

Universität  
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Traditio et Innovatio



# Design and Fabrication of Thermoelectric Generators for Implantable Medical Devices

Dissertation to obtain the academic degree of

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University of Rostock

in cooperation with Jade University of Applied Sciences

submitted by

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Rostock, January 13, 2025



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*Dedicated to my family*

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# Abstract

Implantable medical devices (IMDs) have revolutionized healthcare by enabling advanced treatment options and gathering vital physiological data for diagnostics, prognostics, therapies, and biological studies. However, the lifespan of IMDs is often limited by the capacity of their integrated batteries. Implantable thermoelectric generators (ITEGs) present a promising solution by converting internal body temperature differences into electrical energy via the Seebeck effect, potentially extending device lifespan or enabling self-sustaining operation without battery replacements.

This thesis thoroughly investigates ITEGs for powering active IMDs, focusing on design and fabrication. A novel ITEG design using a polyetheretherketone (PEEK) housing was proposed due to PEEK's biocompatibility and low thermal conductivity. Integrated metal heat conductors in the housing reduce thermal resistance between tissue and the TEG.

A simplified tissue model, including skin, fat, and muscle, was developed to simulate the implantation environment and assess ITEG feasibility. This model considers essential bio-heat effects such as metabolic heat generation, blood perfusion, and sweating. An efficient parametric simulation approach was employed using a simplified TEG model with a representative thermopile and decoupled Peltier heat approximation.

To further increase simulation accuracy, detailed human torso and head models with realistic anatomical geometries were developed, applying bio-heat effects across various tissue types. Parametric analysis and design optimization explored the influence of factors such as fill factor, thermolegs quantity, housing size, material, and implant location on ITEG performance. Trade-offs between power, miniaturization, hermeticity, and manufacturability were extensively analyzed.

To maximize power generation and voltage output, ITEGs with free-standing thermolegs of high aspect ratios were identified as optimal. A novel fabrication method using a removable assembly rig was introduced, facilitating the creation of TEGs with varying array configurations. The most complex TEG featured 254 thermolegs with dimensions of  $0.5 \times 0.5 \times 4 \text{ mm}^3$ . Additionally, a compact TEG with an integrated boost converter circuit was designed and fabricated, meeting the spatial and voltage requirements of IMDs. Characterization was performed using custom setups and benchmarked with commercial devices.

Addressing moisture ingress, which affects ITEG safety and longevity, the hermeticity of the PEEK housing was evaluated through helium leak tests and moisture penetration experiments. Applying Parylene C coating to the PEEK surface significantly enhanced hermeticity.





# Zusammenfassung

Implantierbare medizinische Geräte (IMDs) haben die Gesundheitsversorgung revolutioniert, indem sie fortschrittliche Behandlungsmöglichkeiten bieten und wichtige physiologische Daten für Diagnosen, Prognosen, Therapien und biologische Studien sammeln. Die Lebensdauer von IMDs ist jedoch oft durch die Kapazität ihrer integrierten Batterien begrenzt. Implantierbare thermoelektrische Generatoren (ITEGs) bieten eine vielversprechende Lösung, indem sie Temperaturunterschiede innerhalb des Körpers mittels des Seebeck-Effekts in elektrische Energie umwandeln, was die Lebensdauer der Geräte verlängern oder einen energieautonomen Betrieb ohne Batteriewechsel ermöglichen kann.

Diese Dissertation untersucht ITEGs zur Energieversorgung aktiver IMDs umfassend und konzentriert sich auf Design und Fertigung. Ein neuartiges ITEG-Design mit einem Gehäuse aus Polyetheretherketon (PEEK) wurde aufgrund dessen Biokompatibilität und der niedrigen Wärmeleitfähigkeit vorgeschlagen. Integrierte metallische Wärmeleiter im Gehäuse reduzieren den thermischen Widerstand zwischen Gewebe und TEG.

Ein vereinfachtes Geometriemodell, das Haut-, Fett- und Muskelgewebe beinhaltet, wurde entwickelt, um die Implantationsumgebung zu simulieren und die Machbarkeit von ITEGs zu bewerten. Dieses Modell berücksichtigt wichtige Bio-Wärme-Effekte wie metabolische Wärmeerzeugung, Durchblutung und Schwitzen. Ein effizienter parametrischer Simulationsansatz wurde unter Verwendung eines vereinfachten TEG-Modells mit einem repräsentativen Thermopile und entkoppelter Peltier-Wärme-Approximation angewendet.

Zur weiteren Erhöhung der Simulationsgenauigkeit wurden detaillierte Modelle des menschlichen Torsos und Kopfes mit realistischen anatomischen Geometrien entwickelt, die Bio-Wärme-Effekte über verschiedene Gewebetypen anwenden. Parametrische Analysen und Designoptimierungen untersuchten den Einfluss von Faktoren wie Füllfaktor, Anzahl der Thermoelemente, Gehäusegröße, Material und Implantationsort auf die Leistung der ITEG. Kompromisse zwischen Leistung, Miniaturisierung, Dichtigkeit und Herstellbarkeit wurden umfassend analysiert.

Um die Leistungs- und Spannungsausgabe zu maximieren, wurden ITEGs mit freistehenden Thermoelementen und hohen Aspektverhältnissen als optimal identifiziert. Ein neuartiges Fertigungsverfahren mit einem abnehmbaren Montagerahmen wurde eingeführt, um die Herstellung von TEGs mit unterschiedlichen Anordnungskonfigurationen zu erleichtern. Der komplexeste TEG verfügte über 254 Thermoelemente mit Abmessungen von  $0,5 \times 0,5 \times 4 \text{ mm}^3$ . Zusätzlich wurde ein kompakter TEG mit einem integrierten Spannungswandler entworfen und gefertigt, der die räumlichen Randbedingungen und Spannungsanforderungen von IMDs erfüllte. Die Charakterisierung erfolgte mit benutzerdefinierten Aufbauten und kommerziellen Messgeräten.

Um den Feuchtigkeitseintritt, welche die Sicherheit und Lebensdauer von ITEGs beeinträchtigt, zu adressieren, wurde die Dichtigkeit des PEEK-Gehäuses durch Helium-Lecktests

und Feuchtigkeitsdurchdringungsexperimente bewertet. Die Beschichtung der PEEK-Oberfläche mit Parylene C verbesserte die Dichtigkeit erheblich.

## Abbreviations

|             |                                         |
|-------------|-----------------------------------------|
| <b>TEG</b>  | thermoelectric generator                |
| <b>IMD</b>  | implantable medical device              |
| <b>IoT</b>  | internet of things                      |
| <b>ITEG</b> | implantable thermoelectric generator    |
| <b>ICD</b>  | implantable cardioverter defibrillators |
| <b>DBS</b>  | Deep brain stimulator                   |
| <b>SCS</b>  | spinal cord stimulator                  |
| <b>VNS</b>  | vagus nerve stimulator                  |
| <b>PH</b>   | potential of hydrogen                   |
| <b>WPT</b>  | wireless power transfer                 |
| <b>TE</b>   | thermoelectric                          |
| <b>PEN</b>  | polyethylene naphthalate                |
| <b>PEEK</b> | polyetheretherketone                    |
| <b>FEM</b>  | finite element method                   |
| <b>MOR</b>  | model order reduction                   |
| <b>RT</b>   | representative thermopile               |
| <b>pMOR</b> | parametric model order reduction        |
| <b>DoF</b>  | degrees of freedom                      |
| <b>AMOP</b> | adaptive metamodel of optimal prognosis |
| <b>MRI</b>  | magnetic resonance imaging              |
| <b>CT</b>   | computed tomography                     |
| <b>3D</b>   | three-dimensional                       |
| <b>CAD</b>  | computer-aided design                   |
| <b>PDMS</b> | polydimethylsiloxane                    |
| <b>ALD</b>  | atomic layer deposition                 |

|              |                                           |
|--------------|-------------------------------------------|
| <b>EDM</b>   | electro-discharge machining               |
| <b>DBC</b>   | directly bonded copper                    |
| <b>CNC</b>   | computer numerical control                |
| <b>LFA</b>   | laser flash analysis                      |
| <b>PID</b>   | proportional-integral-derivative          |
| <b>AC</b>    | alternating current                       |
| <b>PECVD</b> | plasma-enhanced chemical vapor deposition |
| <b>QSS</b>   | quasi-steady-state                        |
| <b>RH</b>    | relative humidity                         |
| <b>AAF</b>   | accelerated aging factor                  |
| <b>LT</b>    | lifetime                                  |

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# 1 Introduction

Over recent decades, advances in materials, microelectronics, and biotechnology have led to the development of smaller, more versatile, and more effective implantable medical devices (IMDs). This progress comes at a crucial time given that many countries face the challenges of an aging population due to declining birth rates and increased longevity. Projections indicate that by 2050 one in six individuals worldwide will be 65 and older, a significant increase from the one in eleven recorded in 2019 [1]. This demographic shift has contributed to a heightened prevalence of geriatric ailments. In 2022, over one-third of individuals in the European Union faced chronic diseases or health-related issues [2]. This increased prevalence requires innovative solutions. IMDs offer a promising avenue for proactive healthcare management through continuous monitoring and real-time intervention. Additionally, the emergence of the Internet of Things (IoT) will revolutionize the field of IMDs by providing personalized and data-driven healthcare. These factors fuel the demand for IMDs. They also drive ongoing research and exploration.

However, as the demand for intricate functionality and compact size increases, significant challenges related to energy supply arise. Traditional integrated batteries, which often occupy more than half the volume of IMDs, still possess lifespans shorter than the anticipated life expectancy of most patients. Additionally, the batteries are often embedded within these devices to ensure hermeticity, which protects against potential liquid interference with electronics. Consequently, when their energy reservoir is depleted, patients must undergo surgical procedures to replace the entire device, which entails potential health risks and additional financial burdens.

Fortunately, the modest power demands of IMDs make implantable thermoelectric generators (ITEGs) a feasible potential solution. A thermoelectric generator (TEG), functioning as a solid-state device, can convert temperature differences within the human body into electricity. Implanted TEGs can harness the continuous heat flow from the body's core to the external surface, serving as primary or supplementary power sources.

Despite their potential, ITEGs have received limited attention, primarily due to the challenges associated with their design and validation. The performance of implanted TEGs relies on the temperature difference within the human body, which is difficult to measure accurately using invasive methods. Furthermore, the temperature profile within the body may change after implantation or due to design parameter adjustments, which complicates TEG performance prediction.

Challenges also extend to TEG fabrication. Conventional TEGs cannot provide adequate power and voltage output for IMDs. In particular, custom-tailored TEGs for implants must incorporate a substantial quantity of thermoelements with high aspect ratios, introducing complexity into the fabrication process steps, such as assembly and soldering. Additionally, packaging design is crucial because it directly affects the performance,

dimensions, hermeticity, and biocompatibility of the ITEGs.

This chapter will provide a comprehensive overview of the pertinent knowledge relevant to this work. It will begin with an introduction to various IMDs, presenting their applications, sizes, implant locations, lifespan, and power consumption. The discussion on TEGs will start with a description of their working principle and relevant thermoelectric (TE) effects. This foundational knowledge will support a robust comprehension of the subsequent investigations within this thesis, particularly concerning TEG design and fabrication. Furthermore, the applications of TEGs on human body heat harvesting will be reviewed and discussed. The human body, tightly associated with ITEGs, will be considered primarily in the context of heat transfer. An investigation into the human body's bioheat effects and the corresponding modeling approaches will be provided. The main results and a brief outline of the structure of this thesis will be provided at the end of this chapter.

### 1.1 Implantable Medical Devices

IMDs are introduced into the body through surgical procedures or clinical interventions to fulfill specific functions. IMDs are broadly categorized as active or passive, based on their power requirements.

Passive IMDs lack an internal power source and often encompass prosthetic devices, such as replacements for damaged bone, muscle, or joint fusion hardware. They can also be used to enhance aesthetics (e.g. dental, breast, or pectoral implants) or serve functional roles (e.g. intraocular lenses, contraceptive intrauterine devices).

The IMDs that will be discussed in this thesis are categorized as "active", meaning that they rely on an external source of electrical energy or power for their operation. This definition was established in a 1990 European Union directive that aimed to harmonize national legislation on these devices [3].

#### 1.1.1 Applications, Sizes, Implant Locations, Lifespan and Power Consumption of IMDs

Active IMDs have a history dating back to the first cardiac pacemaker in 1958 [4]. Subsequent decades have seen the development of various active IMDs, including cochlear implants, neurostimulators, and drug pumps (see Figure 1). These developments have significantly expanded treatment options for numerous medical conditions.

Pacemakers and implantable cardioverter defibrillators (ICDs) are typically implanted below the collarbone, with leads extending into the heart. They deliver electrical impulses to regulate cardiac rhythm, treating conditions such as bradycardia and arrhythmias. Traditional pacemakers are 4–5 cm in diameter and 0.5–1 cm thick [10, 11], while ICDs are slightly larger, with dimensions of  $7 \times 5 \times 1.2$  cm<sup>3</sup> [12].

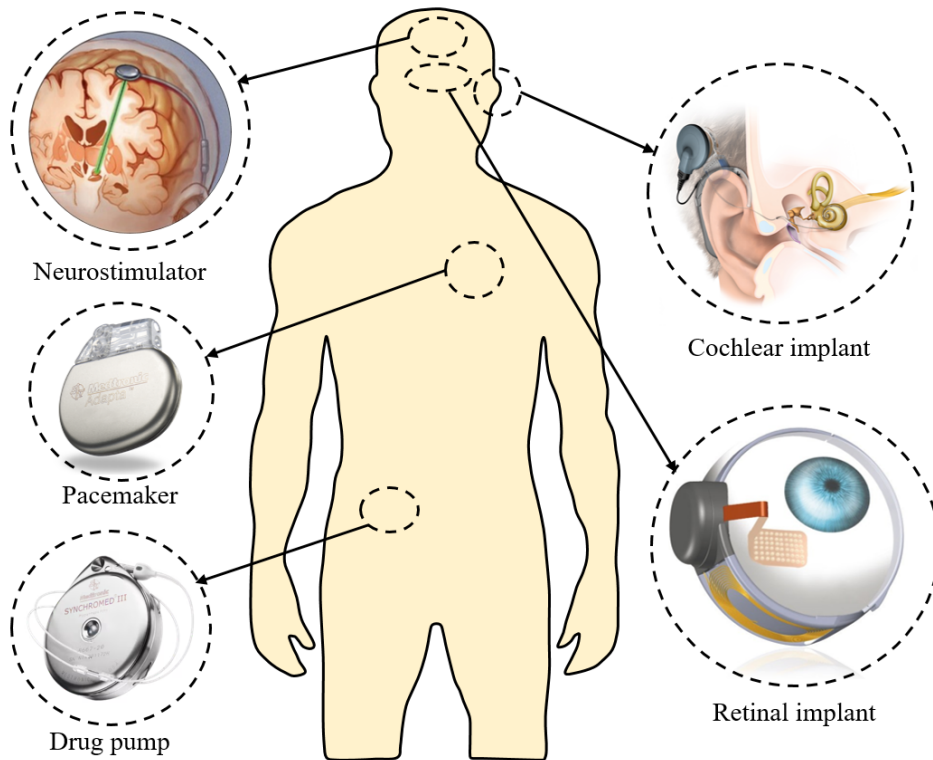


Figure 1: IMDs applications and implant locations. A deep brain neurostimulator consists of two parts: an electrode implanted in the brain to stimulate the target brain area and a pulse generator implanted in the chest (image sourced from [5]). A pacemaker is implanted below the collarbone to regulate cardiac rhythm (image sourced from [6]). A drug pump is implanted in the abdomen for controlled medication delivery (image sourced from [7]). A cochlear implant consists of an internal component implanted subcutaneously behind the upper ear and an external component worn behind the ear (image sourced from [8]). A retinal implant is implanted at the back of the eye to treat retinitis pigmentosa and age-related macular degeneration (image sourced from [9]).

Neurostimulators modulate nerve activity to manage various neurological conditions. These devices typically consist of an electrode and a stimulator. Deep brain stimulators (DBS) employ an electrode implanted within the brain parenchyma to alleviate movement disorders. Spinal cord stimulators (SCS) utilize an electrode positioned near the spinal cord to manage chronic pain. Vagus nerve stimulators (VNS) involve an electrode wrapped around the left vagus nerve in the cervical region to address epilepsy and depression. The stimulators of these devices are often implanted subcutaneously in the chest, with dimensions comparable to pacemakers, measuring approximately  $4 \times 6 \times 0.8 \text{ cm}^3$  [13].

Drug pumps are typically implanted in the abdomen and enable controlled medication delivery for conditions such as diabetes, cancer, and chronic pain. These devices are relatively larger, measuring 8–9 cm in diameter and are 2–3 cm thick [14, 15].

Cochlear implants stimulate the auditory nerve to improve hearing [16]. The internal component, implanted subcutaneously behind the upper ear, is approximately 2 cm in diameter and 5 mm thick [17, 18]. An external component processes acoustic input and sits behind ear. Retinal implants, targeting retinitis pigmentosa and age-related macular degeneration, are significantly smaller (30–500  $\mu\text{m}$ ) and implanted at the back of the

eye [19–21].

The power consumption of IMDs ranges from microwatts to hundreds of milliwatts [22–24] (see Figure 2), primarily attributed to communication, sensing, signal processing, and actuation [22, 23, 25, 26]. Integrated sensors monitor device status and physiological parameters, including heart rate, brain activity, muscle activity, pressure, acceleration, temperature, glucose, oxygen, and potential of hydrogen (pH) [27–31]. Similar biosensors can also be implanted independently for clinical research or may be integrated into passive IMDs for personalized healthcare.

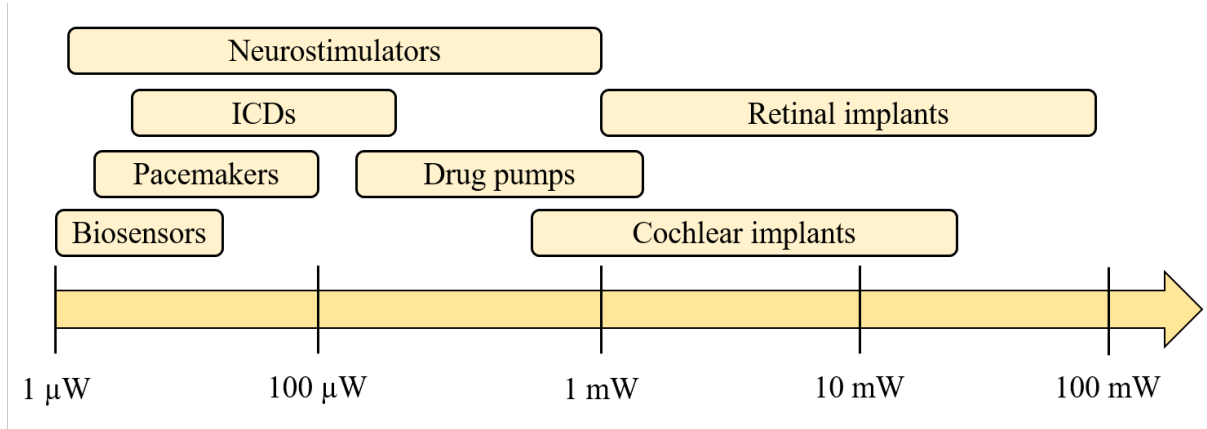


Figure 2: Power consumption of typical IMDs and biosensors.

Currently, lithium batteries (e.g., lithium-iodine, lithium-silver vanadium oxide, and lithium-ion) are the predominant power source for IMDs. Their power capability and energy density scale with volume [32, 33]. These batteries consistently offer 1–4 V to IMDs [34].

While cochlear and retinal implants often house replaceable or rechargeable batteries in their external components, other IMDs typically integrate batteries internally, which occupy a significant portion of the device volume (e.g., 60–75% for pacemakers and ICDs [35]).

Despite advances in power-efficient electronics [36], the lifetime of these IMDs remains limited, for example:

- Pacemakers/ICDs: 5–10 years [37, 38].
- Neurostimulators: 3–8 years [39].
- Drug pumps: 5 years [40].

These batteries are seamlessly integrated within IMDs for hermeticity. Therefore, when the batteries deplete, the patients need to undergo surgical intervention to replace the entire implant. This exposes patients to potential complications, discomfort, and increased healthcare costs.

### 1.1.2 Battery-free Power Supply for IMDs

Research on alternative battery-free power solutions is required to mitigate the drawbacks of batteries inside IMDs. As shown in Figure 3, these approaches can be classified into two categories: wireless power transfer (WPT) and energy harvesting [22–24, 32, 33].

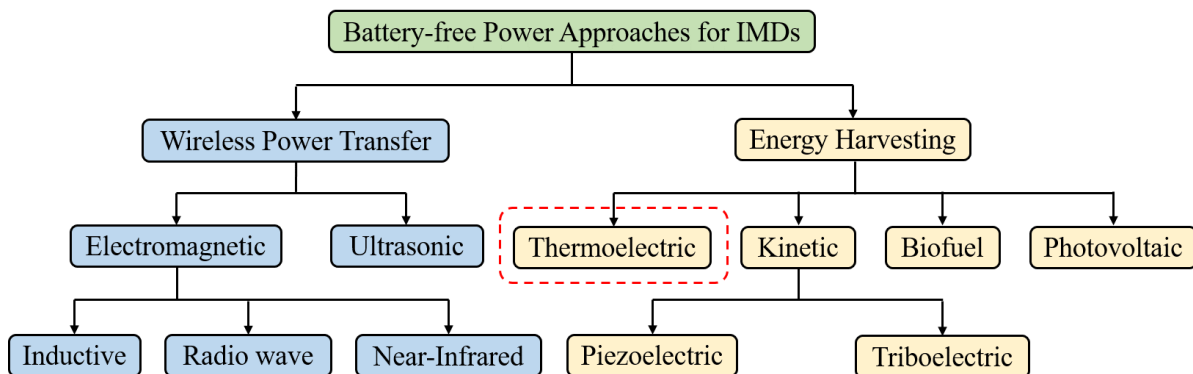


Figure 3: Battery-free power approaches for IMDs, which avoid the need to replace batteries. Wireless power transfer recharges the batteries in IMDs remotely. Energy harvesting utilizes the energy of the human body to generate electricity. This work focuses on studying ITEGs, which transform thermal energy in the human body into electrical energy using TE effects.

A WPT system to power IMDs typically uses a transmitter outside the human body and a receiver integrated within the implant. The receiver can collect energy, such as electromagnetic or ultrasonic waves, to achieve remote charging [33, 41, 42]. Electromagnetic WPT exists across a vast spectrum of frequencies, such as near-field inductive [43, 44], radiative [45], and near-infrared [46]. Ultrasonic WPT transfers power by propagating pressure waves for more efficient coupling [47, 48].

While WPT technologies hold promise to provide power at the milliwatt level to IMDs, they are still in the early clinical stage for most IMDs and their limitations require further investigation. Issues such as energy transfer reliability (influenced by coil alignment and distance) can lead to unsuccessful charging, overcharging, or localized overheating, particularly in life-support IMDs [49].

Efforts to address the power constraint of IMDs also include energy harvesting technologies, which transfer the energy of the human body into electrical power to prolong the lifetime of IMDs or even achieve self-sufficient operation. Different energy harvesters can be classified based on different energy harvesting mechanisms; for example, piezoelectric and triboelectric energy harvesters harness kinetic energy from periodic heartbeats or breathing, biofuel cells utilize living organisms to produce electricity, TEGs convert thermal energy into electricity by utilizing temperature differences in the human body, and photovoltaic cells under the skin transform light into electricity.

The piezoelectric energy harvesters typically offer tens of nanowatts, as shown by in vivo experiments with animals [50–52]. An advanced triboelectric generator provided a power density of  $0.8 \mu\text{W}/\text{cm}^2$  from the breath of a living rat [53]. A biofuel cell implanted

in mice provided a power density of  $0.3 \text{ mW/cm}^2$  [54]. A photovoltaic-driven pacemaker generated a power density of  $0.95 \text{ mW/cm}^2$  from outdoor sunlight [55].

Despite the great potential of these energy harvesters in power generation, they face challenges for long-term implantation. The flexible polymers in these devices can be degraded over time by bodily fluids or enzymatic processes. This degradation can release byproducts that further contribute to biocompatibility issues. Additionally, piezoelectric and triboelectric materials can experience fatigue and degradation over time due to periodic stress. This can reduce their effectiveness and potentially lead to device failure.

TEGs offer several advantages over other energy harvesting technologies, making them well-suited for powering IMDs, as follows:

- **Continuous and Scalable Direct Voltage:** TEGs provide a continuous direct voltage based on the body’s natural temperature gradient. Additionally, the voltage output of TEGs can be scaled by adjusting the configuration of the TE elements within the device.
- **Stable and Quiet Operation:** Their solid-state design ensures stable and silent operation, potentially leading to a longer lifespan and eliminating the need for maintenance.
- **Enhanced Biocompatibility Potential:** TEGs demonstrate robust functionality within embedded systems, even when enclosed within biocompatible materials. This inherent capability helps to alleviate the potential biocompatibility concerns that are often associated with other energy harvesters.

## 1.2 Thermoelectric Generator

TEGs have gained significant attention for their ability to convert heat into electricity, making them valuable in a wide range of applications [56–60]. They are utilized in waste heat recovery from power plants, industrial processes, and vehicle engines. TEGs also provide power generation solutions in situations where accessing electricity is difficult, such as in portable power supplies, aerospace missions, and biomedical devices.

This thesis focuses on the low-temperature and low-power application of TEGs—specifically, human body heat harvesting.

### 1.2.1 Thermoelectric Effects

The TE effects consist of three phenomena: the Seebeck effect, the Peltier effect, and the Thomson effect.

## Seebeck Effect

The Seebeck effect describes the conversion of a temperature difference into an electric voltage within a conductor or semiconductor. At the heart of this phenomenon is the behavior of charge carriers, which are electrons in metals, or electrons and holes in semiconductors.

These charge carriers are not stationary but constantly move and vibrate around their average positions in the crystal lattice. When a temperature gradient is applied across the material, the charge carriers on the hot side possess higher average kinetic energy compared to those on the cold side. This increased kinetic energy causes the charge carriers on the hot side to move more rapidly and with higher intensity.

