeve prototype was developed considering the usage of a thermoplastic polyurethane (TPU) as the main structural material. This choice grants the finger device with flexibility and high elasticity. Moreover, TPU can be used as a 3D printable material in fused filament fabrication (FFF) technique, which allows fast prototype iterations. All these properties make this material the ideal candidate for its purpose.

The prototype allows adaptability to patients' physiognomy since it has adjustable straps. As a result, the device can be adapted to different finger shapes (Figure 2).



2.jpg

Figure 2. Schematic illustration of the wearable device prototype (finger sleeve) worn by an index finger.

As the PVDF/IL haptic actuators were previously characterised ???, as well as the finger sleeve was properly designed ?, an *in-silico* numerical model was developed on Dassault Systemès Abaqus ? to study the device's behaviour in real scenarios. Afterwards, real interaction scenarios between the user and the haptic feedback device were simulated using a FEM approach.

FEM Model

The numerical model was defined in Abaqus 2022 ?. The elements needed to perform the *in-silico* user study case were defined, i.e., the mechanical properties, interactions between parts, the boundary conditions (BCs) and the mesh.

The physiognomy of the finger model was based on literature ??? and drawn in Dassault Systemès SolidWorks 2020 ?. It is composed of bone, soft tissue and skin with corresponding properties based on the literature ????. The wearable haptic device was modelled considering PVDF/IL soft actuators (with 6 mm width, w, 12 mm total length, l_t, 9 mm free length, l_f, and 0.06 mm thickness, t) and a TPU finger sleeve. The properties of those materials were also based on the literature ????. Different tip shapes for the actuation were considered, as shown in Figure 3a, such as rectangular, circular (with radius, R, of 4.3 mm centred at the middle of the free length) and triangular (with different tip pronunciations, started at length l=0, l=1/4, l=2/4, l=3/4 and finally l=4/4 from the free tip). Table 1 summarises the properties of the parts used to simulate the *in-silico* user study case.

The FEM model was prepared and assembled in Abaqus, considering the wearable haptic device and the finger model (Figures 3b and 3c). Since the finger sleeve glove is not an active part of the simulation, it was not considered, resulting in a faster numerical simulation. Moreover, the contact and interaction properties, as well as the node ties were defined among all the active parts (PVDF/IL actuators ⇔ skin - contact/interaction; skin ⇔ soft tissues - tie; soft tissues ⇔ bone). The BCs were applied in the anterior and posterior sections of the finger model and on both PVDF/IL actuators. Since the finger sleeve glove is not present in the simulation, the BCs applied on the actuators satisfy the interaction with the glove. The electrical inputs were also applied in both PVDF/IL soft actuators, assuming a square pulse with different voltage intensities with a time duration of 10 seconds (Figure 3d). The active elements/parts of the simulation were then meshed. The finger model was meshed considering 3D stress hexagonal mesh elements (C3D8R - 8 node linear brick, reduced integration). On

Table 1. Properties of the parts used in the numerical model in Abaqus.

Parts	Density, ρ (kg/m^3)	Young Modulus, E (MPa)	Poisson Ratio, ν (adimensional)	Elect. Conductivity, σ_{AC} (S/m)	Piezoelectric Const., d_{33} (pm/V)	Piezoelectric Const., d_{31} (pm/V)
Bone **?**	1900	17x10^3	0.3	-	-	-
Soft Tissue **?**	1000	0.08	0.4	-	-	-
Skin **??**	-	2.5	0.48	-	-	-
TPU finger sleeve **?**	-	2.41x10^3	0.3847	-	-	-
PVDF/IL soft actuators **???**	1425.9	127.0	0.18	1.98x10^{-07}	0.6	3.75x10^5

another hand, the PVDF/IL soft actuators were defined using piezoelectric tetrahedral mesh elements (C3D4E - 4 node linear piezoelectric brick). The model's number of elements and nodes changed according to the tip geometry (rectangular, 15303 nodes and 21728 elements; circular, 18225 nodes and 30958 elements; or triangular, 16913 nodes and 26673 elements). Figure 3e shows the mesh applied to a specific tip geometry situation.

Results and Discussion

The *in-silico* user study case was carried out to understand the effectiveness of the touch sensation provided by the finger sleeve prototype on the skin to be used in the context of re-educational therapy. Both PVDF/IL soft haptic actuators were input simultaneously under different initial conditions, such as the tip geometries (rectangular, circular and triangular; initiated at the middle of the free part of the actuator) and pulse intensities (2.5, 5.0, 7.5 and 10.0 V). The numerical results achieved were, then, compared against the minimum touch sensation felt by human fingers described in the literature.

FEM Simulation

The FEM simulation of the *in-silico* user study case, considering different voltage intensities and tip geometries, evidenced results according to our expectations. Figure 4 shows the results for the different tip geometries tested, considering the maximum voltage intensity (10.0 V). The best touch sensation provided by the finger sleeve prototype was achieved for the triangular tip geometry of the PVDF/IL soft actuators. According to the simulation, that value was 578.0 Pa on the skin finger model (Figure 4c).

The results showed that the touch sensation provided by PVDF/IL soft actuators increased with the voltage intensity. For the minimum voltage intensity tested, 2.5 V, the simulation did not show any contact pressure for all the different situations. The relation between the contact pressure and the voltage intensity is justified by the proportionality of the electrical input applied and the consequent mechanical deformation induced. It means that when higher voltage intensities are applied to the samples, higher polarization is induced on them, which increases the outputs, such as the contact pressure **?** and displacement **?**. The same cause/effect is observed in other studies available in the literature. Raza et al **?** studied micro-structured porous electrolytes for highly responsive ionic soft actuators, using PVDF-co-hexafluoropropylene (PVDF-co-HFP) as the main active element. They observed a direct influence of the

