POLITECNICO DI TORINO

Master's Degree in Biomedical Engineering





Master's thesis

Design and fabrication of a biodegradable coil for inductive power transfer

Supervisors: Prof. Danilo DEMARCHI Prof. Diego GHEZZI Candidate: Filippo QUAGLIA 275511

Co-supervisors: Eng. Adele FANELLI Eng. Gian Luca BARBRUNI

March 2022

Abstract

Neuromodulation is a fascinating and promising way to restore a lost function or treat neurodegenerative diseases. The fundamental idea is to generate a stimulus which is capable to activate a response from the neurons. Several types of stimuli can be used to achieve neural activation and one of the most used is to deliver electrical impulses. Usually, neural interfaces (NIs) exploited for this application require invasive surgery for their implantation. This can potentially reduce the pool of eligible patients and increase risks of device failure due to strong inflammatory reaction. Moreover, the presence of cables could complicate such procedures and require further operations to replace broken or infected leads. Recent efforts to overcome these limitations suggest an alternative way to reach the neural tissue by placing the neural interface inside blood vessels adjacent to the target area, via standard and minimally invasive angioplasty procedures. In addition, the use of transient materials avoids the need for further surgeries for device removal and reduce inflammatory reaction. However, such devices still need cables to be powered. In this work, an electrical circuit for a wireless inductive power transfer (WIPT), realized exploiting transient materials which could be integrated into such devices, is exploited. The circuit is an ensemble of three elements: an inductor coil, a diode and a capacitor. In this groundwork, the main focus is paid to the design, fabrication and preliminary test of the coil inductor element. Degradation and penetration tests are reported to understand the degradation profile of the device and techniques are proposed to enhance the functioning time of the device. Finally, suggestions for materials and possible fabrication techniques of the diode element and the capacitor are reported.

Sommario

La neuromodulazione è un modo affascinante e promettente per ripristinare una funzione persa o trattare le malattie neurodegenerative. L'idea fondamentale è quella di generare uno stimolo che sia in grado di attivare una risposta dei neuroni. Diversi tipi di stimoli possono essere utilizzati per indurre l'attivazione neurale e uno dei più utilizzati è quello di fornire impulsi elettrici. Di solito, le interfacce neurali (NIs) sfruttate per questa applicazione richiedono un intervento chirurgico invasivo per il loro impianto. Ciò può potenzialmente ridurre il pool di pazienti eleggibili ed aumentare i rischi di guasto del dispositivo a causa di una forte reazione infiammatoria. Inoltre, la presenza di cavi potrebbe complicare tali procedure e richiedere ulteriori operazioni per sostituire i cavi rotti o infetti. Recenti sforzi per superare queste limitazioni suggeriscono un modo alternativo per raggiungere il tessuto neurale posizionando l'interfaccia neurale all'interno dei vasi sanguigni adiacenti all'area target, attraverso procedure di angioplastica standard e minimamente invasive. Inoltre, l'uso di materiali transitori evita la necessità di ulteriori interventi chirurgici per la rimozione del dispositivo e riduce la reazione infiammatoria. Tuttavia, tali dispositivi hanno ancora bisogno di cavi per essere alimentati. In questo lavoro viene proposto un circuito elettrico per un trasferimento di potenza induttivo wireless (WIPT), realizzato sfruttando materiali transitori che potrebbero essere integrati in tali dispositivi. Il circuito è un insieme di tre elementi: un induttore, un diodo e un condensatore. In questo lavoro preliminare, l'attenzione è principalmente rivolta alla progettazione, alla fabbricazione ed ad un primo test dell'elemento induttore. Vengono riportati test di degradazione e penetrazione per comprendere il profilo di degradazione del dispositivo e vengono proposte tecniche per migliorare il tempo di funzionamento del dispositivo. Infine, vengono riportati suggerimenti sui materiali e sulle possibili tecniche di fabbricazione del diodo e del condensatore.

Contents

Al	bstra	ct		ii
So	omma	ario		iii
A	crony	\mathbf{ms}		ix
1	Intr	oducti	on	1
	1.1	Endov	ascular Interfaces	2
	1.2	Transi	ent electronics	4
	1.3	Wirele	ess Power transfer	5
	1.4	Overv	iew of the master thesis project	6
2	The	ory of	the Inductive Power Transfer	8
	2.1	Safety	concern and ISM bands	8
	2.2	Coil M	fodel	10
		2.2.1	Inductance	10
		2.2.2	Loss Term R	11
		2.2.3	Parasite Capacitance	13
		2.2.4	Self Resonance Frequency	13
		2.2.5	Quality Factor	14
	2.3	Induct	ive link	15
	2.4	S-para	meters and experimental measure of Q	16
3	Apr	olicatio	on constraints and water barrier tests	19
	3.1	Applic	ation Constraints and Trade offs	19
	3.2	Degra	dation test and Penetration test	21
		3.2.1	Degradation test of encapsulated Mg	21
		3.2.2	Penetration test	24
		3.2.3	Tailoring of degradation profile: ideas for future implementations	25
		3.2.4	Corrosion test: proof of concept for water barrier enhancement .	29
4	Ind	uctive	Coil: Fabrication and Design optimization	32
	4.1	Fabric	ation: Optimization of the procedure	32
		4.1.1	Conductive elements: spiral and L-electrode	32
		4.1.2	PDMS wafer: support for the construction of the device	35
		4.1.3	Substrate and Encapsulation layer	35
		4.1.4	Interlayer	37

		4.1.5 Assembling and connection of the inductor coil	38
	4.2	Simulation and Design	40
5	Indu	active Coil: Experimental testing and Discussion	46
	5.1	Network Analyzer measures	47
		5.1.1 L-maximizing coils: Results	47
		5.1.2 Q-maximizing coil: Results	48
	5.2	Discussion	49
		5.2.1 Preliminary IPT test	52
6	Futi	are Perspectives	56
	6.1	Coil contacts and evaporated Mg	56
		6.1.1 Conductive wax, leads protection and oxide inter-layer	56
		6.1.2 Evaporated Mg	57
	6.2	Diode: Principle application and materials	60
	6.3	Capacitor: Principle application and materials	64
	6.4	Water barrier enhancement	65
7	Con	clusion	66
Ac	lditio	on material	68
	7.1	Coil Parametrization ANSYS	68
	7.2	Matlab script	68

List of Figures

1.1	Example of IMDs. Reported from [35]	1
1.2	Stentrode and SSS	3
1.3	Fanelli et al. NI. Reported from [20]	3
1.4	Materials for transient electronic. Reported from [11]	5
1.5	WPT technologies. Reported from [4].	6
1.6	Electric circuit for IPT	7
2.1	General IPT system. Reported from [68]	8
2.2	ICP coil types. Reported from [68]	10
2.3	Equivalent electric circuit coil. Reported from [68] and modified	10
2.4	Current crowding effect. Reported from [88]	12
2.5	Skin effect. Reported from [68]	12
2.6	Impedance graph of an ideal (red) and real (blue) coil. Reported from [17].	14
2.7	Inductive link. Reported from [4]	15
2.8	2-port Network. Reported from [74]	17
2.9	Example of scattering parameter. Reported from [9]	18
3.1	Geometry of the simplified coil.	21
3.2	Degradation test	22
3.3	Degradation test results	23
3.4	PCL sample in colored saline solution.	25
3.5	Penetration test results	26
3.6	Corrosion test results at different time steps	31
4.1	Coil Inductor Model	33
4.2	First design of reference coil	33
4.3	Laser cutter parameter and results for 10 μ m thick foil	34
4.4	Failing extraction of Mg spirals 10 μ m	34
4.5	Laser cutter parameter and results for 20 μ m thick foil \ldots \ldots \ldots	35
4.6	Glass wafer with PDMS	36
4.7	Dose-Defocus test	36
4.8	Inter-layer geometry and obtained result	37
4.9	Inter-layer melting	38
4.10	Assembling and connection process	39
4.11	Example of simulated coil.	40
4.12	Trends: varying width and gap	42
4.13	Trends: number of turns-width	43

4.14	Best simulated coils: Q & L maximization	4
4.15	Difference between Q factors and L	5
5.1	Tested coil	:6
5.2	Q factor plots of L-maximizing coils	8
5.3	L plots of L-maximizing coils	9
5.4	Q factor plot of Q-maximizing coil	0
5.5	L plots of Q-maximizing coil	0
5.6	Q factor and L values of fabricated coil for ISM bands of interest 5	1
5.7	Receiving coil and matching capacitor	3
5.8	Receiver LC tank characteristics	3
5.9	IPT measure	4
6.1	Evaporated Mg test	8
6.2	Evaporated Mg circular traces	9
6.3	Materials for diode over PCL	2
7.1	Input parameters and conversion equations	8

List of Tables

2.1	ISM bands	9
3.1	Resistance measures before and after first immersion	23
3.2	Penetration depths	25
3.3	Techniques for improving the water barrier capability	27
3.4	Deposition techniques for silicon oxides and nitrades	28
4.1	Range of parameters variation	41
$6.1 \\ 6.2$	MS connection types	61 63

Acronyms

IMD Implantable Medical Device **DBS** Deep Brain Stimulator **VNS** Vagus Nerve Stimulator SSS Superior Sagittal Sinus PCL PolyCaproLactone **PEDOT:PSS** PolyEthyleneDiOxyThiophene:PolSstyrene Sulfonate **EEG** ElectroEncephaloGraphy **NI** Neural Interface PLA PolyLactic Acid **BED** Bioresorbable Electronic Device **TE** Transient Electronic **MEMS** Micro Electro-Mechanical Systems **WPT** Wireless Power Transfer **IPT** Inductive Power Transfer **CPT** Capacitive Power Transfer **APT** Acoustic Power Transfer **RF** Radio Frequency PCB Printed Circuit Board **MEA** Multi-Electrode Array **AC** Alternating Current **PDL** Power Delivered to the Load **PTE** Power Transfer Efficiency FCC Federal Communications Commission

SAR Specific Adsorption Rate

ISM Industrial, Scientific and Medical

ITU Internation Communication Unit

WWC Wire Wound Coil

PSC Printed Spiral Coil

SRF Self Resonance Frequency

 ${\bf P}{\bf A}$ Power Amplifier

 ${\bf NA}\,$ Network Analyzer

ALD Atomic Layer Deposition

PDMS PolyDiMethylSiloxane

MLA MaskLess Aligner

PGMEA Propylene glycol methyl ether acetate

 $\mathbf{D}\mathbf{I}$ Deionized Water

PET PolyEthylene Terephthalate

 ${\bf SMA}$ SubMiniature version A connector

 ${\bf SiNM}$ Silicon Nano
Membrane

 \mathbf{PVD} Physical Vapor Deposition

 ${\bf CVD}\,$ Chemical Vapor Deposition

LPCVD Low Pressure Chemical Vapor Deposition

PECVD Plasma Enhance Chemical Vapor Deposition

WF Work Function

Chapter 1

Introduction

Neural modulation consists in a tailored stimulation of the neural system through electrical, electromagnetic, chemical or optogenetic methodologies in order to obtain a certain response from the interested area of stimulation [47].

Neural stimulation has an enormous potential, with its fields of application varying from treatments of mental disorders and neurodegenerative diseases to motor-sensory recovery after severe injuries [6].

In this context, implantable medical devices (IMDs) are one of the most clear and evident examples of technology that exploits neural modulation for restoring or replacing a function of the body. The most common IMDs are: retinal prosthesis [5], cochlear implants [86], deep brain stimulator (DBS) [56], vagus nerve stimulator (VNS) [25] and cardiac stimulator such as implantable pacemaker and defibrillator [39]. Examples of IMD can be seen in fig. 1.1



Figure 1.1: Example of IMDs. Reported from [35].

Even if these IMDs are at the cutting edge of their research, they can face some drawbacks that limits their clinical use:

• invasive surgery for their insertion [54];

- secondary surgery for removal of the implants [10];
- presence of batteries or external wires to connect stimulating-recording electrodes to power supply [13].

In the last decades, researchers in the filed of neural interface have tried to tackle these problems using different and complementary approaches. In particular:

Endovascular Interfaces reduce the burden of the surgery procedure and decrease the recovery time [18];

Transient electronics avoid the need of a second surgery [10];

Wireless Power Transfer solve the problem of internal batteries and external wire;

1.1 Endovascular Interfaces

From 70's to end of 90's, the endovascular interfaces consisted in a singular guide wires with a conductive tip from which it was possible to record or stimulate the touched tissue. These devices were exploited to measure the amplitude of evoked somatosensory stimuli [72], detect of epilectic foci [58], and to ensure the reduction of seizure after drug assumption [23]. Even if these experiments proved the possibility of endovascular recording with wire electrodes, they show some limitations like the lack of multiple recording points, the high movement noise effect and safety concerns like uncontrolled movements of the wire into the vessel.

At the beginning of the 21st century, the technological development and the consequent miniaturization of electrodes brought to the development of multiple nano-wire electrodes and the first application of conducting polymer instead of platinum [53]. Moreover, catheters with multi electrode array were started to be used for electrophysical recording, such as electroencephalography (EEG). The first venous catheter composed of 16 microelectrodes was implanted in the Superior Sagittal Sinus (SSS) of pigs, verifying that the magnitude, shape and spatiotemporal evolution of EEG (epileptiform activity) were similar to those collected by a subdural electrodes grid [7], validating the endovascular approach as valid alternative of more invasive subdural electrodes.

The wires and catheters technologies pave the way to endovascualr recording and stimulation but they also have some limitations as the low spatial resolution and the short-time application [18]. An extraordinary leap forward was given by the development of the Stentrode (Synchron, Inc) which consists in a stent-shaped multielectrode endovascular device for recording and stimulation. The stent-electrode array, due to its self-opening architecture and the presence of multiple electrodes, overcomes the aforementioned issues [19]. Moreover, the quality of the recorded signal was validated to be comparable to the one obtained with subdural and epidural interfaces [38]. The Stentrode performances were validated in sheep [60][59] and human [61]. Fig.1.2 show the Stentrode and placing site for human and sheep implantation.

The major downside of the Stentrode is its metallic scaffold which gives potential way for shortcuts phenomena [60]. Moreover, the contact between the metallic scaffold and the internal endothelium of veins and arteries induces a high inflammatory response [46][63].



Figure 1.2: (a) Stentrode and catheter system. Scale bar 3 mm; (b) SSS in human and sheep brain. Scale bars of 3 cm and 1 cm respectively. Reported from [60].

These aspects motivated Fanelli *et al.* [20] to create a transient stent-electrode array based on polymeric materials. In fact, bioresorbable stents have been demonstrated to have a better interaction with tissue, lowering the inflammatory response and allowing the tissue to regrow at the implant site [46][26]. Fig. 1.3 show this transient Neural Interface (NI).



Figure 1.3: Fanelli *et al.* NI. Reported from [20].

The stent-shaped scaffold is made of poly- ε -caprolactone (PCL), which is a biodegradable synthetic polymer extensively used for tissue engineering and medical devices [81]. For the conductive element poly(3,4-ethylenedioxythiophene):polystyrene sulfonate (PE-DOT:PSS) was used. This is a conductive and biocompatible polymer usually exploited as coatings of metallic electrodes for neural interfaces or in organic photovoltaic cells. Such device, similarly to the Stentrode, could potentially be used for recording or stimulation of excitable tissue from within a blood vessel.

Teplitzky *et al.* has been demonstrated that five out of seventeen targets of DBS stimulation therapy could be treated with an endovascular approach [73]. This can have a tremendous impact because neurological disorders such as Parkinson, Alzheimer, Tremor, Depression and Obsessive-Compulsive Disorders could be treated with a min-

imally invasive approach using an endovascular interface [73]. The application of this type of devices could be interesting for those patients who could benefit from neuromodulation therapy but the invasiveness of the surgical implantation and the burden of the post-operation couldn't overcame the benefits.

1.2 Transient electronics

The advantage of the IMDs is given by the proximity of their implantation to the target organs which increase the bio-sensing accuracy and their therapeutic efficacy compared to the non-invasive technique [11].

However, the composing element of the commercially available IMDs are made of abiotic materials which can cause a negative immune response [49]. For that reason, after a certain time they have to be removed from the body, causing a state of physical and mental stress to the patient.

Using bioresorbable materials can avoid the burden of a second surgery thanks to their ability to dissolve and be absorbed harmlessly in the human body.

Bioresorbable passive element such as suture wires, glues and stent are already available in the market. They are usually made with natural or synthetic polymers like polylactic acid (PLA), polycarbonate and salicylic acid [71].

For what concerns Bioresorbable Electronic Devices (BEDs) or Transient Electronics (TE), they are still on the research level but the researchers are briskly working on this topic, as the increasing numbers of paper depicts [50].

Usually, BEDs are composed of a conductive element (for conductive traces, electrodes, harvesters, etc...) which lays on the substrate and is encapsulated by an encapsulation layer. Depending on the application, there can also be present semi-conductive materials (for capacitor, diodes, etc...).

The more exploited materials for BEDs are alkaline (Mg) and transitional metals (Mo,W) for conductive elements, inorganic materials for semi-conductive element (SiO₂, ZnO, Si nanomembrane) and Metal Oxide (MgO), natural polymer (silk) or artificial polymer (PLA, PLGA, PCL) for substrate/encapsulation layer. The following image from [11] greatly resume the most used bioresorbable materials:

The choice of the materials is crucial because each of them has different degradation rates. The degradation pathway is also an important aspect to be considered, as well as the physiologically allowed local concentration threshold of the degraded material.

The two main degradation processes that happen inside the body are:

- **Hydrolysis** reaction between implanted material and water which brings to decomposition of both substance and water;
- Metabolic degradation disintegration of implanted material through metabolic pathway, generally with an enzymatic reaction;

Usually, the degradation process is a combination of both mechanisms and leads to waste elements which biocompatible. With an appropriate and careful choice of the material, the duration of the BED can be tailored in an effective manner.

In the last decade, a wide number of BEDs was proposed: temperature sensors [67], pH sensors [78], pressure sensors [42], brain signal sensor [85] and neural stimulator [45]. The realization of these BEDs required a re-adaptation of the exiting micro

	Туре		Materials		
Conductor	Alkalin	e metal	Mg, Ca		
	Transitio	on metal	Fe, Zn, Mo, W		
	Al	oy	Mg Alloy		
	Metalli	c glass	Mg-Zn-Ca		
Semiconductor	Inorg	ganic	Si nanomembrane, ZnO		
	Natural Organic		Indigo, Melanin		
	Artificial Organic		Polypyrrole, PDPP-PD		
Insulator/encapsulation	Metal nitride		Si ₃ N ₄		
	Metal oxide		SiO ₂ , MgO		
	Natural polymer	Polysaccharide	glucose, cellulose, alginic acid, chitin, levan		
		Polypeptide	silk, collagen, biotin, keratin		
		Lipids	phospholipid, triglycerides		
	Artificial polymer	Polyester	PLA, PLGA, PCL, PHB		
		Polyanhydride	PSA, PCPP:SA		
			copolymers		

Figure 1.4: Materials for transient electronic. Reported from [11].

electro-mechanical systems (MEMS) fabrication techniques (such as photolitograpy and shadow-masking). For example, conductive metallic material couldn't stand process where water is present owing to their high reactivity and natural-synthetic polymers often cannot withstand high temperature processes because of their low melting temperature ($50^{\circ}C \leq T_m \leq 100^{\circ}C$). Alternative fabrication techniques are emerging specifically avoid these problems: laser cutter [13], evaporation [67], transfer printing [31].

1.3 Wireless Power transfer

Wireless power transfer (WPT) techniques have gained increasing interest in the past decades for being an effective way to avoid the presence of batteries and transcranialtranscutaneous wires of implantable devices.