Driven by this temperature gradient, charge carriers tend to diffuse from the hot side to the cold side. However, as they diffuse, they constantly interact with atoms in the material, dissipating some of their energy through scattering, which results in resistance to their flow. Eventually, an equilibrium is reached when the driving force of diffusion, pushing charge carriers towards the cold side, is balanced by the buildup of an opposing electric field and the resistance caused by these interactions. The distribution of charge carriers creates an electric field, which in turn generates a voltage difference (electromotive force) across the material, which is the Seebeck voltage and is written as:

$$U_{Seebeck} = \alpha \Delta T \quad (1.1)$$

where  $\alpha$  is the Seebeck coefficient, which reflects how efficiently the material converts a temperature difference into voltage.  $\Delta T$  is the temperature difference across the material.

To pursue a higher Seebeck coefficient, most TE materials used in TEGs are semiconductors. These TE materials are divided into n-type and p-type, which differ in their majority charge carriers (electrons vs. holes).

Figure 4 (a) illustrates a thermocouple consisting of a p- and n-type TE element. When a temperature difference exists across the thermocouple, holes in the p-type TE element and electrons in the n-type TE element diffuse from the hot side to the cold side. Consequently, the Seebeck voltage is accumulated when p-type and n-type TE elements are connected. Typical TEGs have multiple thermocouples for sufficient voltage output.

## Peltier Effect

The Peltier effect is the reverse of the Seebeck effect, where an applied voltage creates a temperature difference. As illustrated in Figure 4 (b), when a voltage source is applied across a thermocouple it creates an electric field that drives the holes in the p-type TE element and electrons in the n-type TE element from the bottom side to the top side.

The n-type material has a higher Fermi level closer to the conduction band due to the presence of additional donor atoms. The p-type material has a Fermi level closer to the valence band due to the presence of acceptor atoms. Therefore, there is a slight energy

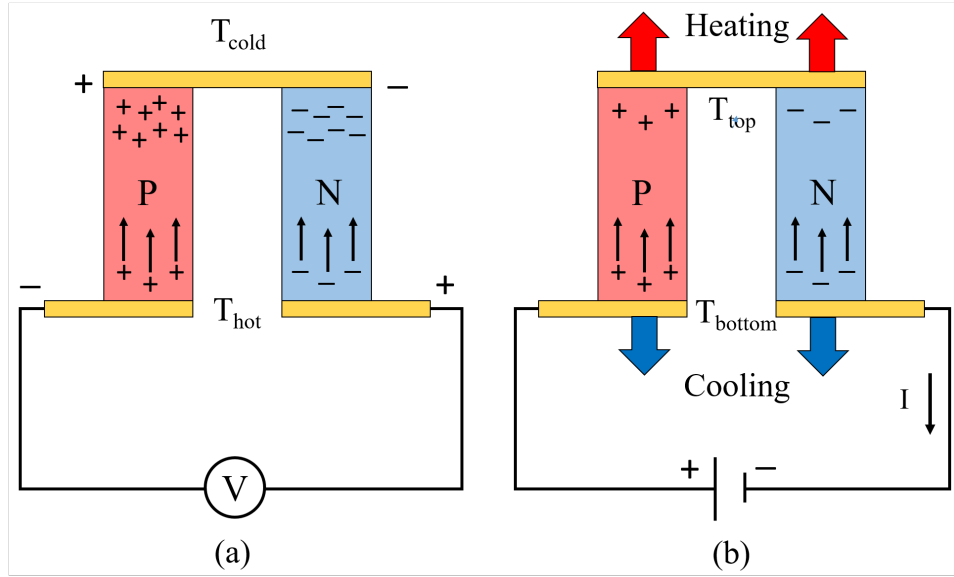


Figure 4: Working principles of TE devices. (a) A TE generator with one thermocouple transforms temperature difference into electricity based on the Seebeck effect. (b) A Peltier module with one thermocouple transforms electricity into heat transfer based on the Peltier effect.

barrier at the junction due to the difference in Fermi levels between the two materials.

On the top side, the electrons in the n-type material try to overcome the energy barrier and enter the p-type material. They lose some of their energy through collisions with the crystal lattice of the n-type material, which manifests as heat released at the top junction.

On the bottom side, when an electron from the p-type material jumps across the junction to enter the n-type material it gains energy to bridge the gap between the conduction and valence bands. This energy gain comes from the lattice vibrations of the p-type material. This transfer of energy from the lattice to the electron translates to cooling at the bottom junction.

The Peltier heat that is liberated on the top junction and absorbed on the bottom junction is written as:

$$Q_{\text{Peltier}} = (\alpha_p - \alpha_n)TI \quad (1.2)$$

Based on the Peltier effect, TE devices can also be used as Peltier modules that are controlled by the current or voltage for cooling or heating applications. Notably, the Peltier heat generation requires the change of the Seebeck coefficient of the TE material and current through TE material. Therefore, an operated TEG, fulfilling both requirements, also generates Peltier heat. Since the Peltier effect is the reverse of the Seebeck effect, the Peltier heat from the TEG is absorbed on the hot side and liberated on the cold side, attempting to reverse the existing temperature difference across the TEG.

### Thomson Effect

The Thomson effect is the continuous version of the Peltier effect, which describes the heating or cooling within a current-carrying TE material with a temperature gradient.



The Seebeck coefficient of the TE material is temperature-dependent. The temperature gradient leads to a gradient in the Seebeck coefficient. Thus, the TE material can be regarded as inhomogeneous, with temperature-dependent Seebeck coefficients. Peltier heat is generated at the junctions between intervals when current flows through the TE material. The Thomson heat, which is regarded as a continuous version of "Peltier heat" in TE material, is expressed as:

$$Q_{Thomson} = -T \frac{d\alpha}{dT} I \nabla T \quad (1.3)$$

## 1.2.2 Thermoelectric Properties

To maximize the effectiveness of a material in converting thermal energy into electrical energy, a high Seebeck coefficient, low electrical resistivity, and low thermal conductivity are required [61].

### Seebeck Coefficient

The relationship between the Seebeck coefficient and charge carrier has been proven both experimentally and theoretically [62, 63]:

$$\alpha = \frac{8\pi^2 k_B^2}{3eh^2} m^* T \left(\frac{\pi}{3n}\right)^{2/3} \quad (1.4)$$

where  $K_b$  is the Boltzmann constant,  $T$  is temperature,  $e$  is the electron charge,  $h$  is the Planck constant,  $m^*$  is the effective mass of the charge carrier, and  $n$  is the carrier concentration.

### Electrical Resistivity

The electrical resistivity ( $\rho$ ) of the TE material represents the property of the material to resist the charge carrier motion and is related to  $n$  and the carrier mobility  $\mu$ :

$$\rho = \frac{1}{ne\mu} \quad (1.5)$$

### Thermal Conductivity

Thermal conductivity ( $\kappa$ ) in TE material is a combination of charge carrier thermal conductivity ( $\kappa_e$ ) and lattice thermal conductivity ( $\kappa_l$ ), which reflect the ease of electrons and holes heat transporting, and phonons (quantized lattice vibration) traveling through the lattice, respectively:

$$\kappa = \kappa_e + \kappa_l \quad (1.6)$$

$\kappa_e$  is related to  $n$  and  $\mu$ , which is written as:

$$\kappa_e = ne\mu LT \quad (1.7)$$

where  $L$  is the Lorenz factor of free electrons.

### Figure of Merit

The material properties of the TE material are conflicting and cannot be solely optimized. Considering  $\alpha$ ,  $\rho$ , and  $\kappa$  comprehensively, the figure of merit ( $ZT$  or  $Z$ ) is used to evaluate the performance of TE materials at a specific temperature, which is expressed by:

$$ZT = \frac{\alpha^2 T}{\rho \kappa} \quad (1.8)$$

or

$$Z = \frac{\alpha^2}{\rho \kappa} \quad (1.9)$$

where  $\alpha^2/\rho$  is the power factor, which signifies the ability of the TE material to generate electricity.

Despite the discovery of various TE materials, only a few exhibit high  $ZT$  near room temperature [64]. Among these, bismuth telluride ( $\text{Bi}_2\text{Te}_3$ ) remains the most efficient TE material for low-temperature applications [65]. Additionally, novel magnesium antimonide-based ( $\text{Mg}_2\text{Sb}_3$ ) materials are receiving attention as promising alternatives [64].

### 1.2.3 Applications on Human Body Heat Harvesting

The application of TEGs for harvesting heat from the human body can be categorized into two main types: wearable and implantable. Wearable TEGs are designed to be worn externally on the body, converting the temperature difference between the skin and the environment into electrical power for portable devices or sensors. Due to the high thermal contact resistance between the TEG and the skin, these generators are often made flexible to better conform to the skin surface, thus reducing thermal resistance and enhancing power output. Heat sinks are sometimes employed to improve heat dissipation from wearable TEGs to the environment.

Significant progress has been made in the field of wearable TEGs, driven by advances in materials [66], structural design [67], and thermal optimization [68]. Figure 5 illustrates various examples of wearable TEGs with different structural designs.

Lee et al. developed a wearable TEG that achieves high TE performance and excellent conformability [69]. The TEG consists of two polyethylene naphthalate (PEN) substrates and 440  $\text{Bi}_2\text{Te}_3$  thermolegs ( $1 \times 1 \times 2 \text{ mm}^3$ ). The substrates are coated with silver nanowire interconnects and are embedded with silver-coated nickel heat conductors. This TEG achieved a maximum power of 7 mW and an open-circuit voltage of 2 V at a temperature difference of 40 K.

Karthikeyan et al. presented a 500  $\mu\text{m}$  thick flexible TEG using thin-film technologies. This device features a flexible polyimide substrate, 50 nm thick aluminum interconnects,

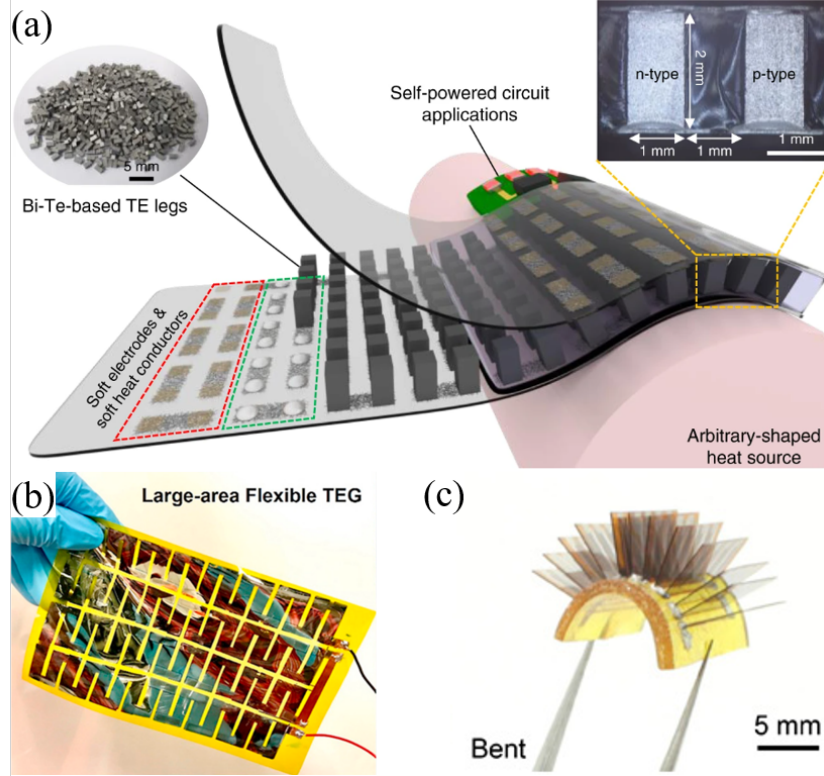


Figure 5: Wearable TEG utilizing skin-ambient temperature difference to generate electricity. (a) A bulk TEG features flexible substrates embedded with silver-coated nickel heat conductors, enabling high TE performance and conformability (image adapted from [69]). (b) A thin-film TEG consists of a 500  $\mu\text{m}$  thick flexible polyimide substrate and 100 nm thick TE elements, which offers unique possibilities for integration into clothing or other wearable devices (image adapted from [70]). (c) A soft TEG with a plugin structure for high stretchability, self-healability, and recyclability (image adapted from [71]).

and 100 nm thick n-type PbTe and p-type SnTe thermolegs. It demonstrated an open-circuit voltage of 250 mV at a temperature difference of 120 K. While its power output was lower (366 nW), its thin profile offered unique possibilities for integration into clothing or other wearable devices.

Ren et al. created a wearable TEG that is notable for its high stretchability, self-healability, and recyclability [71]. The TEG has a soft plugin structure, incorporating flexible dynamic covalent polyimide substrates, liquid-metal electrical wiring, and rigid TE chips with coated thin film Bi and Sb chalcogenides. At room temperature, the TEG, when attached to a forearm, generated an average power density of 45 and 83  $\text{nW}/\text{cm}^2$ , and an average output voltage of 25 and 33  $\text{mV}/\text{cm}^2$  while the wearer was sitting and walking, respectively. To boost outdoor performance, the cold-side surface of the TEG was coated with a glass-polymer hybrid metamaterial film, which has low solar absorptivity and high infrared emissivity.

Despite the boom of wearable TEGs, ITEGs (which have significant promise for applications within the human body) have not yet received a commensurate level of attention when compared to their wearable counterparts. In addition, doubts regarding the ability

of ITEGs to generate the requisite levels of power and voltage output for IMDs due to low-temperature differences within the human body continue to persist. Furthermore, ITEGs require stricter biocompatibility compared to wearable TEGs because they are in direct contact with inner tissue and body fluids.

### 1.3 Bio-heat Effects in Human Body

Mathematical models representing the human thermal system with its thermal and thermoregulatory mechanisms are typically divided into two systems:

- **Passive System:** This system represents human geometry and considers both heat generation by basal metabolism and heat transport by basal blood perfusion under thermal neutrality conditions.
- **Active System:** This system controls the passive system under transient environmental conditions. It uses skin temperature and core body temperature as control parameters. When these temperatures deviate from the target state, various mechanisms (e.g., shivering, sweating, vasoconstriction, and vasodilation) are activated to release, store, or produce heat.

The classification of body models is based on whether the model consists of a single segment or multiple segments. Further differentiation is made by the number of nodes. Single-node models describe body segments as a single unit, while two-node models divide body segments into two concentric layers, such as the body core and skin. Multi-node models are extensions of two-node models.

The first mathematical models for human thermoregulation were proposed by [72], [73], [74], and [75]. [76] developed an equation to describe temperature evolution in the human body, accounting for blood circulation (known as Pennes' bioheat equation). This equation is based on Fourier's law and assumes infinitely fast heat propagation. Studies such as [77] and [78] have highlighted that heat propagation in biological tissues has a finite velocity and wave character. They introduced the thermal wave model of bioheat transfer, which more accurately represents real processes in biological tissues. The limitations of Pennes' equation and some potential solutions are discussed in [79].

A three-dimensional (3D) transient multielement model incorporating the human thermoregulatory system, blood circulation, and respiration was developed by Smith [80]. This model simulates human temperature distribution under highly inhomogeneous external temperature conditions. One of the most successful and widely used multi-node models of human thermoregulation was developed by Fiala [81]. Similar to Smith's model, Fiala's model considers mechanisms of heat release, storage, and production in the human body.

In the passive system, human body parts are geometrically described using spherical or cylindrical elements. Each geometric object represents a body part, with corresponding

thicknesses for layers such as bone, muscle, fat, and skin. Other tissue types that are considered include the brain, lungs, and internal viscera. The geometric objects are divided into sectors to define different environmental conditions for various body regions. For the passive system, the bioheat equation is solved by Pennes' bioheat equation [76].

The thermoregulatory system of the human body involves complex mechanisms that allow humans to maintain a core body temperature of 37 °C. The thermoregulatory center, which is located in the hypothalamus, analyzes temperature information from skin receptors and activates thermoregulatory mechanisms. Fiala's model also distinguishes the human thermal system into an active and a passive system. The passive system describes basal perfusion and metabolism, while the active system manages thermoregulatory functions such as vasoconstriction, vasodilation, sweating, and shivering. Despite its success, Fiala's model has several limitations. It is quasi-two-dimensional, assuming that heat propagation in the horizontal direction dominates over the vertical direction. Heat transfer between horizontal planes is simplified. Most models use artificial geometries to simulate temperature distribution and thermoregulatory mechanisms, which are typically constructed from concentric cylinders. Only a few studies, such as [82], consider realistic human geometries, albeit with limited tissue types.

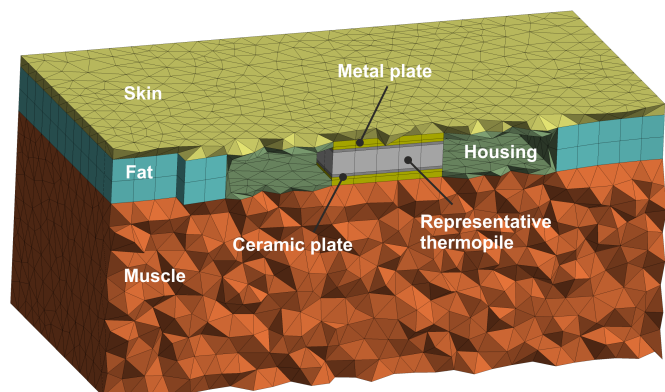
Further development of models for human thermoregulatory mechanisms is complex and requires interdisciplinary collaboration among experts in physiology, thermodynamics, and mathematical modeling. Coupling numerical bioheat models with complex thermoregulatory mechanisms is crucial for medical applications, such as hypothermia or hyperthermia treatments, and in the field of implant development.

## 1.4 Main Results

### Modelling Approaches for Efficient Parametric Simulation of ITEGs

A simplified finite-element tissue model considering the main bioheat effects was developed as a virtual thermal implantation environment to investigate ITEGs. The TEG model embedded in the simplified tissue was simplified and solved using solely thermal analysis. This simplification involved replacing the detailed thermologs with a representative thermopile, where changes in the configurations of the thermologs

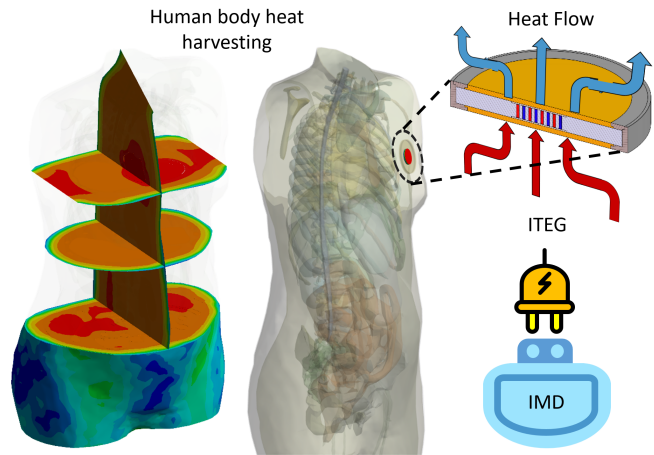
were reflected through alterations in the material properties of the representative thermopile. The decoupling approach approximated the Peltier heat as conductive heat, which



allowed the prediction of ITEG performance using solely thermal analysis. These methods aimed for time-efficient and accurate simulations, which are described in Chapter 2.

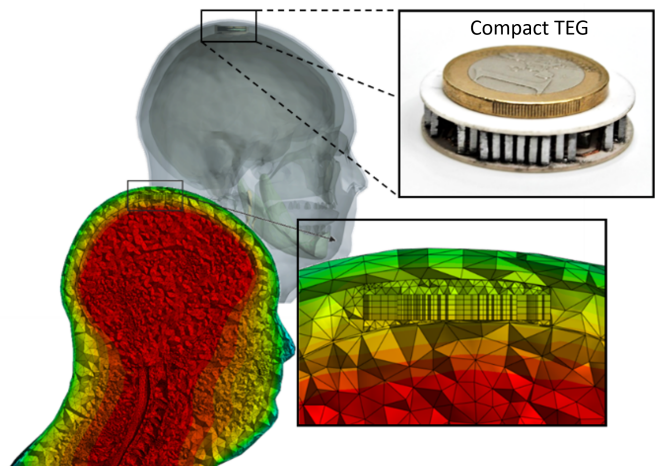
### Comprehensive Analysis of ITEGs Design Based on Detailed Human Body Models

Detailed human body models were developed for ITEG analysis in more realistic implantation environments. Parametric analysis and design optimization were conducted within these models, examining the influence of packaging design, TEG design, and implantation location on ITEG performance. Analyzing the simulation results achieved a deeper understanding of the operating mechanism, TE behavior, and thermal management of ITEGs embedded in the human body. Trade-offs were discussed regarding the ITEG's power output, miniaturization, hermeticity, and manufacturability. This analysis is presented in Chapter 3.



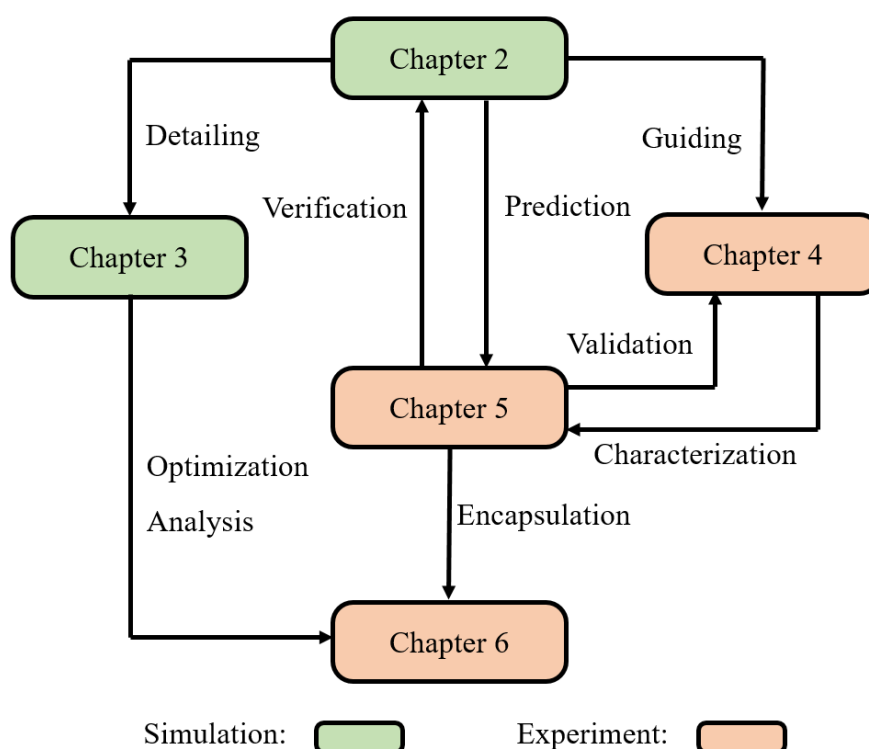
### Fabrication Strategies for ITEGs with Free-standing High Aspect Ratio Thermolegs

ITEGs should feature free-standing high-aspect-ratio thermolegs to provide maximum power and sufficient voltage to power IMDs. A fabrication strategy was proposed that used a removable assembly rig to secure the thermolegs during a single-round soldering process. This approach enabled a TEG with 254 thermolegs to be manufactured, each with dimensions of  $0.5 \times 0.5 \times 4 \text{ mm}^3$ . Additionally, a compact TEG with an integrated boost converter was fabricated, where the voltage was up-converted and regulated to meet the requirements of the IMDs. The fabrication process of these TEGs is detailed in Chapter 4. Chapter 5 presents their characterization results. Chapter 6 investigates the packaging of ITEGs and conducts hermeticity tests.



## 1.5 Thesis Outline

Figure 6 illustrates the outline of this thesis. Chapter 2 describes the efficient simulation of the ITEG and presents a prototype design. It also includes cross-dependencies as simulations to predict the performance of the ITEG. Chapter 3 utilizes detailed human body models based on the work in Chapter 2 to conduct comprehensive design optimization and parametric analysis of packaged ITEGs. Chapter 4 describes how the prototype design was used to guide the fabrication of the ITEGs. Chapter 5 presents the characterization results of the fabricated TEGs and validates the fabrication approach. The characteristics obtained from Chapter 5 are then used to verify the model (e.g., material properties, mesh, and approximation accuracy). Following a study of the performance of the ITEGs, Chapter 6 investigates the encapsulation of the ITEGs, focusing on their biocompatibility and safety.



Chapter 2: ITEG Embedded in Simplified Tissue Model

Chapter 3: ITEG Embedded in Detailed Human Body Model

Chapter 4: ITEG Fabrication

Chapter 5: ITEG Characterization

Chapter 6: ITEG Packaging

Figure 6: Thesis outline and links between chapters, where the simulation works are highlighted in green, and experimental works are highlighted in red.

## 2 ITEG Embedded in Simplified Tissue Model

This chapter will build upon the previously discussed concepts and will introduce an initial design for an ITEG and conduct proof-of-concept simulations. These simulations utilize a simplified tissue geometry model that incorporates essential bioheat effects via the finite element method (FEM). The TEG model will undergo simplification, parameterization, decoupling, and reduction to enable computationally efficient parametric analysis.

Integrating the TEG into the tissue model facilitates parametric analysis, which reveals the influence of varying parameters on the performance of the TEG. Optimization of the key parameters guides the design process, aiming to minimize the dimensions of the device while achieving the desired power output. Based on the simulation results, a prototype TEG design will be proposed for fabrication in Chapter 4.

### 2.1 Methods Review

ITEGs harness the temperature differences within the human body to generate electrical power. The human body is a complex thermal system, where processes such as thermal conduction, metabolism, blood perfusion, and heat dissipation maintain a stable core temperature (normothermia) of 37 °C [83]. Implanting a TEG affects the heat flux distribution and introduces additional heat sources (i.e., Peltier and Joule heating), which alters the temperature distribution at the implantation site. Although understanding these temperature gradients is crucial for designing efficient ITEGs, direct measurement is challenging. In particular, in vivo experiments offer accuracy but raise safety concerns, are often costly, and may not be feasible in early development stages.

Numerical simulation techniques offer a cost-effective and informative alternative to accelerate the design process. Thermal tissue modeling was initially focused on thermal therapy investigations [84–90], which accurately described the body’s thermal behavior.