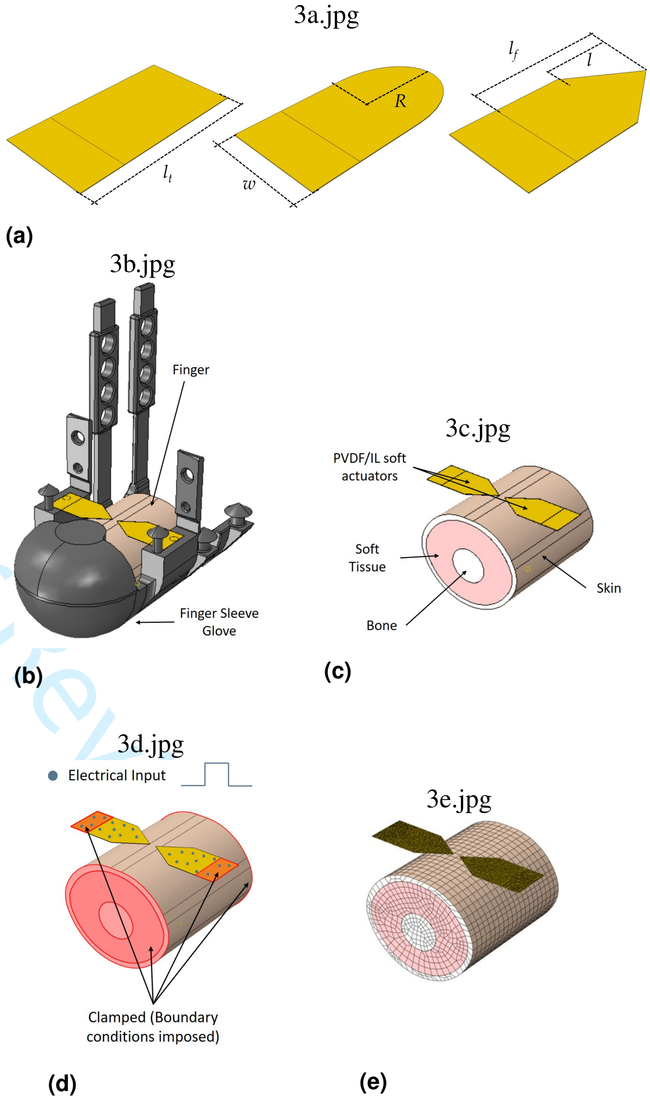


Figure 3. (a) Tip shapes tested: rectangular (left), circular (centre) and triangular (right). (b-e) Numerical FE models of the wearable haptic device in Abaqus: (b) schematic illustration of the finger sleeve prototype for the user study case; (c) FE model simulated in Abaqus, composed by the finger and the PVDF/IL soft actuators; (d) Electrical input and boundary conditions applied; (e) mesh applied to the numerical model.

electrical input intensity on the mechanical deformation of the samples.

In addition, the geometry of the contact surface also had influence on the touch sensation provided by the finger sleeve prototype, in particular, by both PVDF/IL soft actuators. Independently of the input voltage applied, smaller contact

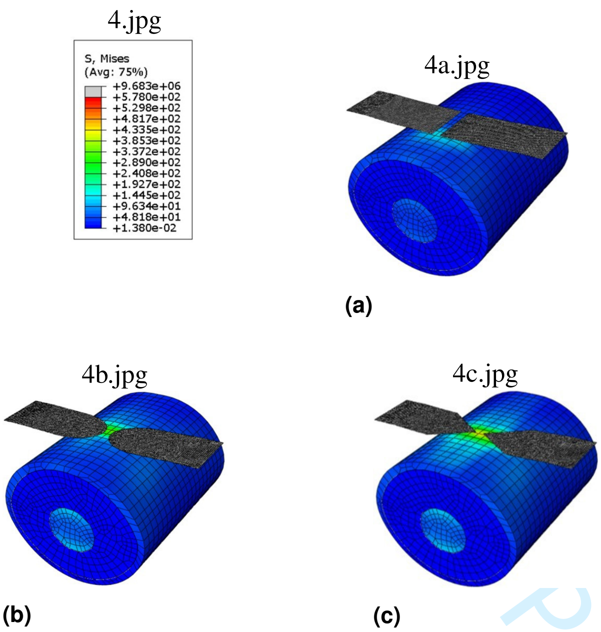


Figure 4. Numerical simulation considering the maximum voltage intensity (10 V) for all tip geometries tested. (a) Rectangular tip geometry; (b) Circular tip geometry; (c) Triangular tip geometry.

areas showed higher pressure contact. From the different geometrical tips tested, the rectangular tip resulted in the smallest touch sensation. As the contact areas decreased, when the actuators were changed from rectangular to triangular (rectangular $\Rightarrow$ circular $\Rightarrow$ triangular), the contact pressure increased. This fact can be easily explained considering the mathematical equation for pressure, $p = F/A$, where p means the contact pressure, F is the force provided by the samples and A is the contact area. When similar values of force are applied to a perpendicular area or surface, the resultant pressure is inversely proportional to the contact area between the two surfaces.

The results of the tests simulated are shown in Figure 5 and Table 2.

Table 2. Contact pressure of each tip shape tested, considering different input voltage intensities.

	Contact Pressure (Pa)		
Input Voltage Intensity (V)	Rectangular Tip	Circular Tip	Triangular Tip
2.5	0	0	0
5.0	8.0	33.7	51.2
7.5	35.1	274.8	313.1
10.0	62.1	509.7	578

The perception of touch can change considering different factors, such as age, gender, body region touched and health conditions **??**. However, in literature, Meissners corpuscles, responsible for transmitting the sensations of

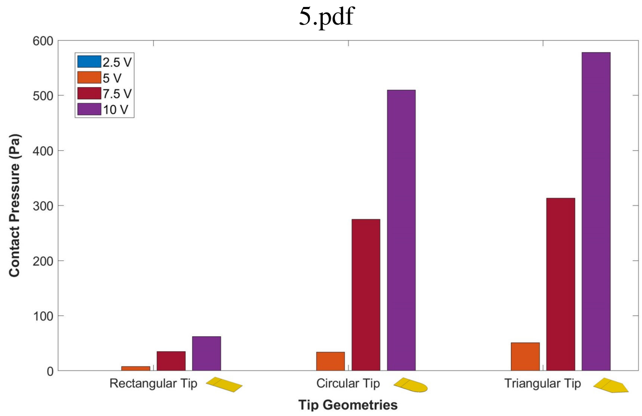


Figure 5. Numerical results considering the initial conditions tested in the *in-silico* user studies case, such as the tip geometry and the voltage intensity. For 2.5 V, the pressure was 0 Pa for all tip geometries.

fine, discriminative touch and vibration, as well as allowing Braille reading in blind people, were reported as having a minimum sensitivity to skin indentation (displacement) around 10 μm **??**. In the conditions of this study, the simultaneous actuation of two PVDF/IL soft actuators on the skin of an index finger resulted in different indentation results for different tip geometries, similar to what happened with the touch sensation (contact pressure) study. Considering the maximum voltage intensity, 10 V, the actuators with rectangular tip geometry indented the skin 0.80 μm, while circular and triangular tip geometries of the actuators indented 3.19 and 2.75 μm, respectively. Although quite similar, the actuators with circular tip geometry indented slightly more the finger skin than the actuators with triangular tip geometries, in contrast to the contact pressure. These values might be justified by the amount of piezoelectric material available to be actuated near the free tip, in this case, higher for the circular tip than the triangular tip.

Considering the triangular tip geometry as an example, the results for skin indentation change according to the pronunciation of the vertices. Figure (6) illustrates the results obtained for different triangular tips. There are minimum and maximum threshold geometries, in which the best results are achieved considering a particular tip shape.

When comparing the best result obtained considering this FEM study with the minimum value reported in the literature, it is around three times lower. This difference might be justified by the lower number of actuators used in the *in-silico* model of the device since it is expected to have higher skin indentation when more actuators are used over the same area. Finally, although a preliminary study about tip geometries was made, the optimised shape of the soft actuator might not have been achieved, which could also justify the results obtained.

Conclusions and Future Works

In conclusion, the present *in-silico* user study case allowed the authors to define some important guidelines for the development of a wearable haptic device to be used in re-educational scenarios.

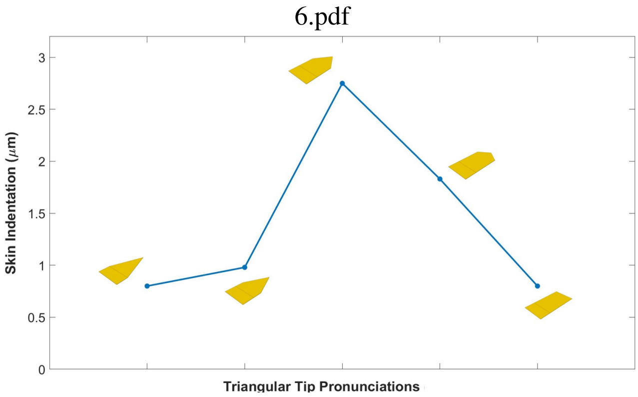


Figure 6. Numerical results considering different triangular tip pronunciations, from totally pronounced triangular tip geometry to rectangular tip geometry, respectively 0.80 μm, 0.98 μm, 2.75 μm, 1.83 μm and 0.80 μm of skin indentation.