The main techniques in the field are Inductive Power Transfer (IPT), Capacitive Power Transfer (CPT), Acoustic Power Transfer (APT) and the radio frequency (RF) radiation in mid and far field. Their characteristics in terms of achievable penetration depth, receiver dimension and link efficiency (transmitter power/power delivered to the load) are depicted in the following image [4]:

Each of them has advantages and disadvantages. RF radiation in mid and far field are used mainly for implants of big dimensions (cm scale or tens of mm): for smaller devices the transfer efficiency strongly decreases and the increasing attenuation given by the tissue absorption demands for higher operating frequencies. APT can achieve high penetration depths and high power transfer thanks to the lower attenuation of the acoustic signals and the higher limit of allowed exposure. The main drawback is that it is usually needed a printed circuit board (PCB) as support for the composing elements, increasing the overall dimension of the device. CPT is less affected by the transmitter-receiver misalignment and angular rotation compared to the APT and IPT. It is also applicable to flexible substrates. The drawback is that its efficiency is strictly dependent on the operating frequency. In fact, the higher the frequency, the higher the efficiency but consequently the tissue attenuation will increase and the penetration depth will reduce. Moreover, the miniaturization is still challenging and limits the



Figure 1.5: WPT technologies. Reported from [4].

application to low penetration depths. Finally, IPT is the most studied and used WPT. As for the CPT, its efficiency is linked to the frequency but the miniaturization is easier and more reliable. The IPT method is preferred for intermediate depth, smaller and multiple implants [4].

The majority of the devices that implement a WPT consists in multi-electrode arrays (MEAs) or free-standing electrodes with the wireless enabling elements (coil for IPT, plate for CPT, piezoresistor for APT) embedded in a CMOS technology substrate. In the last decade, other substrate materials have been considered, such as flexible and polymeric platform. For example, the research group of John Rogers (Northwestern University, Illinois, US) developed various IPT stimulators using polymeric substrate such as poly(lactic-co-glycolic acid) (PLGA) [13] [27] [45] and polyurethane [14],

1.4 Overview of the master thesis project

The three approaches presented in the latter sections drove the concept of this project: a wireless and transient endovascualr interface for brain stimulation. In particular, starting from the transient neurovascular interface developed by Fanelli *et al.* [20], the idea is to design and create a transient electrical circuit composed by an ensemble of three elements that could be able to wirelessly deliver a stimulating impulse to the brain tissue around the SSS through IPT induced by an external transmitter. The three circuital elements are:

Coil indictor which generates a current proportional to the variation of magnetic flux that flows through it;

Diode which rectifies the generated signal;

Capacitor which operates a smoothing effect and matching role;

Fig. 1.6 gives a representation of the general electrical circuit (a) and the placement of the elements (b) in Fanelli et al. neural interface.



Figure 1.6: Electric circuit for IPT.

The attention of this work is focused on the design and optimization of the coil element.

It is started from the evaluation of the selected materials deterioration and permeation behavior in water solution. Then, the design of receiving coil is studied in ANSYS HFSS. The most promising geometries are fabricated and their quality is compared to the simulations. After selecting the best coil, a preliminary WIPT test is performed. Moreover, preliminary studies of materials and fabrication techniques for the realization of the diode and capacitors are reported.

Due to the challenging objectives that this project wants to achieve, an accurate work of finding the best trade off has to be carried out. The most challenging aspect are:

- reduced dimension of the wireless system: 4x4 mm² for the receiving coil element;
- choice of bioresorbable-transient material;
- ad-hoc fabrication technique.

Chapter 2

Theory of the Inductive Power Transfer

An IPT is a wireless delivery of power that happens when two inductive coils, a Transmitting coil Tx and a Receiving coil Rx, are sufficiently close so that they are magnetically coupled.



Figure 2.1: General IPT system. Reported from [68].

Referring to fig. 2.1, the IPT is possible when an AC signal (i_{in}) with a certain operation frequency (f_p) is applied to Tx. The flowing current generates a magnetic field which is intercepted by Rx. Considering the Faraday's law, this varying magnetic field induces an alternate electrical signal (i_{out}) in the Rx.

The general goal of a WPT is to deliver a sufficient power with an acceptable efficiency for avoiding an excessive heating of the tissue. For this, the two main parameters used for evaluating the goodness of an IPT are the Power Delivered to the Load (PDL) and the Power Transfer Efficiency (PTE). These are strongly influenced by the Quality factors (Q-factor or Q) of the coils and the coupling of the two coils forming the inductive link [66].

2.1 Safety concern and ISM bands

The majority of the medical applications that exploits the IPT relies on the RF in the near-field region (from kHz to few GHz). The usage of this band is strictly regulated as [34] reports. The Federal Communications Commission (FCC) reports that the max-

imum averaged Specific Adsorption Rate (SAR) for the human head is 1.6 W/Kg for 1 g of tissue mass measured during 6 minutes of exposure. The SAR can be defined as:

$$SAR = \frac{\sigma |E_{rms}|^2}{\rho} \tag{2.1}$$

where: σ is the electrical conductivity of the tissue, ρ is the tissue density and E_{rms} is the round value of the induced electric field. Due to the fact that the coils inductors are magnetically coupled through a magnetic field B, the electric field E can be extracted from Maxwell's equation:

$$\nabla \times E = -\frac{\partial B}{\partial t} \propto \omega_p I_1 \tag{2.2}$$

where I_1 is the alternated current that flows in the receiver L_1 with a angular operating frequency of $\omega_p = 2\pi f_p$. It is clear that $SAR \propto (f_p \times I_1)^2$ [16]. This proportion is interesting because it states that the current flowing in the primary coil can be increased and at the same time the SAR can be kept constant with an appropriate decrease of the operating frequency f_p . In this way, the PDL can be improved without overcoming the SAR limit, if the PTE remains high for low f_p s [4]. In any case, an f_p decrease should not reach the sub-MHz range in order to avoid an excessive heating of the tissues due to ohmic losses [51].

The frequency at which IPT operates, i.e f_p , has to be selected by the user between ranges that are reserved for Industrial, Scientific and Medical applications, also knows as ISM bands. They are defined by the International Communication Unit (ITU) for non-telecommunication purposes [75]. The Tab. 2.1 shows them.

Available ISM bands
6.765 MHz - 6.795 MHz
13.553 MHz - 13.567 MHz
26.957 MHz - 27.283 MHz
40.66 MHz - 40.70 MHz
83.996 MHz - 84.004 MHz
167.992 MHz - 168.008 MHz
433.05 MHz - 434.79 MHz
886 MHz - 906 MHz
2.4 GHz - 2.5 GHz
5.725 GHz - 5.875 GHz
24.0 GHz - 25.25 GHz
61.0 GHz - 61.5 GHz
122 GHz - 123 GHz
244 GHz - 246 GHz

2.2 Coil Model

The coils employed in the IPT are usually of two kind: Printed Spiral coil (PSC) and Solenoid Wire Wound Coil (WWC). The PSC, also known as PCB spiral for their usual substrate material, can be further divided based on their geometry such as squared, circular, hexagonal, etc...The main difference is that for WWCs the turns have a constant radius and they are not planar while for PSCs the radius changes and they are planar. A representation of these two configurations is presented in Fig. 2.2.



Figure 2.2: ICP coil types. Reported from [68].

Moreover, Fig. 2.2 displays the principal geometrical characteristics for WWS and PSC which have a key role in the definition of their electromagnetic properties. The electrical equivalent circuit of a coil is presented in Fig. 2.3



Figure 2.3: Equivalent electric circuit coil. Reported from [68] and modified.

2.2.1 Inductance

The inductance is the most important electrical characteristic of a coil. A large number of papers were published trying to express L as a function of geometrical features of the coil [44][57]. In this chapter, only the circular PSC configuration was considered due to

our application. An effective way to express the inductance of circular coils is based on the fill factor $\varphi = (d - d_{in})/(d + d_{in})$ and the average diameter $d_{avg} = (d + d_{in})/2$ [57]:

$$L = \frac{\mu n^2 d_{avg}}{2} \left[\ln \left(\frac{2.46}{\varphi} \right) + 0.2\varphi^2 \right]$$
(2.3)

with $\mu = \mu_0 \mu_r$ is the permeability of the conductive metal.

2.2.2 Loss Term R

Series Loss Term

The Series Loss R_s is the combination of three loss phenomena: resistive loss, the proximity effect and the skin effect.

$$R_s = R_{DC} + R_{prox} + R_{skin} \tag{2.4}$$

 R_{DC} is the resistive loss due to the resistivity of the conductive material. Its value depends on the geometry and length:

$$R_{DC} = \rho_c \frac{l_c}{A_c} \tag{2.5}$$

where ρ_c is the electrical resistivity of the material, $l_c = \pi \left(d - \frac{(w+s)n}{2} \right) n$ is the length of the conductive trace and $A_c = wt_0$ is the cross-section area of the conductor with width w and thickness t_0 .

 R_{prox} is the resistive loss caused by the proximity effect, also known as the current crowding effect. When considering a coil, the flow of an AC produces a magnetic field (H) around the conductor trace. H interacts with the adjacent loop and eddy currents are generated within them, in a direction which is perpendicular to variation of the magnetic field. Since also the other loops are crossed by a current, the eddy currents obstruct the current flow in the region of the trace near to the other. In summary, the H-field generated by a conductor interferes with the current density of adjacent loops limiting the volume in which the current can flows and so increasing the resistive loss. Fig. 2.4 gives a visual representation of the current crowding effect.

Kuhn and Ibrahim [48] elaborate an approximation of the R_{prox} based on an analytical model which resulted in the equation (2.6)

$$R_{prox} = \frac{R_{DC}}{10} \left(\frac{\omega}{\omega_{crit}}\right)^2 \tag{2.6}$$

where $\omega_{crit} = \frac{3.1}{\mu_0} \frac{(w+s)}{w^2} R_{sheet}$ is the critical frequency at which the proximity effect starts to occur and R_{sheet} is the sheet resistance of the conductor.

 R_{skin} is the series loss term that takes into account the skin effect. It consists in the tendency of an alternating current to flow into a conductor with a non-uniform distribution. In particular, the current density is higher near the external surface of the conductor and decreases reaching the center of the trace. As for the proximity effect, this brings the current to flow in only a fraction of the available conductor volume



Figure 2.4: Current crowding effect. Reported from [88].

and leads to an overall increase of the resistance. Fig. 2.5 displays the skin effect in a conductor. Wheeler *et al.* [79] formalized an expression for R_{skin} for rectangular



Figure 2.5: Skin effect. Reported from [68].

cross-section traces:

$$R_{skin} = R_{DC} \frac{t_0}{\delta \left(1 - e^{-\frac{t_0}{\delta}}\right)} \frac{1}{1 + \frac{t_0}{w}}$$
(2.7)

where R_{DC} is defined in (2.5), t_0 is the thickness of the rectangular conductive trace and δ is the skin depth expressed as a function of angular frequency ω :

$$\delta(\omega) = \sqrt{\frac{2\rho_c}{\omega\mu}} \tag{2.8}$$

with ρ_c being the conductivity and μ the permittivity of the metal trace. The skin depth δ is the thickness from the external surface in which the majority of the current flows and, as can be easily seen from (2.8), it decreases for higher frequencies bringing the R_{skin} to higher values.

Parallel Loss Term

 R_p is a parallel resistance that keeps into account the phenomenon related to the dielectric loss δ_i of each material *i* that surrounds the coil (i.e. tissue, top coating,

air, bottom coating and substrate). Pieters *et al.* [62] modeled R_p as a function of loss tangent $tan(\delta_i)$, the relative dielectric constant of each dielectric layer $\varepsilon_{r,i}$, angular frequency ω and the complete elliptic integral of the first kind K(k):

$$\frac{1}{R_p(\omega)} = G_p(\omega)$$

$$= \frac{\omega\varepsilon_0}{2} \left[\varepsilon_{r1} \tan \delta_1 \frac{K(k_1')}{K(k_1)} + (\varepsilon_{r2} \tan \delta_2 - \varepsilon_{r1} \tan \delta_1) \frac{K(k_2')}{K(k_2)} + \varepsilon_{r3} \tan \delta_3 \frac{K(k_3')}{K(k_3)} + (\varepsilon_{r4} \tan \delta_4 - \varepsilon_{r3} \tan \delta_3) \frac{K(k_4')}{K(k_4)} + (\varepsilon_{r5} \tan \delta_5 - \varepsilon_{r4} \tan \delta_4) \frac{K(k_5')}{K(k_5)} \right]$$
(2.9)

 R_p is significant at low frequencies and for materials with small δ_i and for this reason could be neglected for Rx.

2.2.3 Parasite Capacitance

The parasite capacitance (C_p) in the electric model of the coil (Fig. 2.3) represents the capacitive effect that is generated when conductive materials are nearly placed and are divided by a dielectric one. C_p relies mainly on the spacing and the thickness of the metal trace as well as the encapsulation and the environmental materials (e.g. air, human tissues or fluids) which have different physical properties. A complete modelization of C_p was proposed in literature [41], which is based on the evaluation of the effective relative dielectric constant (ε_{r-eff}) of the multilayer structure (substrate, coil, coating, tissue and air) and complete elliptic integral of the first kind. A simpler formulation is given in [40] which considers only FR4 as substrate material for the coil and air as environment:

$$C_p = (\alpha \varepsilon_{rc} + \beta \varepsilon_{rs}) \varepsilon_0 \frac{t_0}{s} l_g \tag{2.10}$$

where $l_g = 4(d - w \cdot n)(n - 1) - 4s \cdot n(n + 1)$ is the length of the gap, ε_{rc} is the relative dielectric constant of coating material, ε_{rs} is the relative dielectric constant of substrate and α - β are constants equal to 0.9-0.1 for air-FR4 materials. C_p is the main responsible for the resonant behavior of the coil as shown in the next part.

2.2.4 Self Resonance Frequency

An ideal inductor L, a coil without any loss and parasitic term, would have an impedance equal to $Z = sL = j2\pi fL$. It can be easily understand that in this case the impedance is directly proportional to the frequency and its magnitude bode diagram is a line with a constant angular coefficient. In Fig. 2.6, the impedance of an ideal inductor is shown as a red line.

As reported above, the coil presents a parasitic capacitance which is expressed by the term C_p . This capacitance, which in the electrical equivalent circuit is in parallel to the L, changes the impedance that in turn becomes:

$$Z = \frac{sL}{s^2 L C_p + 1} \tag{2.11}$$



Figure 2.6: Impedance graph of an ideal (red) and real (blue) coil. Reported from [17].

In this case, the resulting impedance has again a zero for f = 0 but it appears a double pole for a particular frequency:

$$f_{SRF} = \frac{1}{2\pi\sqrt{LC_p}} \tag{2.12}$$

 f_{SRF} is the Self Resonance Frequency (SRF) of the coil and it is the frequency at which the inductor L and the parasite capacitance C_p of the coil resonate resulting in a high impedance. The coil behaves like an inductor for frequencies lower than SFR while for higher frequencies its behavior is ruled by the capacitance. Moreover, at SRF the quality factor of the coil goes to zero. For that reason the operating frequency f_p has to be chosen well under the SRF. As a rule of thumb, the effect of C_p can be neglected for $f_p \leq f_{SRF}/10$.

2.2.5 Quality Factor

The Q-factor expresses the ability of a coil inductor to generate a magnetic field when it is crossed by an AC signal or to generate an alternated current when it is crossed by a varying magnetic flux. Q is a dimensionless quantity expressing how similar the behavior of the real under-study coil is compared to the ideal case of a perfect inductor which is loss-free. In other words, it tells the extent to which the coil is prone to losses. High Q means low dissipation while low Q reflects a high dissipation. Since all the components of the coil model are defined, the Q factor can be expressed from the equivalent impedance Z of the lumped circuit of Fig. 2.3 as the ratio of the imaginary part over the real part:

$$Q = \frac{Im(Z)}{Re(Z)} \tag{2.13}$$

$$Z = \frac{R_s + j\omega L}{(R_s + j\omega L)(G_p + jwC_p) + 1}$$

$$(2.14)$$

As mentioned before, for Rx the effect of R_p can be neglected bringing to the simplified equation of Q:

$$Q = \frac{\omega L - \omega C_p (R_s^2 + \omega^2 L^2)}{R_s} \tag{2.15}$$

From the previous discussion, it is easy to understand that the Q-factor is strongly dependent on the geometry and material of the coil as well as the encapsulation materials.

2.3 Inductive link

Now that all the parameters for defining the electromagnetic behavior of the coils are given, the inductive link is presented.



Figure 2.7: Inductive link. Reported from [4].

Fig. 2.7 shows a general model of an inductive link between two coils. Starting from the left, a voltage V_s is generated by the power source which is usually amplified by a Power Amplifier (*PA*) that has an intrinsic resistance R_s . The PA is connected to the Transmitting Coil which is magnetically coupled to the Receiving coil through M_{12} . The subscripts 1 and 2 refer respectively to the primary circuit (PA and Tx) and the secondary circuit (Rx and Load). The transmitted signal is then delivered to the load R_L .

 C_{s1} and C_2 are additional capacitors called matching capacitors. It is demonstrated [2] that to maximize the PTE and PDL, C_{s1} -Tx tank and Rx- C_2 tank should be tuned in order to have the same resonance frequency in correspondence of the wanted operational frequency f_p . It is important to underline that the tanks resonating frequency is and must be different from the SRFs of the single coils.

The last parameter to be defined is the mutual inductance M_{12} between the Tx and Rx. M_{12} depends again on the geometry and distance. It can be evaluated making use of complete elliptic integrals of first kind K(k) and complete elliptic integrals of second kind E(k) [52]:

$$M_{12} = \theta \sum_{i=1}^{N_1} \sum_{j=1}^{N_2} M(x_i, y_j, d_{12})$$
(2.16)

$$M(x, y, d_{12}) = \mu_0 \sqrt{xy} \left[\left(\frac{2}{\gamma} - \gamma \right) K(\gamma) - \frac{2}{\gamma} E(\gamma) \right]$$
(2.17)

$$\gamma = \sqrt{\frac{4xy}{(x+y)^2 + d_{12}^2}} \tag{2.18}$$

where x and y are the spatial coordinates and d_{12} is the distance between the coils. The magnetic coupling is heavily influenced by the distance between the two coils and the angular-linear displacement [33] because they reduce the amount of magnetic flux which passes in-between the receiver and elicit a response.

Finally, the equation for PTE (η) and PDL (P_L) can be expressed as [29]:

$$\eta(\omega) = \frac{k_{12}^2 Q_1 Q_{2L}}{1 + k_{12}^2 Q_1 Q_{2L}} \frac{Q_{2L}}{Q_L}$$
(2.19)

$$P_L(\omega) = \frac{V_s^2}{2R_1} \frac{k_{12}^2 Q_1 Q_{2L}}{(1 + k_{12}^2 Q_1 Q_{2L})^2} \frac{Q_{2L}}{Q_L} = \frac{V_s^2}{2R_1} \frac{\eta}{1 + k_{12}^2 Q_1 Q_{2L}}$$
(2.20)

in which $k_{12} = M_{12}/\sqrt{L_1L_2}$ is the coupling coefficient, $Q_L = R_L/(\omega L_2)$ is the quality factor of the load and Q_{2L} is defined as $Q_{2L} = (Q_2Q_L)/(Q_2 + Q_L)$.

Another important aspect is that, as it was demonstrated in [33], the PTE and PDL can be maximized if the R_L is equal to the parallel resistance of $Rx - C_2$ tank R_{2p} . The optimal value of the load resistance is:

$$R_{L,opt}(\omega) = \omega_p L_2 Q_{L,opt} = \omega_p L_2 Q_2 = R_{p2}$$

$$(2.21)$$

Usually for biomedical implants, R_{p2} is around 100 Ω while R_L can reaches thousands of Ω [40]. Thus, the Load resistance should be decreased to $R_{L,opt}$ value and this can be achieved by using an L-match network with a series inductor.