Yang et al. pioneered a simplified tissue geometry model for ITEG performance estimation. The tissue geometry uses three layers of homogeneous thickness for the skin, fat, and muscle tissue. While the authors considered blood perfusion and metabolic heat generation primarily in the muscle layer, they modeled the TEG without packaging [91]. Jadhav et al. later added metabolism and perfusion across all tissue layers and also included the TEG housing [92]. Importantly, studies such as Yuan et al. highlighted the TEG fill factor as a significant influence on the power output of implanted devices [93]. However, the effects of Peltier and Joule heat were often neglected, which could potentially lead to inaccuracies, especially in situations where current is delivered to an electrical load. Therefore, further research is needed to fully understand how the design parameters influence the performance of the ITEG.



## 2.2 Initial Design

ITEGs are designed to serve as primary or supplementary power sources for IMDs, such as pacemakers and neurostimulators. These IMDs contain various electronic components, including sensors, actuators, and microcontrollers that are responsible for signal processing, data transmission, and power management. Additionally, batteries or large capacitors may be integrated within the IMD for energy storage or backup power.

To ensure practical relevance and a successful design, the initial TEG development phase must consider the device together with its package, along with further essential IMD components. A housing is proposed for this purpose (see Figure 7). Since this housing is in direct contact with human tissue, biocompatibility is an essential requirement. Polyetheretherketone (PEEK) is chosen for the housing material due to its biocompatibility and its lower thermal conductivity when compared to conventional options such as ceramics or titanium. The low thermal conductivity of the housing reduces thermal shunting (i.e., a parasitic heat path bypassing the actual TEG), thereby enhancing TEG performance. Additionally, two gold-coated copper heat conductors are attached to the TEG, which improve the thermal contact between the ITEG and tissue.

The packaged ITEG is positioned in the fat layer beneath the skin layer, where maximum temperature gradient and thus maximum power output can be expected [91, 92, 94]. To maximize the temperature difference across the TEG, it is advantageous to adjust the height of the housing to match the thickness of the fat layer, fully utilizing the space within the fat layer.

Despite the advantages of flexible substrates in wearable TEGs to reduce thermal contact resistance, they present significant drawbacks for long-term implantation. First, most flexible polymeric substrates have poor hermeticity, which raise safety concerns when used within the body. Second, polymers generally have lower thermal conductivity when compared to metal heat conductors, which negatively impacts the heat transfer through the ITEG.

## 2.3 Computational Model

This section elaborates the computational models that have been developed based on the initial design. The models were created using FEM in ANSYS (2021 R2) and were simulated with steady-state analysis. The modeling approaches that were applied to the tissue and TEG will be introduced separately.

### 2.3.1 Simplified Tissue Model with Bioheat Effects

A three-layer simplified tissue finite element model was set up, featuring a cross-section of  $100\text{ mm} \times 100\text{ mm}$ . This model consists of a 2 mm thick skin layer, an 8 mm thick fat

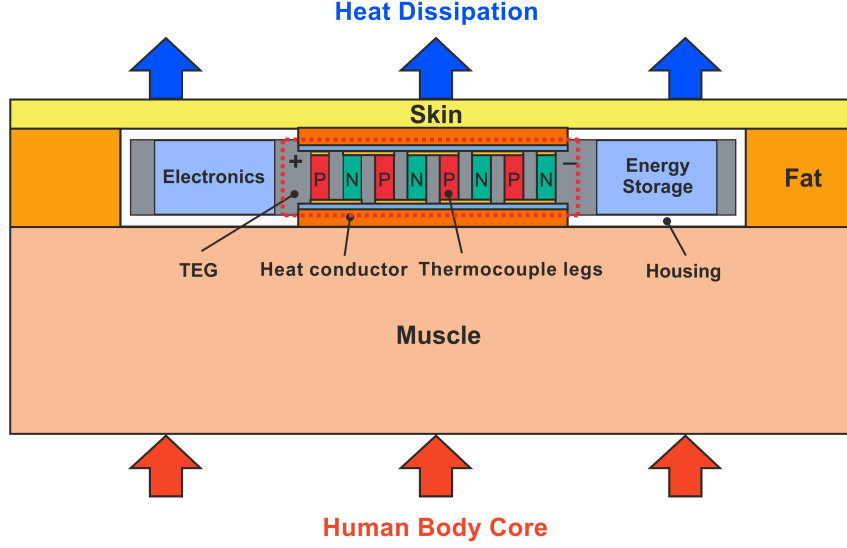


Figure 7: The initial design of an ITEG. A PEEK housing encapsulates the TEG and essential components within the IMDs (e.g., electronics and energy storage devices). Two heat conductors are embedded in the upper and lower walls of the housing, which contact the tissue and TEG. The maximum temperature in the human body exists in the human body core, which is around 37 °C. The minimum temperature in the human body exists on the skin due to its heat dissipation effects. Therefore, the TEG is implanted in the fat layer to ensure a high-temperature difference across the TEG, and thus high electrical power generation. Reprinted from [95].

layer, and a 35 mm thick muscle layer. The effects of blood perfusion and metabolism were considered in all tissue layers.

The steady-state heat transfer in the tissue can be expressed by the well-known Pennes' bioheat equation [76]:

$$\nabla(\kappa \nabla T) + \underbrace{\rho_b c_b \omega (T_a - T)}_{q_b} + q_m = 0 \quad (2.1)$$

where  $\kappa$  represents the thermal conductivity of tissue. The terms  $q_b$  and  $q_m$  describe the effects of blood perfusion and metabolic heat generation, respectively. Additionally,  $\rho_b$ ,  $c_b$ , and  $T_a$  represent the density, specific heat, and temperature of arterial blood, respectively. The term  $\omega$  represents the perfusion rate.

In Equation 2.1, the first term represents heat conduction. The heat flux  $q_b$  depends on the perfusion rate  $\omega$  and the tissue temperature  $T$ . In the simulation,  $T_a$  is assumed to be 37 °C. For tissue temperatures below  $T_a$ ,  $q_b$  functions as heat generation. For temperatures above  $T_a$ ,  $q_b$  operates as heat dissipation. No heat contribution comes from  $q_b$  when  $T$  equals  $T_a$ . The term  $q_m$  represents metabolic heat generation, which is constant but specific to the tissue type.

The boundary conditions of the tissue model were defined as follows. The temperature on the bottom surface of the muscle layer was fixed at 37 °C to simulate the human body core. The effects of convection, radiation, and evaporation were simulated as heat flux

through the skin surface [96]:

$$q_{skin} = \underbrace{h_c(T - T_{amb})}_{q_{conv}} + q_{rad} + q_{eva} \quad (2.2)$$

where  $q_{conv}$ ,  $q_{rad}$ , and  $q_{eva}$  represent the convective, radiative, and evaporative heat fluxes at the skin surface, respectively.  $T_{amb}$  denotes the ambient temperature and  $h_c$  is the convective heat transfer coefficient.

The convective heat transfer coefficient,  $h_c$  (in W/m<sup>2</sup>/K), depends on the air velocity ( $v_{air}$ ) in m/s:

$$h_c = \begin{cases} 3.1 & \text{when } 0 < v_{air} \leq 0.2 \\ 8.3v_{air}^{0.6} & \text{when } 0.2 < v_{air} < 0.4 \end{cases} \quad (2.3)$$

Due to the small temperature difference between the skin and the ambient, the radiative heat transfer coefficient  $h_{rad}$  was obtained by expanding  $q_{rad}$  as a Taylor series around  $T_{amb}$  [97]:

$$q_{rad} = \underbrace{4\sigma\epsilon T_{amb}^3}_{h_{rad}}(T_{skin} - T_{amb}) \quad (2.4)$$

where  $\sigma = 5.67 \times 10^{-8}$  W/m<sup>2</sup>/K<sup>4</sup> is the Stefan–Boltzmann constant and  $\epsilon = 0.95$  is the emissivity of human skin.

The evaporative transfer coefficient  $h_e$  is defined as:

$$q_{eva} = 1.65h_c\omega_{skin}\left[\exp\left(18.956 - \frac{4030.18}{T_{skin} + 235}\right) - \phi\exp\left(18.956 - \frac{4030.18}{T_{amb} + 235}\right)\right] \quad (2.5)$$

where  $\omega_{skin}$  and  $\phi$  represent skin wetness<sup>1</sup> and relative humidity, respectively.

In the simulation,  $T_{amb}$  was assumed to be 21 °C and  $v_{air}$  was set to less than 0.2 m/s. The parameters  $\phi$  and  $\omega_{skin}$  were assigned values of 0.4 and 0.07, respectively. Subsequently,  $h_c$  was calculated as 3.1 W/m<sup>2</sup>/K based on Equation 2.3, while  $h_{rad}$  was calculated as 5.48 W/m<sup>2</sup>/K using Equation 2.5. The tissue properties were sourced from publicly available data [98] and are listed in Table A1.

### 2.3.2 Simplified Thermal TEG Model

#### TEG Simplification

A detailed TEG model was constructed based on the geometry of a commercial device (TEC-20-33-31, Wakefield Thermal), as illustrated in Figure 8. This device comprises a thermopile, multiple interconnects, and two substrates. The thermopile consists of 31 p-type and 31 n-type cubic bismuth telluride elements, each measuring  $1.4 \times 1.4 \times$

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<sup>1</sup> $\phi$  determines how wet the skin is, which is expressed as  $\phi = E/E_{max}$ , where  $E$  denotes actual evaporative rate,  $E_{max}$  is maximum evaporation possible in those conditions [96].

$1.4 \text{ mm}^3$ . The copper interconnects measure  $3.8 \times 1.4 \times 0.25 \text{ mm}^3$ , while two aluminum oxide ceramic plates ( $20 \times 20 \times 0.7 \text{ mm}^3$ ) serve as substrates and thermal contacts. The finite element model of this TEG has 78630 degrees of freedom (DoF).

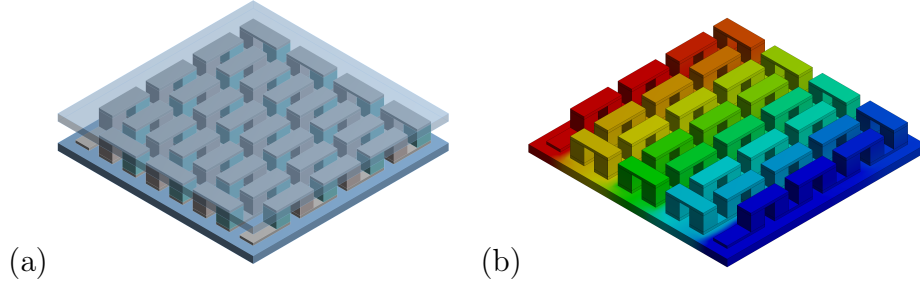


Figure 8: A detailed TEG model, which consists of a thermopile, substrates, and interconnects. (a) Geometry. (b) Qualitative electrical potential distribution. Reprinted from [95].

The material properties of both p- and n-type TE elements were assumed to be identical and temperature-independent due to the low-temperature difference within the human body. Neglecting the effects of the interconnects and solder joints, the Seebeck coefficient and electrical resistivity were derived from measurements of the commercial device, yielding values of  $162 \mu\text{V/K}$  and  $1.1 \times 10^{-5} \Omega \cdot \text{m}$ , respectively. Notably, the sign of the Seebeck coefficient for p- and n-type elements is the opposite. The thermal conductivity of the TE elements was assumed to be  $1.6 \text{ W/m/K}$  [99].

The detailed TEG model was validated by comparing experimental data from the commercial device with the simulation results of the detailed TEG model. To mimic physiological conditions, the hot and cold side temperatures of the TEG were fixed at  $34^\circ\text{C}$  and  $33^\circ\text{C}$ , respectively. A variable resistive load was attached to the TEG for load matching. The simulation results closely match the experimental data, with an average error of 2.8% in voltage output and 5.7% in power output (see Figure 9). Notably, when the load resistance is below  $0.4 \Omega$ , a significant error in power output is observed, which is attributed to the limited accuracy of the load resistor value.

While highly accurate, the detailed TEG model has some significant limitations. First, it is not conducive to parametric analysis (e.g., in altering the number of TE elements). Even though geometric parameters such as the width and height of the TE elements can be adjusted, changing the number of elements necessitates rebuilding the geometry of the entire thermopile array, which is both cumbersome and time-consuming. Second, the detailed model imposes constraints on computational efficiency. While modifying parameterized geometric parameters may not always require complete mesh regeneration, employing mesh adaptivity methods still prolongs the preprocessing time for simulation. Additionally, mesh adaptation alters the system’s matrices, which potentially complicates the applicability of model order reduction (MOR) techniques, which will be discussed in Section 2.3.2.

A simplified TEG model was developed to address these issues, as depicted in Fig-

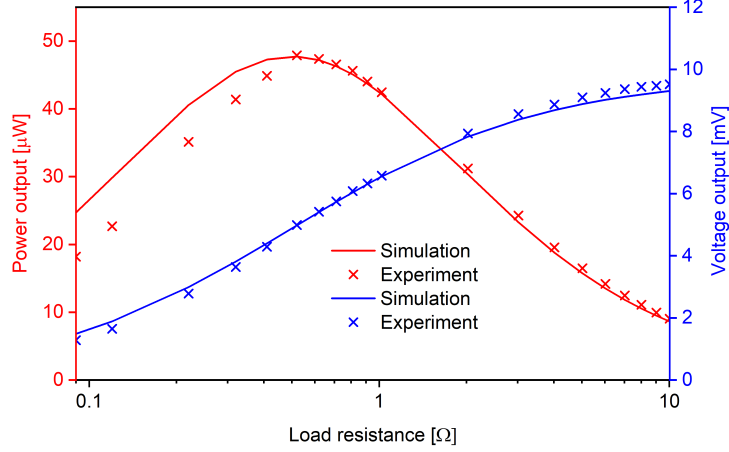


Figure 9: Power and voltage output of the commercial TE device dependent on the load resistance, with a hot side temperature of 34 °C and a cold side temperature of 33 °C. Reprinted from [95].

ure 10. The simplifications are presented below:

- The interconnects were omitted. Due to their thin profile relative to the TE elements in the bulk TEG, the interconnects exhibit much lower thermal resistance and have negligible impact on the performance of the TEG.
- The detailed TE elements were replaced with a representative thermopile (RT) [93], which simplifies the model's geometry and mesh during parametric simulation. While the prior work calculated an equivalent thermal conductivity  $\kappa'$  [100], this model also derives equivalent parameters for the Seebeck coefficient  $\alpha'$ , and electrical resistivity  $\rho'$  from the detailed thermopile's properties.

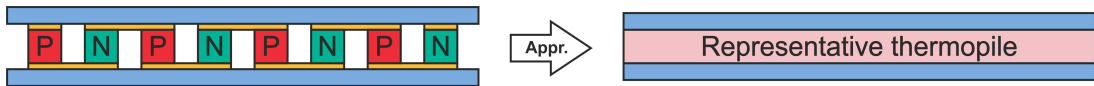


Figure 10: The simplified TEG model, where the interconnects are omitted and the detailed thermopile is replaced by a RT. Reprinted from [95].

According to the Seebeck effect, the open-circuit voltage of a TEG can be expressed as:

$$U_{oc} = n \cdot \frac{\alpha_p - \alpha_n}{2} \cdot \Delta T \quad (2.6)$$

where  $n$  is the number of TE elements,  $\alpha_p$ , and  $\alpha_n$  are the Seebeck coefficients of p- and n-type TE elements, respectively, and  $\Delta T$  is the temperature difference across the TE elements.

The equivalent Seebeck coefficient of the RT, denoted as  $\alpha'$ , can be expressed as:

$$\alpha' = n \cdot \frac{\alpha_p - \alpha_n}{2} \quad (2.7)$$

The TE elements are thermally connected in parallel and are electrically connected in series, as illustrated in Figure 11. Based on this configuration, the thermal conductance ( $K$ ) and electrical resistance ( $R$ ) of the thermopile are given by:

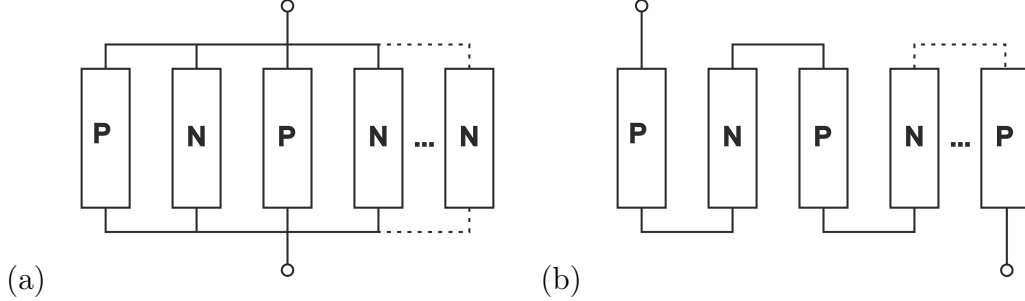


Figure 11: Connection of the TE elements in the TEG. (a) Thermal connection. (b) Electrical connection. Reprinted from [95].

$$K = n \cdot \frac{\kappa_p + \kappa_n}{2} \cdot \frac{a}{L} = \eta \cdot \frac{\kappa_p + \kappa_n}{2} \cdot \frac{A}{L} \quad (2.8)$$

$$R = n \cdot \frac{\rho_p + \rho_n}{2} \cdot \frac{L}{a} = \frac{n^2}{\eta} \cdot \frac{\rho_p + \rho_n}{2} \cdot \frac{L}{A} \quad (2.9)$$

where  $\kappa_p$  and  $\kappa_n$  are the thermal conductivities of p- and n-type TE elements, respectively;  $\rho_p$  and  $\rho_n$  denote the electrical resistivities of p- and n-type TE elements, respectively;  $a$  is the cross-sectional area of a single TE element;  $\eta$  is the fill factor indicating the ratio of the cross-sectional area of total TE elements to TEG substrate;  $A$  is the cross-sectional area of the RT; and  $L$  is the height of the RT.

The equivalent thermal conductivity ( $\kappa'$ ) and electrical resistivity ( $\rho'$ ) of the RT can thus be expressed as:

$$\kappa' = \eta \cdot \frac{\kappa_p + \kappa_n}{2} \quad (2.10)$$

$$\rho' = \frac{n^2}{\eta} \cdot \frac{\rho_p + \rho_n}{2} \quad (2.11)$$

The simplified TEG model contains 38610 DoF, which is 40020 less than the detailed model. The equivalent material properties of the RT in the simplified model are calculated using Equations 2.7, 2.10, and 2.11, where  $\eta$  and  $n$  are 30.4% and 62, respectively. For validation purposes, both models were solved using TE analysis in ANSYS with the same boundary conditions: the temperature at the bottom substrate is fixed to 34 °C and an outward heat flow of 1 W is applied to the top substrate. The simulation results of the two models closely match, with the power output varying with load resistance exhibiting an average error of 0.8% (See Figure 12).

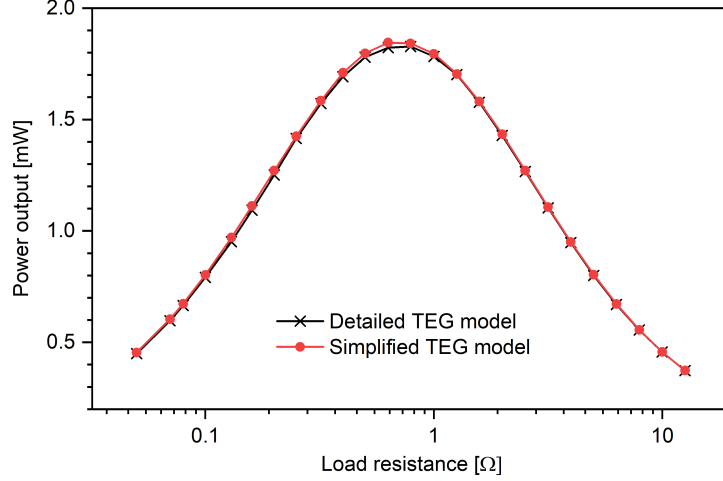


Figure 12: Comparison of the power output between the simplified and detailed TEG models. Reprinted from [95].

### Advanced Thermal TEG modeling

Although the simplification of the TEG model facilitates parametric simulation, its impact on reducing simulation time remains limited. Embedding the TEG model into the tissue model adds complexity, especially when using detailed anatomical geometry, as will be introduced in Chapter 3. Furthermore, design optimization and parametric simulation in such models significantly increase computational efforts.

MOR is a potential solution to address this challenge. Conventional MOR algorithms [101–103] project the original full-scale FE model onto a lower subspace that is constructed by orthonormalized Krylov-subspace, which substantially reduces the model’s DoF while maintaining accuracy. Additionally, multivariate moment-matching algorithms [104–106] enable parametric model order reduction (pMOR), which streamlines design exploration by avoiding the need to repeatedly generate reduced models with changed parameters.

However, applying pMOR to the TE model poses challenges due to the coupled physics in thermoelectricity. Yuan et al. attempted pMOR on a thermal human torso model with an embedded TEG [97]. However, since the full-order model neglects TE effects, the simulation results could not accurately predict the TEG’s performance, despite significantly reduced computation time. To address this issue, a solely thermal TEG model was developed in this work, which incorporates current-dependent Peltier heat by using effective thermal conductivity, as discussed in [107–109].

The development of the thermal TEG model was based on the understanding of the TE theoretical framework. The heat flux in a TE material is driven by both the temperature gradient (Fourier’s law) and the electrical current (Peltier effect):

$$q = \alpha T j - \kappa \nabla T \quad (2.12)$$

The heat transfer in a TE material is given by:

$$\nabla(\kappa \nabla T) + \rho j^2 - T \frac{d\alpha}{dT} j \nabla T = 0 \quad (2.13)$$

Equations 2.12 and 2.13 govern TE phenomena [110].

Assuming homogeneous heat flow and current density through the RT, the temperature of the RT can be calculated using a one-dimensional (1D) model, as depicted in Figure 13.

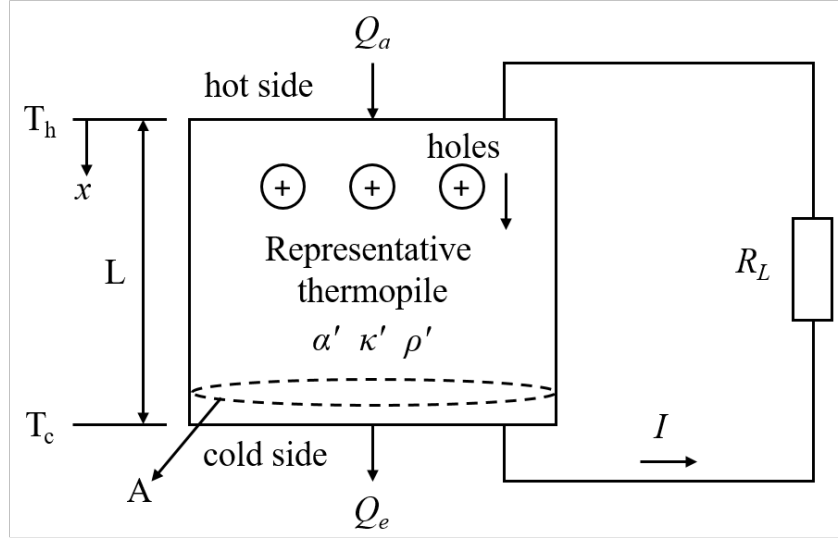


Figure 13: One-dimensional model of the RT, assuming that the heat flow and current density through the RT are homogeneous.

Since the Seebeck coefficient is independent of temperature, Equation 2.13 can be written as:

$$\frac{d}{dx}(\kappa' A \frac{dT}{dx}) + \frac{I^2 \rho'}{A} = 0 \quad (2.14)$$

With boundary conditions  $T_{x=0} = T_h$  and  $T_{x=L} = T_c$ , Equation 2.14 has the solution:

$$T = -\frac{I^2 \rho'}{2\kappa' A^2} x^2 + \left( \frac{I^2 \rho' L}{2\kappa' A^2} - \frac{T_h - T_c}{h} \right) x + T_h \quad (2.15)$$

$$\left. \frac{dT}{dx} \right|_{x=0} = \frac{I^2 \rho' L}{2\kappa' A^2} - \frac{T_h - T_c}{h} \quad (2.16)$$

Based on Equation 2.12, the heat absorbed by the hot side of the RT is given as:

$$Q_a = \alpha' T_h I - \kappa' A \left. \frac{dT}{dx} \right|_{x=0} \quad (2.17)$$

With Equation 2.16, Equation 2.17 is rewritten as:

$$Q_a = \underbrace{\frac{1}{R_{th}} \Delta T}_{Q_c} + \underbrace{\alpha' T_h I}_{Q_{Ph}} - \underbrace{\frac{1}{2} I^2 R_{in}}_{Q_{Jh}} \quad (2.18)$$



Similarly, the heat emitted at the cold side of RT is calculated as:

$$Q_e = \underbrace{\frac{1}{R_{th}}\Delta T}_{Q_c} + \underbrace{\alpha' T_c I}_{Q_{Pc}} + \underbrace{\frac{1}{2} I^2 R_{in}}_{Q_{Jc}} \quad (2.19)$$

where the first term  $Q_c$  is the conductive heat through the RT;  $\Delta T = T_h - T_c$  is the temperature difference across the RT;  $R_{in}$  is the internal electrical resistance of the RT, equal to  $\rho' L/A$ ;  $R_{th}$  is the thermal resistance of the RT, equal to  $L/\kappa' A$ ;  $Q_{Jh}$  and  $Q_{Jc}$  are the Joule heat generated at the hot and cold sides, respectively;  $Q_{Ph}$  is the Peltier heat absorbed at the hot side;  $Q_{Jc}$  is the Peltier heat emitted at the cold side; and  $I$  is the electrical current flowing through the RT.

By introducing the figure of merit ( $Z$ ) of RT and electrical load factor ( $m$ ):

$$Z = \frac{\alpha'^2}{\kappa' \rho'} \quad (2.20)$$

$$m = \frac{R_L}{R_{in}} \quad (2.21)$$

Equations 2.18 and 2.19 can be written as:

$$Q_a = Q_c \left[ 1 + \frac{Z T_h}{1+m} - \frac{Z \Delta T}{2(1+m)^2} \right] \approx Q_c \left( 1 + \frac{Z T_h}{1+m} \right) \quad (2.22)$$

$$Q_e = Q_c \left[ 1 + \frac{Z T_c}{1+m} + \frac{Z \Delta T}{2(1+m)^2} \right] \approx Q_c \left( 1 + \frac{Z T_c}{1+m} \right) \quad (2.23)$$

Here, the temperature difference across the implanted TEG is only a few Kelvins, which is much less than the temperature of the implantation environment (around 310 K). Thus, the Joule heat in Equations 2.22 and 2.23 can be neglected.