The numerical study of different tip geometries highlighted the importance of the actuator's shape. As observed, the tendency for better results in terms of contact pressure and skin indentation is strongly related to the contact area between the actuator and the user. The feedback response for triangular tip geometries was the most promising, followed by circular and rectangular shapes. Nevertheless, when different triangular tip geometries were tested, the numerical model showed an optimal shape for achieving the best results, excluding the most extreme triangular shape. Moreover, the *in-silico* study also evidenced the importance of the input voltage intensity in the results. Higher input values lead to better feedback responses. However, it is important to be aware of the limitations of numerical studies such as the current one.

First of all, characterising piezoelectric materials in a proper way to be tested in *in-silico* conditions is extremely difficult, since some properties, such as the piezoelectric matrix, are not easily obtained in literature or experimentally. The piezoelectric matrix used in this study, despite limited, gives numerical displacement results similar to the experimental displacements observed on the samples' tips.

However, in the authors' opinion, the results reported are quite promising and point out in the right direction towards the future viability of haptic technologies in rehabilitative therapies, despite more developments being needed. In future works, it is important to test the wearable haptic prototype in real scenarios, through user study cases with volunteers. Only with the people's feedback is possible to have some reliable data on the effectiveness of the device, in terms of pleasantness and touch sensation.

With this study, the authors hope to have given a step forward in the analysis and characterisation of the applicability of piezo soft materials, such as PVDF-based materials, in therapeutic rehabilitative approaches.

Author Contribution

Conceptualisation, A.D.A, P.M.; Formal analysis, A.D.A; Investigation, A.D.A; Methodology, A.D.A, P.M.; Software, A.D.A, M.P.; Supervision, P.M., M.P.; Writing - original draft, A.D.A.; Writing - review & editing, P.M., M.P. All authors have read and agreed to the published version of the manuscript.

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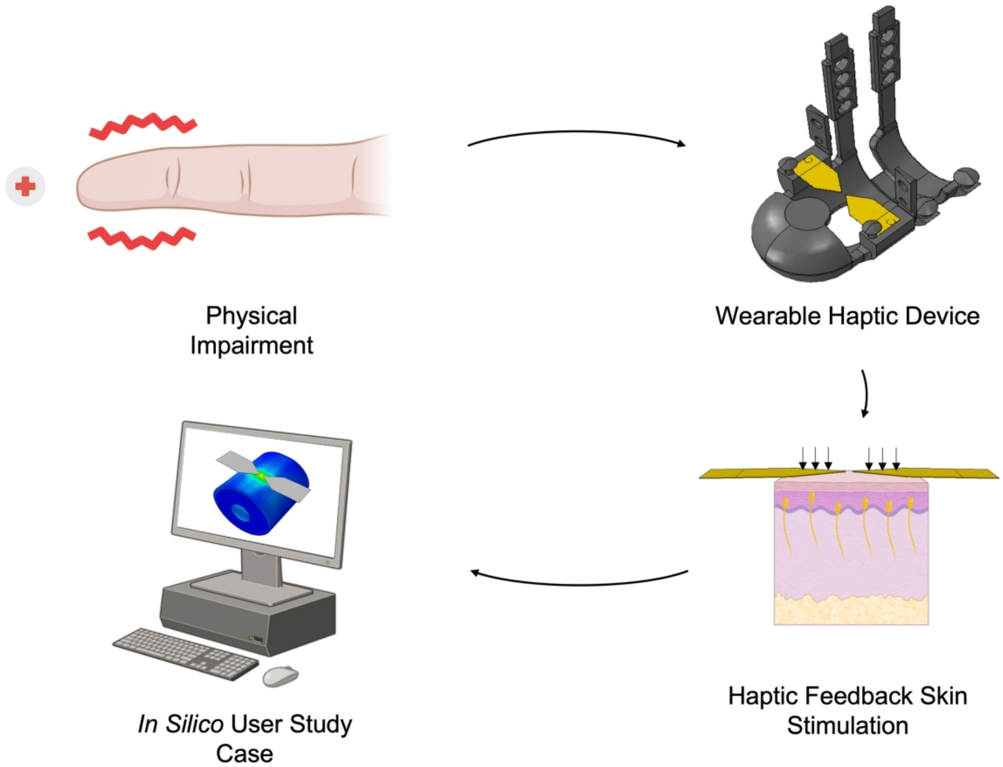
Conflict of Interest

The authors state that they have no financial, professional or other personal involvement in any product, service and/or company that would possibly affect their stance.



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Abstract

Soft smart materials have useful properties for addressing everyday problems affecting human health and well-being, having a positive societal impact. For instance, these materials can serve as sensors for breath monitoring or as soft actuators to stimulate muscles impaired by injury or illness. A notable example of their versatility lies in piezoelectric materials, which can function both as passive elements (utilising the direct piezoelectric effect) and as active elements (employing the converse piezoelectric effect). This dual functionality showcases the broad potential of smart materials in various applications. The present study is an *in silico* simulation of a wearable piezoelectric material (polyvinylidene fluoride - PVDF), using finite element analysis (FEA) to evaluate the effectiveness of the touch sensation provided by the haptic device on human skin, using different actuators geometries and voltage input intensities. Moreover, the main active element, a PVDF-based soft actuator, was fully characterised in terms of the piezoelectric matrix, using an inverse finite element approach. In conclusion, the findings point to promising results when using this haptic technology for re-educational therapies.

Keywords

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Introduction

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During the last decades, the medical and research community have been trying to mitigate this problem and accelerate the recovery of those who are suffering from lack of mobility (3). As a result, society's efforts have led to a trend of developing rehabilitative wearable devices, with the possibility to assist without tethering the patient to a specific location. However, when based on soft materials, those devices evidence limited ability to perform motion tasks (2), being essentially developed for haptic stimuli solutions on human skin.

Skin is the largest organ of the body with different functions associated. It is the first barrier against germs, helps to regulate the body temperature and, most importantly, provides the main and the first sensory channel of the body to perceive external stimuli, based on touch.

Through the skin, people can interact with the surrounding objects, feeling their properties, such as weight, temperature, textures and motion (4). The interaction between skin and objects results in nerve stimuli, which are primarily received by different receptors in the skin and later sent to the brain. Pacinian corpuscles, Meissner corpuscles, Merkel complexes, Ruffini corpuscles, and C-fiber LTM (low threshold mechanoreceptors) are the mechano-receptors that can detect even innocuous stimuli (5), for example, responding to displacements of the skin on the order of micrometres (μm). Different stimuli, such as mechanical, electrical and thermal can feedback those receptors and transmit different sensations and information through the nerves (4).