2.4 S-parameters and experimental measure of Q

The aforementioned theory is useful for understanding the principal concepts of the IPT and how the electro-magnetic properties of the coil can be obtained from geo-

metrical and environmental characteristics. Another way to evaluate the parameters that describe the coil (i.e. Q) and the IPT, is to exploit the scattering parameters (S-parameters). The S-parameters can be extracted from an RF simulation software like Ansys HFSS or connecting the coils to a Network Analyzer (NA). They measure the amount of power that is reflected or transmitted given a certain input signal in a port.



Figure 2.8: 2-port Network. Reported from [74].

A single coil can be seen as a 1 port network and the S_{11} parameter can be evaluated as:

$$S_{11} = \frac{reflected \, power}{incident \, power} \tag{2.22}$$

 S_{11} , also known as the return loss, depends on the frequency and for values around f_p it should be minimized so that a low amount of power would be reflected and the majority would be transmitted. The $S_{11} - f$ graph is usually a curve with one or more minima. These points represent the optimal working frequency for the under-study coil because for those the reflection of the input signal is in a minimum. The minimum can be shifted to the wanted f_p adding a capacitor in series or in parallel.

When an inductive link is studied, it can be considered as a two-ports network. In this case the forward transmission S_{21} is more interesting. This is in fact the ratio of transmitted power over the incident one. S_{21} should be therefore maximized for f_p . Fig. 2.9 shows an example of scattering parameters for an inductive link where for the working frequency of 993 MHz the S_{11} is in the minimum while the S_{21} is in the maximum.

Starting from the S-parameter, both real and imaginary parts of the impedance Z of the coil can be extracted. Once Z is known, the Q factor can be calculated with 2.13 and the inductance as:

$$L(\omega) = \frac{Im(Z)}{\omega} \tag{2.23}$$



Figure 2.9: Example of scattering parameter. Reported from [9].

Chapter 3

Application constraints and water barrier tests

3.1 Application Constraints and Trade offs

The realization of the elements which constitute the electrical circuit for the stimulation is not trivial, as introduced in Chapter 1. In particular, the main constraints are:

- The dimension of the receiving coil should stay inside 4 mm x 4 mm square due to the fact that the SSS has a diameter of 3.5-4 mm and the usually exploited catheters have internal diameter of 1.6 mm, leading to a circumference of ≈ 5 mm.
- Usage of biodegradable-transient materials which usually have lower electrical properties compared to the non-biodegradable ones;
- Ad hoc fabrication techniques which require try-and-error approach.

A series of trade offs has to be decided in order to meet these constraints.

Starting form the first point, a careful design of the coil has been carried out in order to optimize the available space and the electrical properties of the coil. In fact, bigger receiving coils can be manipulated in an easier way and usually have higher power transfer efficiency. However, in this application the space is limited and a trade off between the number of turns, width of the conductor and space gap between the traces has to be evaluated in order to maximize the inductive properties.

Concerning the materials, metals usually exploited in standard non-transient electronics are gold or copper, thanks to their high electrical conductivity, $4.11 \times 10^7 S/m$ and $5.96 \times 10^7 S/m$ respectively. Unfortunately, these materials are not biodegradable and thus other conductive materials must be picked. The pool of available materials is reported in the first chapter in Fig. 1.4. The Alkaline metals (such as magnesium, calcium, beryllium, etc...) have usually a higher conductivity compared to the transitional ones, however they are more reactive with water and air and durability of the device could be compromised. For this, another possibility could be to investigate the use of a conductive polymer such as PEDOT:PSS, but its conductivity of several orders of magnitude lower than metals limits its application for the antenna element. Therefore, it is decided to use the Magnesium (Mg) because it has a higher electrical

conductivity compared to the transient metal $(22.6 \times 10^7 S/m$ against $1.89 \times 10^7 S/m$ of W and $18.7 \times 10^6 S/m$ of Mo) and its degradation pathway was demonstrated to be unharmful [28]. The main drawback is its high susceptibility to oxidation in air and fast corrosion and dissolution in water. In fact, it has a dissolution rate in Deionized (DI) water of 0.07 μ m/h against 0.0003 μ m/h of W and Mo while a corrosion rate of 0.05-0.5 μ m/h against 0.02-0.06 μ m/h of W and 0.0002-0.001 μ m/h of Mo [83]. In any case, the metals elements must be protected by an encapsulating layer, which has to be carefully designed to reduce leakage at the least.

As substrate and encapsulation materials, there are plenty of possibilities [10]. The two main groups are biodegradable polymers and metal derived materials such as nitrides and oxides. The latter are usually employed as insulation layers for devices which need an isolated part (e.g. insulating layer for capacitor [45]) or as thin protective layers against the corrosion [15]. The drawbacks are that their deposition process is usually slow, preventing the realization of thicker layers needed for a longer protective action, or they require high processing temperatures which could not be sustainable for the other materials of the device. Degradable polymers are instead extensively used as encapsulations thanks to their ease of handling and manipulation. Even if natural polymers, especially silk, were used at the raising of the TE [87], now the research is moving towards the synthetic counterparts, due the possibility to easily tailor mechanical and physical properties by adjusting the polymeric chain composition [37]. Another important aspect to be considered is how the polymers degrade in water solutions. The degradation process can be a bulk process, in which the polymer lets the fluid cross all its volume before it starts to degrade, or a surface process, where the degradation begins at the surface before starting to spread into the sample. Examples of bulk degradation polymers are the PCL and PLGA while surface degradation polymers are polyanhydrides and poly(ortho esters) [76]. For encapsulation and water barrier purposes, the surface degradation polymers could be a better choice but usually they require advanced production methods ad surface modification processes. On the other hand, the bulk degradation polymers are readily available in the market and do not require particular advanced manipulation processes. For these reasons, the PCL was chosen as encapsulation material for our device. Moreover, the PCL has the slowest degradation rate among degradable polyesters, that could therefore ensure a longer operational lifetime of the device.

Finally, the number of usable microfabrication techniques is severally reduced because the high temperature and the aggressive environments of standard processes are not tolerated by TE materials. For example, the Mg cannot stand process in which oxygen or water are present; the PCL, due to its low melting temperature, can not bear process with temperatures higher than 60°C. These limitations bring to the exploitation of processes like laser cutting or sputtering through stencil masks. Compared to MEMS techniques which are well established and optimized after years of implementation, the new processes are less known and the final results can vary in function of the sample properties and geometries.

3.2 Degradation test and Penetration test

The PCL was demonstrated to be an excellent material for tissue engineering scaffolds and drug delivery system thanks to its low degradation rate in aqueous environment [81]. In the application of the present work, the PCL is not only intended to give structural rigidity but also to be a barrier to the water penetration inside the device. Given that a minimal water leakage is expected from degradable polymers such as PCL, we tested how long would it take water to pass through the PCL layers and attack the encapsulated magnesium. Furthermore, we analysed in time the water penetration depth inside PCL sheets of different thickness.

3.2.1 Degradation test of encapsulated Mg

The liquid barrier capability of the PCL layer is tested through a test in which a simplified coil geometry is encapsulated between two PCL layers (40 μm) and is immersed in saline water at 37 °C at pH 7 and pH 12. To evaluate the degradation of the Mg coil induced by penetration of saline, the resistance R of the coil is measured at different time points .

Fabrication

For this test, a simplified coil geometry is laser cut (WS Turret Laser Micromachining Workstation, Optec) from an Mg foil (Goodfellow Corp., USA) of 10 μm thickness. The geometry is realized in AutoCAD and is reported in the Fig. 3.1.



Figure 3.1: Geometry of the simplified coil.

The substrate and encapsulation layers are made of PCL through replica molding technique. Two molds are realized: one with protruding pattern of the simplified coil, used to pattern corresponding grooves on the substrate layer; an engraved mold of the coil which draws protruding traces on the encapsulation layer to tightly close the device. The mold are fabricated using SU8 GM1070 photoresisit spin coated on 4-inch Si

wafers (1500 rpm, 35 μm) and via photolitographic process (MLA150, Heidelberg Instruments). After the realization of the moulds, a release layer of polystyrene sulfonate (PSS 18 wt% in H_2O , Sigma Aldrich) is spin coated over them and they are baked at 145 °C for 10 minutes. The PCL (MW 50000, 25090, Polyscience, Inc.) is dissolved in chloroform (C2432, Sigma-Aldrich) with a proportion of 30% w/v and than spin coated over the molds (800 rpm, 40 μ m). The molds are put in the oven for 30 min at 80 °C in order to remove internal stress of the PCL induced by the spin coating process and than put under the hood at room temperature to let the chloroform solvent evaporate completely. The patterned PCL pieces are than removed from the molds using deionized (DI) water and let dry. The final device is obtained by placing the Mg simplified coil over the PCL substrate and closed with the encapsulation layer. The sealing between the two PCL layers is obtained with their melting over a hot plate at a temperature of 60 °C. For the devices with cables for resistance measurements, Teflon coated stainless steel wires (114- μm diameter; Z-790500, Science Products GmbH) are connected to the two ends with silver paste (G3692 Acheson Silver DAG 1415, PLANO GmbH). To facilitate the connection to the multimeter (Keysight Technologies), additional wire of bigger diameter ($\phi=0.5$ mm, Habia) are soldered. The final devices are shown in Fig 3.2.



Figure 3.2: (a) Set up of the experiment; (b) Encapsulated coils.

Experiment and Results

Two coils with wires for R measurements and two coils without wires are fabricated. The ones without wires are realized in order to assess the influence of the contact area Mg coil-wire over the permeability of encapsulation. In fact, the presence of wires produces a non perfect sealing of the PCL which could facilitate the penetration of saline solution. Two saline solutions (0.9% NaCl) are prepared: one with pH 7 to simulate physiological condition, and one with pH 12 (1% v/v NaOH) to accelerate the degradation process. Then, they are placed over a hot plate (SuperNuova+, Thermo Scientific) and they are kept at 37 °C by a temperature measurement and adjustment system. The devices with wires are immersed partially to avoid that the contact areas are submersed while the devices without wire are totally immersed in the solutions. The resistances of the wired-devices are measured before the after the first immersion in solutions to ensure the quality of the encapsulation and the absence of holes which

could bring to ad accelerate degradation of the Mg trace in saline. Five R-measures are taken and resumed in the following table:

	Measure 1	Measure 2	Measure 3	Measure 4	Measure 5
R no Saline pH 7 (Ω)	36.37	36.39	36.39	36.40	36.39
R Saline pH 7 (Ω)	36.64	36.53	36.53	36.52	36.53
R no Saline pH 12 (Ω)	38.28	34.78	38.31	35.31	38.37
R Saline pH 12 (Ω)	36.43	35.25	35.08	35.10	35.88

Table 3.1: Resistance measures before and after first immersion.

The variation of the resistances between the no Saline and Saline condition are comparable with the variation between the different measures of the same condition, meaning that the devices are well sealed (One way ANOVA, p=0.15). The R values and the device morphology are evaluated at different time steps: after 4h, 19h and 21h. Fig.3.3 presents the evolution of degradation in the aforementioned steps.



Figure 3.3: Degradation test results. (a) and (b) coils after 4h in saline pH 7 and pH 12; (c) and (d) coils after 19h in saline pH7 and pH 12; (e) and (f) coils after 21 h in saline pH7 and pH 12.

The degradation after 4h is already visible, in particular for the wired samples with the appearance of white spots over the Mg trace. In fact, when the saline solution penetrates in the PCL and reaches the trace it starts to degrade it through hydrolysis. The increase of the resistance is considerable since the values reaches 150 k Ω for both samples.

After 19h, the degraded areas increase their extensions and it is also clear for the unwired samples. It is also important to notice that as suspected, the samples in pH 12

solution show areas of bigger degradation. The measured resistances have experienced another increment, with values around 4 M Ω for both pH.

Finally, after the 21h the samples exhibit again a rise of white areas but the resistance of both samples remains in the order of 4 M Ω , meaning that the Mg trace are interrupted and the conductive path is compromised.

This test demonstrates that the saline solution is able to penetrate the PCL already after 4h but it would be interesting to know the extent of the penetration. This motivated us to pursue a water penetration test.

3.2.2 Penetration test

The penetration test is carried out in order to evaluate the extension of the penetration of the fluids (saline solution) in samples of bulk PCL. The effect of the fabrication processes on the permeability is also evaluated.

Fabrication

The PCL (MW 50000, 25090, Polyscience, Inc.) is mixed with chloroform (C2432, Sigma-Aldrich) with a proportion of 30% w/v and then is poured into a beaker glass to obtain a thickness of about 1 cm. Then, the baker is placed in the oven for 2 h at 80 °C with the aim of helping the evaporation of the chloroform but avoiding the formation of a hard PCL cap on the top that would prevent an homogeneous solidification. At that point, the beaker is put under the hood overnight for a complete solidification and evaporation of the remaining solvent. After the evaporation of the solvent, the remaining thickness will be approximately 2 mm. The PCL disk is then removed form the beaker and cut in the desired shape for the experiment.

Experiment and Results

The experiment consists in measuring the penetration depth of a colored solution at different time steps. Rhodamine B (R6626-25G, Sigma-Aldrich) in a dissolved Saline solution (0.9% NaCl) at 0.05% w/v and a dark-red solution is obtained. Then, two pieces of PCL are trimmed from the PCL disk and one is directly dipped in the dye solution while the second one before the immersion is melted at 60°C and re-solidified at room temperature. This is done with the purpose of assessing if the melting process used for sealing the substrate and the encapsulation influences greatly the penetration profile.

After 4h and 6h of immersion, the samples are taken from the solutions, dried and put in the freezer at 20 °C for 1h. This is done to avoid the movement of Rhodamine along the cutting surface. After that time, the PCL samples are cut along the thickness and the penetration profile is analyzed with the DVM6 digital microscope (Leica Microsystem). Fig.3.5 shows the results of this experiment.

As can be seen from the figure, the penetration is not uniform in the PCL crosssection and the penetration depth increases after 6h, as expected. For what concerns the difference between the remelted and non remelted samples, the penetration depth is not highly different. The table 3.2 resume the penetration depths for the two case, reporting the higher and the lower value of penetration.



Figure 3.4: PCL sample in colored saline solution.

Table 3.2: Penetration depths.

	Penetration depth 4h $[\mu m]$	Penetration depth 6h $[\mu m]$
Unmelted	43-84	57 - 124
Melted	40-50	90-160

This experiment confirms that after 4 hours the saline solution is able to penetrate into the PCL for a depth that is higher than the thickness of the substrate and encapsulation layer used for sealing the Mg traces (40 μ m) inducing their degradation.

3.2.3 Tailoring of degradation profile: ideas for future implementations

The most interesting and appealing aspect of the TE is the ability to degrade in physiological conditions. The degradation rate of the device must comply with its desired life time. In this way the effectiveness of the treatment provided by the BED can be ensured for the desired time. In the case of this project, the electrical stimulation has to be guaranteed from weeks to months. This time cannot be ensured using bare PCL as water barrier for the Mg trace. In literature there were proposed several methods to increase the life time of the transient devices and improve the water barrier capability. Each of them have drawbacks and advantages. The table 3.3 resume the main techniques as well as the cons in the use of them.

The adhesion strength between PCL-PLC layers and PCL-Mg is an important property of the polymer because a poor value of this feature could lead to the detachment of the two polymeric layers or the Mg trace from the substrate with the consequent failure of the entire device in a water solution. Moreover, it is found that the breakdown of the BED is induced not only from the water penetration through the polymer encapsulation material (out-of-plane water penetration) but also from the water at the polymer-polymer interface or polymer-metal interface (in-plane water penetration) which can find a weak barrier due to weak interfacial bonds or holes between material



Figure 3.5: Penetration test results. (a) and (b) Penetration of unmelted PCL after 4h immersion; (c) and(d) Penetration of melted PCL after 4h immersion; (e) and (f) Penetration of unmelted PCL after 6h immersion; (g) and (h) Penetration of melted PCL after 6h immersion.

layers [14]. Therefore, the increase of the adhesion strength, which is strictly linked to the chemical bond strength between the device materials, would lead to an improvement of the water barrier capability. The adhesive strength between PCL and Mg was measured by Degner *et al.* [82] to be 1.85 MPa. In the same work, they compared this value with the one corresponding to the poly (l-lactic acid) (PLLA)-Mg interface which resulted to be higher (4.1 MPa). They claimed that this evident difference is due to the higher presence of oxigen in the polymeric chain of the PLLA which gives more electrostatic interaction with the Mg [82]. From the literature [36], it is demonstrated that plasma treatments could effectively modify and activate the surface of PCL, increasing the number of oxygen species. If on one hand the plasma treatment theoretically may improve the adhesion strength, on the other it is proved that it enhances the wettability of the PCL surface and so decrease the water barrier efficacy [70]. In addition, the plasma treatment has to be non-thermal (e.g. dielectric barrier discharge or low pressure plasma), otherwise given the temperature reached well above T_m of PCL, could damage the sample.

The microstructure of the PCL can influence the water barrier behavior and degradation of the final device. In particular, the water adsorption and the consequent swelling of the PCL happen in two steps [21]: the water diffusion begins in the amorphous regions which are less organized and facilitate the penetration; after that, the majority of the amorphous regions are swelled, the diffusion starts also in the crystalline domains from the margins to the center. Jenkins and Harrison [37] demonstrated that both molecular weight (the length of polymer chain) and grade of crystallinity (percentage of the crystalline domain over the amorphous) if increased can lead to reduction of the degradation of the PCL. In fact, it is seen that longer polymeric chains slow down the entry of water in the polymer as well as the diffusion of products of the degradation from the bulk [37]. There are several techniques to enhance the grade of crystallinity such as imposing a high pressure during cooling or cooling the sample slowly [3]. The drawback of using PCL of higher molecular weight and with an higher degree of crystallinity is that there is a decrease of the elastic behavior and an increase
Methodologies	Implementations	Drawbacks		
Improve layers adhesion	Plasma treatment.	Temperature issues & Re- duction of hydrophobic- ity.		
Work on micro-structure	Higher molecular weight & Increase the crys- tallinity of PCL.	Increase of stiffness.		
Increase the thickness	Spin coat thicker PCL and use thicker Mg foil.	Increase of stiffness and encapsulation issues.		
Surface treatment of Mg	Oxidation of the surface.	Change in electromag- netic properties.		
Natural waxes layer	Encapsulate the device with dipping or spin coat of natural waxes.	Change in electromag- netic properties and In- crease of stiffness.		
Surface degradation polymer	Spin coat of polyanhy- drides and poly(ortho es- ters).	More difficult synthesiza- tion and Increse of stiff- ness.		

Table 3.3: Techniques for improving the water barrier capability.

of the stiffness of the PCL layer. This could reduce the mechanical properties like bendability which is important for the stent technologies. Moreover, the use of higher molecular weight PCL reduces the number of free ends (carboxyl group) in the polymer chains, decreasing the number of electrostatic interaction sites useful for enhancing polymer-Mg adhesion [82].

One of the easiest way to retard the corrosion of the metal encapsulated in the PCL layers is to use thicker layer of the polymer. In this way, the path that the fluid has to do before reaching the Mg increases and consequently the life time of the device rises. This method has the disadvantage that thicker layers bring to higher stiffness of the layer and the whole device will be more rigid. On the other hand, a thicker Mg foil can be used. The use of thicker Mg trace can theoretically improve the duration of the device in the sense of increasing the time required for the water solution to corrode the Mg in a way in which the electrical properties are completely damaged. However, the exploitation of thicker Mg traces can induce encapsulation issues if the PCL layers are not thick enough. Moreover, the thickness of metal conductor should be equal or less than the skin depth of Mg for working frequencies between 10 MHz $\leq f_p \leq 433$ MHz is 33 $\mu m \geq \delta_{skin} \geq 5 \ \mu m$ meaning that the thickness should not overcome these limits [28].

Another effective way to increase the life time of a BED is to deposit a layer of oxide or nitride over the device [15]. The purpose of this method is to protect the metals with a further insulating layer made out a of material with a lower degradation rate. The most common inorganic materials for insulating and used as biofluid barrier are: silicon oxides (SiO₂), silicon nitrades (Si₃N₄) and magnesium oxides (MgO) [28].