Due to the small difference between  $T_h$  and  $T_c$ , the Peltier heat at both sides was approximated to be conductive passing through RT (see Figure 14). The effective conductive heat  $Q'_c$  incorporates the intrinsic conductive heat and the approximated Peltier heat, which is formed as:

$$Q'_c = Q_c \left( 1 + \frac{Z T_{avg}}{1+m} \right) \quad (2.24)$$

where  $T_{avg} = (T_c + T_h)/2$  is the average temperature of the RT. The effective thermal conductivity ( $\kappa_e$ ) of the RT is expressed as:

$$\kappa_e = \kappa' \left( 1 + \frac{Z T_{avg}}{1+m} \right) \quad (2.25)$$

This approximation allows the dependence of the Peltier heat on current to be decoupled, which facilitates the implementation of pMOR. The accuracy of this approximation was validated by transforming the simplified TEG model into a solely thermal TEG model

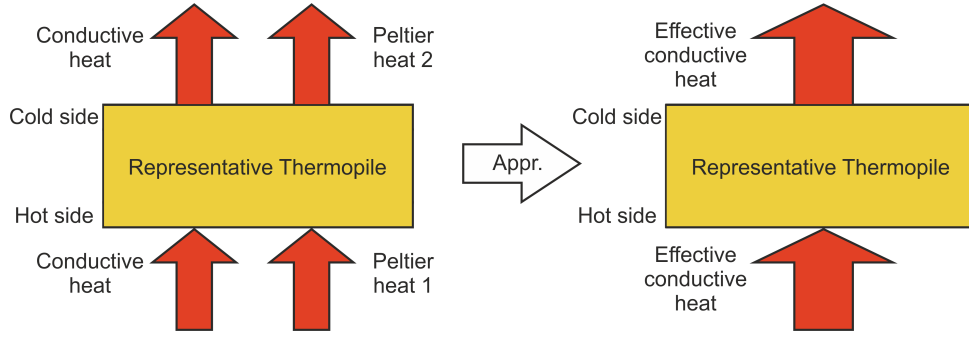


Figure 14: Approximation of the Peltier heat in the thermal TEG model using effective thermal conductivity. The Peltier heat at both sides of the RT is modeled as a part of the effective conductive heat. Reprinted from [95].

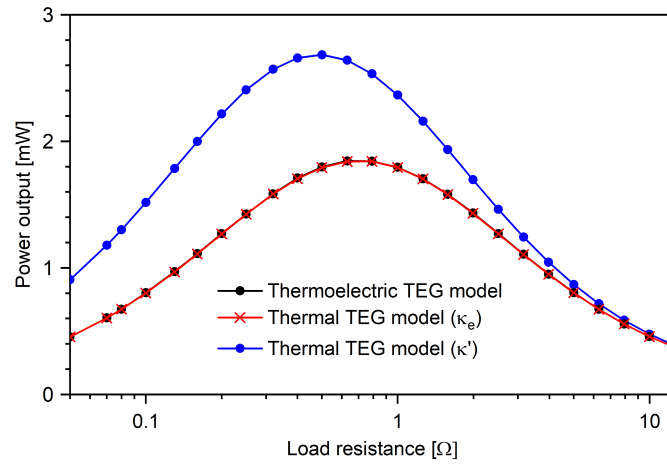


Figure 15: Comparison of the power output between the TE and the thermal TEG model. Reprinted from [95].

and then comparing the results using the boundary conditions introduced in Figure 12. The validation results, depicted in Figure 15, show an average error in power output between the TE and thermal TEG models of 0.6%. Meanwhile, neglecting the Peltier effect in the thermal model (using  $\kappa'$  instead of  $\kappa_e$ ) would overestimate the power output by approximately 30%.

## 2.4 Simulation Results

The implementation of both the tissue and simplified TEG models has been discussed in the preceding sections. Here, the simplified TEG was embedded into the tissue model and then reduced using pMOR for efficient parametric simulation. Following this, parametric simulation was conducted on the implant and tissue to explore their influence on the TEG's power output. An optimization scheme was applied to the model for further investigation. Finally, a prototype TEG design was prepared for experimental investigations.

### 2.4.1 Validation of the Tissue Model Embedded with TEG

Following the initial design outlined in Section 2.2, the simplified TEG was encapsulated by a PEEK housing and integrated into the simplified tissue, as depicted in Figure 16. The square housing measures 60 mm in width and features four chamfered edges with a radius of 6.8 mm, which was designed to mitigate tissue damage. The housing wall thickness is set at 1 mm, while the height of the housing spans 8 mm, occupying the entirety of the fat layer's thickness. The ceramic plates of the TEG are dimensioned at  $20 \times 20 \times 2 \text{ mm}^3$ , making direct contact with the 4 mm high RT and tissue layers. Notably, the heat conductor is not implemented in this configuration. The internal space of the housing is assumed to be devoid of thermal conductivity. This finite element model comprises 54942 elements and 82956 nodes.

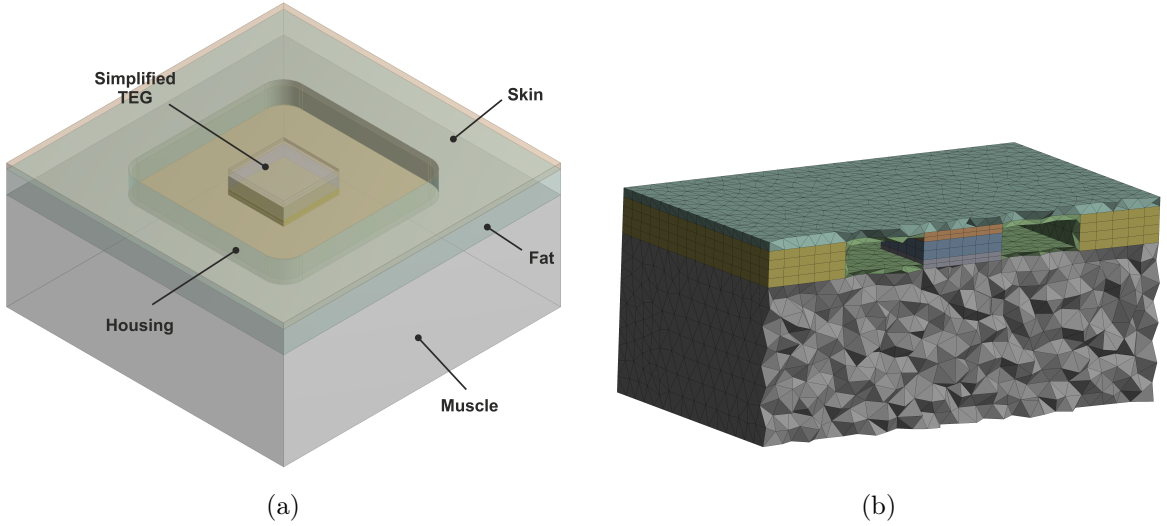


Figure 16: The simplified TEG is packaged with a PEEK housing and embedded in the fat layer of the simplified tissue model. (a) Geometry. (b) Mesh. Reprinted from [95].

As elaborated in Section 2.3.2, the thermal TEG modeling is contingent on the temperature difference being sufficiently small. Consequently, the Joule heat can be neglected. Additionally, the approximation assumes equal Peltier heat at both the hot and cold sides, treating it as if it were conducted through the TEG. To validate this approximation, a TE model was solved in ANSYS.

As depicted in Figure 17(a), the variation in  $\Delta T$  across the TEG, as a function of fill factor and load resistance, is smaller than 8 K, representing less than 3% of the temperature on the hot side of the TEG. Consequently, the Joule heat exerts a minimal impact on the temperature profile and TEG performance. An increase in the fill factor of the TEG causes a decrease in its thermal resistance, leading to a reduction in  $\Delta T$ . Conversely, as the load resistance increases, the effective thermal conductivity ( $\kappa_e$ ) of the RT decreases, which results in a higher  $\Delta T$ . The relative error between the Peltier heat at the hot and cold sides follows a similar trend as  $\Delta T$ , peaking at a maximum value of 2.4% (see

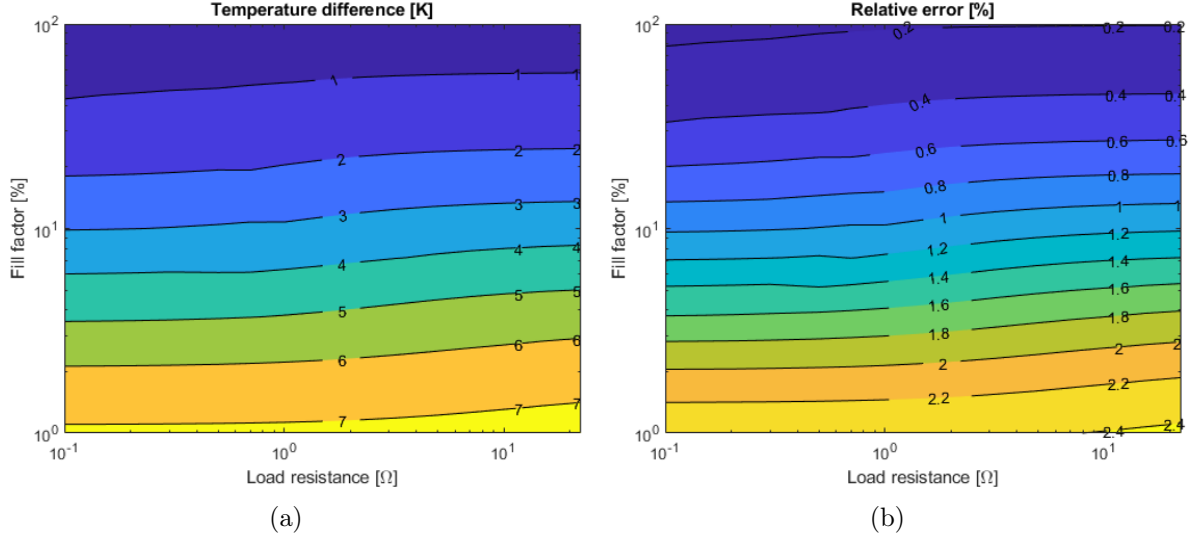


Figure 17: Practicality analysis of the thermal TEG modeling. (a) Temperature difference across the TEG dependent on the fill factor and load resistance. (b) Relative error between the Peltier heat at the hot and cold sides of the TEG. Reprinted from [95].

Figure 17(b)). This observation suggests that the TEG coupling is reasonable in this case.

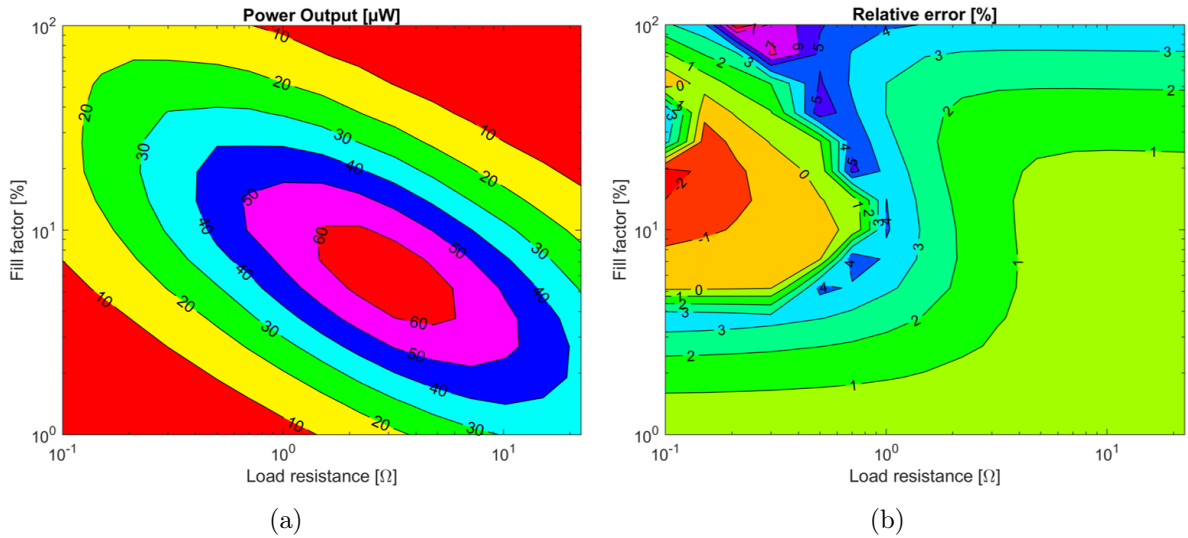


Figure 18: Comparison between the TE model and the thermal model. (a) Power output of the TE model dependent on the fill factor and load resistance (b) Relative error between the power output of the TE and the thermal model. Reprinted from [95].

The solution of the TE model serves as a reference (referred to as the "TE model"). A total of 225 parametric points were computed by sweeping the fill factor ( $\eta$ ) and load resistance ( $R_L$ ), with  $\eta$  selected from  $\{1, 1.4, 1.9, 2.7, 3.7, 5.2, 7.2, 10, 13.9, 19.3, 26.9, 37.2, 51.8, 71.9, 100\}\%$ , and  $R_L$  chosen from  $\{0.1, 0.15, 0.21, 0.3, 0.5, 0.7, 1, 1.5, 2.2, 3.2, 4.8, 7.1, 10.2, 15.1, 22.4\}\Omega$ . The parametric simulation results are presented in Figure 18(a). An optimal fill factor exists, which maximizes power output under electrical

load matching. A higher fill factor corresponds to lower internal resistance of the TEG, resulting in a decrease in the optimal load.

Subsequently, the TEG was replaced by the thermal TEG model in the tissue model (referred to as the "thermal model") and underwent the same parametric simulation using thermal analysis. Upon implantation of the TEG into the tissue, both  $\Delta T$  and the heat flow through the TEG undergo variations due to changes in the parameters. In these scenarios, the interaction between  $\kappa_e$  and  $T_{avg}$  in Equation 2.25 becomes prominent. Consequently, the thermal model utilizing the effective thermal conductivity was iteratively solved, initiating with  $\kappa_e$  evaluated at  $T_{avg} = 300$  K. Throughout the iterative process,  $\kappa_e$  was updated until convergence was achieved.

The power output of the thermal model in this parametric study is consistent with the data of the TEG model (see Figure 18(b)). The relative error in power output between the models ranges from -2.4% to 8.8%, with an average relative error across all parametric points calculated as 1.5%.

Based on the full-order thermal model (84,093 DoF), a reduced-order model <sup>2</sup> with 10 DoF was created using pMOR. In calculations performed on an Intel® Xeon® processor E5-4650, the simplified model simulated 4,500 times faster than the full model. The parametrization shown in Figure 18 was performed on the reduced model to verify its accuracy. The relative error ranges from -4.0 to 7.1% and the average absolute value of these errors is 0.1%.

## 2.4.2 Parametric Analysis and Design Optimization

Previous simulation results have underscored the importance of optimizing the fill factor and electrical load resistance to achieve maximum power output. The optimal electrical load should align with the inner resistance of the TEG, while the optimal fill factor is closely tied to the external thermal load, representing the implantation environment. Changes in the parameters affecting the implantation environment may necessitate adjustments to the optimized values of the fill factor and electrical load resistance.

Table 1: Default parameter values in the TEG embedded tissue model.

| Parameter            | Default value (mm) |
|----------------------|--------------------|
| Fat thickness        | 8                  |
| Muscle thickness     | 35                 |
| Housing width        | 60                 |
| Heat conductor width | 20                 |
| Implantation depth   | 0                  |

To expand upon the model depicted in Figure 16, additional parameters were param-

<sup>2</sup>The reduced model was developed by Chengdong Yuan, the second author in [95].

eterized. Their initial values are provided in Table 1. The implantation depth signifies the distance between the bottom surface of the skin and the upper surface of the implant. Heat conductors, as illustrated in Figure 7, were introduced to improve thermal contact between the simplified TEG and tissue. The thickness of the heat conductors was set to 1.5 mm, while the thickness of the ceramic plates remained fixed at 0.5 mm.

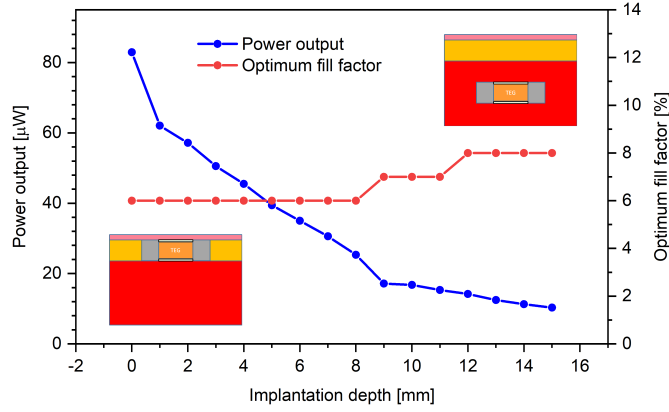


Figure 19: Power output and optimum fill factor dependent on the depth of the implantation. The load resistance equals the TEG’s inner resistance during the fill factor sweep. Reprinted from [111].

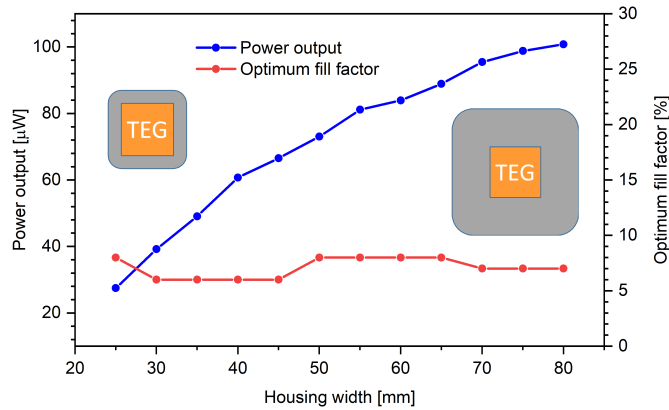


Figure 20: Power output and optimum fill factor dependent on the width of the housing. The load resistance equals the TEG’s inner resistance during the fill factor sweep. Reprinted from [111].

The material properties of the TP were refined through iterative loops, involving the simulations described in this chapter, the fabrication processes detailed in Chapter 4, and the subsequent characterization in Chapter 5. The updated values for the equivalent Seebeck coefficient ( $\alpha'$ ), electrical resistivity ( $\rho'$ ), and thermal conductivity ( $\kappa'$ ) of the TP were determined as  $207 \mu\text{V/K}$ ,  $1.2 \times 10^{-5} \Omega \cdot \text{m}$ , and  $1.5 \text{ W/m/K}$ , respectively.

The influence of housing design on power output was examined by varying implantation depth, as well as the dimensions of the housing and heat conductor. Each design necessitated the determination of a respective optimal fill factor. Maximum power was achieved when the TEG was implanted directly beneath the skin (refer to Figure 19), consistent with findings in previous studies [91, 92, 94]. As the separation between the skin

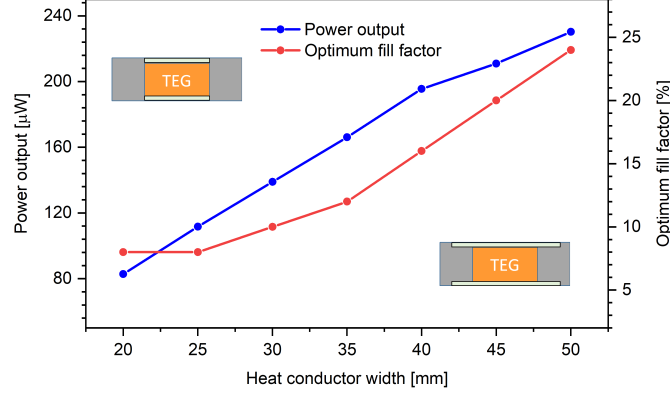


Figure 21: Power output and optimum fill factor dependent on the width of the heat conductor. The load resistance equals the TEG’s inner resistance during the fill factor sweep. Reprinted from [111].

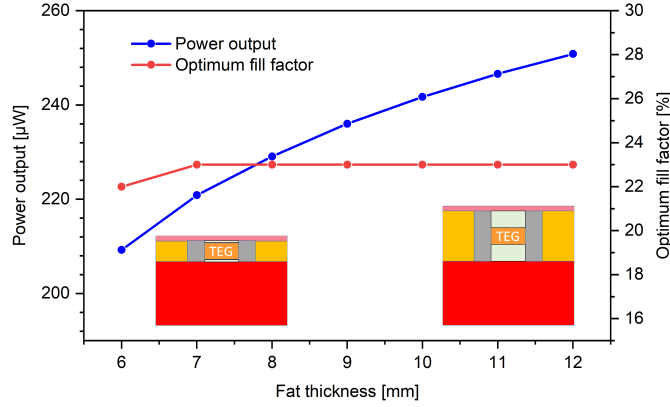


Figure 22: Power output and optimum fill factor dependent on the thickness of the fat. The load resistance equals the TEG’s inner resistance during the fill factor sweep. Reprinted from [111].

and device surface increases, the temperature difference, and consequently power output, decreases. Deeper implantation locations exhibit lower temperature gradients, resulting in a smaller temperature difference across the TEG. Meanwhile, the optimum fill factor needs a slight adjustment to maintain thermal resistance matching. The power output rises with housing size due to its higher thermal insulation, as depicted in Figure 20. The optimum fill factor remains nearly unchanged.

Larger heat conductors enhance the heat flow through the TEG, thereby improving its performance when implanted (see Figure 21). In comparison to the impact of housing position and size, the dimensions of the heat conductors significantly affect power output and the required optimum fill factor. In the scenario presented in Figure 21, with a 60 mm wide housing and 50 mm wide heat conductors, the optimal fill factor of the 4 mm-high TE elements is 24%, yielding a power output of 230  $\mu$ W in a matched resistive load.

The influence of tissue layer thickness on power output was investigated, considering variations between individuals and across different body locations. While the skin thickness remained fixed at 2 mm, the width of the metal plate was set to 50 mm. To fully utilize the thickness of the fat layer, the height of the housing and metal plates

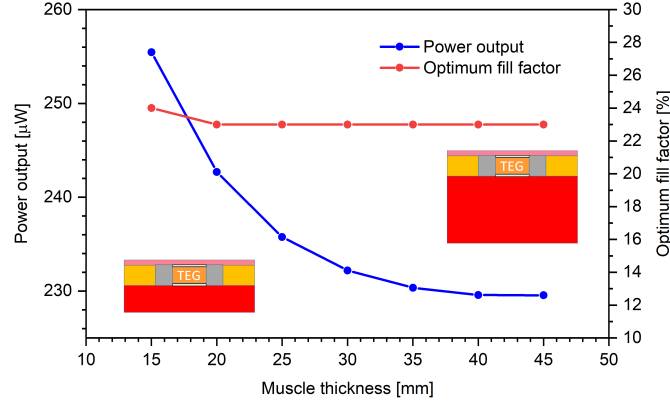


Figure 23: Power output and optimum fill factor dependent on the thickness of the muscle. The load resistance equals the TEG’s inner resistance during the fill factor sweep. Reprinted from [111].

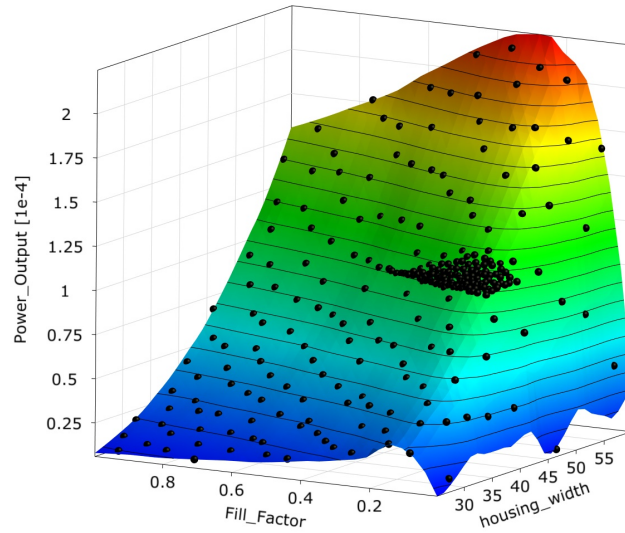


Figure 24: Response surface for power output [W] dependent on the fill factor and housing width [mm] generated using AMOP. The optimization objective is to minimize the housing width and obtain at least  $100 \mu\text{W}$  of power. More design candidates (black dots) are sampled around the optimal design. Reprinted from [111].

were adjusted accordingly, providing an optimal thermal scenario. Thermal and electrical resistance matching was applied to the fill factor and the load resistor, respectively, for each design.

Power output was found to increase for higher fat layer thickness (see Figure 22), while a thicker muscle layer resulted in decreased power output (see Figure 23). However, these variations only had a minimal impact on the optimal fill factor.