To take advantage of the high sensitivity of skin, different technologies have been developed, using different powered-up soft materials, for rehabilitation and re-educational therapies through the use of external devices. As reviewed by André et al (6), different soft actuators can be used to provide mechanical, electrical or thermal feedback on the skin, such as electroactive materials (piezoelectric), magnetic responsive materials, thermally responsive materials or even photo-responsive materials. All these smart materials, in particular piezo polymers due to their versatility, can play an important role in actuation for tactile sensation since they are light, flexible and biomimetic.

Piezoelectric materials, in particular polymers such as polyvinylidene fluoride (PVDF), have been largely studied in rehabilitation solutions since they can be used as sensors (7; 8) or actuators (9; 10). For example, Pan *et*

[1]Associated Laboratory of Energy, Transports and Aeronautics (LAETA), Biomechanic and Health Unity (UBS), Institute of Science and Innovation in Mechanical and Industrial Engineering (INEGI). Campus da FEUP, Rua Dr. Roberto Frias, 400, 4200-465, Porto, Portugal
[2]Faculty of Engineering of University of Porto (FEUP). Rua Dr. Roberto Frias, s/n, 4200-465, Porto, Portugal
[3]i3A, Universidad de Zaragoza, Zaragaoza. C. de Mariano Esquillor Gómez, s/n, 50018, Zaragoza, Spain

Corresponding author:
Pedro Martins
Email: palsm@fe.up.pt

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al. (11) developed a mechanomyography (MMG) sensor using PVDF's piezoelectric properties for lower limb rehabilitation exoskeleton. They concluded that the approach used improved the sensitivity in driving the device. Gariya *et al.* (12), developed a pneumatic soft actuator with a PVDF membrane for sensing the bending deformation of the actuator, which was tested for the medical application of human finger rehabilitation.

Nevertheless, the use of PVDF, as a haptic feedback element, has been presented in the literature with distinct purposes. Ege and Balikci (13) introduced a haptic interface using transparent thin films of PVDF actuators to feedback touch displays. Maeda *et al.* (14) developed a wearable haptic augmentation system using a skin vibration sensor made from PVDF. In their studies, they pointed to an increase in haptic sensation of, approximately, 5% when compared to no feedback condition.

The present study aims to perform an *in-silico* user study, as proof of concept using the finite element method (FEM), of a wearable device for re-educational therapies, using PVDF/ionic liquid (IL) as haptic actuators. These feedback elements, with distinct tip geometries (rectangular, circular and triangular), will be fed with different voltage intensities, to study the influence of input voltage on the touch perception. With this approach, the authors hope to understand if the developed methodology can be effectively applied in rehabilitation scenarios and if the skin can be stimulated through gentile touch sensations provided by PVDF/IL's actuators. This work uses the index finger as a reference model, due to its high sensibility (15).

Inverse Finite Element Analysis

André et al (16; 17) characterised PVDF/IL haptic actuators experimentally, in terms of porosity and microscopic defects (high resolution scans were obtained); mechanical properties (i.e. Young modulus (E) and yield stress (σ_{Yield})); chemical properties (i.e. degree of crystallinity (χ) and electroactive phases (β phase)); and electrical, piezo and conductive properties (i.e. dielectric permittivity (ϵ'), AC conductivity (σ_{AC}) and piezoelectric constant (d_{33})). However, those properties are not sufficient to define a material model able to reproduce the behaviour observed experimentally, namely the electromechanical performance (17).

To better mimic the experimental phenomenon observed, an inverse finite element analysis approach was used to evaluate the piezoelectric matrix.

Piezoelectric Constitutive Equations

The standard form of the piezoelectric constitutive equations can be presented in four different forms by taking either two of the four field variables as independent. Considering a tensorial representation of the strain-electric displacement form and the components of stress and electric fields as the independent variables (18; 19; 20), it comes

$$S_{ij} = s^E_{ijkl} \cdot T_{kl} + d_{kij} \cdot E_k \tag{1}$$

$$D_i = d_{ikl} \cdot T_{kl} + \epsilon^T_{ik} \cdot E_k \tag{2}$$

where the index i, j, k, l represent the different components or directions in the tensor equations, S_{ij} are the strain components, D_i are the electric displacements, s^E_{ijkl} are the elastic compliance constants, T_{kl} are the stress components, d_{kij} are the piezoelectric constants of the material, ϵ^T_{ik} are the permittivity constants and finally E_k is the electric field component.

In a matrix form, the equations (1) and (2) are given as (21),

$$\begin{bmatrix} S \\ D \end{bmatrix} = \begin{bmatrix} s^E & d^t \\ d & \epsilon^T \end{bmatrix} \cdot \begin{bmatrix} T \\ E \end{bmatrix} \tag{3}$$

where the superscripts E and T denote that the respective constants are evaluated at constant electric field and constant stress, respectively, and t represents the transpose.

Assuming the study of a piezoelectric thin film, such as the case under analysis, matrix 3 can be further simplified. If the thin structure is assumed as a thin beam, based on the Euler-Bernoulli beam theory or Rayleigh beam theory (21), the stress components T_{22}, T_{33}, T_{23}, T_{13} and T_{12} are negligible ($T_{\neq 11} = 0$), since only the one-dimensional bending stress (T_{11}) has a non negligible value. Moreover, if the electrodes are placed perpendicular to the 3-direction, equation (3) becomes

$$\begin{bmatrix} S_{11} \\ D_{33} \end{bmatrix} = \begin{bmatrix} s^E_{11} & d_{31} \\ d_{31} & \epsilon^T_{33} \end{bmatrix} \cdot \begin{bmatrix} T_{11} \\ E_{33} \end{bmatrix} \tag{4}$$

Matrix (4) shows that the piezoelectric constant, d_{31}, is crucial and also needed to characterise properly the piezo polymer PVDF/IL thin film.

Piezoelectric Matrix

Overall, the piezoelectric matrix of PVDF is given by equation (5), since the material evidences anisotropy (22; 23). That fact implies that the piezoelectric properties change with direction (22).

$$d_{ij} = \begin{bmatrix} 0 & 0 & 0 & 0 & d_{15} & 0 \\ 0 & 0 & 0 & d_{24} & 0 & 0 \\ d_{31} & d_{32} & d_{33} & 0 & 0 & 0 \end{bmatrix} \tag{5}$$

where the subscript i refers to the direction of plane polarisation, while j is the direction of the induced strain.

However, in the literature, the piezoelectric properties of PVDF more commonly used to characterise the polymeric material are d_{31} and d_{33} (24; 25). The d_{31} constant is the transverse coefficient, which defines the mechanical strain created in the perpendicular direction to the applied electric input; while d_{33} is the longitudinal coefficient, which defines mechanical strain in the same direction as the applied stress (26). With a good approximation, the two coefficients are enough to define the phenomenological behaviour observed experimentally. Considering that fact, for this particular study case, matrix (5) could be simplified into matrix (6), ignoring d_{32} coefficient and the rotational and shear components (d_{24} and d_{15}).

$$d_{ij} = \begin{bmatrix} 0 & 0 & 0 \\ 0 & 0 & 0 \\ d_{31} & 0 & d_{33} \end{bmatrix} \tag{6}$$

Inverse-FEM Algorithm

In recent years, the identification and characterisation of the piezoelectric matrix using inverse FEM algorithms have grown and gained popularity (27). Figure (1a) summarises the steps adopted to approximate numerically the piezoelectric constant d_{31}.