These passivation layers can be deposited with different methodologies which generate different quality of layers. The quality of the generated layer, in terms of density and stoichiometry, highly influences the degradation rate. In fact, higher density and a favorable stoichiometry reduced the degradation rate of the covered metals [43]. Examples of deposition methods for silicon oxides and nitrides are summarized in the table 3.4 [43].

	Deposition methods	Dissolution rate $[nm/day]$	Stoichiometry	Density $[g/cm^3]$
C:O	Thermally grow	0.003-0.01	1:2	2.34
510_2	e-beam	0.1 10	1:2 1:2.2	$2.1 \\ 1.9$
$\rm Si_3N_4$	LPCVD PECVD	$\begin{array}{c} 0.16 \\ 0.85 \end{array}$	3:3.9 3:3.3	$3.1 \\ 2.5$

Table 3.4: Deposition techniques for silicon oxides and nitrades.

The techniques which guarantee a higher quality usually are slower and require higher operational temperatures. Other interesting deposition techniques are the atomic layer deposition (ALD) and the sputtering which implement lower temperatures (<100 °C) and so they can be used also for deposition on delicate polymers [28]. Finally, it is shown that a deposition of alternated oxides and nitrades can improve the water barrier capability thanks to the fact that possible defects on the deposited surface can be covered by the upper one [43]. The drawback of these techniques is that these material are dielectric and so they could change the electromagnetic properties of the device.

One common method adopted in transient electronics for insulating the conductive material from the water solution is the exploitation of natural waxes. They are used because of their biodegradability and easiness in the usage [80]. Different natural waxes are available for this purpose such as bees, carnauba, candelilla, soy and myrtle wax. Won *et al.* [80] tested the water barrier of the latter three, by encapsulating an Mgtrace (43 mm length, 300 μ m width and 200 nm thick) in a 300 μ m wax layer. From that test, it is depicted that candellila wax offers the higher resistance to the water penetration, giving a change in the resistance of the Mg trace after more than 7 days which is significantly different compared to the less than 1 day duration for the other two. It is also important to underline that the penetration barrier is highly influenced by the thickness of the wax layer. In fact, higher the thickness of the wax layer, slower the penetration of the water and consequently longer the life time of the device. In the same paper, the authors show that the resistance of the Mg trace is doubled after 4,7 and 11 days for a thickness of 100, 200 and 300 μ m respectively [80]. The encapsulation techniques consist in dipping the sample in the fused wax or spin coat the melted wax over the device [84]. Drawbacks are the melting temperature of the wax that can be too high for the polymer substrate and the general increment of the stiffness of the device due to the rigid behavior of the wax.

The last way to increase the duration of a device in a water ambient could be to use as encapsulation layer a polymer with a surface degradation process. For this type of polymers, the water starts to degrade the surface instead of penetrating into the whole substrate-encapsulation layers. Two examples of this kind of polymers are the polyanhydrides and poly(ortho esters). Choi *et al.* [12] demonstrated that a particular polyanhydride polymer (PBTPA) retards the corrosion of the Mg trace (300 nm thickness) up to 80 h compared to the few hours of other bulk erosion polymer (PLGA), for a thickness of 600 μ m. This time decreases for thinner layers (100 μ m protects for 13 h) while it increases for thicker Mg traces (6 day for 1.1 μ m). Moreover, the authors realized a bioresorbable wireless system to power an LED which, with a Mg receiving coil 50 μ m thick, keeps the LED working for more than 1 month [12]. The main drawbacks are the complex synthesis of such polymers and the fact that for becoming to express the surface degradation process these polymers have to overcome a critical thickness length (L_{crit}) which is 75 μ m for polyanhydrides and 600 μ m for poly(ortho esters) [76].For that reason, the L_{crit} could limit the use of these polymer if thinner layers are required.

All the presented methods are valuable alternatives for augmenting the water barrier capability of the device. Moreover, some of them can be combined in order to obtain superior characteristics. Although, particular attention has to be paid to the substrate upon which the metal conductor (i.e. Mg) lies. In fact if the substrate is a polymer, as in our case the PCL, it might not stand temperature higher than the its melting temperature $(T_m=60 \ ^{\circ}C)$. Another challenging aspect is the high reactivity of the Mg which limits the exploitation of techniques that require oxygen that could purposelessly oxidase the surface. Finally the stiffness of the final device has to be taken into account, especially for a device which needs to be bent (stent).

3.2.4 Corrosion test: proof of concept for water barrier enhancement

As mentioned before, there are several techniques to increase the water barrier of the device. A preliminary experiment for testing the retarding of the corrosion due to the contact of the water with the metal conductor is presented in this section. The experiment consists in create two samples, a reference one and a modified one, soak the them in a Saline solution with pH 7 and look at the degradation through microscope images at different time steps. In particular, two modifications are implemented and applied to the same sample: sputtering of 150 nm of SiO₂ over both sides of a squared Mg piece and the use of thicker PCL layers of 80 μ m as substrate and encapsulation.

Fabrication

Two rectangular pieces of 3x2 mm are obtained from an Mg foil of 10 μ m thickness (Goodfellow Corp., USA) using the laser cutter WS Turret Laser Micromachining Workstation (Optec). One sample, is sputtered with a 150 nm of SiO₂ on both sides using the AC450 sputtering machine (Alliance Concept, FR), while the reference one is left pristine. Then, the PCL layers are realized with the same process described in section 3.2.1. For testing higher thicknesses of PCL, the 30% w/v solution of PCL is spin coated at 150rpm to obtain 80um thick layers. The thickness has been measured by the profilometer DektakXT (Bruker). The layers are released via PSS dissolution in water. Finally,the reference Mg piece is encapsulated between two PCL layer of 40 μ m while the sputtered one in-between two 80 μ m thick PCL pieces as described in section 3.2.1.

Experiment and Results

The samples are dipped in the saline solution which is kept at 37 °C thanks to the hot plate and the temperature sensing and adjusting system (SuperNuova +, Thermo Scientific). Pictures of the modified and reference sample are taken at different time points: pre-dipping (0h), 1h, 2h, 3h, 4h, 5h, 6h, 7h and 24h. The images are shown in Fig. 3.6.

As can be seen from Fig. 3.6, the corrosion for the reference sample is already visible after 1h. On the other hand, the modified sample exhibit corrosion after 3h of immersion. During the subsequent time steps, the corrosion of the samples increased as expected, evidenced by the growing white areas which correspond to the release of H_2 and the accumulation of magnesium hydroxide (Mg(OH)₂).

This test verifies the effectiveness of these two techniques in increasing the water barrier effect. Tailoring the thickness of the SiO_2 and the thickness of the PCL could bring to an higher blocking effect.

For realizing the composing parts of wireless system (the coil) discussed in the next sections, the bare PCL and the Mg traces without SiO2 coating were used, as the focus of experiments was primarily the design. In this way, a starting point is set which enables a comparison for further improvements.



Figure 3.6: Corrosion test results at different time steps. Left sample: reference one without SiO₂ and PCL layers of 40 μ m. Right sample: modified one with 150 nm SiO₂ and PCL layers of 80 μ m.

Chapter 4

Inductive Coil: Fabrication and Design optimization

4.1 Fabrication: Optimization of the procedure

The main focus of this thesis is the development of a coil inductor made with bioresorbable elements. As presented in the previous chapters, the fabrication using transient materials is not trivial and requires specific techniques for avoiding any damage during the processes. In particular, the techniques used for realizing the device need to be tailored and adjusted using a try-and-error approach because satisfactory fabrication results are strongly dependent on the exploited materials characteristics such as the thickness, the oxidation rate and oxidative state of the surface and the dimensions of the device.

For these reasons, it is decided to start fabricating a reference coil in order to study the feasibility of the process, the critical fabrication steps and have a clearer idea of the parameters which make sense to vary in the design optimization section.

The inductor coil consists of two pieces of magnesium separated by an insulating inter-layer; this ensemble is encapsulated between two pieces of PCL. The two magnesium traces have different shapes: one is a circular and planar spiral, representing the main coil, while the other one has an L-shaped geometry. The electrical continuity beetween the two is given by a contact in the center of the spiral trace, where an opening of the inter-layer is cut out between the two metallic elements. The opening in the inter layer guarantees that the electrical contact is only in the center of the spiral and not where the L-electrode passes over the circular trace. Fig. 4.1 shows an out-of-scale representation of the stacked device from different points of view.

4.1.1 Conductive elements: spiral and L-electrode

The conductive elements are obtained from Mg foils (Goodfellow, US). For the first attempt, an Mg foil of 10 μ m thickness is selected. This thickness is in fact lower than the skin depth of Mg for frequencies spanning from 10 MHz to 100 MHz, where biological tissue attenuate less the power transmission. This will reduce the loss as presented in Chapter 2.2.2, especially the R_{skin} values.

An initial design of the two elements is drawn in AutoCAD. Fig. 4.2 depicts their





Figure 4.1: Coil Inductor Model. (a) Encapsulation model; (b) Front view: stacked composition of the device.

geometry.



Figure 4.2: First design of reference coil.

For the spiral element, the external diameter maximizes the available space and is equal to 3.3 mm. The width of the trace (w) and the gap between the loops (s) are equal to 50 μ m. The two values are decided to keep equal, as commonly done and reported in literature work [14] [31] [27] [45] [13]. The number of turns (n) is set as 7 to test the feasibility of a more complicated geometry compared to a single loop. For the L-electrode, a width of 200 μ m is used.

Once the geometry is created, the Mg foil is cut using the laser cutter WS Turret Laser Micromachining Workstation (Optec). The two main parameters to be set are the power of the laser (as a percentage of the full available power) and the number of times in which the laser repeats the cut (repetition object). Several combinations of these two parameters are tested in order to obtain good results. Fig. 4.3 shows a table of the parameters combinations and the respective results in terms of cutting profile. The images depict the spiral trace as it represents the more challenging geometry in the device fabrication.

The experimental results show that higher power and a lower number of repetitions reduce the duration of the process and the times in which the Mg is hit by the laser beam, decreasing the stress applied to the foil and avoiding its burning. In addition,



Figure 4.3: Laser cutter parameters and results for 10 μ m thick foil. (a) Table with parameters combination tests; (b) Results in terms of cutting profile.



Figure 4.4: Failing extraction of Mg spirals 10 μ m.

the Mg foil may slightly shift during the cut due to a non-perfect attachment of it to the hosting platform and this would result in an incorrect cut. With a faster process, also this phenomenon is less probable. The best combination of the parameters is: Power=8 and Rep.Obj=7.

Even if a good profile cut of the spiral is achieved, the spiral trace looks darker in correspondence of the loops, suggesting that the laser may induce a modification of the surface and a light burning effect. Once the best combination of the parameters is identified, the L-electrode is easily detached from the foil, while the coil requires time-consuming handling before release due to electrostatic forces. Even though extra cutting lines have been added to the design to facilitate the extraction (as can be seen from Fig. 4.3 b), the overall structure is too fragile to be safely manipulated without breaking or heavily deform it. Example of failing extractions are reported in Fig. 4.4

The burning effect is thought to be induced by the excessive proximity of the cutting lines in the loops $(50\mu\text{m})$ while the fragility is thought to be caused mainly by the thickness of the used foil and the *w-s* dimensions which are too small for allowing an easy handling. For these reasons, it is decided to use an Mg foil of 20 μm thickness and to increase to 60 μm the width *w* and the gaps *s*. At this point, the selection of the parameters for a good cut is faster thanks to the acquired experience. The table and the figure with the result of the cutting profile are presented in Fig. 4.5.

The optimized parameters values for obtaining a satisfying cut are: Power=8 and Rep.Obj=11. The spiral trace and the L-electrode piece obtained from the Mg foil of 20 μ m with these parameters are perfectly shaped and easily detachable from the Mg foil. Furthermore, the increased thickness and the greater width facilitate the handling for the next fabrication steps.



Figure 4.5: Laser cutter parameters and results for 20 μ m thick foil. (a) Table with parameters combination tests; (b) Results in terms of cutting profile.

4.1.2 PDMS wafer: support for the construction of the device

The coil inductor device has a layered structure and its construction involves subsequent steps in which the composing elements (PCL substrate, spiral trace, inter-layer, L-electrode and encapsulation layer) are overlapped one after the other. At some points during the fabrication, melting processes are required to fix and encapsulate the underlying elements. In order to be able to see clearly the melting of the PCL and do not overexpose the sample to the melting temperature, the device is fabricated over a transparent glass wafer.

Moreover, to ensure the stability of the stacked structure during the assembling of the device, the glass wafer is covered by 100 μ m thick Polydimethylsiloxane (PDMS) layer, which ensures a good adhesion with the PCL substrate layer thanks to its sticky behavior. Fabrication of the PDMS-covered glass wafer starts with the realization the PDMS soultion, by mixing ratio base-hardening agent of 9:1. After a mixing the polymer base and its curing agent at a ratio of 9:1, respectively (Sylgard 184 Dow Corning). After mixing in an automatic mixer and degassing (Thinky, USA), the solution is poured over the 4-inch glass wafer it is spin coated at 660 rpm for 30 s. Then, the wafer with the PDMS layer is cured for 2 h at 75°C. In this way, a good support is guaranteed for the fabrication of the coil. An example of glass wafer with a spin coated PDMS layer is presented in Fig. 4.6.

4.1.3 Substrate and Encapsulation layer

To facilitate the positioning and the alignment of the coil on the substrate and firmly encapsulate it between two PCL layers, a replica molding technique is used to pattern both substrate and encapsulation with the coil geometry. The molds are fabricated in SU8 GM1070 photoresist (Gersteltec, Switzerland), which is spin coated at 1500 rpm over 4-inch wafer previously cleaned with plasma for 10 min at 800 W (Pink, Germany). After a soft bake (2 °C temperature ramp until 65 °C, 15 min at 65 °C, 2 °C temperature ramp until 95 °C, 2 h at 95°C and cooling again to ambient temperature), the wafer is exposed using MLA 150 machine (Heidelberg Instruments). The MLA 150 allows a mask-less exposition using a 405 nm laser controlled and modulated by a digital micromirror device [8]. In order to assess the best values of exposition dose and focus given the specific design of the coil, a test is carried out varying the dose from 150 to 900 mJ/cm² with steps of 50 mJ/cm² and the defocus is varied from -10 to 10 with steps of 2. Fig. 4.7 shows the CAD design created for the substrate mold and few examples of



Figure 4.6: Glass wafer with PDMS. Adhesion between PDMS and a PCL encapsulating a rectangular piece of Mg.

the SU8 patterning achieved after the post-exposure bake, development in Propylene glycol monomethyl ether acetate (PGMEA) and the final hard bake.



Figure 4.7: Dose-Defocus test. (a) Geometry of the substrate pattern; (b) $250 \text{ mJ/cm}^2 \& 4 \text{ defocus}$; (c) $450 \text{ mJ/cm}^2 \& 0 \text{ defocus}$; (d) $500 \text{ mJ/cm}^2 \& -10 \text{ defocus}$; (e) $800 \text{ mJ/cm}^2 \& 4 \text{ defocus}$.

The best combination of parameters is found to be: Dose=250 mJ/cm² and Defocus=4. Once the mould is ready, a PSS release layer is spin coated and cured over it. Subsequently, the 30% w/v PCL-Chloroform solution is spin coated over it (150 rpm, 80 μ m) and baked for 30 min at 80 °C to remove the internal stress. After the complete evaporation of the solvent under the hood, the patterned PCL is detached from the mold with DI water.

At this point, positioning of the spiral coil in the obtained grooves is attempted but some problems occur. Although the substrate grooves have a height of 40 μ m and the Mg has a thickness of 20 μ m, they are not able to keep it in position in a satisfactory way. Moreover, the alignment of the conductive trace in the patterned grooves is really hard due to the electrostatic force that develop between the spiral trace and the PCL substrate layer. Plus, even though using a thicker Mg foil improved the coil extraction from the foil and its handling, occasional deformations occur, which hinder the coil fitting inside the patterned grooves of the PCL. If a good alignment cannot be ensured, the presence of the grooves is completely useless and even dangerous for the structural stability. In fact, without a perfect alignment of coil and substrate, alignment of the encapsulation layer is difficult, allowing for the creation of entrapped air during the melting process. These areas induce a weakening of the structure and would theoretically reduce the water barrier capability introducing preferential paths for the water penetration. Furthermore, the mold preparation is a long and time-consuming process which would limits its application in the case several geometry wanted to be tested.

For all these reasons, replica mold technique won't be used and the tested coils will be encapsulated between two not patterned PCL layers. The fabrication of these layers is the same as mentioned before for the patterned one, the only difference is that it is not spin coated over the SU8 mold but over a simple 4-inch Si wafer.

4.1.4 Interlayer

For avoiding the contact between the two conductive traces (the spiral and the Lelectrode) in a position other than in the center of the spiral, an inter-layer is exploited. Fig. 4.8 shows a representative example of the inter-layer.



Figure 4.8: Inter-layer geometry and obtained result. (a) Inter-layer geometry; (b) Cut PCL inter-layer.

The geometry is created on AutoCAD in order to cover completely the loops of the trace but leaves uncovered the center of the spiral where the circular pad is placed for the contact with the L-electrode. Initially, an inter-layer of thickness equal to 20 μ m is used to minimize the step height between the connection pad of the spiral and the L-electrode. This decision is limited by the achievable thicknesses using highly viscous PCL-based solutions. Moreover, it is motivated from the fact that the Mg spiral trace tends to partially sink in the PCL substrate layer during the melting process so an

inter-layer height of 20μ m should be enough. To obtain such thickness value, PCL-Chloroform 30% w/v solution is spin coated with a angular velocity of 2000 rpm over a 4-inch Si Wafer, which is pre-treated with the PSS release layer for the detachment in DI. Then, the inter-layer is generated through the cutting of the PCL spin coated layer using the AutoCAD design and the laser cutter machine (Optec). The detachment of the circular ring inter-layer is possible using DI water, dissolving the PSS.



Figure 4.9: Inter-layer melting. (a) 20 μ m thick; (b) 40 μ m thick.

However, Fig. 4.9 (a) shows that 20μ m are not enough to ensure correct insulation of the spiral trace, as can be seen from the shiny color of the Mg trace suggesting that it is uncovered. Therefore, a 40 μ m inter-layer might be preferred and it can be easily obtained by adjusting the spin coating rpm to 800rpm, as done already previously. The result of the insulation is reported in Fig. 4.9 (b).

4.1.5 Assembling and connection of the inductor coil

Now that all the constituting parts of the device are obtained, the assembling and the connection are presented.

Fig. 4.10 summarizes the steps of the procedure. The process starts with placing the PCL substrate layer on the glass wafer with the spin coated PDMS. Then, the spiral trace is positioned over it and the inter-layer is aligned so that it covers correctly the turns (Fig 4.10 a). At this point, the first melting is performed to fix the structure and avoid the movement of the coil, limiting the risks of additional deformation of the trace (Fig 4.10 b). The substrate-spiral trace-inter layer stack is kept in position during the melting with a Polyethylene terephthalate (PET) foil which exerts a strong adhesion with the PDMS layer of the glass support wafer. Moreover, before the melting process the glass wafer is put in a vacuum machine (EasyVac Pro, Solis) for improving the contact between the layers of the stack and reducing the amount of air between them. In fact, if the layers are not well attached to each other and to the PET foil there would be an incorrect sealing due to the creation of air bubbles which introduce areas of mechanical instability. The melting is performed by putting the glass wafer with the stacked structure over a preheated hot plate at 60°C. The process can be



Figure 4.10: Assembling and connection process. (a) Inter-layer positioning ; (b) Inter-layer melting; (c) L-electrode connection; (d) final encapsulation (front); (e) fibal encapsulation (back); (f) SMA connector connection.

easily controlled by looking at the melting PCL. For ambient temperature, the PCL has a light white color, while when it is melting it starts to become transparent. The sealing procedure can be considered finished when the two pieces of PCL are completely transparent. Usually, for the thicknesses used in this work 10-20 s are necessary over the hotplate. After that, the glass wafer is removed from the hot plate and let cool down at ambient temperature until the PCL regains the light white color. The PET layer is removed and the connection of the L-electrode with the spiral trace is done (Fig 4.10 c). The electrical connection is ensured by few drops of silver conductive paste (G3692 Acheson Silver DAG 1415, PLANO GmbH) which are manually placed using an insulin syringe (BD Micro-fine +, 30G). This is the only non-resorbable element of the device. The Ag paste is let dry until the complete evaporation of the solvent. Then, the encapsulating PCL layer is placed on top of the stack, letting the two unconnected extremities of the spiral trace and L-electrode uncovered for the further connection with the SubMiniature version A (SMA) connector. The second and last melting is carried out with the same procedure explained before (Fig 4.10 d-e). Finally, the SMA connector is connected to the device using silver paste (Fig 4.10 f).