Common IMDs typically require power inputs ranging from tens of microwatts to a few milliwatts [112]. To identify a packaged TEG design with minimal dimensions that can still provide  $100 \mu\text{W}$ , an optimization scheme was established using ANSYS optiSLang [113]. The adaptive metamodel of optimal prognosis (AMOP) was employed for this purpose. The width of the heat conductors was set to be 5 mm less than the



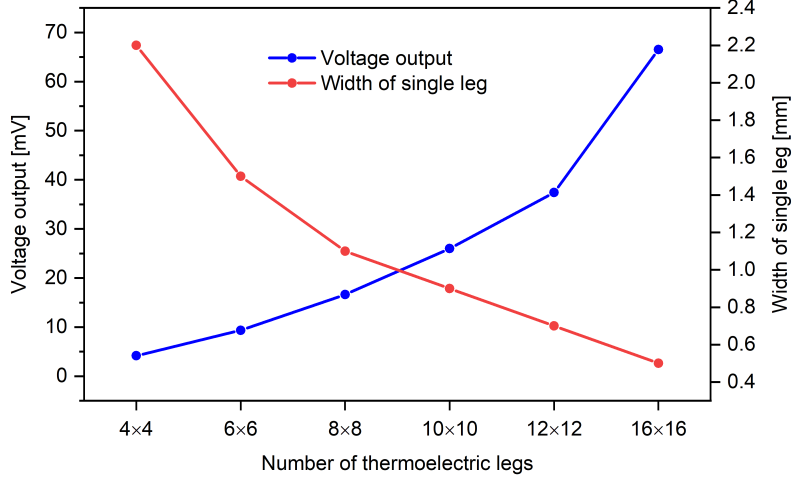


Figure 25: Voltage output and width of a single leg dependent on the number of TE legs. The TEG is packaged in a 42.3 mm wide housing. The fill factor and load resistance are kept optimized. Reprinted from [111].

width of the housing. Polynomial regression [114] was utilized to initially construct a response surface of power output dependent on the fill factor and housing width across an extended parameter space. Moving least squares [115] and Kriging [116] approaches were subsequently used to sample multiple design candidates around the optimal design for more accurate prediction. The optimization revealed that the width of the housing could be reduced to 42.3 mm with an optimum fill factor of 19.3% (see Figure 24).

For a given fill factor, the number of TE elements does not affect power but does impact the voltage level. Maintaining the optimum fill factor, the voltage output dependent on the number of TE legs is depicted in Figure 25. Increasing the number of TE legs leads to higher output voltages. However, to accommodate the increasing number of legs within the size of the TEG, the width of a single leg must be reduced. When the leg width is reduced to 0.5 mm, the voltage output increases to 65 mV. Nevertheless, most implantable devices operate within a voltage range of 2 to 3 volts, requiring the use of voltage boost converters. These circuits require at least 20 mV of input voltage [117], necessitating a leg width of less than 1 mm. Consequently, with a leg height of 4 mm, the aspect ratio (height-to-width) of a single leg is expected to exceed 4:1, which presents a challenge for the TEG manufacturing process.

### 2.4.3 Prototype TEG Design

In the previous sections, the influence of various parameters on TEG performance has been thoroughly examined. Leveraging these findings, a packaged TEG design for the experiment was developed that balanced considerations such as performance, dimension, geometry complexity, assembly requirements, and manufacturability. This design is illustrated in Figure 26(a).

In comparison to the original design, the housing has been adapted to a disc shape to

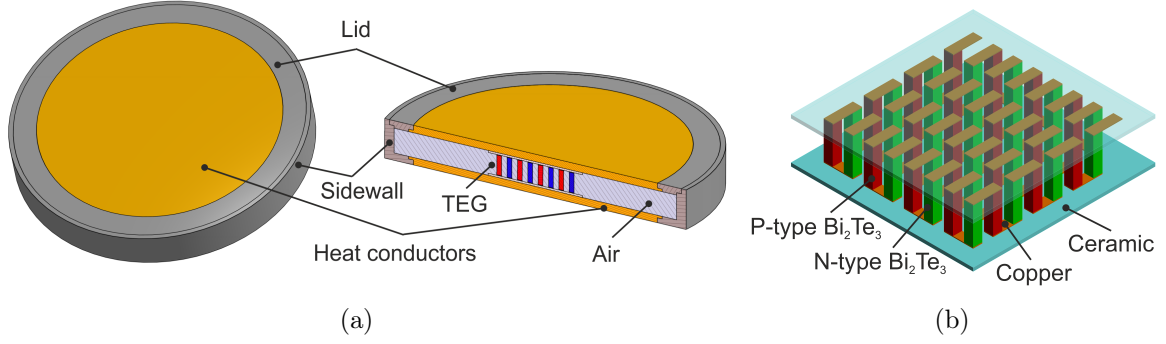


Figure 26: (a) Prototype TEG design, where the TEG is sandwiched by two heat conductors (gold-coated brass) and packaged by a disk-shape housing (PEEK). (b) Optimized TEG featuring 62 thermolegs ( $1 \times 1 \times 4 \text{ mm}^3$ ) within a cross-section of  $20 \times 20 \text{ mm}^2$ . Figure (a) is reprinted from [118]. Figure (b) is reprinted from [111].

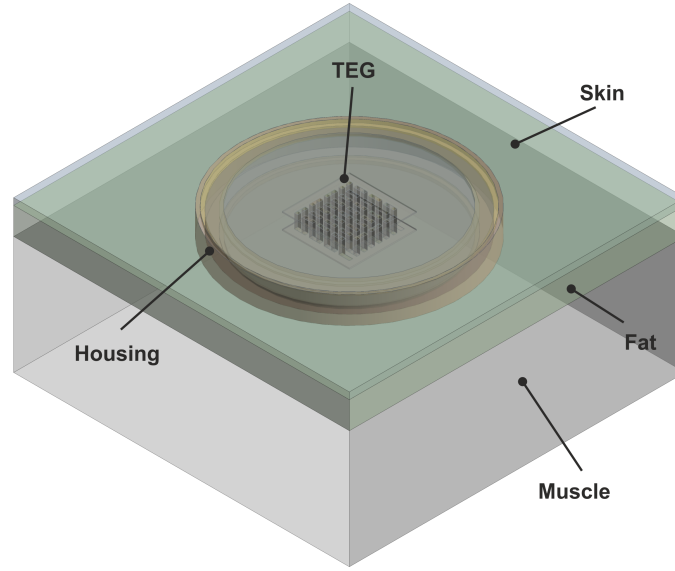


Figure 27: Model of the fabrication-ready TEG design embedded in the simplified tissue.

enhance compatibility with the implantation environment and to reduce manufacturing difficulty. The housing, with a diameter of 65 mm, consists of a lid, a side wall, and two heat conductors, with a total height of 8 mm. The lid and side wall are manufactured from PEEK, while the two heat conductors are made from gold-coated brass. Each heat conductor has a diameter of 55 mm and a thickness of 1.5 mm, and the side wall has a thickness of 2 mm. The interior of the housing is filled with air.

This design, with detailed geometry, was simulated in the simplified tissue model, as depicted in Figure 27. The array configuration of the TEG was optimized to maximize power output while ensuring sufficient input voltage to the boost converter. This optimization resulted in a near-optimum fill factor of 15.5% within a cross-section of  $20 \times 20 \text{ mm}^2$ . The TEG is equipped with 62 p- and n-type  $\text{Bi}_2\text{Te}_3$  thermolegs, featuring high aspect ratio dimensions of  $1 \times 1 \times 4 \text{ mm}^3$  (see Figure 26(b)). When a matched resistive load of  $3.2 \Omega$  is connected to the TEG, it provides  $153 \mu\text{W}$  of power at a voltage of 22 mV.

The simulation results reveal that the hot and cold side temperatures of the TEG

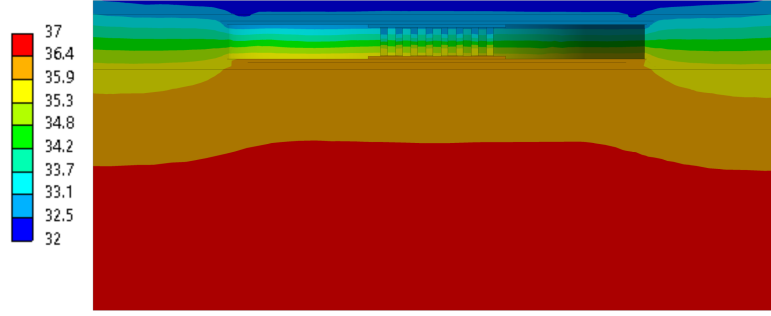


Figure 28: Temperature distribution of the TEG embedded tissue model, where the temperature difference across the TEG is 3.3 K.

are 35.9 °C and 32.6 °C, respectively. These temperatures will serve as the basis for the TEG characterization in Section 5.6.3 to establish a realistic thermal environment in this implantation scenario. The temperature distribution of the tissue and TEG is illustrated in Figure 28. The temperature difference across the TEG is larger than the temperature difference across the surrounding tissue. Due to the high thermal conductivity of the heat conductors, the temperature of the heat conductors is homogeneous. This suggests that the packing density and shape of the thermolegs have limited impacts on the thermal profile and TEG performance when the fill factor is fixed.

## 2.5 Summary

This chapter presents a preliminary investigation into ITEGs for powering IMDs. The initial design considerations include application scenarios, focusing on performance and biocompatibility. Consequently, a novel packaging design was proposed that uses PEEK for the main body of the housing and gold-coated brass for the heat conductors.

To address key questions regarding the feasibility and potential of ITEGs, the following aspects were considered:

- The power and voltage generation capability of the ITEGs and their adequacy for IMDs.
- The influence of the design parameters on the performance of the TEG and opportunities for optimization.
- The dimensions and array configuration of optimized TEGs, ensuring practicality and feasibility for fabrication.

Given the absence of clinical data, numerical simulations are deemed necessary. However, ensuring reliable results requires iterative refinement and cross-validating simulations with experimental data, which is a time-consuming process. Thus, the primary focus of this chapter was on conducting accurate and time-efficient proof-of-concept simulations, as follows:

- A simplified tissue model was developed, which accounts for the main bioheat effects.
- A simplified thermal TEG model was constructed, which facilitates parametric analysis and MOR techniques.
- Parametric analysis and design optimization were done within the tissue model with embedded TEG.

A fabrication-ready TEG design was optimized in the simplified tissue model and will serve as the foundation for the experimental work, detailed in Chapter 4. The housing has a diameter of 65 mm and a height of 8 mm. The optimized TEG features an  $8 \times 8$  array of  $\text{Bi}_2\text{Te}_3$  legs ( $1 \times 1 \times 4 \text{ mm}^3$ ) within a cross-section of  $20 \times 20 \text{ mm}^2$ . The results of the simulation indicate that this TEG can provide  $153 \mu\text{W}$  of power to a resistive load at a voltage of 22 mV, which demonstrates the potential for powering IMDs with a boost converter for voltage up-conversion.

### 3 ITEG Embedded in Detailed Human Body Model

While the simplified tissue model (Chapter 2) provides a valuable starting point for ITEG research, its simplification limits accuracy due to the omission of complex tissue structures and organ interactions. This chapter aims to improve simulation accuracy by developing anatomically realistic human torso and head models. These highly detailed models will enable a more comprehensive analysis of how design parameters influence TEG performance within a realistic human body environment. The simulations in these models enable a deeper understanding of real-world TEG behavior, including predictions of power output, optimum dimensions, hermeticity, and manufacturability. Ultimately, this chapter seeks to establish design guidelines for ITEGs, which balance performance requirements with the practical realities of fabrication and implantation.

#### 3.1 Methods Review

In human body heat research, the classification of thermal and thermoregulatory human body models lacks a fixed criterion. All bioheat effects in the human body contribute to thermal regulation to maintain normothermic temperatures. Thermal models typically encompass steady-state models, focusing on major bioheat effects like metabolism, blood perfusion, and external heat exchange through the skin. In contrast, thermoregulatory models are transient, besides the main bioheat effects incorporating periodic effects (e.g., respiration and circadian rhythm) or responses to external environmental stimuli (e.g., transpiration, vasodilation, vasoconstriction, and shivering).

Recent advancements have led to the development of sophisticated thermal and thermoregulatory models with high resolutions [123]. These models derive their geometry from magnetic resonance imaging (MRI) or three-dimensional (3D) surface scanning, enabling them to represent realistic anatomical geometries. For instance, Nelson et al. [119] created a thermoregulatory voxel model based on anatomical data to investigate body temperature responses in seated adult humans exposed to warm or hot environments. Silva et al. [120] developed a thermal body model from a computed tomography (CT) scan to study brain cooling procedures, particularly for therapeutic hypothermia after ischemia. Unnikrishnan et al. [121] constructed a 3D virtual human thermoregulatory model to predict temperature distributions under conditions of physical exertion and environmental thermal stress. Castellani et al. [122] devised a thermoregulatory female model to explore human body temperature distribution in non-uniform and transient environments.

These models significantly enhance temperature prediction accuracy within the human body due to their anatomically realistic geometries, as demonstrated in Figure 29. However, to date, none of these models have been specifically applied to study ITEGs. There exists a notable gap in the literature regarding a comprehensive design study for ITEGs.

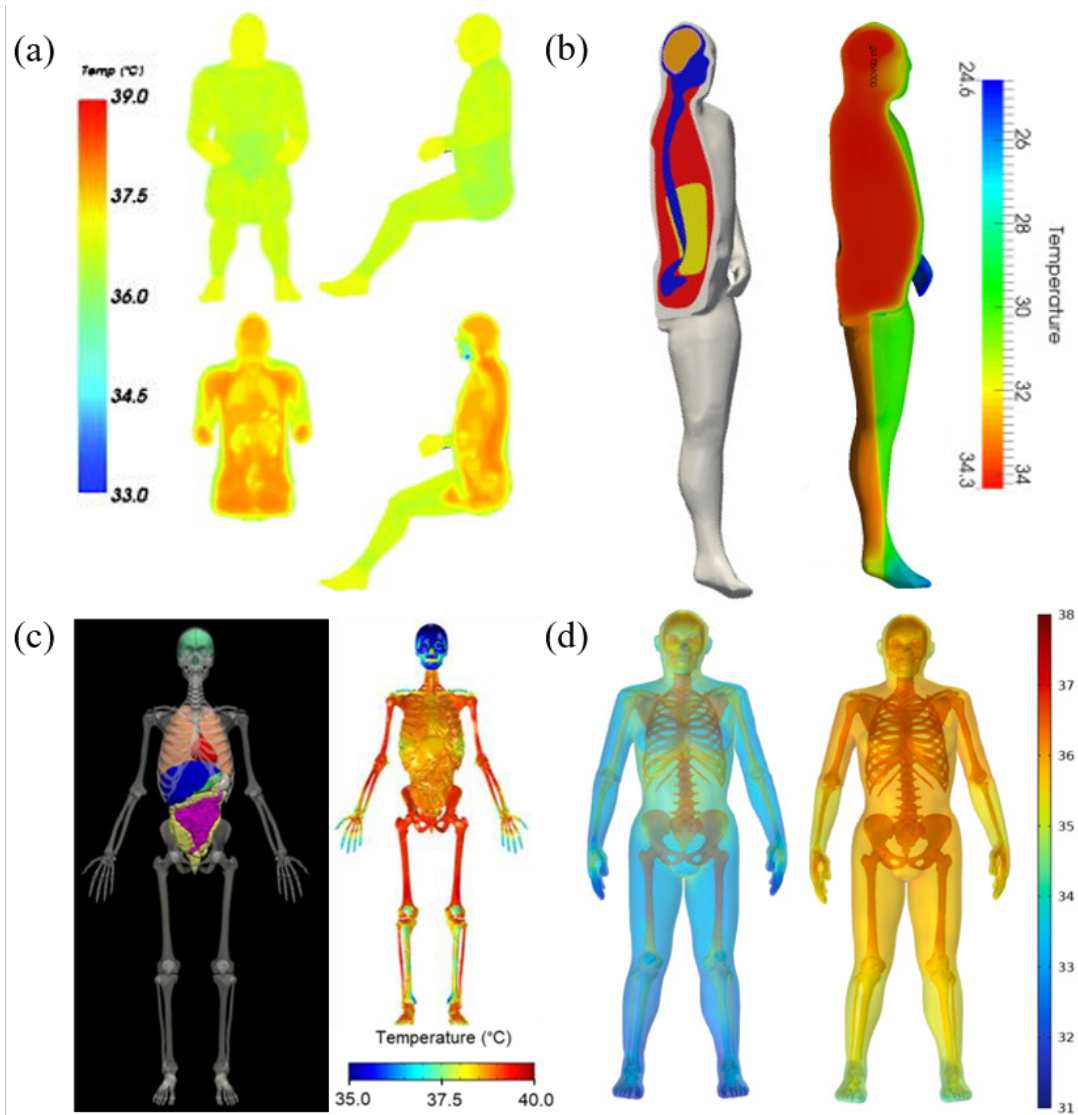


Figure 29: Overview of the high-resolution thermal or thermoregulatory human body models. (a) Temperature distribution of a seated adult human in an extremely hot exposure protocol (one-hour exposure at 28.1 °C and 43% relative humidity, followed by two-hour exposure at 47.8 °C and 27% relative humidity). The temperature above is the skin surface view, and the temperature below is the body core midplane section view. Image adapted from [119]. (b) Temperature distribution of an adult human after 24 hours of whole-body cooling treatment (right-hand panel). The model's geometry is halved due to the symmetry, which incorporates the brain, viscera, and lungs with simple shapes (left-hand panel). Image adapted from [120]. (c) Temperature distribution of a male human model after 2 hours of exercise with T-shirt/shorts at 36 °C and 30% relative humidity (right-hand panel). The model's geometry incorporates the major organs, including the brain, heart, lungs, liver, intestines, and the skeletal system (left-hand panel). Image adapted from [121]. (d) Temperature distribution of the skin and skeleton in a female human model in different environmental conditions: 25.6 °C and 50% relative humidity (left-hand panel); and 50 °C and 17% relative humidity (right-hand panel). Image adapted from [122].

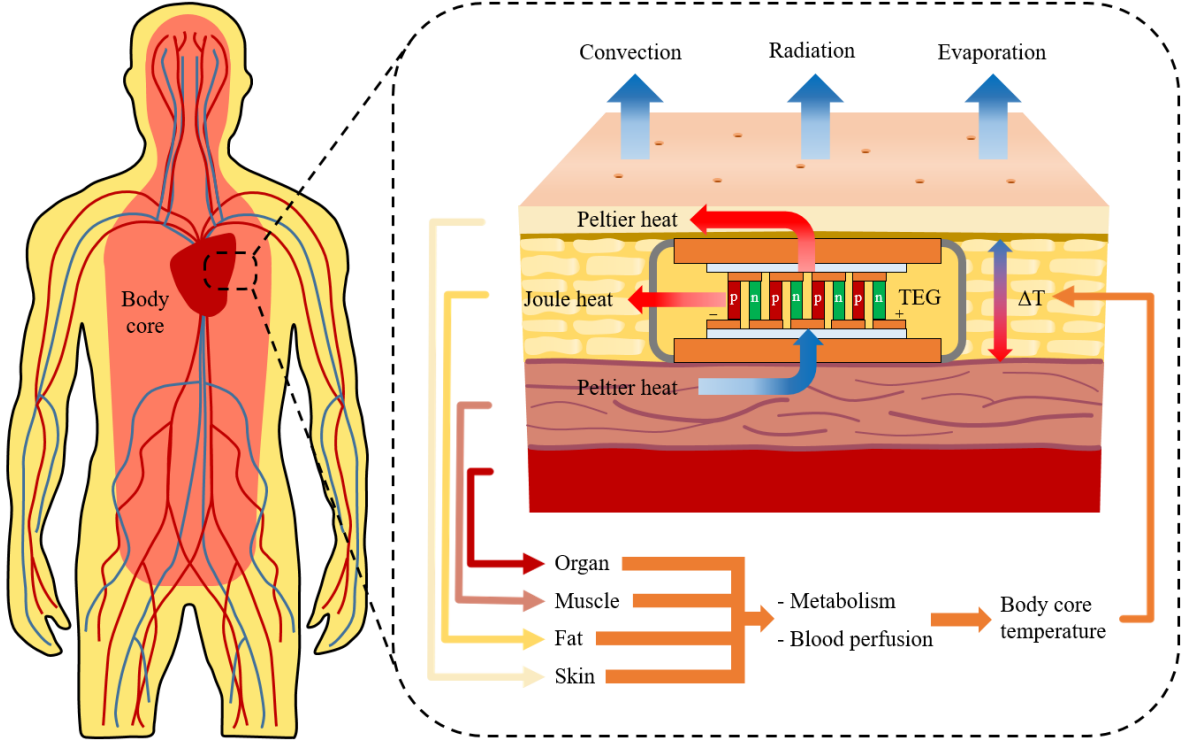


Figure 30: The human body maintains a core temperature of around 37 °C through metabolism and blood perfusion in organs and tissues, while skin temperature is typically lower. This temperature difference can be harnessed by an implanted TEG to convert heat into electricity. At the skin's surface, heat dissipates through convection, radiation, and evaporation. A TEG's thermopile liberates Joule heat as electricity is generated. Additionally, the Peltier effect causes heat absorption at the TEG's hot side (near the body core) and heat generation at its cold side (near the skin). Reprinted from [124].

## 3.2 Computational Model

To simulate the implantation scenarios of TEGs, finite element models of a human torso and a human head were developed. Figure 30 depicts the heat transfer mechanism within these human body models with embedded TEGs. These bioheat thermal models incorporate conductive heat transport, along with the thermal effects of metabolism and blood perfusion in all organs and tissue regions. Additionally, convection, radiation, and evaporation effects were incorporated into the skin surface to account for heat loss mechanisms, influenced by environmental conditions such as ambient temperature, air velocity, relative humidity, and skin wetness. Parameterized TEGs were integrated into the body tissue and their power delivery was assessed under a resistive load. The simulations accounted for the Joule and Peltier heat generation caused by current flow within the operating TEGs. The simulations were conducted with steady-state TE analysis in ANSYS (2023 R2).



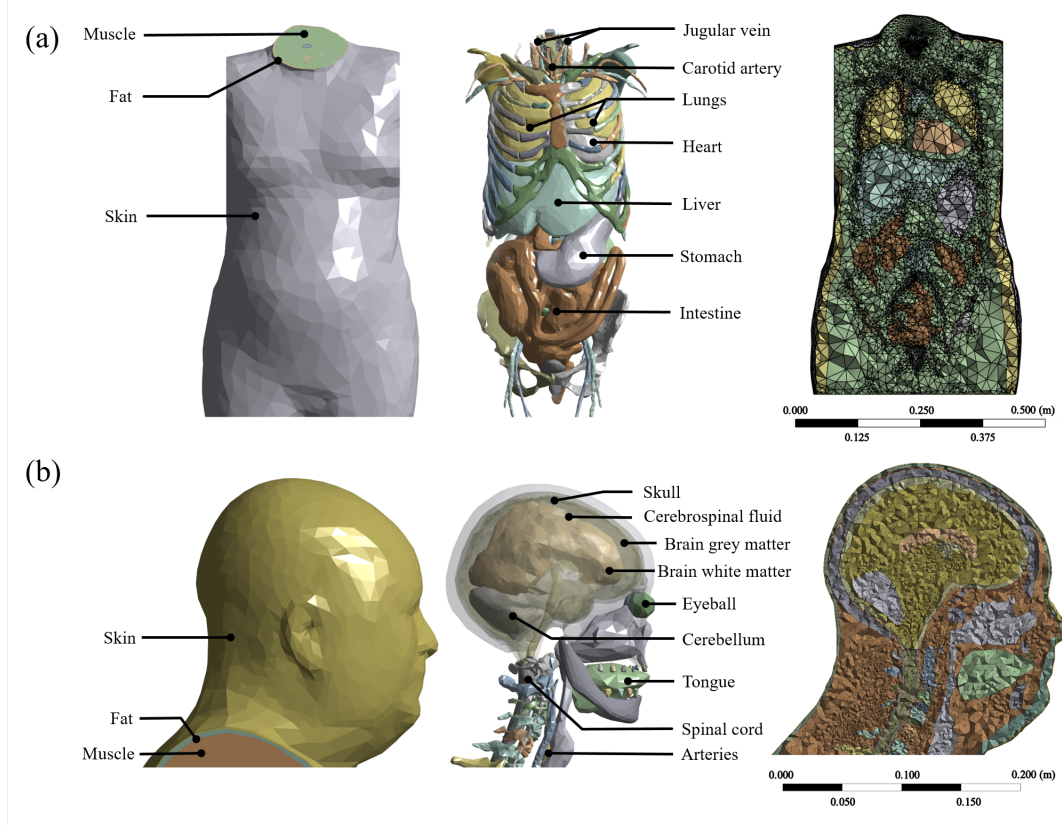


Figure 31: Anatomical geometry and mesh of the human body models: (a) torso and (b) head. Reprinted from [124].

#### 3.2.1 Human Body Model

The human body geometry model was derived from an open-source computer-aided design (CAD) model based on a high-resolution cryosection image dataset [125]. Preprocessing was performed on the geometry to adapt the model for the simulation:

- The artifacts in the model (e.g., self-intersections, holes, and overlapping domains) were removed or improved.
- The body shell was separated into individual tissue types (i.e., skin, fat, and muscle). The muscle section is an entire part that wraps around the organs and bones.
- The solid heart was adapted to present a cavity filled with blood surrounded by the heart muscle.
- The torso and head sections were separated from the entire body to reduce the computational complexity when incorporating TEG sub-models.

The torso and head models comprise various tissues, bones, and major organs (see Figure 31). The torso and head meshes consist of roughly 0.8 and 1.2 million four-node tetrahedral elements, respectively, ensuring a high level of geometric detail. The skin of the human body models incorporates heat dissipation effects as described in Section 2.3.1,



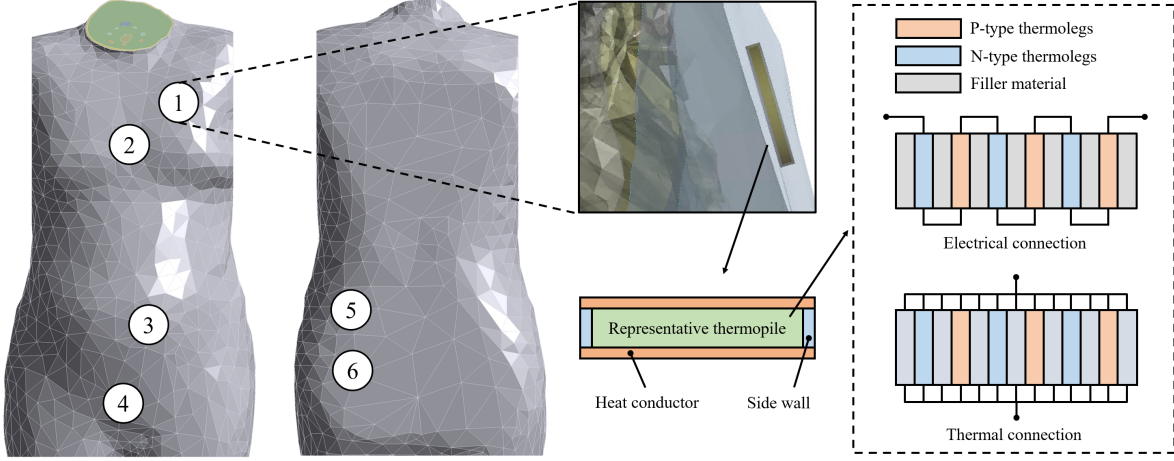


Figure 32: A packaged TEG is embedded into the upper chest region of the human torso model. The TEG’s interconnects and substrates are neglected. The package consists of a side wall around the TEG and two heat conductors attached to the TEG. The detailed TEG geometry is simplified into a representative thermopile, which represents the electrical series connection of the thermocouple legs and the thermal parallel connections of the legs and filler material (insulating but heat-conducting). Reprinted from [124].

with ambient temperature set at 21 °C, air velocity maintained below 0.2 m/s, relative humidity at 0.4, and skin wetness set to 0.07. Unlike the simplified tissue model, where a fixed temperature of 37 °C was applied to the bottom surface of the muscle section, no Dirichlet boundary condition was applied to the human body models. The material properties of the tissue and organs were sourced from publicly available literature [98], including thermal conductivity, blood perfusion rate, and metabolic heat generation rate, as presented in Table A1.