In general, the optimisation algorithm runs until achieving a d_{31} value that grants similar numerical behaviour when compared to the experimental behaviour. The experimental displacement (Displ_{exp}) measured at the tip of the PVDF/IL sample, when electrically stimulated with a square pulse of 10 V for 10 seconds, was 1.86 mm (17). The optimisation algorithm optimised the initial d_{31} parameter guess (15x10^{-12} pm/V (28)) until the exit optimisation criteria is satisfied: numerical displacement (Displ_{num}) is equal to the Displ_{exp} more or less the error (Err = 15%) (Figure 1b).

Applying an iterative refinement method (29), and after 8 iterations, the optimised d_{31} was 3.75x10^{-07} pm/V, with a Displ_{num} of 1.87 mm (error of 0.8%).

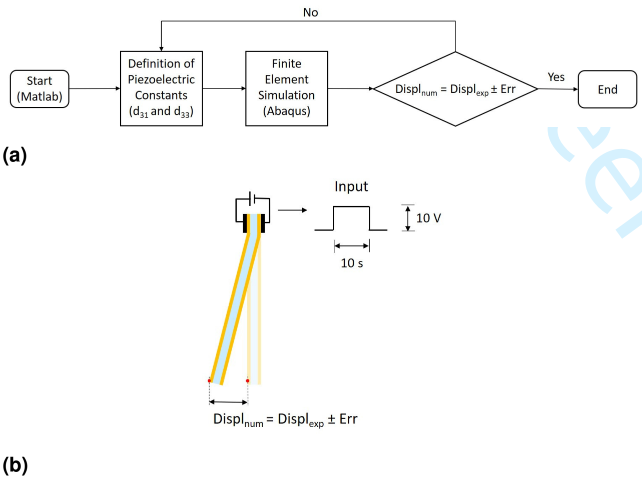

Figure 1. (a) Flow chart for d_{31} optimisation; (b) Exit optimisation criteria when Displ_{num} = $\mathrm{Displ}_{exp} \pm$ Err.

In-Silico User Study Case

A numerical model of the wearable haptic device was developed to simulate the stimuli contact and interaction between an index finger and the device, through gentile and soft haptic touch sensations.

The finger sleeve prototype used was first designed by André et al (17). Briefly, the finger sleeve prototype was developed considering the usage of a thermoplastic polyurethane (TPU) as the main structural material. This choice grants the finger device with flexibility and high elasticity. Moreover, TPU can be used as a 3D printable material in fused filament fabrication (FFF) technique, which allows fast prototype iterations. All these properties make this material the ideal candidate for its purpose.

The prototype allows adaptability to patients' physiognomy since it has adjustable straps. As a result, the device can be adapted to different finger shapes (Figure 2).

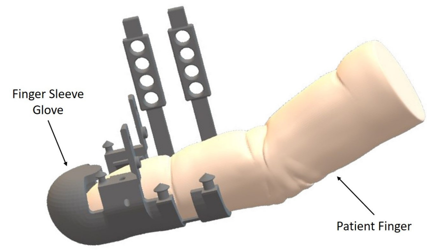

Figure 2. Schematic illustration of the wearable device prototype (finger sleeve) worn by an index finger.

As the PVDF/IL haptic actuators were previously characterised (16; 17; 30), as well as the finger sleeve was properly designed (17), an *in-silico* numerical model was developed on Dassault Systemès Abaqus (31) to study the device's behaviour in real scenarios. Afterwards, real interaction scenarios between the user and the haptic feedback device were simulated using a FEM approach.

FEM Model

The numerical model was defined in Abaqus 2022 (31). The elements needed to perform the *in-silico* user study case were defined, i.e., the mechanical properties, interactions between parts, the boundary conditions (BCs) and the mesh.

The physiognomy of the finger model was based on literature (32; 33; 34) and drawn in Dassault Systemès SolidWorks 2020 (35). It is composed of bone, soft tissue and skin with corresponding properties based on the literature (32; 36; 37; 38). The wearable haptic device was modelled considering PVDF/IL soft actuators (with 6 mm width, w, 12 mm total length, l_t, 9 mm free length, l_f, and 0.06 mm thickness, t) and a TPU finger sleeve. The properties of those materials were also based on the literature (16; 17; 39; 30). Different tip shapes for the actuation were considered, as shown in Figure 3a, such as rectangular, circular (with radius, R, of 4.3 mm centred at the middle of the free length) and triangular (with different tip pronunciations, started at length l=0, l=1/4, l=2/4, l=3/4 and finally l=4/4 from the free tip). Table 1 summarises the properties of the parts used to simulate the *in-silico* user study case.

The FEM model was prepared and assembled in Abaqus, considering the wearable haptic device and the finger model (Figures 3b and 3c). Since the finger sleeve glove is not an active part of the simulation, it was not considered, resulting in a faster numerical simulation. Moreover, the contact and interaction properties, as well as the node ties were defined among all the active parts (PVDF/IL actuators ⇔ skin - contact/interaction; skin ⇔ soft tissues - tie; soft tissues ⇔ bone). The BCs were applied in the anterior and posterior sections of the finger model and on both PVDF/IL actuators. Since the finger sleeve glove is not present in the simulation, the BCs applied on the actuators satisfy the interaction with the glove. The electrical inputs were also applied in both PVDF/IL soft actuators, assuming a square pulse with different voltage intensities with a time duration of 10 seconds (Figure 3d). The active elements/parts of the simulation were then meshed. The finger model was meshed considering 3D stress hexagonal mesh elements (C3D8R - 8 node linear brick, reduced integration). On

Table 1. Properties of the parts used in the numerical model in Abaqus.

Parts	Density, ρ (kg/m^3)	Young Modulus, E (MPa)	Poisson Ratio, ν (adimensional)	Elect. Conductivity, σ_{AC} (S/m)	Piezoelectric Const., d_{33} (pm/V)	Piezoelectric Const., d_{31} (pm/V)
Bone (36)	1900	$17\text{x}10^3$	0.3	-	-	-
Soft Tissue (32)	1000	0.08	0.4	-	-	-
Skin (37; 38)	-	2.5	0.48	-	-	-
TPU finger sleeve (39)	-	$2.41\text{x}10^3$	0.3847	-	-	-
PVDF/IL soft actuators (16; 17; 30)	1425.9	127.0	0.18	$1.98\text{x}10^{-07}$	0.6	$3.75\text{x}10^5$

another hand, the PVDF/IL soft actuators were defined using piezoelectric tetrahedral mesh elements (C3D4E - 4 node linear piezoelectric brick). The model's number of elements and nodes changed according to the tip geometry (rectangular, 15303 nodes and 21728 elements; circular, 18225 nodes and 30958 elements; or triangular, 16913 nodes and 26673 elements). Figure 3e shows the mesh applied to a specific tip geometry situation.

Results and Discussion

The *in-silico* user study case was carried out to understand the effectiveness of the touch sensation provided by the finger sleeve prototype on the skin to be used in the context of re-educational therapy. Both PVDF/IL soft haptic actuators were input simultaneously under different initial conditions, such as the tip geometries (rectangular, circular and triangular; initiated at the middle of the free part of the actuator) and pulse intensities (2.5, 5.0, 7.5 and 10.0 V). The numerical results achieved were, then, compared against the minimum touch sensation felt by human fingers described in the literature.