4.2 Simulation and Design

The fabrication procedure identifies some limitations to the achievable geometries. For example, it depicts the need of an Mg foil which has a thickness of 20 μ m. Moreover, the smaller width and gap realizable with a good resolution and without burning the Mg foil is found to be 60 μ m. Based on the fabrication limitations, simulations using the software ANSYS HFSS are performed to extract the best geometries for the device.

The quality of the coils is evaluated upon three main aspects:

- Quality factor Q;
- Inductance value L;
- Self resonance frequency f_{SRF} .

In particular, as presented in Chapter 2, the Q factor is one of the most important parameter for evaluating the goodness of a coil inductor. It resumes the electromagnetic properties of the coil and higher its value, better its power transfer behavior. The second parameter, the value of the inductance, is important because an high inductance implies a good magnetic coupling and also allows for a smaller value of matching capacitor. Finally, the value of the self resonance frequency, f_{SRF} , has to be considered because it imposes a limit to the higher possible working frequency that can be used.

The simulation environment consists in a cubic box of air (500 mm side) to which the radiation boundary condition is applied. The planar circular coil is placed at the center of this box and it is totally encapsulated in a layer of PCL with a total thickness of 200 μ m. In order to evaluate the scattering parameters of the simulated coil, a lumped port excitation (R_{in} = 50 Ω) is applied to the two ends of the coil and frequencies from 10 MHz to 5 GHz are evaluated. Under the PCL layer, a piece of FR4 is placed with the aim to simulate the support upon which the device will lie for the experimental measurements. An example of the simulated coil is reported in Fig.4.11.



Figure 4.11: Example of simulated coil.

For all the simulations, the thickness of the coil and the outer diameter is kept constant and with a value of 20 μ m and 3.3 mm, respectively. The geometrical properties varied in the simulation are: the number of turn (n), the width of the trace (w) and the gap between the loops (s). The variations of the parameters are summarized in table 4.1. The incorrect coils generated from an unfavorable combination of n, w and

Parameter	Range [start:step:end]
\boldsymbol{n}	3:1:9 turns
$oldsymbol{w}$	0.06:0.02:0.14 mm
S	0.06:0.02:0.2 mm

 Table 4.1: Range of parameters variation.

s are excluded using the Matlab script reported in the Additional material, as well as the parameterization implemented to create the coil.

Analyzing the results coming from the simulations, some trends can be seen. In particular, for all the coils with fixed n:

- increasing the width w leads to a shift of the f_{SRF} to higher frequency and a reduction of the induction L;
- increasing the gap s leads to a shift of the f_{SRF} to higher frequency and a reduction of the induction L;
- for coils with a high number of turn (n=6,7,8,9), the quality factor is bigger for traces with smaller widths (w=0.06-0.1 mm);
- for coils with a lower number of turn (n=3,4,5), the quality factor is bigger for traces with bigger widths (w=0.12-0.14 mm).

A graphical representation of these trends is shown in Fig. 4.12 and Fig. 4.13. Ansys HFSS is able to calculate the scattering parameters of the coil and, using the equation and theory presented in Chapter 2, all the other parameters such as the real and imaginary part of the impedance, the inductance and the Quality factor can be derived.

In order to highlight the shifting of the f_{SRF} and the reduction of the inductance, the simulated values of the inductance L of two representative coils are reported. In particular, Fig. 4.12 a shows the L plots for a coil with 7 turns (n=7), a constant gap equal to 0.06 mm (s=0.06 mm) and a varying width. On the other hand, Fig. 4.12 b depicts L plots for a coil of 7 turns but this time the width is constant with a value of 0.06 mm (w=0.06 mm) and the gap is varying. The zero-crossing point corresponds to the f_{SRF} value while the inductance values are reported in the zoomed graph. The value of the inductance L changes in function of the frequency but is almost constant for $f < f_{SRF}/10$. For that reason, it is decided to display the L in correspondence of $f < f_{SRF_{min}}/10$ with $f_{SRF_{min}} = 1.07GHz$ the lowest self resonance frequency of the considered coils which for both plots is the coil with n=7, s=0.06 mm and w=0.06mm. As anticipated, the increase of the width (from 0.06 mm to 0.12 mm) brings to a shift of the SRF (from 1.07 GHz to 1.33 GHz) and a reduction of L (from 150 nH to



Figure 4.12: Trends: varying width and gap. (a) Fixed number of turn and gap, varying width; (b) Fixed number of turn and width, varying gap.

81.3 nH) as well as the rise of the gap (from 0.06 mm to 0.14 mm) induces a shift of the SRF (from 1.07 to 1.53 GHz) and a decrease of L (from 150 nH to 79.3 nH).

For what concerns the last two trends, Fig. 4.13 shows examples of these behaviors. The plots in this figure report the Quality factors in function of the frequency (x-axis) and the width (y-axis) for a fixed gap. More specifically, the plot a presents the Q for the coil with n=8, s=0.06 mm; the b shows the Q for n=7 and s=0.06; c depicts Q for the coil n=4 and s=0.12 mm; d displays Q for n=3 and s=0.1 mm. It is clear that the first peak of the Q factor, the one that occurs before the SRF and so the one in which we are interested, is higher for small widths for coil with higher number of the turns while is higher for big width for coil with low number of turns.

These general trends are valuable for all the simulated coils but punctual considerations must be done in order to find the best geometries in the meaning of the higher Q factor and inductance. Due to the high number of simulations, it is decided to separate the results considering the best geometries for coils with a certain number of turn and



Figure 4.13: Trend: number of turn-width. (a) and (b) coil with high number of turns; (c) and (d) coil with low number of turns.

than compare these to find the more interesting ones. The results of the best simulated coils are summarized in the Fig. 4.14.

In particular, this table shows: the dimensions of the gap and width of the coils with fixed number of turns which maximize the Q factor and the inductance for a working frequency of 433 MHz, which is the suggested one for maximize the power transfer and respect the SAR conditions [1]; the values of the Q factor in correspondence of f = 433 MHz; the self resonance frequency f_{SRF} of the coil; the value of inductance (L) in correspondence of f = 433 MHz. For the coil with 9 turns, the geometry which maximizes the Q factor is the same that maximize the L.

As formerly mentioned, an optimized coil would have: the highest Q factor for a good wireless power efficacy; a high inductance to improve the coupling and especially to reduce the matching capacitance needed to match the receiver tank resonance with the transmitter ones; a f_{SRF} high enough to guarantee the possibility to use as working frequency a frequency up to hundreds of MHz. Regarding the f_{SRF} , all the simulated coils show a self resonance frequency higher or approximately equal to 1 GHz, ensuring the exploitability of a working frequency of hundreds of MHz. Unfortunately, the coils which maximize the quality factor are also the same with the lowest inductance. In fact, the coils which express the highest Q are the ones with low n and a high w, while the inductance is maximized for coil with a high n and small w and s. This could be

n turns	Maximization	Dimensions [mm]	Q factor @ 433 MHz	fsrf [GHz]	L @ 433 MHz
9	Q factor & Inductance	s=0.06 & w=0.06	57.47	0.98	227 nH
0	Q factor	s=0.1 & w=0.06	60.39	1,24	129 nH
0	Inductance	s=0.06 & w=0.06	59.82	0.98	205 nH
7	Q factor	s=0.06 & w=0.08	65.52	1.15	141 nH
/	Inductance	s=0.06 & w=0.06	62.87	1.07	178 nH
C	Q factor	s=0.1 & w=0.1	71.43	1.66	72.9nH
D	Inductance	s=0.06 & w=0.06	64.09	1.02	147 nH
c	Q factor	s=0.1 & w=0.1	72.82	1.78	65 nH
5	Inductance	s=0.06 & w=0.06	63.77	1.13	113 nH
4	Q factor	s=0.1 & w=0.14	76.94	2.18	40.9 nH
4	Inductance	s=0.06 & w=0.06	57.58	1.43	73.6 nH
3	Q factor	s=0.1 & w=0.14	81.00	2.32	30.9 nH
	Inductance	s=0.06 & w=0.06	61.11	1.63	50.1 nH

Figure 4.14: Best simulated coils: Q & L maximization. The green rectangles show the selected geometries.

explained by the fact that the increase of the inductance is linked to an increase of the number of turns but more loops imply a longer traceas well, producing higher R_{DC} , R_s and C_p and so decreasing the Q factor.

Moreover, if it is considered the Q factors difference between Q maximized geometries and L maximized geometries for a fixed number of turns (Δ_Q) at 433 MHz, it can be seen that for a high number of turn (9, 8 and 7 turns) this value is small while for a low n (6, 5, 4, and 3) the difference increase. The Fig. 4.15 shows these findings.

In particular, for coils with high n, the differences between the Q factors for the geometries which maximize the Q and the L are small (lower than 2.65) while the variations of L (Δ_L) are important (around 76 nH) and so the geometry which maximize the L should be selected for this kind of coils. On the other hand, the coils with a low number of turns shows a bigger difference in terms of Q factors between the geometries which maximize the Q and L (from 9.05 to 19.89) and so they are not comparable in terms of Q factors. For this reason, geometries which maximize Q should be chosen for the coils with low number of turn.

In conclusion, since the geometries which maximize the Q and L do not match, in this work both the conditions are fabricated and tested to find the most promising for the IPT.

More specifically, it is decided to realize:

Q maximization coil with 3 turns, width=0.14 mm and gap=0.1 mm;

L maximization coils with 7,8,9 turns, width=0.06 mm and gap=0.06 mm.

n turns	Maximization	Dimensions [mm]	Δα @433 MHz	Δ∟@ 433 MHz	
9	Q factor & Inductance	s=0.06 & w=0.06	١	١	
0	Q factor	s=0.1 & w=0.06	0.57	76 nH	
8	Inductance	s=0.06 & w=0.06	0.57		
7	Q factor	Q factor s=0.06 & w=0.08		27 - 11	
/	Inductance	s=0.06 & w=0.06	2.65	37 NH	
c.	Q factor	s=0.1 & w=0.1	7.24	74.1 nH	
0	Inductance	s=0.06 & w=0.06	7.34		
r	Q factor	s=0.1 & w=0.1	0.05	40 -11	
5	Inductance	s=0.06 & w=0.06	9.05	48 NH	
4	Q factor	s=0.1 & w=0.14	10.20	22.7 -11	
4	Inductance	s=0.06 & w=0.06	19.36	32.7 nH	
3	Q factor	s=0.1 & w=0.14	10.00	10.2 -11	
	Inductance	s=0.06 & w=0.06	19.89	19.2 nH	

Figure 4.15: Difference between Q factors and L.

Chapter 5

Inductive Coil: Experimental testing and Discussion

Starting from the optimized geometries, coils are fabricated following the protocol described in section 4.1.5. Fig. 5.1 shows the fabricated coils.



Figure 5.1: Tested coil. (a) coil 7 turns; (b) coil 8 turns; (c) coil 9 turns; (d) coil 3 turns. Red rectangle: L-maximizing coils; green rectangle: Q-maximizing coil.

5.1 Network Analyzer measures

After the connection of the two leads of the coils to the SMA connector, the coils are ready to be tested. The first test consists in the evaluation of the quality factor, the self resonance frequency and the inductance of the coils. These measures are taken via the ZVL Network Analyzer (Rohde&Schwarz, Germany) by connecting the fabricated coil to the port one. The network analyzer (NA) is calibrated using the calibration kits ZV-Z132 and ZN-Z135. This kind of measures ensures the extraction of the sparameter (S11) and the corresponding derivation of the impedance of the coils as well as the quality factors and the inductances.

5.1.1 L-maximizing coils: Results

The fabricated L-maximizing coils (7,8 and 9 turns) are tested scanning frequencies from 5 kHz to 5 GHz. The results in terms of quality factors are reported in Fig. 5.2. In particular, only Q factors relative to frequencies from 5 kHz to 1 GHz are plotted to be able to better appreciate the behavior of the coils for frequency under the simulated self resonance frequency which is for all these coils approximately equal or less than 1 GHz. In any case, for frequencies bigger than the f_{SRF} the inductor coil starts to behave like a capacitor and so they are useless for the IPT application.

Looking at the plots in Fig. 5.2, it is clear that the simulated curves are greatly different from the experimental ones. For what concerns the f_{SRF} , the measured values are:

- coil 7: $f_{SRF_{fab}} = 293$ MHz against $f_{SRF_{sim}} = 1.069$ GHz;
- coil 8: $f_{SRF_{fab}} = 297$ MHz against $f_{SRF_{sim}} = 976$ MHz;
- coil 9: $f_{SRF_{fab}} = 280$ MHz against $f_{SRF_{sim}} = 984$ MHz;

Moreover, the peak of the Q factor curve for the coil with 7 turns is approximately one order of magnitude less than the simulated one and the situation is even worse for the coils with higher n.

Regarding the inductances, the plots in Fig. 5.3 shows the obtained values for the fabricated coils against the simulations, with a zoom spanning from 0 to 200 MHz. The $f_{SRF_{fab}}$ is lower than the simulated one so the inductance values should be compared in correspondence of frequencies lower than 300 MHz (433 MHz is not an option anymore because is $> f_{SRF_{fab}}$).

In order to understand the effectiveness of the number of turns on the variation of L for the fabricated coil, it is decided to consider an operating frequency lower than $f_{SRF_{fab}}/10$. In particular, it is decided to consider a frequency equal to 27 MHz which is included in the ISM bands. A higher available frequency could be used (168 MHz or 80 MHz) but for them the value of the inductance is greatly influenced by the self resonance and its increase could be due to this effect rather than effective benefits induced by a higher number of turns. As depicted from the plots, the inductance for the coil with 7 turn is bigger than the simulated one: 177 nH against 150 nH. On the contrary, when the number of turns increases there is not the rise that would be expected. In fact, the inductances for the fabricated 8 and 9 turns coils have a value



Figure 5.2: Q factor plots of L-maximizing coils: 7, 8 and 9 turns.

of 147 nH and 140 nH against the simulated values of 170 nH and 187 nH respectively. In essence, the increase of the number of turns does not correspond to a rise of the inductance for the considered frequency.

5.1.2 Q-maximizing coil: Results

The realized Q-maximizing coil (3 turns) is tested via the NA analyzing a range of frequency from 5 kHz to 5 GHz, as for the L-maximizing coils. In this case, all the scanned frequencies are reported in Fig.5.4 thanks to the higher $f_{SRF_{fab}}$ of this coil.

From this graph, it is evident that even in this case the two curves (simulated and experimental) are different but some observations can be done. More in detail, the two curves of Q factors have the peak in the same frequency range (800-900 MHz) and the value for the fabricated coil is 19 while for the simulated one is 89. In addition, the self resonance frequency of the realized $(f_{SRF_{fab}})$ coil is more in accordance with the



Figure 5.3: L plots of L-maximizing coils: 7, 8 and 9 turns with their respective zooms.

simulation $(f_{SRF_{sim}})$ in respect to the L-maximizing coils. In fact, the $f_{SRF_{fab}}$ is equal to 2.064 GHz against $f_{SRF_{sim}}$ which is 2.324 GHz.

Talking about the inductance, the graph of the simulated and the fabricated Q-maximizing coil is reported in Fig. 5.5.

Thanks to the fact that the self resonance frequency of the fabricated coil is around 2 GHz, the inductance of the coil is almost flat for the frequencies range in which we are more interested for selecting the working frequency (from 13.5 MHz to 433 MHz). In particular for a working frequency $f_o=433$ MHz, the inductance of the fabricated coil is 42.45 nH against 30.39 nH of the simulation.

5.2 Discussion

The first thing that capture the attention is the difference that exists between the simulation and the fabricated coil. Assuming that the simulated coils are a simplified







Figure 5.5: L plot of Q-maximizing coil: 3 turns with its respective zoom.

versions of the fabricated one and so a perfect match is not possible (no L-electrode and no SMA), it would have been expected to have a better concordance between the simulation and the experimental results, especially for the L-maximizing coils. Some motivations are presented to try to explain this discordance.

Firstly, the fabricated coils are obtained from the Mg foil which may present an oxide layer on the surface which can decrease its conductivity and consequently its electromagnetic behavior. In addition, the contact between the spiral trace and the L-electrode is made exploiting the conductive silver paste while for the simulation it is supposed to be with magnesium. This contact could be not perfect (ohmic contact) and induce some loss effects increasing the C_p . The same problem is identifiable for the contacts between the two ends of the fabricated coils and the SMA connectors. Furthermore, the silver paste is dissolved in an alcholic solution (MEK/Diacetone alcohol 2:1) which may slightly corrode the Mg traces. The handling of the spiral traces of the coils is not trivial, especially for the L-maximizing coils which present several turns and small width and gap, which results in a less rigid and more deformable structure. This difficulty is translated in fabricated coils with geometries that are not exactly equal to those simulated, especially for the gap values as depicted in Fig. 5.1. The last individuated problem could be the thinning of the PCL layers during the melting. As mentioned in section 4.1.5, the stacked structure of the coil is compressed with

a vacuum machine in order to improve the contact between elements and reduce the formation of air bubble. During the melting process, this squeeze brings to the thickness reduction of few microns (1-10 μ m) of the device which is not considered in the simulation.

Given these motivations, it is also understandable why the simulation for the Q-maximizing coil is more in accordance with the experimental results. In fact, this coil has just 3 turns, a larger width equal to 0.14 mm and a 0.06 mm gap which guarantee a more rigid structure and avoid the deformation of the geometry during the handling and the encapsulation.

The purpose of the simulations is to have a general idea about the behavior of the spiral, the next step is to understand which of the realized spirals could be the most promising for the inductive power transfer.

In order to do this, a comparison of the Q factors and the inductances between the tested coil is performed. As previously mentioned, both the Q factor and the L are function of the frequency so it is also necessary to identify the best working frequency f_o for the coils. The main aspect that limits the f_o is the f_{SRF} of the coil. In fact, the f_o should be selected among the ISM bands which are sufficiently small in respect to the self resonance of the coil. In this way, the coil behaves like an inductor and not as a capacitor. Usually, f_o is selected as $f_{SRF}/10$ or in any case it should be chosen a value for which the effect of the resonance is negligible. The Fig. 5.6 resumes the Q factor and the L values of the tested coils for the frequency in the available ISM bands of interest. For the L-maximizing coils, frequencies higher than 27 MHz are neglected because the resonance effect becomes important as depicted from the increment of L in Fig.5.3. For the Q-maximizing coil, it is possible to consider the frequency up to 433 MHz due to the fact that its self resonance frequency is higher than 2 GHz. In fact, even if 433 MHz is not one order of magnitude less than f_{SRF} , the effect of the resonance is not important as depicted from the L plots in Fig.5.5.