### 3.2.2 TEG Integration

Before embedding the TEG into the human body, a protective housing is employed to encapsulate the TEG, safeguarding it from post-implantation complications and providing protection against external factors such as contaminants, physical damage, and moisture. The package comprises a sidewall and two metal heat conductors that attach to the TEG. The inner space of the package accommodates the TEG and optional filler material.

A standard TEG comprises thermocouple legs made of thermoelectric material, electrical interconnections, and substrates for mechanical support. To facilitate parametric simulation and reduce computation time, certain reasonable simplifications were applied to the implants (see Figure 32). The interconnects and substrates were omitted because they have minimal impact on the thermal profile and TEG performance. Within the device’s inner space, the thermocouple legs are electrically connected in series, while the thermocouple legs and filler material are thermally connected in parallel. Considering this arrangement, the inner space of the implant was substituted with an RT to account for the

thermal properties of the filler material and the TE properties of the thermocouple legs. The Seebeck coefficient and electrical resistivity of the RT are described using Equations 2.7 and 2.11, respectively. The thermal conductivity of the RT is defined as:

$$\kappa' = \eta \cdot \frac{(\kappa_p + \kappa_n)}{2} + (1 - \eta) \cdot \kappa_{fill} \quad (3.1)$$

where  $\kappa_p$  and  $\kappa_n$  are the thermal conductivities of p- and n-type bismuth telluride, while  $\kappa_{fill}$  represents the thermal conductivity of the filler material.  $\eta$  is the fill factor, indicating the ratio of the cross-sectional area of the total thermocouple legs to the RT ( $A$ ). The accuracy of this simplification has been demonstrated in Section 2.3.2.

The prolonged presence of implants within the human body induces the development of fibrous tissue, commonly known as fibrous encapsulation, around biocompatible implants, thereby enhancing the connection between the tissue and implants [126]. As a result, the thermal contact resistance between the ITEG and the human body is significantly lower than that observed between the wearable TEG and the human body. Notably, the model did not consider the thermal contact resistance between the ITEG and the human body due to the absence of literature describing this aspect.

Initially, the sidewall of the housing was composed of PEEK, which is a material that is widely employed in implants due to its commendable biocompatibility and structural strength. PEEK also contributes to the implanted TEG's performance by providing effective heat insulation. The heat conductors were assumed to be crafted from gold-coated copper to enhance heat collection and dissipation. Air served as the filler material within the inner space of the packaging. The initial values for the geometry parameters of the packaged TEG are listed in Table 2.

Table 2: Initial values for the geometry parameters of the packaged TEG.

| Parameter                              | Default value |
|----------------------------------------|---------------|
| Height of the thermolegs ( $h$ )       | 4 mm          |
| Diameter of the implant device ( $d$ ) | 50 mm         |
| Implantation depth ( $D$ )             | 5 mm          |
| Height of the side wall                | 4 mm          |
| Thickness of the side wall             | 1 mm          |
| Height of the heat conductors          | 1 mm          |

The material properties of the substances utilized in the TEG model are detailed in Table A1. The material properties of bismuth telluride were derived from measurements conducted in Chapter 5, where  $(\alpha_p - \alpha_n)/2$  is 207  $\mu\text{V/K}$ ,  $(\rho_p + \rho_n)/2$  is 1.2  $\mu\Omega\cdot\text{m}$ ,  $(\kappa_p + \kappa_n)/2$  is 1.5  $\text{W/m/K}$ .

### 3.2.3 Governing Equations and Equivalent Circuit

The steady-state heat equation describes the spatial temperature distribution within the simulation domain due to Pennes' bioheat terms [76] in the tissue, Peltier and Joule heat in the TE material, and conductive heat transfer as well as heat loss mechanisms. It is expressed as:

$$\nabla(\kappa \nabla T) + \underbrace{\rho_b c_b \omega (T_a - T)}_{q_b} + q_m + \rho j^2 - T \frac{d\alpha}{dT} j \nabla T = 0 \quad (3.2)$$

where  $\alpha$ ,  $\kappa$  and  $\rho$  are the material's Seebeck coefficient, thermal conductivity, and electrical resistivity, respectively.  $q_b$  and  $q_m$  describe the effects of blood perfusion and metabolic heat generation. Additionally,  $\rho_b$ ,  $c_b$ , and  $T_a$  represent the density, specific heat, and temperature of arterial blood. The term  $\omega$  represents the perfusion rate and  $j$  signifies the electrical current density.

The first three terms in Equation 3.2 have been introduced in Equation 2.1.  $\rho j^2$  describes the Joule heat due to the electrical current flowing in the TE material during TEG operation. The last term accounts for the heat generated from thermoelectricity in the TE material. Due to the low-temperature difference in the implantation environment, the Seebeck coefficient was assumed to be temperature-independent in the model. Therefore, no Thomson heat was involved in the last term, although it contributes to Peltier heat at the interfaces of the TE material.

The electric behavior of the TE material is governed by Ohm's law and the Seebeck effect, which can be represented as:

$$\nabla \phi + \alpha \nabla T + \rho j = 0 \quad (3.3)$$

where  $\phi$  denotes the electric potential distribution.

A conceptual schematic of the equivalent thermal and electrical circuits, which illustrates the TEG mechanism in the implantation scenario, is presented in Figure 33. This circuit neglects lateral heat conduction between the implant and tissue, which provides a concise visualization of the coupling effect and the relationships between different parameters.

The TEG harnesses temperature gradient within the body to generate electromotive force. In the model, the open-circuit voltage of the TEG can be expressed as the product of  $\alpha'$  and the temperature difference ( $\Delta T$ ) across RT:

$$U_{oc} = \alpha' \cdot \Delta T \quad (3.4)$$

When an electrical load ( $R_L$ ) is connected to the TEG, an electrical current leads to the generation of Peltier and Joule heat, impacting the thermal profile within the human body.

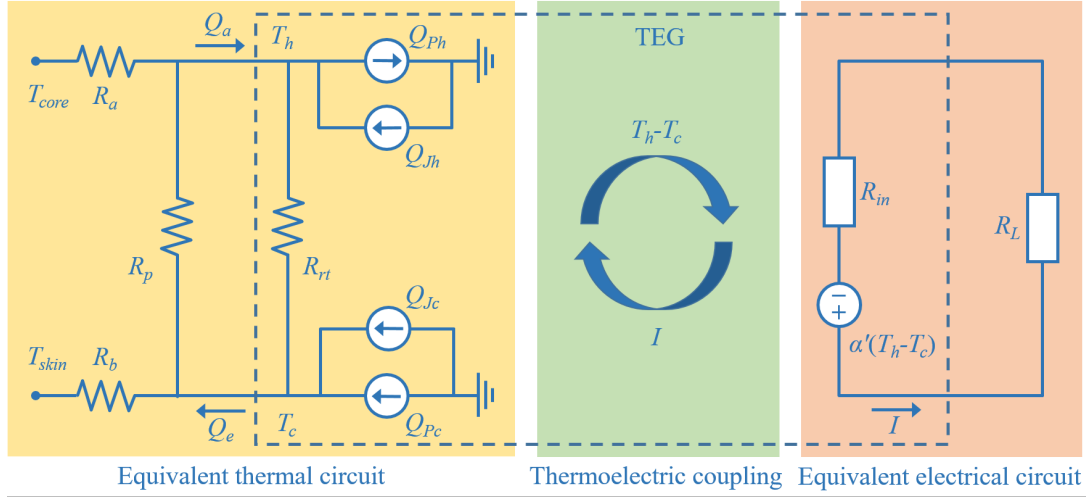


Figure 33: Conceptual schematic of the equivalent thermal and electrical circuits of the TEG embedded within the human body model with coupling interactions. Notations: Skin temperature ( $T_{skin}$ ); body core temperature ( $T_{core}$ ); temperatures at the hot side ( $T_h$ ) and cold side ( $T_c$ ) of the thermopile; thermal resistance below ( $R_b$ ) and above ( $R_a$ ) the thermopile; heat absorbed ( $Q_a$ ) and emitted ( $Q_e$ ) at the thermopile; thermal resistance of the thermopile ( $R_{th}$ ); lateral thermal resistance surrounding the thermopile ( $R_p$ ); Joule heat generation at the hot ( $Q_{Jh}$ ) and cold sides ( $Q_{Jc}$ ); Peltier heat absorption at the hot side ( $Q_{Ph}$ ); Peltier heat emission at the cold side ( $Q_{Jc}$ ); internal electrical resistance of the thermopile ( $R_{in}$ ); current through the thermopile ( $I$ ); and electrical load ( $R_L$ ). Reprinted from [124].

Given the temperature at the hot side ( $T_h$ ) and the temperature at the cold side ( $T_c$ ) of RT, the heat absorbed at the hot side ( $Q_a$ ) is given by Equation 2.18. The heat emitted at the cold side ( $Q_e$ ) follows Equation 2.19.

The electrical power output of the TEG is formulated as:

$$P_{out} = Q_a - Q_e = \alpha' I \Delta T - I^2 R_{in} \quad (3.5)$$

when  $R_L$  is equal to  $R_{in}$ ,  $P_{out}$  can be rewritten as:

$$P_{out} = \frac{(\alpha_p - \alpha_n)^2 Q_c \Delta T}{4(\kappa_p + \kappa_n)(\rho_p + \rho_n)} \quad (3.6)$$

The influence on  $P_{out}$  is given by a combined result of the temperature difference and heat flow in the thermal domain, where  $\Delta T = T_h - T_c$ ,  $Q_c$  is the conductive heat through the RT (detailed in Equation 2.18 and 2.19).

The highest temperature difference within the human body exists between the skin temperature ( $T_{skin}$ ) and body core temperature ( $T_{core}$ ). However, the complete temperature difference cannot be utilized by the implanted TEG because a significant portion is present across the thermal resistance below ( $R_b$ ) and above ( $R_a$ ) the thermopile.  $R_b$  and  $R_a$  encompass the resistance of tissue and heat conductors.

Similarly,  $Q_c$  does not reach its maximum possible value due to the thermal resistance ( $R_p$ ) laterally surrounding the thermopile, causing parasitic heat transfer.  $R_p$  comprises

the thermal resistance of the housing sidewalls and filler material. Since the heat conductors across  $R_{rt}$  and  $R_p$  exhibit high thermal conductivity, it is assumed that the cold side and hot side of  $R_{rt}$  and  $R_p$  maintain uniform temperatures.

It is important to note that Equation 3.6 is calculated using specified values for  $T_h$  and  $T_c$  because the FE models do not impose fixed temperature differences or constant heat flux at the TEG's boundary. Therefore, Equation 3.6 is employed solely for quantitative analysis to aid the interpretation of the observed phenomena in the simulation results.

### 3.3 Simulation Results

Recognizing the pivotal role of the thermal environment of the human body in the design and analysis of ITEGs, this study will initially discuss the simulation results of the human body. Subsequently, the examination of ITEGs is organized into three distinct steps:

1. The housing design and implantation environment were kept constant to investigate the influence of the TEG's parameters on its performance. Optimization efforts were directed towards achieving maximum power output and ensuring sufficient voltage.
2. Various parameter values of the package were systematically varied to observe their effect on power output. TEG parameters were optimized respectively during each parameter study, to observe the vary effect of the package parameters.
3. Based on the insights gained from the first two steps, TEGs were designed with optimized parameters. These designs were embedded into different locations in the human body for performance prediction.

#### 3.3.1 Human Body Thermal Profile

Figure 34 depicts the temperature profile of the human torso and human head models without as implanted TEG. It can be seen that the skin surface temperatures are lower in distal parts and areas with high-fat content (e.g., nose, abdomen, and upper chest). In both models, the observed peak temperature demonstrates remarkable consistency. This can be attributed to the active role of blood perfusion, diligently working to establish and maintain internal thermal homeostasis. In the absence of blood perfusion, the calculated core temperature would significantly surpass practical levels. Additionally, the area around the heart and intestine is at a significantly high temperature due to high metabolic heat production. The temperature of the main organs and tissues in the models is listed in Table 3.

The validation of the human thermal model faces challenges due to the difficulty in measuring temperature gradients inside the body, particularly in inaccessible regions, with noninvasive approaches. Additionally, inaccuracies can be introduced by variations in anatomical geometry and physiological functions among subjects compared to the model.

Table 3: Temperature of organs and tissues in the human body models.

| <b>Regions</b>       | $T_{mean}$ (°C) | $T_{max}$ (°C) | $T_{min}$ (°C) |
|----------------------|-----------------|----------------|----------------|
| skin surface (torso) | 33.3            | 35.8           | 31.8           |
| fat (torso)          | 35.0            | 37.4           | 32.0           |
| muscle (torso)       | 37.2            | 37.8           | 33.7           |
| skin surface (head)  | 33.0            | 34.3           | 30.1           |
| fat (head)           | 33.7            | 35.9           | 30.6           |
| muscle (head)        | 36.6            | 37.3           | 32.2           |
| brain                | 37.3            | 37.5           | 36.4           |
| heart                | 37.6            | 37.8           | 37.4           |
| intestine            | 37.3            | 37.5           | 37.1           |
| stomach              | 37.3            | 37.5           | 37.2           |
| liver                | 37.2            | 37.3           | 37.1           |

In this study, the validation of the human torso and human head models was pursued by separately comparing skin surface and body core temperatures. Experimental insights from [127] revealed that, with an average ambient temperature of 21 °C during the day, the average skin temperature varied between 31–35 °C across different body parts. Our human torso model recorded skin temperature in the range of 31.9–35.7 °C, with an average of 33.3 °C. For the human head model, the skin temperature ranged from 30.1–34.3 °C, with an average of 33.0 °C.

Body core temperature, which is broadly defined as the operational temperature of internal organs, is commonly measured using the rectal temperature or radio pill methods [128]. In an experiment described in [129], where all of the rooms were maintained at a controlled temperature of  $20 \pm 2$  °C, the temperature of the intestine was measured using an ingestible radiotelemetric capsule, which showed a range between 36.6 and 37.1 °C. In our human torso model, the average temperature of the intestine and main organs was recorded as 37.3 °C.

Another experiment from [130] aimed to measure brain temperature in the frontal cortex, utilizing proton magnetic resonance spectroscopy. Under ambient conditions of 23 °C and a relative humidity of 45%, the reported brain temperature was 37.2 °C. In our human head model, the brain (comprising the cerebellum, grey matter, and white matter) maintains an average temperature of 37.3 °C.

The observed skin surface temperature and body core temperature in both the human torso and head models closely align with data reported in the available literature. Additionally, the exhibited temperature profiles within the models are considered reasonable. Consequently, these models are deemed reliable and suitable for the initial investigation of ITEGs.

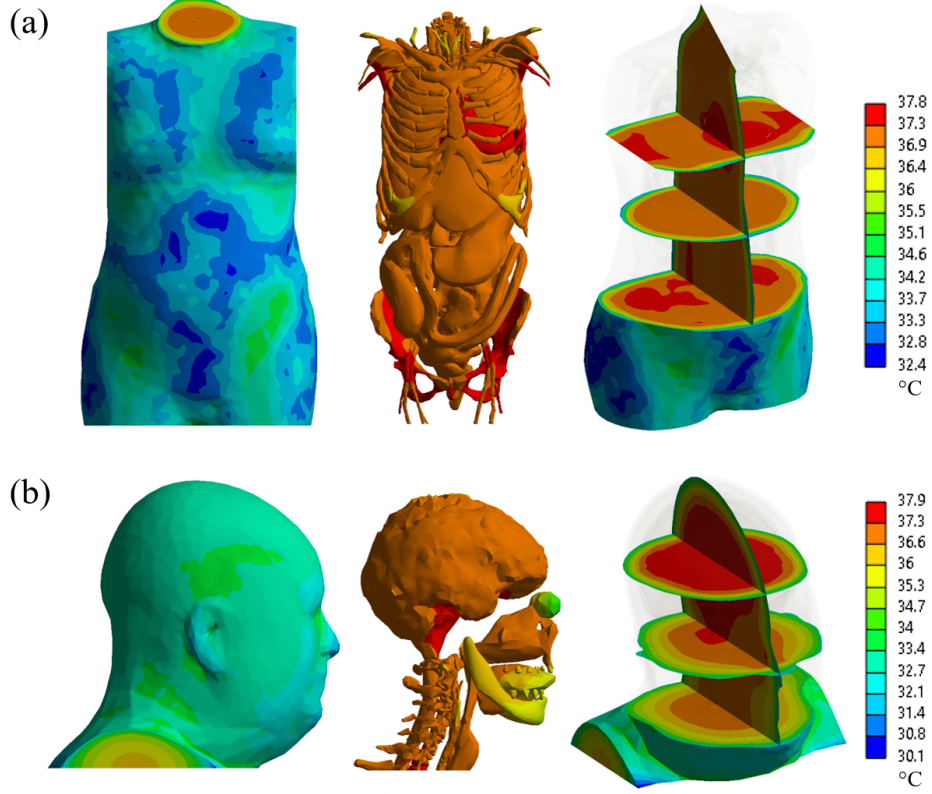


Figure 34: Temperature distribution of human body models without TEG implantation: (a) Torso; (b) Head. Reprinted from [124].

### 3.3.2 Design Optimization of the TEG

This section investigates the impact of TEG performance based on parameters such as the fill factor  $\eta$ , load resistance  $R_L$ , height of thermolegs  $h$ , and the number of thermolegs  $n$ . As described in Step 1, a packaged TEG with parameters according to Table 2 was positioned at location 1 within the human torso model (see Figure 32 for reference). While varying  $h$ , the height of heat conductors was adapted to maintain a constant overall height of the implant device. This approach ensured that changes in geometry were avoided for both the packaging and the surrounding tissue.

The power output of the TEG as a function of the fill factor is presented in Figure 35(a) when a resistive load with optimum resistance is connected to the TEG. The average heat flow through the RT ( $Q_{rt}$ ) increases as the fill factor rises, attributed to the lower  $R_{th}$  and consequently lower parasitic heat flow through  $R_p$ . Simultaneously, the reduction in  $R_{th}$  decreases  $\Delta T$ . As demonstrated in Equation 3.6, both  $\Delta T$  and  $Q_c$  contribute to  $P_{out}$ . Therefore, an optimum fill factor of 7% exists to balance  $\Delta T$  and  $Q_{rt}$ , resulting in a maximum  $P_{out}$  of 31  $\mu$ W. This thermal optimization is referred to as thermal matching between the TEG and its implantation environment. The Joule heat of the RT is identical to  $P_{out}$  when  $R_L$  equals  $R_{in}$ , which is not comparable to  $Q_{rt}$  (79 mW). This is because  $\Delta T$  is very low compared to the absolute temperature within the human body, resulting

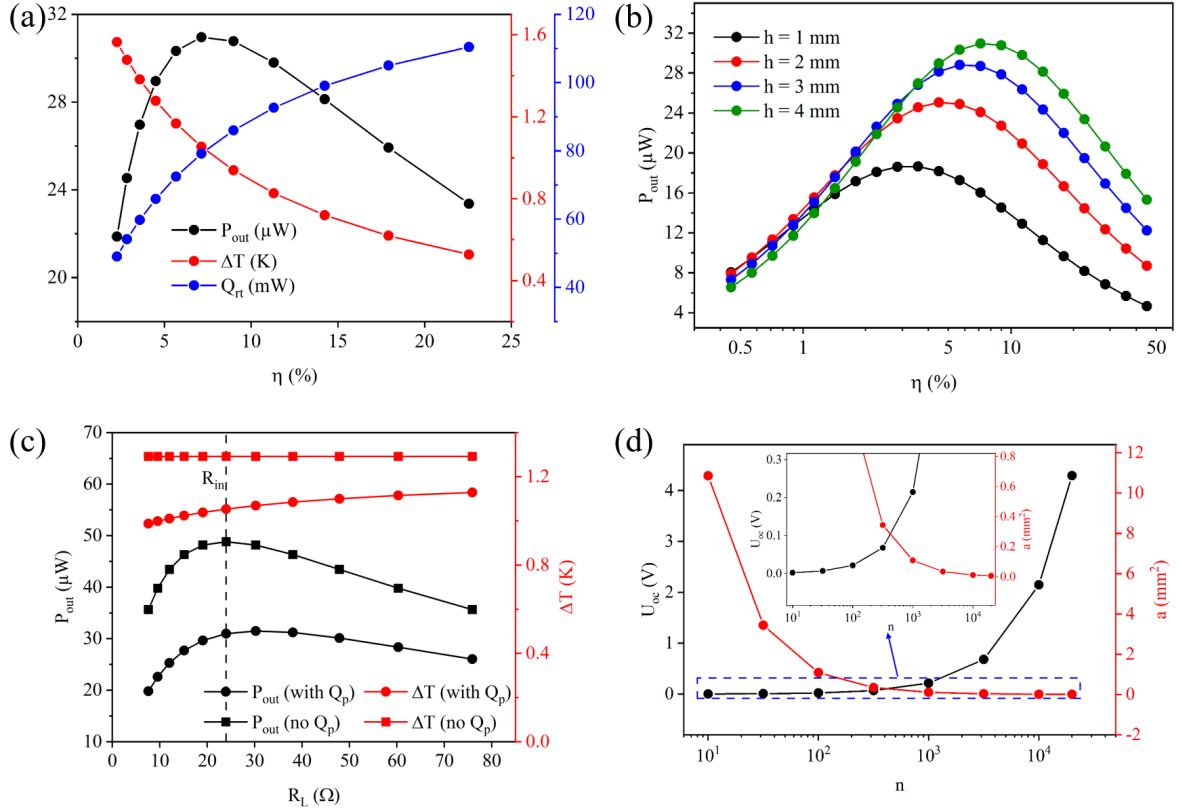


Figure 35: Effects of TEG parameters on power and voltage output: (a) Fill factor, (b) Thermoelectric height, (c) Load resistance, (d) Number of thermoelements. Reprinted from [124].

in minimal impact of the Joule heat on the thermal profile within the body.

The maximum  $P_{out}$  increases from 19 to 31  $\mu$ W as the height of the RT rises from 1 to 4 mm, as illustrated in Figure 35(b). Since  $R_a$  and  $R_b$  remain unchanged during the sweep of  $\eta$ , the increased  $P_{out}$  can be attributed to the elevated values of  $R_p$ , consequently raising  $\Delta T$ . Simultaneously, the optimum  $\eta$  increases from 3.6 to 7% for thermal matching with unchanged  $R_a$  and  $R_b$ .

Subsequently, with  $\eta$  fixed at the optimum value, electrical load matching was conducted, as illustrated in Figure 35(c). Unlike Joule heat, the Peltier heat ( $Q_p$ ) has a significant effect on TEG performance. As the inverse of the Seebeck effect, the Peltier effect works towards reducing the existing  $\Delta T$ . This explains the drop of almost 35% in  $P_{out}$  from 49 to 31  $\mu$ W considering the Peltier heat. When  $Q_p$  is not considered, the TEG can be perceived as a constant voltage source and the optimum  $R_L$  is equal to  $R_{in}$ . However, when  $Q_p$  is taken into account, an increased  $R_L$  results in a lower current, and consequently less  $Q_p$  according to Equations 2.18 and 2.19. This, in turn, leads to an increase in  $\Delta T$  and higher  $U_{oc}$ . This explains why here the optimum  $R_L$  is 30  $\Omega$ , which is larger than  $R_{in}$  (24  $\Omega$ ).

Through Equation 3.7, rewritten from Equation 3.6, it has been observed that for a



given  $\eta$ , the number of thermolegs  $n$  has no impact on  $P_{out}$ .

$$P_{out} = \frac{\eta A(\alpha_p - \alpha_n)^2 \Delta T^2}{8h(\rho_p + \rho_n)} \quad (3.7)$$

According to Equation 2.6,  $U_{oc}$  is proportional to  $n$ , as illustrated in Figure 35(d). Since  $U_{oc}$  is constrained by low  $\Delta T$  within the human body, increasing  $n$ , while maintaining optimum  $\eta$ , becomes a promising approach to elevate  $U_{oc}$  without compromising  $P_{out}$ . With the optimum  $\eta$ ,  $U_{oc}$  is increased from 2 mV to 4.3 V when the number of thermolegs increases from 10 to 20,000. However, as  $n$  increases, the cross-sectional area ( $a$ ) of each leg must decrease, which poses challenges for manufacturing of the legs and TEG assembly. To achieve a  $U_{oc}$  of 2 V, a typical value required by IMDs, at least 10,000 legs with a cross-sectional area of 0.01 mm<sup>2</sup> are necessary.