FEM Simulation

The FEM simulation of the *in-silico* user study case, considering different voltage intensities and tip geometries, evidenced results according to our expectations. Figure 4 shows the results for the different tip geometries tested, considering the maximum voltage intensity (10.0 V). The best touch sensation provided by the finger sleeve prototype was achieved for the triangular tip geometry of the PVDF/IL soft actuators. According to the simulation, that value was 578.0 Pa on the skin finger model (Figure 4c).

The results showed that the touch sensation provided by PVDF/IL soft actuators increased with the voltage intensity. For the minimum voltage intensity tested, 2.5 V, the simulation did not show any contact pressure for all the different situations. The relation between the contact pressure and the voltage intensity is justified by the proportionality of the electrical input applied and the consequent mechanical deformation induced. It means that when higher voltage intensities are applied to the samples, higher polarization is induced on them, which increases the outputs, such as the contact pressure (40) and displacement (17). The same cause/effect is observed in other studies available in the literature. Raza et al (9) studied micro-structured porous electrolytes for highly responsive ionic soft actuators, using PVDF-co-hexafluoropropylene (PVDF-co-HFP) as the main active element. They observed a direct influence of the electrical input intensity on the mechanical deformation of the samples.

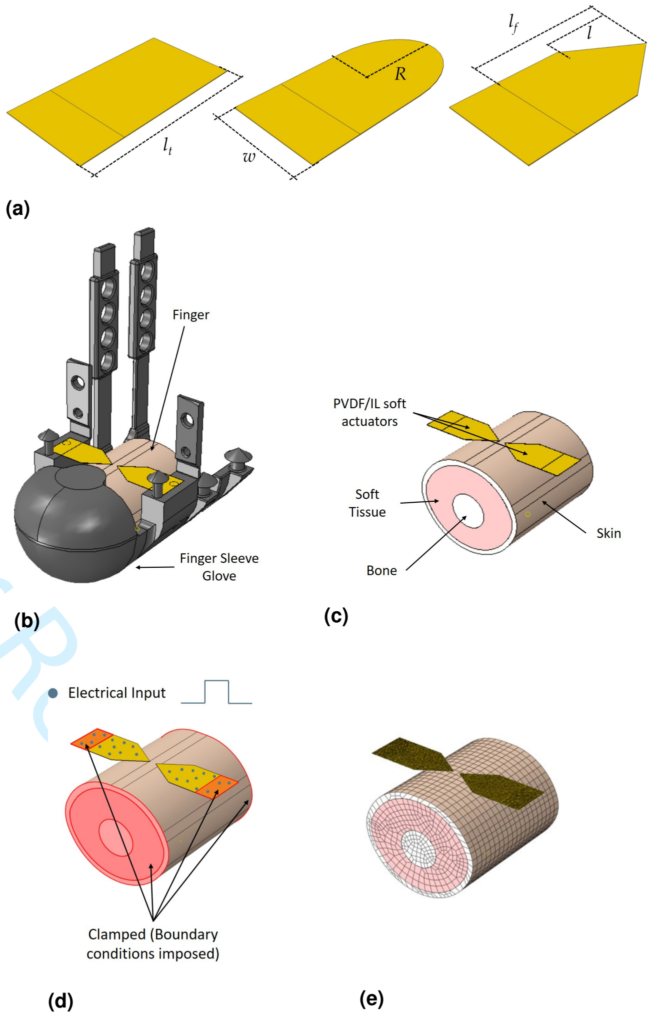

Figure 3. (a) Tip shapes tested: rectangular (left), circular (centre) and triangular (right). (b-e) Numerical FE models of the wearable haptic device in Abaqus: (b) schematic illustration of the finger sleeve prototype for the user study case; (c) FE model simulated in Abaqus, composed by the finger and the PVDF/IL soft actuators; (d) Electrical input and boundary conditions applied; (e) mesh applied to the numerical model.

In addition, the geometry of the contact surface also had influence on the touch sensation provided by the finger sleeve prototype, in particular, by both PVDF/IL soft actuators. Independently of the input voltage applied, smaller contact areas showed higher pressure contact. From the different geometrical tips tested, the rectangular tip resulted in the smallest touch sensation. As the contact areas decreased,

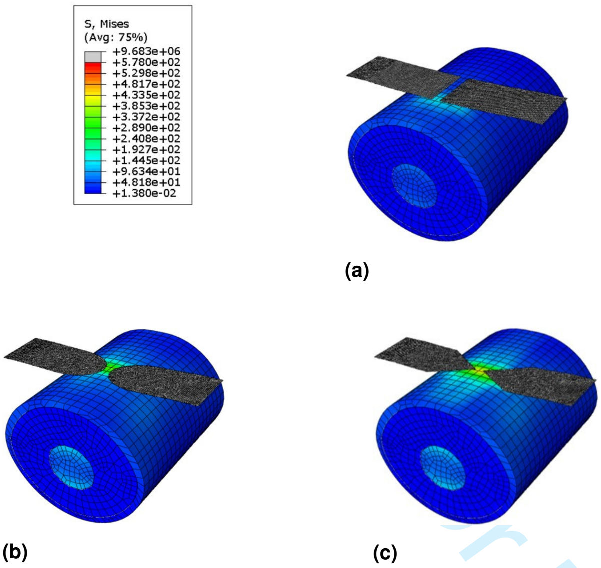

Figure 4. Numerical simulation considering the maximum voltage intensity (10 V) for all tip geometries tested. (a) Rectangular tip geometry; (b) Circular tip geometry; (c) Triangular tip geometry.

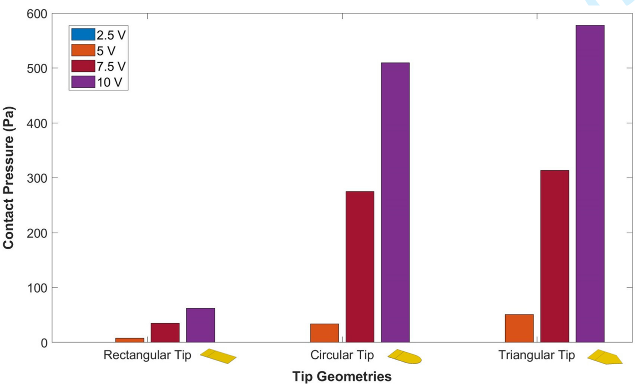

Figure 5. Numerical results considering the initial conditions tested in the *in-silico* user studies case, such as the tip geometry and the voltage intensity. For 2.5 V, the pressure was 0 Pa for all tip geometries.

when the actuators were changed from rectangular to triangular (rectangular ⇒ circular ⇒ triangular), the contact pressure increased. This fact can be easily explained considering the mathematical equation for pressure, $p = F/A$, where p means the contact pressure, F is the force provided by the samples and A is the contact area. When similar values of force are applied to a perpendicular area or surface, the resultant pressure is inversely proportional to the contact area between the two surfaces.

The results of the tests simulated are shown in Figure 5 and Table 2.

The perception of touch can change considering different factors, such as age, gender, body region touched and health conditions (41; 42). However, in literature, Meissner's

Table 2. Contact pressure of each tip shape tested, considering different input voltage intensities.