	Coil	@ 13,5 MHz	@ 27 MHz	@ 40 MHz	@ 84 MHz	@ 168 MHz	@ 433 MHz		Coil	@ 13,5 MHz	@ 27 MHz	@ 40 MHz	@ 84 MHz	@ 168 MHz	@ 433 MHz
	7 turns	1,386	2,719	/	/	/	/		7 turns	177 nH	177 nH	/	/	/	/
0	8 turns	0,148	0,305	/	/	/	/		8 turns	143 nH	147 nH	/	/	/	/
Q	9 turns	0,207	0,454	/	/	/	/	L	9 turns	133 nH	140 nH	/	/	/	/
	3 turns	0,671	1,311	1,844	3,628	6,752	13,63		3 turns	42,98 nH	43,02 nH	42,33 nH	41,74 nH	41,50 nH	42,45 nH

Figure 5.6: Q factor and L values of fabricated coil for ISM bands of interest.

The criteria exploited to evaluate the best coil is to select the coil which has the highest Q factor and the highest inductance for the selected working frequency. The inductance is also evaluated in terms of the corresponding matching capacitor which is required to shift the resonance of the receiver LC tanks (the electrical circuit resulting from the connection of the receiving coil with the matching capacitor) at the selected working frequency. In fact, transmitter LC tank and the receiver LC tank should resonate at the same frequency (the selected working frequency) in order to maximize the IPT efficiency. The LC tanks are obtained placing a capacitor in series or parallel

to the inductor. For the receiving coil, it is needed to add a capacitor in parallel and its dimension is easily calculable as: $C_m = 1/(L_{f_o}(2\pi f_o)^2)$ where L_{f_o} is the inductance of the coil at the working frequency f_o . A low value of the C_m is interesting for microfabricated capacitors because it allows to have capacitor with small dimensions (area of the plates and thickness of the dielectric layer) which for our application is a really important aspect due to the restricted available space.

Talking about the L-maximizing coils (7, 8 and 9 turns coils), the operational frequency at which correspond the maximum of Q and L is 27 MHz. However, the Q factor is limited to 2.719 for the 7 turns coil while for the 8 and 9 turns coils the values are 0.305 and 0.454 respectively. The inductances are decreasing for 7,8 and 9 turns coils with values equal to 177 nH, 147 nH and 140 nH respectively. This decrease of the inductance with the increase of the number of turns is not theoretically expected but it could be explained with fabrication problems.

As previously reported, the geometries of the L-maximizing coils in terms of gaps between the turns are not constant resulting in a not perfectly concentric spiral. This geometrical deformation induces a variation of the parasitic capacitance which is generated from the alternation of the conductive trace and encapsulation material (PCL) and this varies the overall inductance of the coil. The geometry deformation is clearly visible in Fig.5.1 and it is especially present for the 8 and 9 turns coils for which the experimental inductances result smaller than those simulated.

The capacitance of the matching capacitor that would be needed to shift the resonance of the LC receiver tank towards a $f_o=27$ MHz can be calculated for these coils:

- coil 7 turns: $C_m = 196.3 \text{ pF};$
- coil 8 turns: $C_m = 236.4 \text{ pF};$
- coil 9 turns: $C_m = 248.2 \text{ pF};$

Considering now the Q-maximizing coil, it can be seen that the Q factor reaches the value of 13.63 for the available working frequency $f_o=433$ MHz. On the other hand, the experimental inductance of the coil is 42.45 nH. Finally, the theoretical capacitance of the matching capacitor required to bring the resonance of the LC tank to $f_o=433$ MHz is $C_m=3.2$ pF.

Considering these results, the most promising coil is the Q-maximizing coil: n=3, w=0.14 mm and s=0.06 mm. In fact, it allows to use a f_o =433 MHz, it shows the higher Q factor and it requires smaller matching capacitor for the resonance shifting. For all this reasons, it is decided to make the preliminary inductive power transfer test with this coil.

5.2.1 Preliminary IPT test

In order to test the ability of the selected coil to generate an electrical signal with an inductive coupling, the preliminary IPT test is performed.

Firstly, the matching capacitor for the fabricated receiving coil is selected from commercial ones. From calculations reported previously, the theoretical values for the capacitance would be $C_m = 3.2$ pF. The commercially available capacitor with the closest capacitance is 3.3 pF and this one is selected. The next step is to connect the capacitor in parallel to the coil. This is made using a soldering station (Weller) and the tin. Given the high temperature needed for soldering, the capacitor is attached to the SMA and not on the PCL-based device. The Fig. 5.7 shows the coil with the connected C_m .



Figure 5.7: Receiving coil and matching capacitor. (a) Before connection; (b) after connection.

The LC receiver tank is then reconnected to the NA in order to verify the shifting of the resonance. Fig. 5.8 depict the result.



Figure 5.8: Receiver LC tank characteristics: real and imaginary part of the impedance (Z) and return loss (S_{11}) .

As can be seen from 5.8, there is the appearance of a resonance at lower frequency, highlighted by the green circle in the plot. However, the new resonance is at f=350 MHz and not at f=433 MHz, as hoped introducing the C_m . This is thought to be mainly due to:

- use of a matching capacitor which has a different capacitance with respect to the calculated one;
- modification of the coil device during the soldering step;

• not perfect contact between the matching capacitor and the SMA connectors pads.

In fact, even if it is tried to make the soldering process as fast as possible and involving the SMA pads, hundreds of degrees are needed to allow the soldering and the device results slightly damaged in the area near the C_m connection, as depicted in Fig.5.7.

Another interesting aspect is that the return loss S_{11} has a minimum near to the new resonance of the LC tank. On the basis of the theory reported in chapter 2, this is a positive and desired aspect because the use of a frequency corresponding to that minimum coincides with a minimization of the reflected power and consequently an increase in the transmitted one. The exact position of the S_{11} minimum is 218 MHz with a value of -4.2 dB while for the resonance frequency of the receiver tank (350 MHz) is around -2.1 dB.

Although the resonance of the receiver tank is not at 433 MHz, which was the selected working frequency for this test, an explorative test of IPT is carried out. For the transmitter element, a squared single loop coil of 10 mm side and 1 mm wide is used. This was already fabricated for a previous work [24]. The transmitter already presents a matching capacitor for the LC resonance shifting at 433 MHz.

The experimental setup consists in connecting the transmitting coil to the MXG vector signal generator (Keysight, US) to generate the input signal. On the other hand, the receiving coil is connected to InfiiVision DSO-X oscilloscope (Keysight, US) to record the generated signal.

The input signal has an amplitude of 19.00 dBm, which corresponds to ≈ 5.6 V_{pp}. The frequency is varied to verify for which frequency there is the maximum of transmission.

The Fig.5.9 display the manual alignment of the two coils (≈ 1 cm distance) and the received signal measured by the oscilloscope.



Figure 5.9: IPT measure: (a) alignment of the transmitter and receiver; (b) Oscilloscope measure.

As expected, the higher received signal is obtained for an input signal frequency equal to 350 MHz, which corresponds to the receiver LC tank resonance frequency. The amplitude of the received signal is $\approx 280 \text{ mV}_{pp}$. For a frequency of 433 MHz, the voltage decrease to a value of $\approx 200 \text{ mV}_{pp}$. This value could be greatly improved.

First of all, the two LC tanks are not perfectly matched because the receiver's resonance is 350 MHz against 433 MHz of the transmitter. A better matching should be achieved for the future tests, improving the quality of the coil- C_m contact and solving the problem correlated to the deterioration of the coil device induced by the high temperature of the soldering. Secondly, the geometry of the transmitter is not optimized for the designed receiver coil in terms of dimensions, reducing the overall IPT efficiency and making a good alignment difficult. The optimization of the transmitted coil could be done again in Ansys HFSS considering at this time the overall transmission system. This modification and correction can be done in the future development of the project. Moreover, in the feature developments the commercial and non-bioresorbable capacitor will be substitute with a microfabricated and bioresorbable one.

Chapter 6

Future Perspectives

6.1 Coil contacts and evaporated Mg

The realization of the coils brings in evidence some aspects that could be improved. For example, the conductive contact between the coil and the SMA is done via silver paste, whose solvent might damage the underlying Mg and affect measurements; Mg reacts quickly to air and a thin oxide layer on its surface might develop and might impact the experiments results; the variation of the PCL thickness during the melting could introduce some variations, especially for the inter-layer which divides the spiral trace with the L-electrode.

6.1.1 Conductive wax, leads protection and oxide inter-layer

The usage of the silver paste is not positive for two aspects: is not a transient element and it induces a slightly corrosion of the Mg when it is applied. In literature, it is common to exploit conductive wax for precisely connecting parts of electrical elements in bioresorbable devices [13] [45] [14]. The aim consists in the use a biodegradable wax such as Candellilla wax. Normally, the waxes are not conductive so it is necessary to integrate in the wax conductive elements. The most used one are W micro-particles. In this way, varying the volume fraction of these metal particles it is possible to tailor the conductivity of the wax and obtain satisfactory results [80]. The conductive wax could replace the silver paste for the spiral trace-L electrode connection.

Another modification that could be introduced specifically for improve the quality of coil measures is to deposit a protective layer for the coil ends, where the connection with the SMA connector is performed. In the present samples, the ends are left without protection and the silver paste is poured directly on it, covering both the SMA and the Mg traces generating all the aforementioned problems such as oxidation and corrosion of the traces. Instead, protective metallic layer (Titanium or Platinum) of one hundred of nanometers could be deposit through sputtering to protect the Mg from the corrosion action of the Ag paste and the oxidation of the Mg.

It was experimentally seen that the melting process induce a thinning of the PCL. This would effect the measures and introduce a possible deviations from the simulation. In particular, the thinning of the inter-layer could induce stronger capacitance effect between the spiral trace and the L-electrode. In order to have a fixed distance between these two elements, it could be tried to replace the PCL inter-layer with a sputtered oxide layer (SiO_2 or MgO).

6.1.2 Evaporated Mg

The fabrication of the subsequent elements (the diode and the capacitor) will introduce the need of conductive traces for their interconnection. If on one hand the conductive wax are useful for punctual connection, they shouldn't be used for long trace because of their limited conductivity in respect to the metals and the difficulty in obtain thin layers from drop casting. A way would be to use a physical vapor deposition (PVD) of metals over the PCL substrate. Moreover, in order to minimally stress the polymeric substrate, the metal patterning can be achieved by using the stencil mask technique. Due to the fact that the substrate is the PCL, a preliminary test of deposition is performed. More in detail, the deposition of 1 μ m thickness Mg through thermal evaporation over 4 PCL pieces of 1 cm x 1.5 cm and 80 μ m thickness (section 3.2.1 for fabrication process) is performed using the EVA 300 vacuum evaporation machine (Alliance Concept, FR). The PCL pieces are placed over a PDMS spin coated glass wafer (section 3.2.1 for fabrication process) in order to ensure a good support for the The deposition process is influenced by the morphology of the substrate, for test. that reason the following conditions are tested: unmelted PCL, melted PCL, unmelted PCL with 20 nm Ti adhesion layer and melted PCL with 20 nm Ti adhesion layer. The melted-unmented condition is tested because for the first case the morphology of the polymer consists in big crystal grain separated by amorphous regions and this results in a higher roughness ($\approx 1 \ \mu m$) while after the melting process (with the PET compressing layer) the roughness decrease to a value of hundreds of nm and the morphology results smoother without visible crystal grains. Instead, the Ti is deposited in order to understand if a thin metallic layer could be helpful for improve the adhesion. The rectangular shape of the evaporated Mg metal (0.4 mm x 0.8 mm.)is generated from a 25 μ m polyimide (PI, Kapton, DuPont) film patterned through the laser cutter WS Turret Laser Micromachining Workstation (Optec). Once the PCL pieces are placed over the glass wafer, the four stencil mask are positioned and two sample (b and d in Fig. 6.1) are melted on the hotplate at 60 °C with a PET layer over the uncovered PCL. The masks are kept in position thanks to the good adhesion which manifests with the PDMS. When the resolidification is completed, samples a and b (Fig. 6.1) are sputtered (AC450 sputtering machine, Alliance Concept, FR) with Ti and a layer of 20 nm is generated. After that, the Mg is thermally evaporated and the results are shown in Fig.6.1.

As it is depicted from the Fig 6.1, the deposition of the Mg is different between the 4 conditions. In particular it is clear that the best deposition happened for the melted PCL with the Ti adhesion layer (sample ii). The unmelted PCL with Ti shows the second best deposition, even if it is not as homogeneous as the the former case. In absence of the Ti layers, the quality of the deposition decreases dramatically. In fact, if for the no-Ti melted sample the deposition is barely visible, for the no-Ti unmelted sample no deposition is achieved.

This motivated us to use the melting of PCL and the Ti adhesion layer to carry out the other analysis. In order to test the achievable resolution, circular traces are



Figure 6.1: Evaporated Mg test: (a) before the peel off of PI masks; (b) after the peel off of the Pi masks. (i) unmelted PCL and no Ti; (ii) melted PCL and Ti; iii) unmelted PCL and no Ti; (iiii) melted PCL and no Ti.

pattered on 25 μ m PI. In particular, two different width are tested: 300 μ m and 60 μ m. The evaporated traces are realized following the aforementioned method and Fig. 6.2 show the results.

In both cases, the deposition of the Mg looks homogeneous and the traces are electrically conductive. In order to check the quality of the evaporated Mg compared to the Mg foil, Mg traces with the same geometry are obtained from the 10 μ m and 20 μ m thickness Mg foil and a 2 point resistance measures are performed using the multimeter (Keysight Technologies). The resistances coming from this test shows similar values for the 300 μ m width traces (2.8 Ω , 3 Ω and 4.5 Ω for 10 μ m, 20 μ m and 1 μ m thickness respectively) while for the 60 μ m width traces the resistance of evaporated Mg is almost double the values of the other ones (4.9 Ω , 8.1 Ω and 16.1 Ω for 10 μ m, 20 μ m and 1 μ m thickness respectively).

In any case, it is demonstrated that it is possible to deposit a conductive Mg traces over the PCL substrate with good conductive properties, comparable to the Mg foil. In the future steps, 4 point measures could be done in order to extract the sheet resistance of the evaporated Mg and compare this values with the Mg foil.

Positive aspects of the evaporated Mg are the tunability of the thickness of Mg layer, the good resolution achievable and the reduction of the oxidized parts over the trace due to its freshly production.

However, some precautions must be considered. Talking about the thickness, higher



Figure 6.2: Evaporated Mg circular traces: (a) PI mask over PCL; (b) Mg traces after stencil mask peel off. (i) trace 300 μ m width; (ii) trace 60 μ m width.

and lower values than 1 μ m are achievable, but it has to be kept in mind that the evaporation is a process that generates some heat. Even if the heating of the wafer is limited thanks to the strong vacuum condition at which the process is performed $(1.5 \times 10^{-5} mbar)$, longer deposition time could lead to the partial melting of the PCL modifying its surface and preventing a correct deposition. The deposition rate of the Mg was estimated to be 0.1μ m/min based on measurements (DektakXT) and a deposition of 1 μ m thickness Mg (i.e. 10 min deposition) do not lead to any PCL melting. On the contrary, a test performed for a deposition of 10 μ m (100 min deposition) depicts an initial PCL melting, underling the necessity of a carefully study of the temperature rise in the vacuum chamber and on the wafer to fully exploit this deposition technique.

Moreover, the good resolution achievable is correlated to a good adhesion of the stencil mask to the PCL in order to avoid shadowing phenomena induced by the height difference between the mask an the PCL. This effect is reduced by the PI mask-PCL melting which is done before the evaporation. In fact, the stencil mask sinks slightly into the PCL, decreasing the step height and consequently the shadowing effect. After the evaporation, the PI stencil mask can be easily detached from the PCL using a tweezer. This step needs to be done slowly to avoid to damage the deposited Mg.

Finally, the Mg evaporated traces could be fabricated on demand, differently from what happens with the Mg foil traces that are repeatedly extract from the same foil, reducing the time in which the Mg is exposed to air and so decreasing the oxidation of the surface.

The evaporation process of the Mg could be exploited not just for the realization of the interconnection traces but also as a possible alternative for the Mg foil in realizing the coil element, the metallic part of the diode and the conductive plates of the capacitor.

For this purpose, these circular traces could be further tested and analyzed. For example, they could be analyzed with the NA and their results could be compared with the Mg traces coming from the metallic foils.

6.2 Diode: Principle application and materials

The diode is a crucial element for the circuit because is the one responsible for the rectification of the alternated signal generated by the receiving coil. Based on the literature, the diodes which are mostly fabricated and used for transient applications can be divided between diode exploiting a semiconductor material (Si) which is doped for creating the p and n type regions (i.e. homojunction) [64] [65] [32] or diode formed by superimposition of different naturally doped materials (i.e. heterojunction) [30] [22] [77].

In particular, the Silicon nanomembranes (SiNM) diodes are the best example of the first kind of diodes. They consists in ultra thin layer of Si (250-320 nm) which is doped using boron (positive) and phosphorus (negative) ions through ion implants and solid-state diffusion. The thin Si layer is obtained from particular wafer called siliconon-insulator (SOI) wafer which presents this nanomeric Si layer on top of an oxide layer (SiO₂). Once the doping steps are terminated, the Si layer can be released through the erosion of the underlying oxide layer with hydrofluoric acid (HF) and the doped membrane can be transferred onto polymeric layer. The metallic pads are deposited through PVD process. The results obtained with SiNM are promising, as reported from recent paper [13] [45] [14]. The limiting aspects of this technologies are the high price of the SOI wafer and the long process steps required for obtaining the final device.

On the other hand, a cheaper and faster way to realize diodes is the realization of a stacked structure with the use of materials which are 'naturally' n-type or p-type and do not need a doping procedures.

In latter case, the rectifying junction is obtained by the physical contact of the two different layers while for the SiNM the junction is generated inside the Si crystal lattice. This aspect is crucial because the quality of the junction is strictly correlated to the good performances of the diode element. For the SiNM, the exploitation of the Si crystal lattice ensure a good mobility of the carrier species between the different parts of the device and limits the presence of the impurities. For the stacked diode, the junction is strongly dependent on the quality of the surface in contact (presence of oxidative layer and impurities) and the quality of the contact itself (intimate attachment).

While for the SiNM fabrication the process is well established and brings to the realization of a PIN diode (p-intrinsic-n) [65], for the stacked diode the selection of the materials can lead to the realization of different types of diodes (p-n junction, Schottky, etc...).

The selection of the materials is crucial for realizing the diode. In addition, the materials need to be transient and biodegradable, accordingly to the aim of the project. Examples of exploited materials in this field are:

- Metals: Magnesium (Mg), Molybdenum (Mo) and Tungsten (W);
- Semiconductor: Zinc-oxide (ZnO), nanomeric Si layer (SiNM);

Another interesting material is the PEDOT:PSS. It is extensively used for photovoltaic cells and it was demonstrated to be a p-type polymer for realizing diodes [30] [69]. Moreover, its solution composition give the possibility to easily create layer through ambient temperature processes such spin coating. For these reasons, even if it is not a biodegradable element, it is decided to study its behavior in order to understand if such polymers type could be a possible alternatives for transient diodes.

The principal fabrication and deposition techniques are: PVD for the metal compound such as the sputtering, e-beam evaporation and thermal evaporation; sputtering or chemical vapor deposition (CVD) and annealing (ZnO) or the usage of SOI wafer (SiNM) for semiconductor; spin coating or drop casting with micropipette for polymer solutions deposition (PEDOT:PSS).

Due to the high frequency of the generated signal from the receiving coil, the realization of a Schottky diode could be a good alternative to the SiNM for our application. In fact, this type of diode has theoretically a faster switching time and a lower forward voltage (0.15-0.3 V) compared to the typical p-n junction diode (0.6-0.7 V).

The fabrication of the Schottky diode involves having a metal and a semiconductor material in contact (MS contact). In general, the MS contacts can lead to a ohime or rectifying (Schottky) contact, depending on the work functions of the metal (Φ_M) and semiconductor (Φ_S) [55]. The work function is the energy required to extract a electron from the material surface to vacuum and it is a propriety of the surface of the material. It can be expressed as the difference between the Fermi energy (E_F) and the vacuum energy (E_0). From the theory [55], both rectifying and ohmic junction can be realized putting in contact the metal with a p-type semiconductor or an n-type semiconductor if the following relations are respected:

Table 6.1: MS connection types.

	n-type semiconductor	p-type semiconductor
$\Phi_M > \Phi_S$	Rectifying	Ohmic
$\Phi_M < \Phi_S$	Ohmic	Rectifying

A preliminary study of materials that could generate Schottky diodes are reported in terms of depositions over the PCL layer and work functions (WFs).