### 3.3.3 Parametric Analysis on Implant

As outlined in Step 2, this section will investigate the impact of TEG packaging on  $P_{out}$ , taking into account the following four parameters: the diameter of the device ( $d$ ), the depth of the device ( $D$ ) to the skin surface, the thermal conductivity ( $\kappa_{pack}$ ) of the package, and the thermal conductivity ( $\kappa_{fill}$ ) of the filler material (see Figure 36(a)). Based on the simulation results above, the optimum  $\eta$  and  $R_L$  were identified for each parameter change to maintain thermal and electrical matching. Other TEG parameters were kept constant at their initial values. A discussion of the trade-offs between power, miniaturization, hermeticity, and manufacturability of the implant follows. In Figure 36(b),  $P_{out}$  decreases with increasing implantation depth. The maximum  $P_{out}$  of 31  $\mu$ W is achieved when the housing is positioned near the skin layer ( $D = 5$  mm). When  $D$  changes from 5 to 10 mm, the maximum  $P_{out}$  already drops to 12  $\mu$ W. Additionally, the optimum  $\eta$  slightly increases from 7 to 9%. With even deeper implantation depth  $D$  ranging between 15 to 25 mm, where the TEG is fully embedded within the muscle section, the maximum  $P_{out}$  significantly drops to less than 1  $\mu$ W with an unchanged optimum  $\eta$  of 11%. The decrease in  $P_{out}$  can be attributed to the presence of a high temperature gradient in the fat layer compared to lower gradients within the muscle layer, a phenomenon further demonstrated and discussed in Figure 38. The  $\Delta T$  of the TEG is influenced by surrounding tissue due to lateral heat conduction, a factor not considered in Figure 33. Thus, when the TEG is implanted deeper into the muscle section, despite negligible changes in  $R_a$  and  $R_b$ ,  $P_{out}$  still decreases due to lower  $\Delta T$  in surrounding tissue.

Figure 36(c) shows that a larger device dimension leads to higher  $P_{out}$ , stemming from a higher heat flow through the RT. The maximum  $P_{out}$  increases from 4.6 to 47.6  $\mu$ W when  $d$  increases from 20 to 60 mm. Since the thermal resistance of the RT,  $R_a$  and  $R_b$ , scale with their cross-section,  $\Delta T$  should remain nearly constant, as indicated by the red line in Figure 35(a). However, the elevated heat flow implies a rise in the blue line in

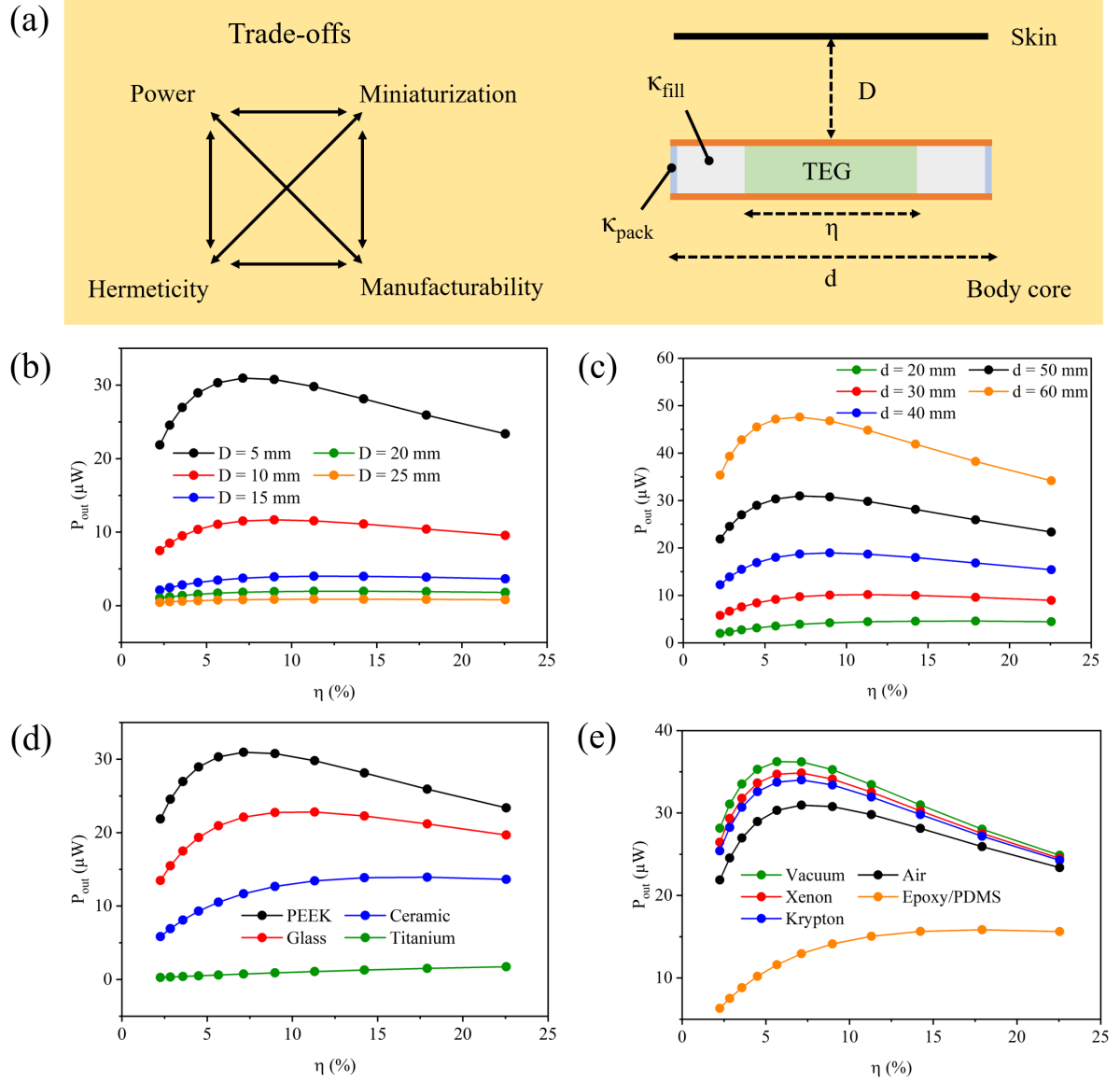


Figure 36: (a) Overview of package parameters and trade-offs between design factors. Effects of package parameters on power output: (b) Implantation depth; (c) Device diameter; (d) Thermal conductivity of package material; (e) Thermal conductivity of filler material. Reprinted from [124].

Figure 35(a), decreasing optimum  $\eta$  from 18 to 7%.

The thermal conductivity of the packaging and filler material should be as small as possible to reduce the impact of parasitic heat flow, as these domains are both thermally connected in parallel to the thermopile. Figure 36(d) presents the results for different bio-compatible housing materials. The PEEK package delivers the highest  $P_{out}$  due to its excellent thermal isolation property. When  $ZrO_2$  ceramic is used as a packaging material, the maximum  $P_{out}$  decreases from 31 to 14  $\mu W$ . Furthermore, the increased thermal shunting through  $R_p$  results in a higher optimum  $\eta$ , increasing from 7 to 18%. This is an effort to mitigate thermal shunting by reducing  $R_{th}$ .  $P_{out}$  delivered from the TEG with a titanium package is very limited, only 1.7  $\mu W$  with an optimum  $\eta$  of 23%.

Like the packaging material, a filler material with a lower thermal conductivity leads to higher  $P_{out}$ , as shown in Figure 36(e). As expected, the TEG without any filler material (vacuum) shows the best performance. A maximum  $P_{out}$  of 36  $\mu W$  is obtained with an optimum  $\eta$  of 5.6%. When filling the package with inert gases such as xenon and krypton, the TEG shows similar performance to a vacuum environment. The maximum  $P_{out}$  drops nearly 15% when filled with air instead of vacuum. Solid filler materials such as epoxy resin and polydimethylsiloxane (PDMS) lead to even lower  $P_{out}$ , which is 16  $\mu W$  with an optimum  $\eta$  of 18%.

In summary, the design of ITEGs should encompass considerations beyond power generation. Firstly, TEGs, akin to other energy harvesters, must strike a balance between power output and size. The size of the implant is limited by the feasibility of surgical procedures, as well as considerations for the comfort and safety of the patient.

Furthermore, the hermeticity of the implant is highly dependent on the material of the package. Despite PEEK having excellent thermal isolation properties among bio-compatible materials, it is not fully hermetic like metals and ceramics. Over prolonged implantation periods, there is a risk of moisture ingress into the package through the PEEK material and along the contact region between PEEK and metal heat conductors [118, 131]. To address this issue, studies have been conducted on the PEEK-based packages. In the realm of connection joints, a study in [132] demonstrated promising improvements in the adhesion between PEEK and aluminum alloy by employing laser surface texturing on the aluminum alloy before friction joining. Additionally, atomic layer deposition (ALD) coating technology has proven effective in preventing moisture ingress through PEEK [131], although achieving an implant with excellent hermeticity remains a challenge. Further details regarding the hermeticity of ITEG will be discussed in Chapter 6.

Glass emerges as an alternative housing material to balance power generation and hermeticity needs. Glass possesses lower thermal conductivity than titanium and ceramic, resulting in an acceptable power drop compared to a PEEK housing. Sacrificing a portion of power can significantly enhance the hermeticity of the implant device, as glass is impermeable to gases and liquids. Ultrafast laser welding enabled a strong and reliable

connection between glass and metal in recent years [133–135].

Last but not least, the manufacturability of the TEG should not be overlooked in practical applications. As explained in Section 3.3.2, achieving both sufficient  $U_{oc}$  and maximum  $P_{out}$  requires a large number of high aspect ratio thermocouples. However, the fabrication of high aspect ratio legs remains challenging due to the brittleness of  $\text{Bi}_2\text{Te}_3$ . Introducing a boost converter to enhance  $U_{oc}$  can reduce the requirement on the aspect ratio of legs but comes at the cost of decreased  $P_{out}$  due to power loss during voltage conversion (see Chapter 5). Additionally, higher  $P_{out}$  is obtained with several gas filler materials, necessitating free-standing assembly of legs onto substrates. Using solid filler materials such as epoxy resin and PDMS may simplify the fabrication process and aid in leg fixation but causes additional thermal shunting and thus an inevitable drop in  $P_{out}$ .

Power delivery, miniaturization, hermeticity, and manufacturability involve trade-off relationships. Holding two factors constant arbitrarily results in a contradiction between the other two. ITEGs should be specifically designed according to different application scenarios, carefully considering the balance between these factors.

#### 3.3.4 Case study

The influence of TEG design and packaging parameters on the system's power delivery has been individually clarified in previous sections. In this section, TEG performance at different implantation locations is investigated in the human torso and head models.

##### **Human Torso Model with Embedded TEG**

As demonstrated in the simulation results above,  $P_{out}$  is enhanced by a low implantation depth, large dimension, and high height of thermocouple legs. To assess the potential of ITEGs in the human torso model, TEGs were strategically positioned within the fat layers at various locations around the chest, abdomen, and lumbar regions. Six implant locations are illustrated in Figure 32. To balance the dimension and performance of the implant device, the diameter of the implant device was chosen to be 50 mm, similar to conventional pacemakers dimensions. To maximize  $P_{out}$ , the implants occupied the entire height of the fat section. The surfaces of two heat conductors were adapted to the skin and muscle sections. The height of RT was selected as high as possible based on the fat thickness and the topography of the skin and muscle. The angle of the horizontal plane of RT was adjusted to balance the angles of the skin and muscle surfaces. Other parameters were kept at their initial values. The optimum  $\eta$  and  $R_L$  values were determined for each implant location.

The TEG implantation at location 1 is illustrated in Figure 37. The temperature distribution of the section view indicates that the tissue below the implant exhibits a lower temperature compared to the surrounding tissue. A comparison of temperature and heat flux with and without TEG implantation along a path perpendicular to the

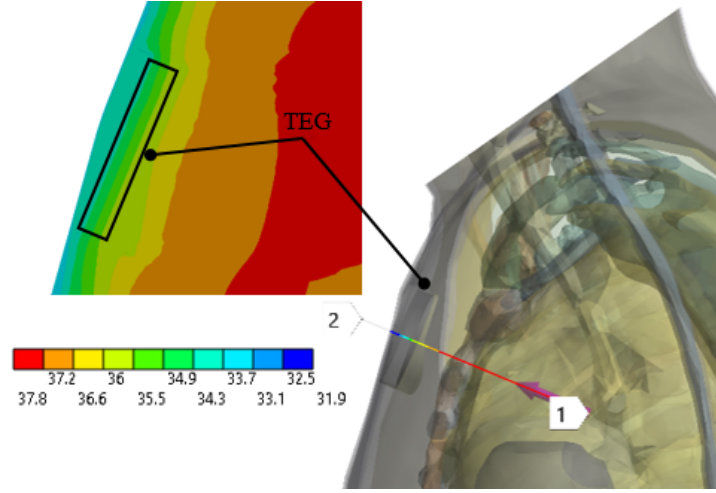


Figure 37: Temperature distribution in a human torso cross-section. The packaged TEG (with optimum fill factor and load resistance) is embedded at location 1. Reprinted from [124].

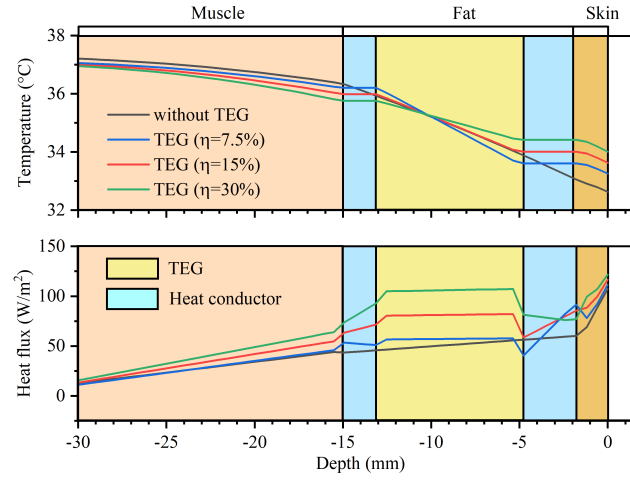


Figure 38: Temperature (above) and heat flux (below) in the human torso model before and after implantation of the optimized TEG. Reprinted from [124].

center point of the RT surface is presented in Figure 38.

The temperature gradient within the body has a pivotal role in power generation. Muscles generally have a higher thermal conductivity than fat. This translates to a steeper temperature gradient within fat tissue and a more favorable environment for TEG operation. Additionally, the temperature gradient decreases towards deeper muscle regions, as organs closer to the body's core experience stronger blood perfusion, maintaining a near-constant temperature of around 37 °C. Consequently, the body's core temperature itself remains largely unaffected by TEG implantation.

Implanting a thermally optimized TEG minimally affects the temperature difference across the existing fat layer. This indicates that the optimized TEG operates at a temperature closely approximating that of the native tissue it replaced. However, skin surface temperature decreases slightly, from 33.6 to 32.6 °C, likely due to the heat conductor's

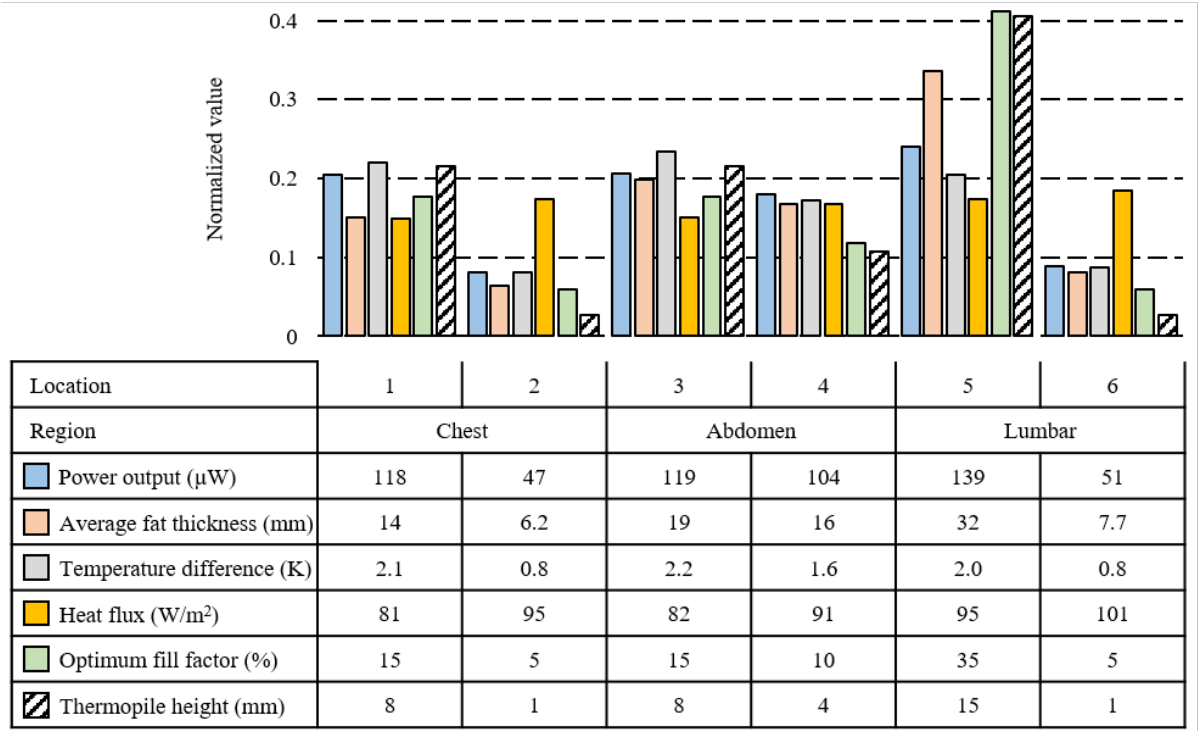


Figure 39: Comparison of TEG performance in different human torso implant locations. Reprinted from [124].

influence. In contrast, heat flux through the implant increases after implantation, rising from 50 to  $81 \text{ W}/\text{m}^2$ .

The case study on different implant locations in the human torso model is depicted in Figure 39. Locations 1 to 6, as shown in Figure 32, were examined. The results indicate that  $P_{out}$  is primarily dependent on the fat thickness at the implant location. Greater fat thickness results in a higher  $\Delta T$  across the optimized TEG, leading to increased  $P_{out}$ . In location 2, situated around the chest region with a fat thickness of 6.2 mm, 47  $\mu\text{W}$  of power can be obtained through a  $\Delta T$  of 0.8 K. The highest  $P_{out}$  of 139  $\mu\text{W}$  is observed in location 5, around the lumbar region with a fat thickness of 32 mm a  $\Delta T$  of 2 K. However, as the fat thickness increases to a certain level, the improvement in  $P_{out}$  diminishes, as evidenced by the comparisons between locations 1 and 2, and locations 5 and 6. In the chest region,  $P_{out}$  increases by 2.5 times when the average fat thickness is increased by a factor of 2.3. In contrast, in the lumbar region,  $P_{out}$  increases by only 2.7 times when the average fat thickness is increased by a factor of 4.2. Compared to  $\Delta T$ , the heat flux through the RT is less sensitive to fat thickness. The optimum  $\eta$  increases with the RT height, as the optimal thermal resistance of the RT, which roughly equals the combined thermal resistance of muscle tissue (from core to fat) and skin ( $R_a$  and  $R_b$  in Figure 33), is consistent across locations. For effective thermal matching, the optimum  $\eta$  should therefore be positively correlated with RT height.

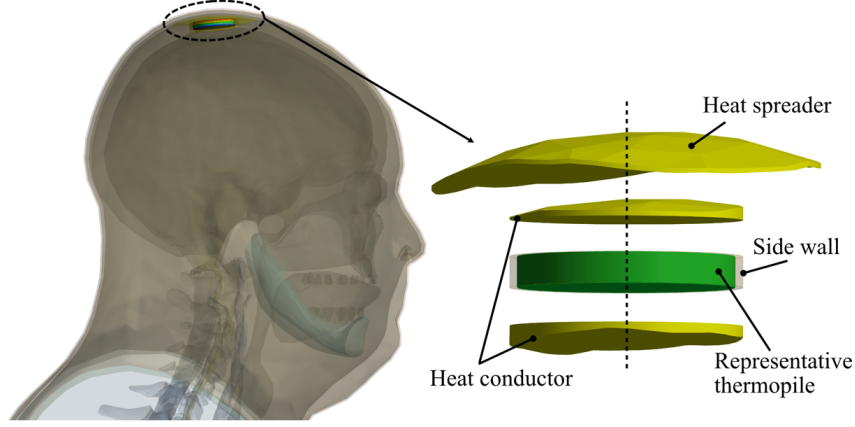


Figure 40: Packaged TEG embedded in the cranial bone of the human head model. Reprinted from [124].

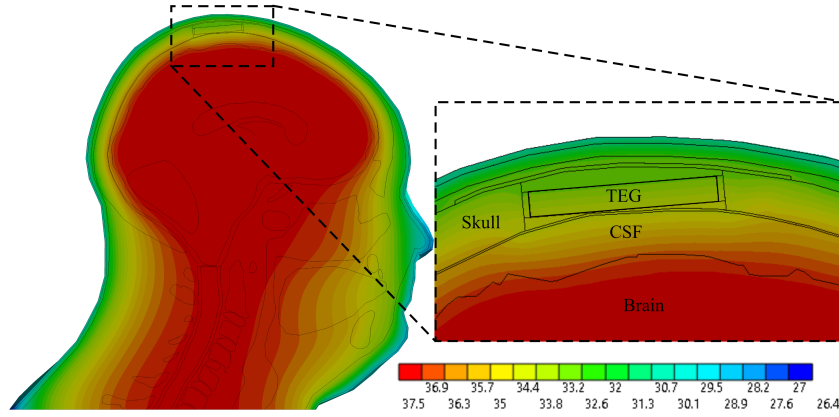


Figure 41: Temperature distribution in the human head model with an embedded optimized TEG. Reprinted from [124].

### Human Head Model with Embedded TEG

In the human head model, the TEG was embedded within the cranial bone, as illustrated in Figure 40. The implant had a diameter of 30 mm. Two heat conductors are designed to conformingly follow the skull's inner and outer surfaces. To enhance the TEG's performance, a 50 mm diameter, 0.6 mm additional thick heat spreaders were introduced, covering the skull surface. The heat spreader contributes to efficient heat distribution and dissipation, thereby enhancing the TEG's energy generation. Other parameters were maintained at their initial values.

The TEG exhibits an optimum  $\eta$  of 30%, resulting in a  $P_{out}$  of 91  $\mu\text{W}$ . Figure 41 depicts the temperature distribution within the human head in the sagittal plane. A temperature difference of 1.5 K exists across the TEG. Notably, the cerebrospinal fluid (CSF) between the TEG and brain demonstrates a lower temperature compared to surrounding CSF, attributable to the heat dissipation from the implant and heat spreader.

The implementation of a heat spreader ( $s = 40$  mm) results in a notable enhancement in  $P_{out}$  for the TEG, with a significant increase from 66 to 84  $\mu\text{W}$ , as illustrated in Figure 42. This improvement can be attributed to the increased heat flow through the TEG.

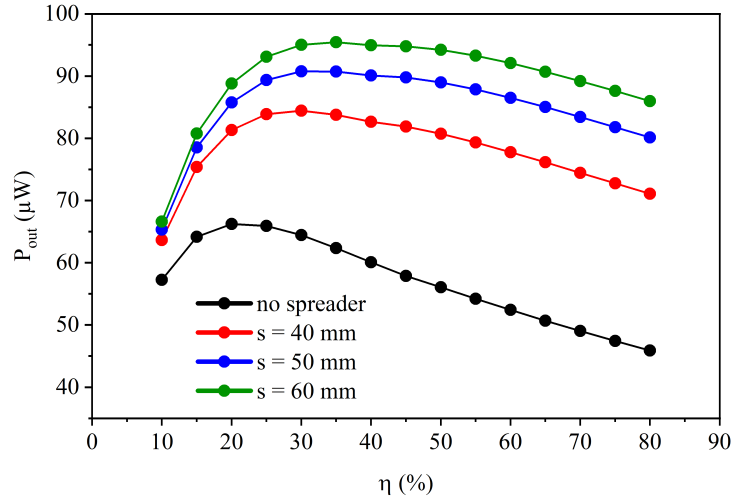


Figure 42: Effect of heat spreader diameter and fill factor on TEG power output. Reprinted from [124].

Concurrently, the optimum  $\eta$  increases from 20 to 30% to match the decreased  $R_a$  due to a higher cross-sectional area. However, the enhancement of a larger heat spreader on  $P_{out}$  is not as significant. When  $s$  is increased from 40 to 60 mm,  $P_{out}$  rises from 84 to 95  $\mu\text{W}$ , with an increased optimum  $\eta$  from 30 to 35%.

### 3.4 Summary

Chapter 2 laid the foundation with a simplified tissue model designed for efficient simulation of ITEGs. Building upon this work, the current chapter elevated the accuracy of performance predictions by employing high-resolution anatomical human torso and head models. These models integrated bio-heat effects within tissues and organs, providing a more realistic representation of the implantation environment. Since numerous parameters influence TEG performance, the study was designed in such a way as to independently investigate TEG design, housing, and implantation location.

Two significant parameters were identified for optimizing the power output ( $P_{out}$ ) of TEGs. In the thermal domain, the fill factor ( $\eta$ ) of thermolegs within the implant device is crucial for achieving thermal matching between the TEG and its environment. In the electrical domain, consideration of Peltier heat and matching between the load resistance and the inner resistance of the TEG are essential for performance optimization.

Peltier heat generated by the TEG significantly impacts its performance by affecting the thermal profile. While the number ( $n$ ) of thermolegs does not influence  $P_{out}$  when maintaining  $\eta$ , a higher  $n$  results in a higher voltage output. Additionally, taller thermolegs benefit both power and voltage output. Thus, TEG design should prioritize numerous high aspect ratio thermolegs to maximize power while ensuring sufficient voltage output.

Larger device dimensions and lower thermal conductivity of packaging and filler ma-



terials positively affect  $P_{out}$  but pose challenges related to miniaturization, hermeticity, and manufacturability. Balancing these factors is crucial in the design and fabrication of ITEGs.

Implant locations near the skin surface are preferred for enhanced TEG performance. In the human torso, implanting TEGs in fat sections yields higher  $P_{out}$ . Locations with higher fat thickness can accommodate devices with increased height, resulting in higher  $P_{out}$ . The temperature difference across an optimized TEG is similar to that of its surrounding tissue. Optimized TEGs, packaged in a PEEK housing with a 50 mm diameter, can generate 47  $\mu\text{W}$  of power from a 0.8 K temperature difference in 6.2 mm thick fat and 118  $\mu\text{W}$  from a 2.1 K temperature difference in 14 mm thick fat.