	Contact Pressure (Pa)		
Input Voltage Intensity (V)	Rectangular Tip	Circular Tip	Triangular Tip
2.5	0	0	0
5.0	8.0	33.7	51.2
7.5	35.1	274.8	313.1
10.0	62.1	509.7	578

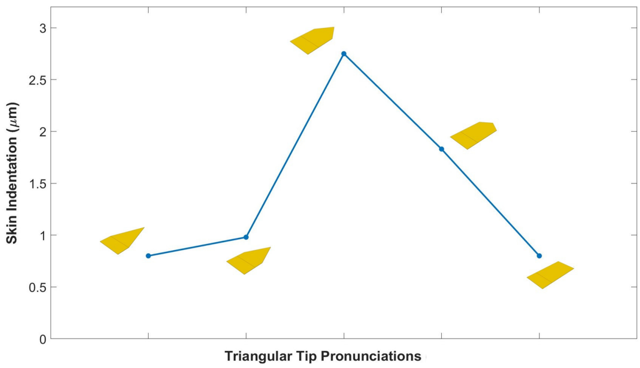

Figure 6. Numerical results considering different triangular tip pronunciations, from totally pronounced triangular tip geometry to rectangular tip geometry, respectively 0.80 μm, 0.98 μm, 2.75 μm, 1.83 μm and 0.80 μm of skin indentation.

corpuscles, responsible for transmitting the sensations of fine, discriminative touch and vibration, as well as allowing Braille reading in blind people, were reported as having a minimum sensitivity to skin indentation (displacement) around 10 μm (43; 44). In the conditions of this study, the simultaneous actuation of two PVDF/IL soft actuators on the skin of an index finger resulted in different indentation results for different tip geometries, similar to what happened with the touch sensation (contact pressure) study. Considering the maximum voltage intensity, 10 V, the actuators with rectangular tip geometry indented the skin 0.80 μm, while circular and triangular tip geometries of the actuators indented 3.19 and 2.75 μm, respectively. Although quite similar, the actuators with circular tip geometry indented slightly more the finger skin than the actuators with triangular tip geometries, in contrast to the contact pressure. These values might be justified by the amount of piezoelectric material available to be actuated near the free tip, in this case, higher for the circular tip than the triangular tip.

Considering the triangular tip geometry as an example, the results for skin indentation change according to the pronunciation of the vertices. Figure (6) illustrates the results obtained for different triangular tips. There are minimum and maximum threshold geometries, in which the best results are achieved considering a particular tip shape.

When comparing the best result obtained considering this FEM study with the minimum value reported in the literature, it is around three times lower. This difference might be justified by the lower number of actuators used in the *in-silico* model of the device since it is expected to have higher skin indentation when more actuators are used over the same area. Finally, although a preliminary study about tip geometries was made, the optimised shape of the soft actuator might not have been achieved, which could also justify the results obtained.

Conclusions and Future Works

In conclusion, the present *in-silico* user study case allowed the authors to define some important guidelines for the development of a wearable haptic device to be used in re-educational scenarios.

The numerical study of different tip geometries highlighted the importance of the actuator's shape. As observed, the tendency for better results in terms of contact pressure and skin indentation is strongly related to the contact area between the actuator and the user. The feedback response for triangular tip geometries was the most promising, followed by circular and rectangular shapes. Nevertheless, when different triangular tip geometries were tested, the numerical model showed an optimal shape for achieving the best results, excluding the most extreme triangular shape. Moreover, the *in-silico* study also evidenced the importance of the input voltage intensity in the results. Higher input values lead to better feedback responses. However, it is important to be aware of the limitations of numerical studies such as the current one.

First of all, characterising piezoelectric materials in a proper way to be tested in *in-silico* conditions is extremely difficult, since some properties, such as the piezoelectric matrix, are not easily obtained in literature or experimentally. The piezoelectric matrix used in this study, despite limited, gives numerical displacement results similar to the experimental displacements observed on the samples' tips.

However, in the authors' opinion, the results reported are quite promising and point out in the right direction towards the future viability of haptic technologies in rehabilitative therapies, despite more developments being needed. In future works, it is important to test the wearable haptic prototype in real scenarios, through user study cases with volunteers. Only with the people's feedback is possible to have some reliable data on the effectiveness of the device, in terms of pleasantness and touch sensation.

With this study, the authors hope to have given a step forward in the analysis and characterisation of the applicability of piezo soft materials, such as PVDF-based materials, in therapeutic rehabilitative approaches.

Author Contribution

Conceptualisation, A.D.A, P.M.; Formal analysis, A.D.A; Investigation, A.D.A; Methodology, A.D.A, P.M.; Software, A.D.A, M.P.; Supervision, P.M., M.P.; Writing - original draft, A.D.A.; Writing - review & editing, P.M., M.P. All authors have read and agreed to the published version of the manuscript.

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Conflict of Interest

The authors state that they have no financial, professional or other personal involvement in any product, service and/or company that would possibly affect their stance.

References

[1] Organisation WH. *Global report on health equity for persons with disabilities - Executive summary*. World Health Organisation, 2022. ISBN 978-92-4-006363-1.

[2] Desplenter T, Zhou Y, Edmonds BP et al. Rehabilitative and assistive wearable mechatronic upper-limb devices: A review. *Journal of Rehabilitation and Assistive Technologies Engineering* 2020; 7: 205566832091787. DOI:10.1177/2055668320917870.

[3] Mavroidis C, Nikitczuk J, Weinberg B et al. Smart portable rehabilitation devices. In *Volume 7: 29th Mechanisms and Robotics Conference, Parts A and B*. IDETC-CIE2005, ASMEDC. DOI:10.1115/detc2005-85517.

[4] Xie X, Liu S, Yang C et al. A review of smart materials in tactile actuators for information delivery. *C* 2017; 3(4): 38. DOI:10.3390/c3040038.

[5] Marzvanyan A and Alhawaj AF. *Physiology, Sensory Receptors*. StatPearls Publishing, Treasure Island (FL), 2023. URL http://europepmc.org/books/NBK539861.

[6] André AD and Martins P. Exo supportive devices: Summary of technical aspects. *Bioengineering* 2023; 10(11): 1328. DOI:10.3390/bioengineering10111328.

[7] Hu Y, Kang W, Fang Y et al. Piezoelectric poly(vinylidene fluoride) (pvdf) polymer-based sensor for wrist motion signal detection. *Applied Sciences* 2018; 8(5): 836. DOI:10.3390/app8050836.

[8] Zhang J, Yao H, Mo J et al. Finger-inspired rigid-soft hybrid tactile sensor with superior sensitivity at high frequency. *Nature Communications* 2022; 13(1). DOI:10.1038/s41467-022-32827-7.

[9] Raza U, Oh S, Tabassian R et al. Micro-structured porous electrolytes for highly responsive ionic soft actuators. *Sensors and Actuators B: Chemical* 2022; 352: 131006. DOI:10.1016/j.snb.2021.131006.

[10] Fook THT, Jeon JH and Lee PS. Transparent flexible polymer actuator with enhanced output force enabled by conductive nanowires interlayer. *Advanced Materials Technologies* 2019; 5(1). DOI:10.1002/admt.201900762.

[11] Pan CT, Chang CC, Yang YS et al. Development of mmg sensors using pvdf piezoelectric electrospinning for lower limb rehabilitation exoskeleton. *Sensors and Actuators A: Physical* 2020; 301: 111708. DOI:10.1016/j.sna.2019.111708.