Since the substrate of the device is the PCL, the deposition techniques must take into account the temperature limitation introduced by the low T_m of PCL.

Usually, the n-type semiconductor elements is constituted by wurtzite structured ZnO (hexagonal crystal structure). The ZnO can be generate from sputtering (Zn target in Ar/O_2 atmosphere) or CVD (zinc acetate in deionized water) but an annealing process at hundreds of degrees (300-400 °C) is required for generate the wurtzite structure. Due to the fact that the annealing temperature is much greater of the melting temperature of the PCL substrate (60 °C), it is tried to use a solution of ZnO nanoparticles (ZnO NPs).

In particular, the following materials are tested: Mo and Mg for the metal element and contacts; ZnO NPs and PEDOT:PSS as n-type semiconductor and p-type polymer respectively.

The Mg is thermally evaporated using the EVA 300 vacuum evaporation machine

(Alliance Concept, FR) while the Mo is sputtered using the AC450 sputtering machine (Alliance Concept, FR). For both metals, a thickness of $\approx 1 \ \mu m$ is achieved. As n-type semiconductor, it is evaluated a ZnO NPs solution (Helios ETL jet, Genesink) while for a possible the p-type polymer it is used the PH1000 PEDOT:PSS solution (Ossila). Both the solutions are deposited through the drop casting technique, using of 5 μ l over the PCL layer.

The morphology of the deposited layers over the PCL (smoothed after melting) are reported in the Fig. 6.3, exploiting the scanning electron microscope images (SU5000, Hitachi).



Figure 6.3: Materials for diode over PCL: (a) Mg; (b) Mo; (b) PEDOT:PSS; (d) ZnO NPs.

Talking about the surface morphology, an ideally perfect surface would be smooth and without imperfections. This would increase the quality of the contact between the superimposed materials. However, the deposited materials shows some problems. The surface of the Mg appears without cracks but with some superficial imperfections like holes. On the other hand, the Mo appears smoother than the Mg but few cracks are visible. Talking about the PEDOT:PSS, the obtained layer looks smooth and without cracks, but the surface is not perfectly flat due to the tendency of the PEDOT to from complex with itself when drop casted and its shrinkage during drying. Finally, the ZnO NPs layers is the most critical one. In fact it appears severally cracked, especially along the border of the ZnO drop.

These problems would limit the diode characteristics and for that reason it should be tried to solve them. Some suggestions could be to reduce the thickness of the Mg
and Mo layer (hundreds of nanometers should be enough for the metal elements) and the application of the spin coating technique that could lead to more homogeneous layers (solve the PEDOT:PSS problem) and help the evaporation of the solvent (solve the ZnO NPs problem).

Even if the deposition of the material could be improved, a first evaluation of the WFs of the deposited material is carried out. The WFs are evaluated singularly for these material deposited on top PCL layer (smoothed after melting). This is made in order to have a general experimental value of the materials work function using the allowable deposition techniques and a common substrate. The evaluation of the WFs is made using the atomic force microscopy (AFM) machine (Bruker), exploiting the Kelvin Probe technique which consists in measure the local contact potential difference (V_{CPD}) between the conductive ATM tip and the sample. Knowing the work function of the tips and the experimental V_{CPD} , the work function of the sample can be extracted as: $WF_{sample} = WF_{tip} - V_{CPD}$.

The obtained values are reported in the following table 6.2:

Table 6.2: Work functions of tested materials.

Material	WF $[eV]$
Mg	4.1
Mo	5.3
ZnO NPs	4.8
PEDOT:PSS	5.3

Given these results, some considerations can be done.

The first one is that the WF_{ZnO} is higher than Mg and lower than Mo, so theoretically it should be possible to create a rectifying contact between the Mo-ZnO NPs layers while a ohmic contact should be generate between the ZnO-Mg junction.

On the other hand, the PEDOT:PSS polymer could be used as a p-type polymer, exploiting his hole carrier capability to create a Schottky junction with the Mg and ohmic contact with Mo.

These are just a general assumptions, in fact the actual values of the Schottky barrier potential (Φ_B), which is the main responsible for the rectifying behavior, should be evaluate for each case. For the n-type Schottky diode, $\Phi_{M_{Mo}}$ and the electron affinity of the ZnO NPs χ_{ZnO} can lead to the evaluation of the Schottky barrier ($\Phi_B = \Phi_{M_{Mo}} - \chi_{ZnO}$) while for the p-type Schottky diode, it would be needed the χ_{PEDOT} , the energy band gap of the PEDOT:PSS ($E_{g_{PEDOT}}$) and the $\Phi_{M_{Mg}}$ ($\Phi_B = E_{g_{PEDOT}}/q + \chi_{PEDOT} - \Phi_{M_{Mo}}$).

Moreover, for the ohmic contacts the potential barrier which is formed between the Mg and ZnO (n-type diode) or Mo and PEDOT:PSS (p-type diode) has to be evaluated in order to understand if in both the biased condition (forward or revere) these are sufficiently low (i.e. if $\Phi_{M_{M_g}}$ is comparable to χ_{ZnO} for n-type diode or $\chi_{PEDOT:PSS}$ is comparable to $\Phi_{M_{M_o}}$ for p-type diode).

This evaluations can be done based on the literature values or experimentally. Due to the fact the le values are strongly dependent from the realization process and substrate morphology, it is not trivial find a perfectly matched experimental condition in literature. For that reason, the experimental evaluation is often performed to extract the Schottky barrier potential and other diode characteristics directly from the I-V curve of the formed MS interface.

Furthermore, the stacked structure implies that the material are deposited totally or partially over materials other than the PCL and this fact can alter the surface morphology and characteristics of the deposited new material.

For these reasons, a good practice would be to retest the WFs of the materials once the deposition on the substrate or different material is achieved. In this way the WFs will represent the values which are really possessed by the surfaces of the materials.

For what concerns the fabrication steps of the whole Schottky diode, it could be used the spin coating (better) or the drop casting for the deposition of the ZnO NPs layers or PEDOT:PSS over a PCL substrate. Subsequently, the metals could be deposited totally or partially overlapped to the ZnO or PEDOT:PSS layer using a pre-patterned stencil mask, in order to create the Schottky contact on one side and an ohmic contac on the other side. These metal parts would be also used as pads for the connection of the diode to the other elements of the circuit.

In any case, further analysis of the effective doping of the n-type semiconductor ZnO NPs layer and the p-type PEDOT:PSS polymer layer should be performed, exploiting for example hall effect or a combination of X-ray diffraction and photoluminescence.

If these materials would result ineffective, alternatives like SiNM or the realization of the wurtzite ZnO layer should be tried. For the former, the high temperature treatment (doping processes) are done over the SOI wafer, avoiding limitation induced by low melting temperature of the biodegradable polymers. However, the transfer technique of the thin doped silicon layer to the PCL should be tested to verify its feasibility. Otherwise, it can be tried to realize a wurtzite ZnO layer on a higher melting temperature polymer (such as PI) and than transfer this layer to PCL substrate (for example: peel off the PI with wurtzite ZnO, flip of the PI and melting of PCL, selective etching of the PI).

6.3 Capacitor: Principle application and materials

The capacitor is the last element that will complete the electrical circuit for IPT. It would be used for two main aspects:

- match the resonance frequency of the receiver coil with the ones of the transmitter in order to maximize the power transfer;
- smooth the rectified signal coming from the diode in order to obtain a continuous (DC) signal.

The capacitors fabricated for resorbable devices consist usually in two metallic plates separated by a dielectric layer. The selection of the materials is again challenging due bioresorbable behave that these elements should present. From literature [28] [45] [43], examples of exploited materials for these two parts are:

- Mg for the metallic plates and conductive pads;
- SiO₂, MgO and Si₃N₄ for the dielectric layers.

The metallic plates could be fabricated exploiting different processes: cut from a metal foil with the desired shape exploiting the laser cutter technique or deposited through PVD methods such as sputtering and e-beam or thermal evaporation.

For what concerns the dielectric layer, the deposition could be achieved through PVD (the same of metals) or CVD (plasma enhance chemical vapor deposition or ALD) directly on top of the metallic plates.

For example, Koo *et al.* [45] used a 50 μ m thickness Mg foil from which they extracted the the metallic plates. Once the plates were ready, they deposited 600 nm of SiO₂ over them. They finished the device laminating the two plates together with SiO₂ layers face-to-face. Moreover, they demonstrated that the capacitance was easily adjustable changing the area of the plates, spanning values from 50 pF to 250 pF with a surface variation of 5 mm² to 25 mm².

Another property of the device that could be varied in order to further tailor the capacitance is the thickness of the dielectric layer. In fact, higher the thickness of the dielectric layer, lower the capacitance of the device, following the equation: $C = \varepsilon A/d$, where ε the electrical permittivity of the metallic plate, A the metal-dielectric contact surface and d the thickness of the dielectric layer.

The fabrication steps of our capacitors could follow the one proposed by Koo *et al.* [45] or could substitute the Mg foil with thermally evaporated Mg. This would remove the need of the lamination process due to the fact that the construction would happen layer-by-layer in a stacked configuration. Firstly, the first Mg plate could be evaporated using a stencil mask to give the wanted shape. Then, the dielectric SiO₂ layer could be sputtered over the deposited Mg, using the same mask (covering the connection pad which will be used for the measurement and connection with the other circuital elements). Finally, a further thermal evaporation could generate the second Mg plate of the capacitor and finish the device fabrication, after the removal of the stencil mask and the PCL encapsulation.

6.4 Water barrier enhancement

As presented in section 3.2.3, there are plenty of possibilities for reducing the water uptake of in the device. The one that should be tested firstly is the use of natural wax. In fact, waxes like Candelilla can easily and effectively increase water barrier efficacy from hours to days. As already presented, its water blocking efficacy is strictly correlated to its thickness so an evaluation of the maximal allowable thickness of the encapsulating wax based on the overall bendability of the device should be carried out.

If the effect of the wax is not enough to increase the water barrier, it could be think to replace the wax with the use a surface degradation polymers for the encapsulation. The difficulty of the fabrication and synthesis is totally rewarded by an important increase of time required for the water to penetrate into the device (up to one month for the PBTPA reported in [12]). Bending test should be carried out as well as adhesion strength test between the polymers.

These two are the first that should be tested because of their great increase of water barrier effect.

In any case, also the other methods and techniques reported in section 3.2.3 could be further evaluated if the first two do not prove sufficiently effective.

Chapter 7

Conclusion

The exploitation of neural modulation has attracted more and more interest due to its beneficial effect in treating lots of health problems spanning from depression to Parkinsonian tremor or recovery after stroke. Among the neuromodulation techniques, one of the most exploited one is the delivery of an electrical signal to the nervous system. In particular, the neural interfaces exploited for the electrical brain neurostimulation often require invasive implantation, second surgery for the removal of the implants after its working period and present batteries or conductive wires for the delivery of the required energy. These three aspects limit their application in larger scale. During the last decades, novel approaches were adopted to tackle these problems. In particular, the usage of endovascular neural interfaces, the exploitation of bioresorbable materials and the development of wireless power transfer devices could greatly reduce these limitations.

In this project, an electrical circuit for WIPT made with bioresorbable material that could be integrated in a transient neurovascular interface is presented. To be able to wirelessly deliver a monophasic electrical impulse to the brain, an ensemble of three elements is proposed. In particular, this ensemble is constituted by an inductive coil, with the aim of generate an electrical signal proportional to the magnetic flux passing through the scalp, a diode, with the objective of rectify the alternated signal coming from the coil and capacitor with the purpose of smooth the rectified signal coming from the diode.

The work of this thesis is mainly focused on the realization of the coil element.

Due to the fact that the device would be placed in a blood vessel (i.e. in a aqueous environment), a first analysis of material behavior in water solution is carried out. From these tests, a water uptake in the PCL encapsulation layer was discovered to happen already after few hours and a penetration test confirms this evidence. Techniques for improving the water barrier efficiency are proposed and a proof of concept using some of these techniques is reported, underlying a positive effect on the retarding the Mg degradation.

Then, the design of the inductor coil is studied, based on the fabrication limitations. From simulations, the best geometries were extracted and fabricated using novel microfabrication approaches.

The fabricated coils were tested exploiting NA and parameters referring to their quality (i.e. Q factors and inductances) were extracted and compared with the simulations. It was found that the simulation results were different from the experimental ones, both in terms of Q factors, inductances and self resonance frequencies. Possible explanations of the discordances are given, such as: presence of oxide layer over the Mg trace; not perfect contact between the traces and corrosion induced by silver paste; geometry deformation of the coil during fabrication; thinning of the PCL (both inter-layer and encapsulation) during melting.

Based on experimental results, the best fabricated coil is decided to be the one with 3 turns, 140 μ m trace width and 60 μ m space gap. A preliminary wireless power transfer test is carried out. From this preliminary test, a WIPT was successfully achieved. The signal generated by the fabricated coil had a amplitude of $\approx 300 \text{ mV}_{pp}$ giving as input to an unoptimized transmitting coil $\approx 5.6 \text{ V}_{pp}$. Strategies for improving the efficiency of the IPT are suggested: make a better matching between receiver and transmitter and geometrically optimize the transmitter coil

Furthermore, improvements for the fabricate coil are proposed: bioresorbable connections, corrosion resistant protective layer for experimental tests, evaporated Mg as alternative conductive metal. Finally, the diode and capacitors are introduced, together with possible materials for their realization and fabrication techniques.

In conclusion, developing a WIPT in such constraints of space, available materials and fabrication techniques is clearly challenging. This thesis represents the first steps towards this goal, highlighting the more impairing limitations and trying to overcome them with different approaches and proposals.

Addition material

7.1 Coil Parametrization ANSYS

The parametrization in Ansys is made exploiting an existing model (PolygonHelix). In this model, the geometrical parameters do not refers to the width of the trace and the space gap between the loops so a conversion to these quantities is proposed. This is done in order to have a better control over the parameters and be able to sweep their values during the simulations. Fig.7.1 a presents the variable input parameters while Fig. 7.1 b shows the conversion equations.

a	Name	Value	Unit	Evaluated V	Туре		Description	Read-only	Hidden	Sweep
	n_turns	7		7	Design					✓
	width_trace	0.06	mm	0.06mm	Design					✓
	r_coil_ext	1.68	mm	1.68mm	Design					✓
	gap_coil	0.06	mm	0.06mm	Design					✓
h	Name	Value				Unit	Evaluated Value	Description		
	Command	CreateUserDefinedPart								
	Coordinate System	Global								
	Name	SegmentedHelix/PolygonHelix.dll								
	Location	syslib								
	Version	1.0								
	PolygonSegments	4					4	Number of cross-section polygon segments, 0 for circle		
	PolygonRadius	width_trace_coil*cos(pi/4)					0.042426407mm	Outer radius of cross-secton polygon		
	StartHelixRadius	r_coil_ext-((n_turns+0.5)*width_trace_coil+(n_turns*gap_coil))					0.81mm	Start radius from polygon center to helix center		
	RadiusChange	width_trace_coil+gap_coil					0.12mm	Radius change per turn		
	Pitch	0				mm	0mm	Helix pitch		
-	Turns	n_turns					7	Number of turns		
	SegmentsPerTurn	ı 36					36	Number of segments per turn, 0 for true surface		
	RightHanded	1					1	Helix direction, non-zero for right handed		

Figure 7.1: (a) Input parameters; (b) conversion equations.

7.2 Matlab script

The following Matlab script allows to identifies the incorrect coil geometries which cannot be generated and simulated in Ansys HFSS. The exclusion criteria are a geometrical one (the coil cannot have even one loop and the external radius is bigger than the fixed one) and a model one (the Ansys model wants the sum of the width and gap to be less than two times the radius of the cross section of the coil). Referring to the output graphs, the allowable geometries are these included between the asterisks and the small circles while the outer ones cannot be simulated.

```
n\_turns = 7;
                                     % number of turns
1
   r_coil_ext = 1.68;
                                     % outer coil radius (mm)
2
   k = 1;
3
   width_coil_vect = [];
4
    for i = (0.06:0.02:0.14)
                                     % thickness of the coil (mm)
\mathbf{5}
6
    width coil = i;
7
    gap\_coil = [0.06:0.02:0.2]; % thickness of the gap (mm)
8
    StartHelixRadius = r\_coil\_ext - ((n\_turns+1)*width\_coil+n\_turns*gap\_coil);
9
10
    g=[StartHelixRadius<2*width_coil] + [StartHelixRadius>r_coil_ext];
11
    new_gap_coil= gap_coil(find(g==0));
                                                      % geometric condition
12
13
        if isempty(new gap coil)~= 1
14
15
        Polygon_Radius = width_coil*cos(pi/4);
16
        Radius_Change = new_gap_coil+width_coil;
17
18
       x=find(Radius_Change>2*Polygon_Radius); % Ansys model condition
19
20
            if isompty (x) \sim = 1
21
22
            gap\_coil\_min(1,k) = new\_gap\_coil(1,x(1,1));
^{23}
            gap coil \max(1,k) = \text{new gap coil}(1,x(1,\text{end}));
24
25
            width_coil_vect = [width_coil_vect i];
26
            k=k+1;
27
            end
28
        end
29
    end
30
31
   figure, plot(width_coil_vect,gap_coil_min, 'o'), hold on,
32
   plot(width_coil_vect,gap_coil_max, '*')
33
   xlabel('width coil'), ylabel('gap coil'),
legend('min gap coil', 'max gap coil'), xticks(width_coil_vect)
34
35
   title (['Coil with ' num2str(n_turns)'' turns'])
36
```

Bibliography

- Nur Ahmadi et al. "Towards a distributed, chronically-implantable neural interface". In: (2019), pp. 719–724.
- [2] Michael W. Baker and Rahul Sarpeshkar. "Feedback Analysis and Design of RF Power Links for Low-Power Bionic Systems". In: *IEEE Transactions on Biomedical Circuits* and Systems 1.1 (2007), pp. 28–38. DOI: 10.1109/TBCAS.2007.893180.
- [3] Cameron Baptista et al. "The effect of temperature and pressure on polycaprolactone morphology". In: *Polymer* 191 (2020), p. 122227.
- [4] Gian Luca Barbruni et al. "Miniaturised Wireless Power Transfer Systems for Neurostimulation: A Review". In: *IEEE Transactions on Biomedical Circuits and Systems* 14.6 (2020), pp. 1160–1178. DOI: 10.1109/TBCAS.2020.3038599.
- [5] Edward Bloch, Yvonne Luo, and Lyndon da Cruz. "Advances in retinal prosthesis systems". In: *Therapeutic Advances in Ophthalmology* 11 (2019). PMID: 30729233, p. 2515841418817501. DOI: 10.1177/2515841418817501. eprint: https://doi.org/ 10.1177/2515841418817501. URL: https://doi.org/10.1177/2515841418817501.
- [6] David Borton et al. "Personalized Neuroprosthetics". In: Science Translational Medicine 5.210 (2013), 210rv2-210rv2. DOI: 10.1126/scitranslmed.3005968. eprint: https://www.science.org/doi/pdf/10.1126/scitranslmed.3005968. URL: https://www.science.org/doi/abs/10.1126/scitranslmed.3005968.
- [7] Mark Bower et al. "Intravenous Recording of Intracranial, Broadband EEG". In: Journal of neuroscience methods 214 (Jan. 2013). DOI: 10.1016/j.jneumeth.2012.12.027.
- [8] Payton Broaddus. "Waveguide fabrication using the Heidelberg MLA150 Maskless Aligner". PhD thesis. Stanford University, 2019.
- [9] Fabián Leonardo Cabrera, Renato S. Feitoza, and Fernando Rangel de Sousa. "Extending the inductor operating frequency for optimally-coupled wireless power transfer systems". In: 2015 SBMO/IEEE MTT-S International Microwave and Optoelectronics Conference (IMOC) (2015), pp. 1–5.
- [10] Gi Doo Cha et al. "Bioresorbable Electronic Implants: History, Materials, Fabrication, Devices, and Clinical Applications". In: Advanced Healthcare Materials 8.11 (2019), p. 1801660. DOI: https://doi.org/10.1002/adhm.201801660. eprint: https:// onlinelibrary.wiley.com/doi/pdf/10.1002/adhm.201801660. URL: https:// onlinelibrary.wiley.com/doi/abs/10.1002/adhm.201801660.
- [11] Gi Doo Cha et al. "Bioresorbable electronic implants: history, materials, fabrication, devices, and clinical applications". In: *Advanced healthcare materials* 8.11 (2019), p. 1801660.
- [12] Yeon Sik Choi et al. "Biodegradable polyanhydrides as encapsulation layers for transient electronics". In: Advanced Functional Materials 30.31 (2020), p. 2000941.

- [13] Yeon Sik Choi et al. "Fully implantable and bioresorbable cardiac pacemakers without leads or batteries". In: *Nature biotechnology* 39.10 (2021), pp. 1228–1238.
- [14] Yeon Sik Choi et al. "Stretchable, dynamic covalent polymers for soft, long-lived bioresorbable electronic stimulators designed to facilitate neuromuscular regeneration". In: *Nature communications* 11.1 (2020), pp. 1–14.
- [15] Yeonsik Choi, Jahyun Koo, and John A Rogers. "Inorganic materials for transient electronics in biomedical applications". In: *MRS Bulletin* 45.2 (2020), pp. 103–112.
- [16] Andreas Christ et al. "Assessing Human Exposure to Electromagnetic Fields From Wireless Power Transmission Systems". In: *Proceedings of the IEEE* 101.6 (2013), pp. 1482–1493. DOI: 10.1109/JPROC.2013.2245851.
- [17] Sagami Elec Co. "Tips for coil user. Part 1". In: (). URL: :%20https://www.sagamielec.co.jp/file/tech/%20coil_doc_100e.pdf.
- [18] James Z. Fan, Victor Lopez-Rivera, and Sunil A. Sheth. "Over the Horizon: The Present and Future of Endovascular Neural Recording and Stimulation". In: Frontiers in Neuroscience 14 (2020). ISSN: 1662-453X. DOI: 10.3389/fnins.2020.00432. URL: https: //www.frontiersin.org/article/10.3389/fnins.2020.00432.
- [19] James Z. Fan, Victor Lopez-Rivera, and Sunil A. Sheth. "Over the Horizon: The Present and Future of Endovascular Neural Recording and Stimulation". In: Frontiers in Neuroscience 14 (2020). ISSN: 1662-453X. DOI: 10.3389/fnins.2020.00432. URL: https: //www.frontiersin.org/article/10.3389/fnins.2020.00432.
- [20] Adele Fanelli et al. "Transient Neurovascular Interface for Minimally Invasive Neural Recording and Stimulation". In: Advanced Materials Technologies (June 2021). DOI: 10.1002/admt.202100176.
- [21] Danyelle Campos França et al. "Hydrolytic and thermal degradation of PCL and PCL/bentonite compounds". In: *Materials Research* 19 (2016), pp. 618–627.
- [22] Yang Gao et al. "Moisture-triggered physically transient electronics". In: Science advances 3.9 (2017), e1701222.
- S. Garcia-Asensio et al. "Technical Aspects of Intra-Arterial Electroencephalogram Recording". In: Interventional Neuroradiology 5.4 (1999). PMID: 20670526, pp. 289– 300. DOI: 10.1177/159101999900500405. eprint: https://doi.org/10.1177/ 159101999900500405. URL: https://doi.org/10.1177/159101999900500405.
- [24] Barbara Gentile. "Development of a Novel Simultaneous Information and Power Transfer System to Inductively Address Miniaturized Neural Implants". PhD thesis. Politecnico di Torino, 2021.
- [25] Mark S. George et al. "Vagus nerve stimulation: a new tool for brain research and therapy". In: *Biological Psychiatry* 47.4 (2000), pp. 287–295. ISSN: 0006-3223. DOI: https: //doi.org/10.1016/S0006-3223(99)00308-X. URL: https://www.sciencedirect. com/science/article/pii/S000632239900308X.
- [26] Maurizio Gulino et al. "Tissue Response to Neural Implants: The Use of Model Systems Toward New Design Solutions of Implantable Microelectrodes". In: Frontiers in Neuroscience 13 (July 2019). DOI: 10.3389/fnins.2019.00689.
- [27] Qinglei Guo et al. "A Bioresorbable Magnetically Coupled System for Low-Frequency Wireless Power Transfer". In: Advanced Functional Materials 29.46 (2019), p. 1905451.

- [28] Aleksi Hänninen. Bioresorbable Wireless Resonance Sensors: Materials and processes. English. Tampere University Dissertations - Tampereen yliopiston väitöskirjat. Tampere University", 2021. ISBN: 978-952-03-2161-1.
- [29] Reid R. Harrison. "Designing Efficient Inductive Power Links for Implantable Devices". In: 2007 IEEE International Symposium on Circuits and Systems. 2007, pp. 2080–2083. DOI: 10.1109/ISCAS.2007.378508.
- [30] N Hernandez-Como et al. "Flexible PEDOT: PSS/ZnO Schottky diodes on polyimide substrates". In: *Microelectronic Engineering* 216 (2019), p. 111060.
- [31] Suk-Won Hwang et al. "High-performance biodegradable/transient electronics on biodegradable polymers". In: Advanced Materials 26.23 (2014), pp. 3905–3911.
- [32] Suk-Won Hwang et al. "Materials for bioresorbable radio frequency electronics". In: Advanced Materials 25.26 (2013), pp. 3526–3531.
- [33] Ahmed Ibrahim and Mehdi Kiani. "A Figure-of-Merit for Design and Optimization of Inductive Power Transmission Links for Millimeter-Sized Biomedical Implants". In: *IEEE Transactions on Biomedical Circuits and Systems* 10.6 (2016), pp. 1100–1111. DOI: 10.1109/TBCAS.2016.2515541.
- [34] "IEEE Standard for Safety Levels with Respect to Human Exposure to Radio Frequency Electromagnetic Fields, 3 kHz to 300 GHz". In: *IEEE Std C95.1*, 1999 Edition (1999), pp. 1–83. DOI: 10.1109/IEEESTD.1999.89423.
- [35] *IMD Shield: Securing Implantable Medical Devices.* Massachusetts Institute of Technology. 2011. URL: https://groups.csail.mit.edu/netmit/IMDShield/index.html.
- [36] Tinneke Jacobs et al. "Plasma treatment of polycaprolactone at medium pressure". In: Surface and Coatings Technology 205 (2011). PSE 2010 Special Issue, S543-S547. ISSN: 0257-8972. DOI: https://doi.org/10.1016/j.surfcoat.2011.02.012. URL: https://www.sciencedirect.com/science/article/pii/S0257897211001125.
- [37] MJ Jenkins and KL Harrison. "The effect of crystalline morphology on the degradation of polycaprolactone in a solution of phosphate buffer and lipase". In: *polymers for Advanced Technologies* 19.12 (2008), pp. 1901–1906.
- [38] Sam John et al. "Signal quality of simultaneously recorded endovascular, subdural and epidural signals are comparable". In: *Scientific Reports* 8 (May 2018). DOI: 10.1038/ s41598-018-26457-7.
- [39] Richard P. Jones and Ronald D. Berger. "Chapter 30 The past, present, and future of implantable cardioverter-defibrillators". In: *Emerging Technologies for Heart Diseases*. Ed. by Udi Nussinovitch. Academic Press, 2020, pp. 669–681. ISBN: 978-0-12-813704-8. DOI: https://doi.org/10.1016/B978-0-12-813704-8.00030-9. URL: https://www.sciencedirect.com/science/article/pii/B9780128137048000309.
- [40] Uei-Ming Jow and Maysam Ghovanloo. "Design and Optimization of Printed Spiral Coils for Efficient Transcutaneous Inductive Power Transmission". In: *IEEE Transactions on Biomedical Circuits and Systems* 1.3 (2007), pp. 193–202. DOI: 10.1109/ TBCAS.2007.913130.
- [41] Uei-Ming Jow and Maysam Ghovanloo. "Modeling and Optimization of Printed Spiral Coils in Air, Saline, and Muscle Tissue Environments". In: *IEEE Transactions on Biomedical Circuits and Systems* 3.5 (2009), pp. 339–347. DOI: 10.1109/TBCAS.2009. 2025366.

- [42] Seung-Kyun Kang et al. "Bioresorbable silicon electronic sensors for the brain". In: *Nature* 530.7588 (2016), pp. 71–76.
- [43] Seung-Kyun Kang et al. "Dissolution behaviors and applications of silicon oxides and nitrides in transient electronics". In: Advanced Functional Materials 24.28 (2014), pp. 4427– 4434.
- [44] David W. Knight. "An introduction to the art of Solenoid Inductance Calculation With emphasis on radio-frequency applications". In: 2016.
- [45] Jahyun Koo et al. "Wireless bioresorbable electronic system enables sustained nonpharmacological neuroregenerative therapy". In: *Nature medicine* 24.12 (2018), pp. 1830– 1836.
- [46] Takashi Kozai et al. "Brain Tissue Responses to Neural Implants Impact Signal Sensitivity and Intervention Strategies". In: ACS Chemical Neuroscience 6 (Jan. 2015), pp. 48–67. DOI: 10.1021/cn500256e.
- [47] Elliot S. Krames et al. "Chapter 1 What Is Neuromodulation?" In: Neuromodulation. Ed. by Elliot S. Krames, P. Hunter Peckham, and Ali R. Rezai. San Diego: Academic Press, 2009, pp. 3-8. ISBN: 978-0-12-374248-3. DOI: https://doi.org/10.1016/B978-0-12-374248-3.00002-1. URL: https://www.sciencedirect.com/science/ article/pii/B9780123742483000021.
- [48] W.B. Kuhn and N.M. Ibrahim. "Analysis of current crowding effects in multiturn spiral inductors". In: *IEEE Transactions on Microwave Theory and Techniques* 49.1 (2001), pp. 31–38. DOI: 10.1109/22.899959.
- [49] Julia Kzhyshkowska et al. "Macrophage responses to implants: prospects for personalized medicine". In: Journal of leukocyte biology 98.6 (2015), pp. 953–962.
- [50] Antonino La Mattina, Stefano Mariani, and Giuseppe Barillaro. "Bioresorbable Materials on the Rise: From Electronic Components and Physical Sensors to In Vivo Monitoring Systems". In: Advanced Science 7 (Feb. 2020). DOI: 10.1002/advs.201902872.
- [51] G. Lazzi. "Thermal effects of bioimplants". In: IEEE Engineering in Medicine and Biology Magazine 24.5 (2005), pp. 75–81. DOI: 10.1109/MEMB.2005.1511503.
- [52] Shuo Liu, Jianhui Su, and Jidong Lai. "Accurate Expressions of Mutual Inductance and Their Calculation of Archimedean Spiral Coils". In: *Energies* 12.10 (2019). ISSN: 1996-1073. DOI: 10.3390/en12102017. URL: https://www.mdpi.com/1996-1073/12/ 10/2017.
- [53] Rodolfo Llinás et al. "Neuro-vascular central nervous recording/stimulating system: Using nanotechnology probes". In: J. Nanoparticle Research 7 (June 2005), pp. 111– 127. DOI: 10.1007/s11051-005-3134-4.
- [54] Melanie R Major et al. "The foreign body response: at the interface of surgery and bioengineering". In: *Plastic and reconstructive surgery* 135.5 (2015), pp. 1489–1498.
- [55] Metal-Semiconductors Contacts. Sept. 2021. URL: https://eng.libretexts.org/ @go/page/367.
- [56] Svjetlana Miocinovic et al. "History, Applications, and Mechanisms of Deep Brain Stimulation". In: JAMA Neurology 70.2 (Feb. 2013), pp. 163-171. ISSN: 2168-6149. DOI: 10.1001/2013.jamaneurol.45. eprint: https://jamanetwork.com/journals/jamaneurology/articlepdf/1391074/nnr120008_163_171.pdf. URL: https://doi.org/10.1001/2013.jamaneurol.45.

- [57] S.S. Mohan et al. "Simple accurate expressions for planar spiral inductances". In: *IEEE Journal of Solid-State Circuits* 34.10 (1999), pp. 1419–1424. DOI: 10.1109/4.792620.
- [58] H. Nakase et al. "An intra-arterial electrode for intracranial electro-encephalogram recordings." In: Acta neurochir 136 (1995), pp. 103–105. ISSN: 0195-6108. URL: https: //doi.org/10.1007/BF01411445.
- [59] Nicholas Opie et al. "Focal stimulation of the sheep motor cortex with a chronically implanted minimally invasive electrode array mounted on an endovascular stent". In: *Nature Biomedical Engineering* 2 (Dec. 2018). DOI: 10.1038/s41551-018-0321-z.
- [60] Thomas Oxley et al. "Minimally invasive endovascular stent-electrode array for high-fidelity, chronic recordings of cortical neural activity". In: *Nature biotechnology* 34 (Feb. 2016). DOI: 10.1038/nbt.3428.
- [61] Thomas Oxley et al. "Motor neuroprosthesis implanted with neurointerventional surgery improves capacity for activities of daily living tasks in severe paralysis: First in-human experience". In: *Journal of NeuroInterventional Surgery* 13 (Oct. 2020), neurintsurg– 2020. DOI: 10.1136/neurintsurg-2020-016862.
- [62] P. Pieters et al. "Accurate modeling of high-Q spiral inductors in thin-film multilayer technology for wireless telecommunication applications". In: *IEEE Transactions on Microwave Theory and Techniques* 49.4 (2001), pp. 589–599. DOI: 10.1109/22.915431.
- [63] Dimiter Prodanov and Jean Delbeke. "Mechanical and Biological Interactions of Implants with the Brain and Their Impact on Implant Design". In: Frontiers in Neuroscience 10 (2016).
- [64] Guoxuan Qin et al. "Flexible microwave PIN diodes and switches employing transferrable single-crystal Si nanomembranes on plastic substrates". In: *Journal of Physics D: Applied Physics* 42.23 (2009), p. 234006.
- [65] Guoxuan Qin et al. "RF model of flexible microwave single-crystalline silicon nanomembrane PIN diodes on plastic substrate". In: *Microelectronics journal* 42.3 (2011), pp. 509– 514.
- [66] Anil Kumar RamRakhyani, Shahriar Mirabbasi, and Mu Chiao. "Design and Optimization of Resonance-Based Efficient Wireless Power Delivery Systems for Biomedical Implants". In: *IEEE Transactions on Biomedical Circuits and Systems* 5.1 (2011), pp. 48–63. DOI: 10.1109/TBCAS.2010.2072782.
- [67] Giovanni A Salvatore et al. "Biodegradable and highly deformable temperature sensors for the internet of things". In: *Advanced Functional Materials* 27.35 (2017), p. 1702390.
- [68] Matthew Schormans, Virgilio Valente, and Andreas Demosthenous. "Practical Inductive Link Design for Biomedical Wireless Power Transfer: A Tutorial". In: *IEEE Transactions on Biomedical Circuits and Systems* 12.5 (2018), pp. 1112–1130. DOI: 10.1109/ TBCAS.2018.2846020.
- [69] Bhupendra K Sharma, Neeraj Khare, and Shahzada Ahmad. "A ZnO/PEDOT: PSS based inorganic/organic hetrojunction". In: *Solid State Communications* 149.19-20 (2009), pp. 771–774.

- [70] Manikandan Sivan et al. "Plasma treatment effects on bulk properties of polycaprolactone nanofibrous mats fabricated by uncommon AC electrospinning: A comparative study". In: Surface and Coatings Technology 399 (2020), p. 126203. ISSN: 0257-8972. DOI: https://doi.org/10.1016/j.surfcoat.2020.126203. URL: https: //www.sciencedirect.com/science/article/pii/S0257897220308720.
- [71] Sebastian Spencker et al. "Comparison of skin adhesive and absorbable intracutaneous suture for the implantation of cardiac rhythm devices". In: *Europace* 13.3 (2011), pp. 416–420.
- P Stoeter et al. "Intracranial electroencephalographic and evoked-potential recording from intravascular guide wires." In: American Journal of Neuroradiology 16.6 (1995), pp. 1214-1217. ISSN: 0195-6108. eprint: http://www.ajnr.org/content/16/6/1214. full.pdf. URL: http://www.ajnr.org/content/16/6/1214.
- [73] Benjamin A Teplitzky et al. "Computational modeling of an endovascular approach to deep brain stimulation". In: *Journal of neural engineering* 11.2 (2014), p. 026011.
- [74] Understanding S Parameters. Rohde Schwarz. 2019. URL: https://www.youtube.com/ watch?v=-PiOUbErHTY.
- [75] International Telecommunication Union. "Radio Regulations. Articles". In: 2020. ISBN: 978-92-61-30301-3. URL: http://handle.itu.int/11.1002/pub/814b0c44-en.
- [76] Friederike Von Burkersroda, Luise Schedl, and Achim Göpferich. "Why degradable polymers undergo surface erosion or bulk erosion". In: *Biomaterials* 23.21 (2002), pp. 4221– 4231.
- [77] Saisai Wang et al. "Physically Transient Diode With Ultrathin Tunneling Layer as Selector for Bipolar One Diode-One Resistor Memory". In: *IEEE Electron Device Letters* 42.5 (2021), pp. 700–703.
- [78] Xu Wang et al. "Food-Materials-Based Edible Supercapacitors". In: Advanced Materials Technologies 1.3 (2016), p. 1600059.
- [79] H.A. Wheeler. "Formulas for the Skin Effect". In: Proceedings of the IRE 30.9 (1942), pp. 412–424. DOI: 10.1109/JRPROC.1942.232015.
- [80] Sang Min Won et al. "Natural wax for transient electronics". In: Advanced Functional Materials 28.32 (2018), p. 1801819.
- [81] Maria Ann Woodruff and Dietmar Werner Hutmacher. "The return of a forgotten polymer—Polycaprolactone in the 21st century". In: *Progress in Polymer Science* 35.10 (2010), pp. 1217–1256. ISSN: 0079-6700. DOI: https://doi.org/10.1016/j.progpolymsci. 2010.04.002. URL: https://www.sciencedirect.com/science/article/pii/S0079670010000419.
- [82] Liping Xu and Akiko Yamamoto. "Characteristics and cytocompatibility of biodegradable polymer film on magnesium by spin coating". In: *Colloids and Surfaces B: Bioin*terfaces 93 (2012), pp. 67–74.
- [83] Lan Yin et al. "Dissolvable metals for transient electronics". In: Advanced Functional Materials 24.5 (2014), pp. 645–658.
- [84] David W York, Stephen Collins, and Mooketsi Rantape. "Measuring the permeability of thin solid layers of natural waxes". In: *Journal of colloid and interface science* 551 (2019), pp. 270–282.

- [85] Ki Jun Yu et al. "Bioresorbable silicon electronics for transient spatiotemporal mapping of electrical activity from the cerebral cortex". In: *Nature materials* 15.7 (2016), pp. 782– 791.
- [86] Fan-Gang Zeng et al. "Cochlear Implants: System Design, Integration, and Evaluation". In: *IEEE Reviews in Biomedical Engineering* 1 (2008), pp. 115–142.
- [87] Bowen Zhu et al. "Silk fibroin for flexible electronic devices". In: Advanced Materials 28.22 (2016), pp. 4250–4265.
- [88] Stan Zurek. "Qualitative FEM study of proximity loss reduction by various winding configurations, Part 1". In: *Transformers Magazine* 3 (Jan. 2016), p. 70.