[12] Gariya N, Kumar P, Prasad B et al. Soft pneumatic actuator with an embedded flexible polymeric piezoelectric membrane for sensing bending deformation. *Materials Today Communications* 2023; 35: 105910. DOI:10.1016/j.mtcomm.2023.105910.

[13] Ege ES and Balikci A. Transparent localized haptics: Utilization of pvdf actuators on touch displays. *Actuators* 2023; 12(7): 289. DOI:10.3390/act12070289.

[14] Maeda T, Peiris R, Nakatani M et al. Wearable haptic augmentation system using skin vibration sensor. In *Proceedings of the 2016 Virtual Reality International Conference*. VRIC '16, ACM. DOI:10.1145/2927929. 2927946.

[15] Jarocka E, Pruszynski JA and Johansson RS. Human touch receptors are sensitive to spatial details on the scale of single fingerprint ridges. *The Journal of Neuroscience* 2021; 41(16): 3622–3634. DOI:10.1523/jneurosci.1716-20.2021.

[16] André AD, Teixeira AM and Martins P. Influence of dmso non-toxic solvent on the mechanical and chemical properties of a pvdf thin film. *Applied Sciences* 2024; 14(8). DOI: 10.3390/app14083356. URL https://www.mdpi.com/2076-3417/14/8/3356.

[17] André AD, Coondoo I, Bdikin I et al. Piezo-ionic actuator for haptic feedback ; Submitted/Unpublished.

[18] 176-1987 ieee standard on piezoelectricity. Electronic resource type: Report.

[19] Serrao PH and Kozinov S. Robust mixed fe for analyses of higher-order electromechanical coupling in piezoelectric solids. *Computational Mechanics* 2023; DOI:10.1007/s00466-023-02407-7.

[20] Schoeftner J and Gahleitner J. Approximate analytical solutions for piezoelectric rectangular beams by using boley-tolins method. *Journal of Physics Communications* 2021; 5(10): 105015. DOI:10.1088/2399-6528/ac2c32.

[21] Erturk A. *Piezoelectric energy harvesting*. Chichester, West Sussex, U.K: Wiley, 2011. ISBN 9780470682548.

[22] Kalimuldina G, Turdakyn N, Abay I et al. A review of piezoelectric pvdf film by electrospinning and its applications. *Sensors* 2020; 20(18): 5214. DOI:10.3390/s20185214.

[23] Mohammadpourfazeli S, Arash S, Ansari A et al. Future prospects and recent developments of polyvinylidene fluoride (pvdf) piezoelectric polymer; fabrication methods, structure, and electro-mechanical properties. *RSC Advances* 2023; 13(1): 370–387. DOI:10.1039/d2ra06774a.

[24] Wang X, Tong W, Chen Y et al. Effective mechanical energy harvesting from pvdf multilayers by head-to-head parallel assembly. *ACS Applied Energy Materials* 2021; 4(10): 11133–11143. DOI:10.1021/acsaem.1c02045.

[25] Xie L, Wang G, Jiang C et al. Properties and applications of flexible poly(vinylidene fluoride)-based piezoelectric materials. *Crystals* 2021; 11(6): 644. DOI:10.3390/cryst11060644.

[26] Sukumaran S, Chatbouri S, Rouxel D et al. Recent advances in flexible pvdf based piezoelectric polymer devices for energy harvesting applications. *Journal of Intelligent Material Systems and Structures* 2020; 32(7): 746–780. DOI:10.1177/1045389x20966058.

[27] del Castillo M and Pérez N. Machine learning identification of piezoelectric properties. *Materials* 2021; 14(9): 2405. DOI: 10.3390/ma14092405.

[28] Bystrov VS, Pullar R, Kholkin AL et al. Modeling of switching and piezoelectric phenomena in polyvinylidenefluoride (pvdf). In *2013 Joint IEEE International Symposium on Applications of Ferroelectric and Workshop on Piezoresponse Force Microscopy (ISAF/PFM)*. IEEE. DOI:10.1109/isaf.2013.6748703.

[29] Chapra SC and Canale RP. *Numerical Methods for Engineers*. McGraw-Hill Education. ISBN 9780073397924.

[30] Razavi S, Iannucci L and Greenhalgh E. Piezoelectric pvdf smart fibre for composite applications. In *Proceedings of the 17th European Conference on Composite Materials ECCM17 - 17th European Conference on Composite Materials, Munich, Germany*. ECCM.

[31] Systemès D. Dassault systemes abaqus. https://www.3ds.com/products/simulia/abaqus. Accessed: 2024-03-12.

[32] Shimawaki S and Sakai N. Quasi-static deformation analysis of a human finger using a three-dimensional finite element model constructed from ct images. *Journal of Environment and Engineering* 2007; 2(1): 56–63. DOI:10.1299/jee.2.56.

[33] Wu JZ, Welcome DE and Dong RG. Three-dimensional finite element simulations of the mechanical response of the fingertip to static and dynamic compressions. *Computer Methods in Biomechanics and Biomedical Engineering* 2006; 9(1): 55–63. DOI:10.1080/10255840600603641.

[34] Ikemura S, Endo T and Matsuno F. Multiple remote vibrotactile noises improve tactile sensitivity of the fingertip via stochastic resonance. *IEEE Access* 2021; 9: 17011–17019. DOI:10.1109/access.2021.3053297.

[35] Systemès D. Dassault systemes solidworks. https://www.solidworks.com/. Accessed: 2024-03-14.

[36] Lai YS, Chen WC, Huang CH et al. The effect of graft strength on knee laxity and graft in-situ forces after posterior cruciate ligament reconstruction. *PLOS ONE* 2015; 10(5): e0127293. DOI:10.1371/journal.pone.0127293.

[37] Li C, Guan G, Reif R et al. Determining elastic properties of skin by measuring surface waves from an impulse mechanical stimulus using phase-sensitive optical coherence tomography. *Journal of The Royal Society Interface* 2011; 9(70): 831–841. DOI:10.1098/rsif.2011.0583.

[38] Wang Q and Hayward V. In vivo biomechanics of the fingerpad skin under local tangential traction. *Journal of Biomechanics* 2007; 40(4): 851–860. DOI:10.1016/j.jbiomech.2006.03.004.

[39] Odenwald S, Schwantiz S and Krumm D. *Tagungsband zum 14. Symposium der Sektion Sportinformatik und Sporttechnologie der Deutschen Vereinigung für Sportwissenschaft*. Universitätsverlag Chemnitz. ISBN 978-3-96100-176-7.

[40] Uchino K. Introduction to piezoelectric actuators and transducers, 2003.

[41] Dargahi J and Najarian S. Human tactile perception as a standard for artificial tactile sensing—a review. *The International Journal of Medical Robotics and Computer Assisted Surgery* 2004; 1(1): 23–35. DOI:10.1002/rcs.3.

[42] Heller MA and Schiff W. *Psychology of Touch*. Taylor Francis Group, 2013. ISBN 9781315799629.

[43] Lechelt EC. Tactile spatial anisotropy with static stimulation. *Bulletin of the Psychonomic Society* 1992; 30(2): 140–142. DOI:10.3758/bf03330421.

[44] Iggo A and Ogawa H. Correlative physiological and morphological studies of rapidly adapting mechanoreceptors in cat's glabrous skin. *The Journal of Physiology* 1977; 266(2): 275–296. DOI:10.1113/jphysiol.1977.sp011768.