In the human head, where the fat layer is thinner than in the torso, the cranial bone emerges as a potential implant location for thermoelectric power generation. Proximity to both the skin surface and brain (with a temperature similar to the body core) allows the implanted TEG to collect more heat flow, making it a promising location. An optimized TEG with a 30 mm diameter yields a  $P_{out}$  of 66  $\mu\text{W}$ . Further enhancement is achieved by employing a 50 mm diameter heat spreader covering the upper skull surface, boosting  $P_{out}$  to 91  $\mu\text{W}$ .

The comprehensive analysis of the simulation results revealed the relationships between power output, optimal fill factor, and various design parameters. This knowledge improves understanding of TEG behavior not only for implantable applications but also for TEGs embedded in diverse environments.

## 4 ITEG Fabrication

The previous simulation studies (Chapters 2 and 3) have highlighted the importance of an optimal fill factor for obtaining thermal matching, maximizing power output, and assembling a sufficiently high number of thermolegs for meeting IMD voltage requirements. Consequently, the ITEG design should prioritize numerous high-aspect-ratio thermolegs. However, this raises fabrication challenges. This chapter proposes a fabrication strategy to realize the optimized TEG described in Section 2.4.3. A novel removable assembly rig was designed to overcome the challenges of thermoleg soldering, which ensures precision and repeatability. The following approaches are employed to further enhance voltage output while maintaining the optimal fill factor:

- The number and the aspect ratio of the thermolegs are further increased for higher voltage output.
- A boost converter is integrated within the TEG for efficient up-conversion to usable voltage levels.

### 4.1 Methods Review

TEGs with high-aspect-ratio thermolegs have previously been developed using several approaches, each presenting unique advantages and challenges. Introducing filler material between the thermolegs can facilitate the assembly of such TEGs. One method involved printing thin-film thermolegs ( $0.64 \times 0.09 \times 5 \text{ mm}^3$ ) on a flexible polyimide substrate [136]. Another approach has utilized self-bonding Kapton tape to assemble thermolegs with an extremely high aspect ratio ( $0.38 \times 0.38 \times 22.86 \text{ mm}^3$ ) into a matrix [137]. However, these methods are not suitable for implantation because the filler material between the thermolegs increases thermal shunting, which significantly decreases the power output in these applications, as illustrated in Figure 36(e).

Alternative approaches obtain free-standing high-aspect-ratio thermolegs from bulk material, which is more feasible and desirable for implantation. Vasilevskiy et al. [138] produced micro-modules with thermolegs measuring  $0.21 \times 0.21 \times 1.27 \text{ mm}^3$  using assembly lines. Meanwhile, Leonov et al. [139] developed a four-stage TEG for wearable devices, with each stage featuring thermolegs measuring  $0.14 \times 0.14 \times 1.25 \text{ mm}^3$ . However, these studies did not provide a detailed description of the fabrication processes. Toan et al. [140] designed a fabrication setup to assemble TEGs for wearable applications, as shown in Figure 43(c). Thermolegs ( $0.4 \times 0.4 \times 2 \text{ mm}^3$ ) were soldered onto the bottom substrate using a stencil. However, the position of the soldered thermolegs may shift upon stencil removal or during the second soldering to the top substrate. In particular, misalignment of several thermolegs was observed after the first soldering (see Figure 43(d)). Wesolowski et al. [141] developed a novel assembly method for these devices. They used sinker electro-discharge

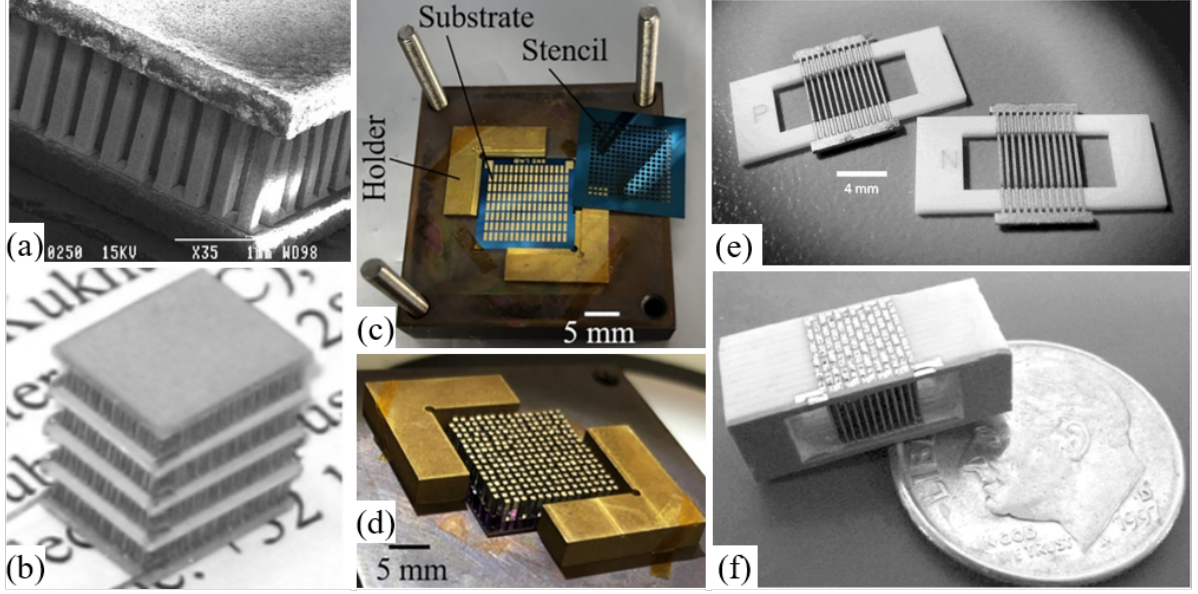


Figure 43: TEGs with high-aspect-ratio thermolegs. (a) A TE module with thermolegs measuring  $0.21 \times 0.21 \times 1.27 \text{ mm}^3$ . Image adapted from [138]. (b) A four-stage TEG with thermolegs measuring  $0.14 \times 0.14 \times 1.25 \text{ mm}^3$ . Image adapted from [139]. (c) An example of the fabrication setup of portable and wearable TEGs. Image adapted from [140]. (d) The thermoleg ( $0.15 \times 0.3 \times 4 \text{ mm}^3$ ) array after soldering using a stencil for the alignment. Image adapted from [140]. (e) Two laser-cut  $\text{Bi}_2\text{Te}_3$  wafers mounted in the slots of alumina carrier plates. Image adapted from [141]. (f) A TEG with thermolegs measuring  $0.15 \times 0.3 \times 4 \text{ mm}^3$ . Image adapted from [141].

machining (EDM) to cut the semiconductor wafers, which were then held by multiple alumina carrier plates (see Figure 43(e)). The TEG with high-aspect-ratio thermolegs ( $0.15 \times 0.3 \times 4 \text{ mm}^3$ ) was then fabricated by stacking and cementing the alumina carrier plates (see Figure 43(f)). This approach eliminates the need for a pick-and-place process for the thermolegs but requires multiple operations (e.g., cuts and bonding) on the alumina frames. It is important to note that the TEGs developed using these methods were not specifically optimized for implantation scenarios. Further investigation is needed to assess the performance of TEGs with high aspect ratio thermolegs for implantation use.

## 4.2 Fabrication Strategy for the ITEGs

Based on the optimized TEG geometry proposed in Section 2.4.3, a TEG (referred to as TEG-1) was fabricated. The overall fabrication process is illustrated in Figure 44.

Solder pasting was conducted by screen printing using a squeegee blade to scrape the solder paste onto the interconnects of the substrates, as shown in Figure 44(a). An assembly rig was developed, consisting of a holder, two spacers, and four alignment fixtures (see Figure 44(b)). This rig was designed to securely hold the thermolegs in place during the soldering process, which prevented any unintended movements or misalignment that could compromise the quality of the solder joints. The thermolegs were placed into the assembly rig using a nozzle that was activated by a vacuum pump (see Figure 44(c)).

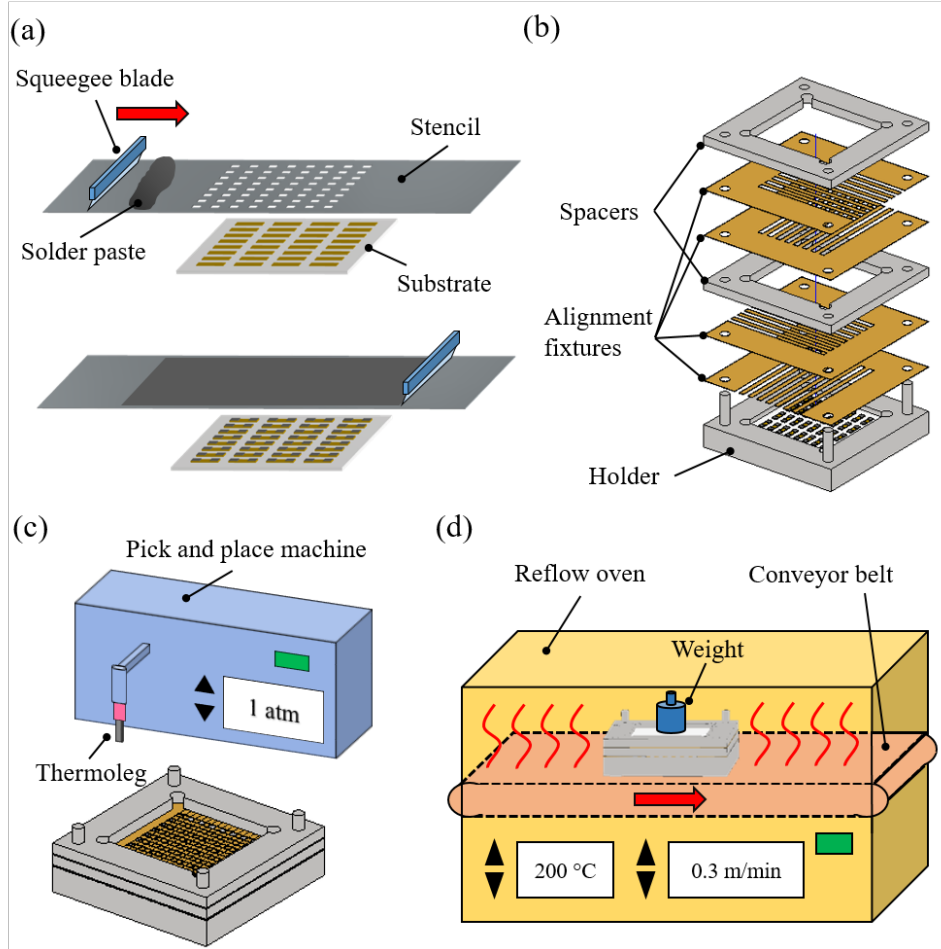


Figure 44: Fabrication process of TEG-1. (a) Solder pasting on the interconnects of the substrate by screen printing. (b) Assembly of a removable assembly rig. (c) Pick and place of thermolegs using a manipulator. (d) Soldering using a reflow oven. Figure (a) and (b) are reprinted from [118].

Subsequently, the top substrate with pasted solder was positioned onto the thermolegs, and the assembly rig was subjected to reflow soldering in a reflow oven. The alignment fixtures were removed from the TEG after soldering (see Figure 44(d)). The fabrication strategy and process are detailed in the following subsections.

### 4.2.1 Solder Pasting

The p-type BiSbTe and n-type BiTeSeS thermolegs, coated with nickel and gold on their top and bottom surfaces, were wire-cut from ingots to dimensions of  $1 \times 1 \times 4 \text{ mm}^3$  (Powercool Technology, Shenzhen, China). The coating on  $\text{Bi}_2\text{Te}_3$  prevents degradation [142] and interdiffusion [143], particularly in high-temperature applications. Degradation can potentially reduce bonding strength. Meanwhile, interdiffusion can lead to the formation of brittle intermetallic compounds [144–148] between  $\text{Bi}_2\text{Te}_3$  and the contacting material, which also affects bonding strength and may alter the electrical properties. Although degradation and interdiffusion are less concerning for ITEGs due to their low applica-

tion temperature, the coating remains essential. The nickel coating increases interface roughness, thereby enhancing bonding [149], while the addition of gold improves coating wettability, which further enhances bonding [150].

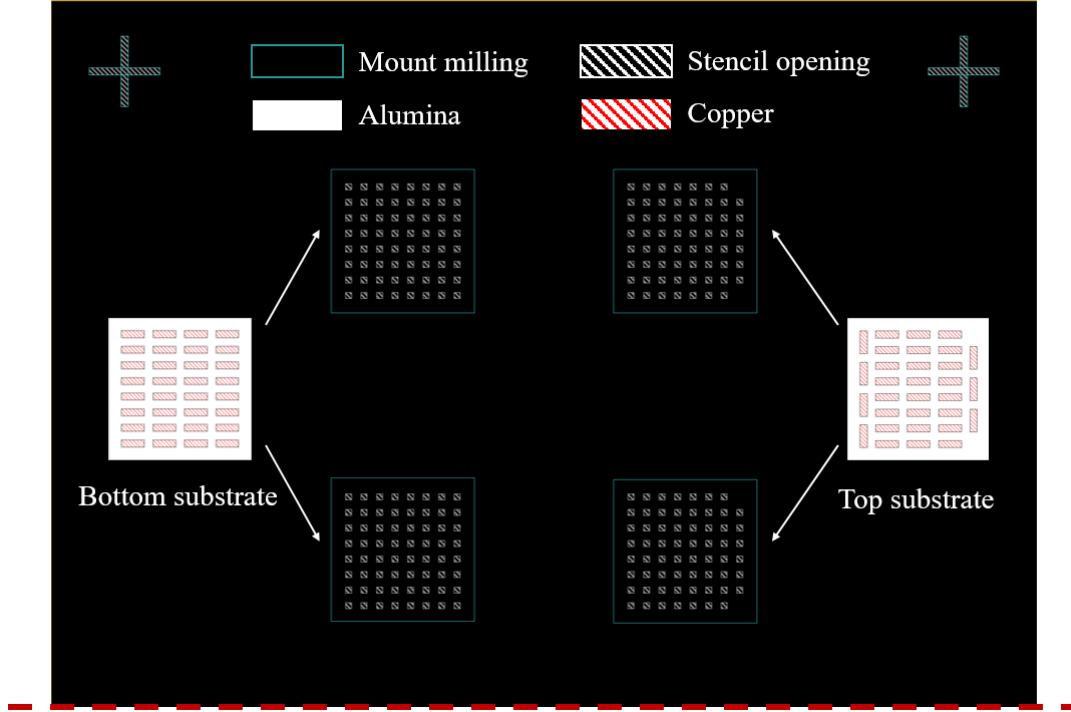


Figure 45: Design of a mount and a stencil for screen printing. The pockets milled on the mount are used to secure the TEG substrates. The openings that were laser-cut in the stencil are prepared for solder pasting. The alignment between the mount and the stencil is achieved by crosshair fiducials.

TEG-1 uses two alumina substrates with directly bonded copper (DBC) interconnects. The alumina substrates ( $20 \times 20 \times 0.4 \text{ mm}^3$ ) were supplied by Tianjin Century Electronics. The interconnecting conductor paths, measuring  $3.2 \times 1 \times 0.1 \text{ mm}^3$ , are designed to connect to the thermolegs, which are assembled in a quadratic array with a pitch of 2.2 mm.

To ensure proper solder application, the alumina substrates must be aligned with the stencil during soldering. This ensures solder paste is applied to both ends of the copper interconnects. A mount and stencil were designed using EAGLE 9.6.2 to facilitate alignment. Figure 45 shows half of the design, highlighting the mount and stencil pattern. The mount is made of a 1.5 mm thick FR4 board, with eight 0.4 mm deep pockets milled to secure the four bottom and four top substrates of TEG-1. These pockets have a 50  $\mu\text{m}$  tolerance for manufacturing variations, ensuring a secure fit. Four 0.4 mm deep crosshair recesses in the corners act as fiducials.

The 100  $\mu\text{m}$  thick stencil was sourced from Rostock Leiterplatten, with laser-cut openings to allow solder flow. These openings were made 0.2 mm narrower than the TE legs to prevent solder spread, which could cause thermal and electrical shunting, reducing TEG performance. A fiducial-shaped area was also laser-cut for alignment between the mount and stencil.

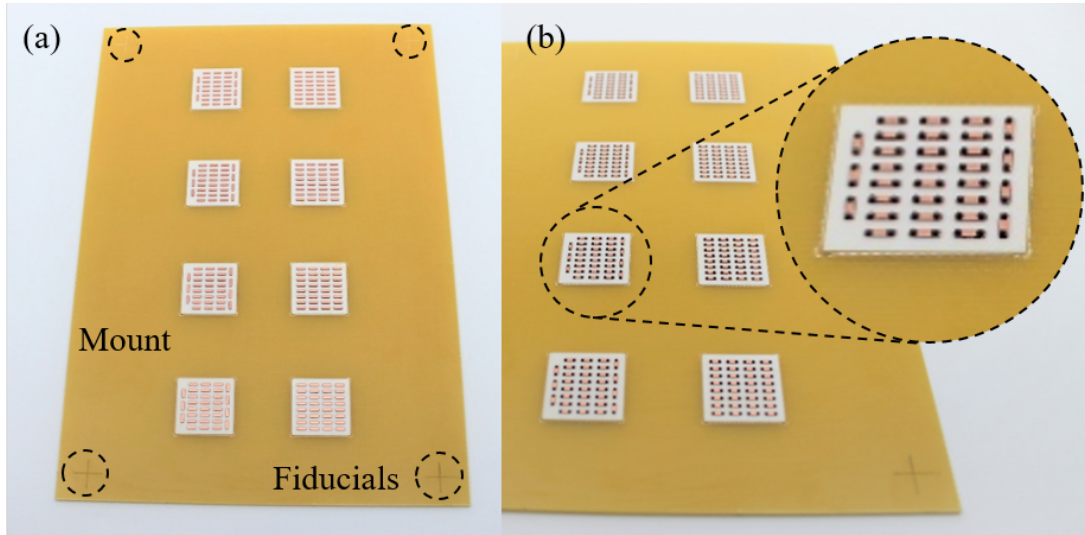


Figure 46: Bottom and top TEG substrates secured by the mount. (a) Before screen printing. (b) After screen printing.

Lead-free solders, typically tin-based alloys, are commonly used in medical implants due to the health risks posed by lead. Even with excellent hermeticity, lead-containing solders are avoided because of lead's toxicity, which presents long-term risks even at low concentrations. To avoid the formation of brittle intermetallic compounds at high temperatures,  $\text{Sn}_{42}\text{Bi}_{57}\text{Ag}_1$  (Heraeus Electronics) solder paste, with a low melting point of 138 to 142 °C, was used in the experiment. The solder paste is typically stored between 0 and 10 °C and was gradually warmed to room temperature before screen printing. This avoided extreme temperature differences between the solder and its contact and also prevented condensation on the solder paste surface, which might lead to thermal shock and poor joints. The solder pasting process using a screen printer (Mechatronic Engineering S20) is illustrated in Figure 44 (a). Figure 46 illustrates the substrates before and after the solder pasting.

### 4.2.2 TEG Assembly

The assembly rig is depicted in Figure 44 (b). The holder and spacers are made from austenitic stainless steel X8CrNiS18-9 and were manufactured using computer numerical control machining (provided by VETEC). Four pins are inserted into the holder using a press-fit connection to guide the spacers and alignment fixtures to their correct positions. The alignment fixtures, featuring open teeth, were laser-cut from 100  $\mu\text{m}$  thick copper sheets using LPKF Protolaser S4.

The assembly process is detailed in Figure 47. Initially, the bottom substrate was positioned and secured within the holder after solder pasting (see Figure 47(a)). Following this, the first two alignment fixtures were mounted on the pins (see Figure 47(b), (c)). The lower spacer was then mounted on the pins, effectively flattening the alignment fixtures (see Figure 47(d)). Subsequently, another two alignment fixtures and the upper spacer



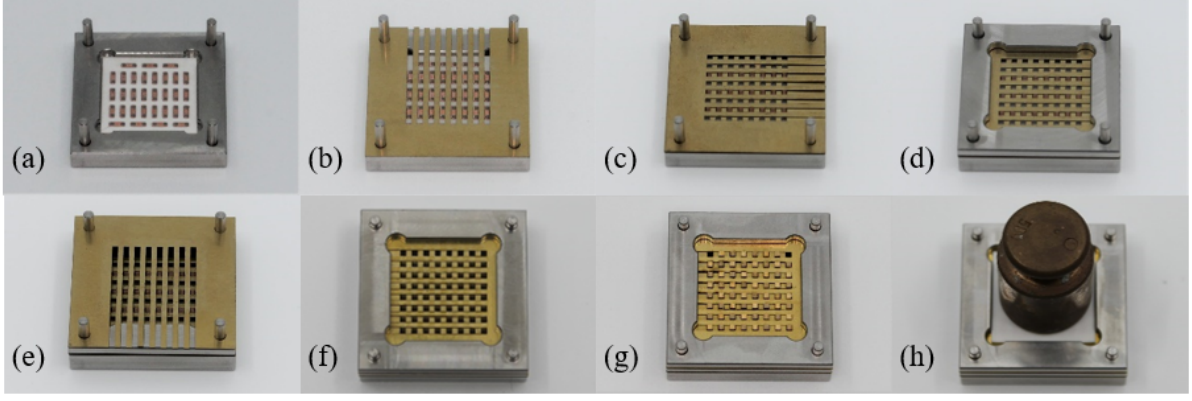


Figure 47: Assembly sequence of the TEG prepared for soldering. (a) The bottom substrate with pasted solder is secured by the holder. (b) The first alignment fixture is mounted on the pins. (c) The second alignment fixture is mounted on the pins. (d) The lower spacer is positioned upon the fixtures. (e) The third and fourth alignment fixtures are placed crosswise upon the lower spacer. (f) The upper spacer is placed. (g) The thermolegs are positioned into the two-layer alignment fixtures. (h) The top substrate with pasted solder is placed onto the thermolegs. Pressure is applied to the substrate using a weight.

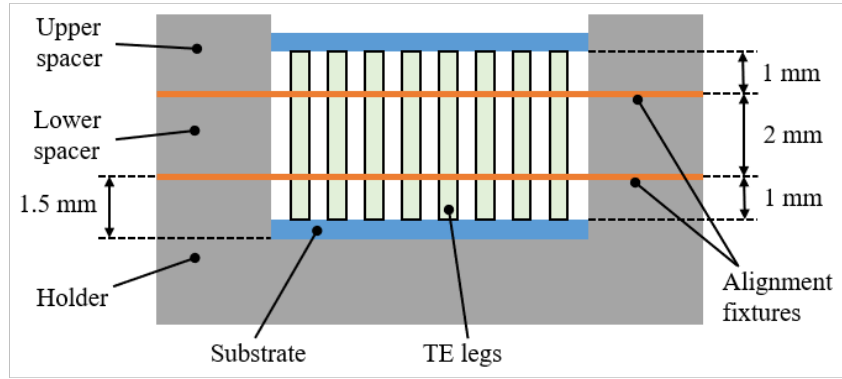


Figure 48: Position of the alignment fixtures, where the two layers are placed symmetrically 1 mm from the center of the thermolegs.

were mounted on the pins (see Figure 47(e), (f)). Notably, the pointing directions of the teeth in each alignment fixture were arranged differently to mitigate any possible asymmetry.

The p- and n-type thermolegs were precisely positioned in the dual-level alignment fixtures (see Figure 47(g)). The legs were arranged in a staggered pattern, ensuring that no two legs of the same type were adjacent. The pick and place operation of the thermolegs was facilitated using a manipulator (Fritsch LM901) equipped with a vacuum nozzle, as depicted in Figure 44(c).

The positioning of the thermolegs in the assembly rig is outlined in Figure 48. Notably, the recess in the holder has a depth of 1.5 mm, while the spacer has a height of 2 mm. The substrate comprises a 0.4 mm thick alumina plate and 0.1 mm thick copper interconnects. The alignment fixtures played a crucial role in securing the thermolegs and maintained a distance of 1 mm from both ends of each leg. This arrangement effectively prevented any potential tilt or rotation of the thermolegs during the assembly process. The tolerances

of the assembly rig were defined through iterative prototyping and testing.

The top substrate with pasted solder was then carefully positioned onto the thermolegs. The position of the substrate was aligned and constrained by the upper spacer. To ensure proper contact between the interconnects and the thermolegs during soldering, a weight of 50 grams was placed on the top substrate, as depicted in Figure 47(h).

### 4.2.3 Soldering

Figure 44(d) illustrates the soldering process using a reflow oven (SMT 400C). The assembly was conveyed through four 0.4 m long heat zones via a conveyor belt with controlled transfer speed and temperature.

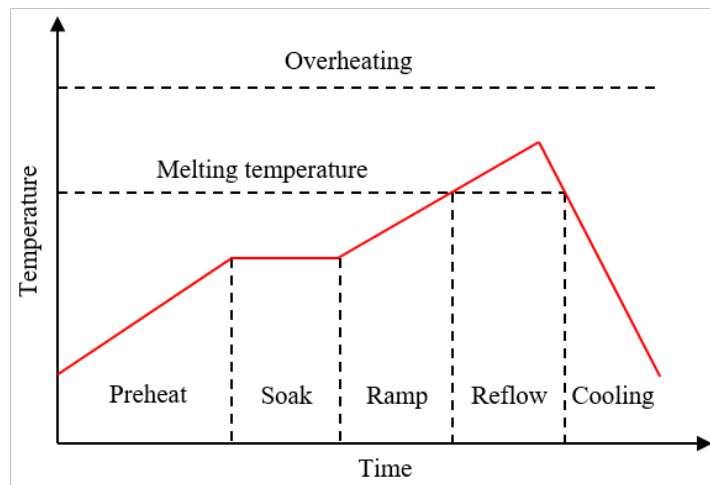


Figure 49: Typical phases of the reflow soldering process.

The reflow soldering process typically has several phases, as depicted in Figure 49. Initially, the assembly undergoes preheating to eliminate volatile solvents and activate the flux in the solder. The flux aids in the removal of metal oxides from the surface of TE legs and copper interconnects, promoting solder wetting. The soak zone ensures uniform temperature distribution across the assembly, reducing the risk of thermal shock. Subsequently, the temperature gradually increases to the solder's melting point during the ramp zone. This is followed by the reflow zone, where the solder melts, creating solder joints. Finally, air cooling solidifies the solder, ensuring good joint quality. It is important to note that the rate of temperature increase during preheating and ramping should be sufficiently slow to prevent thermal shock. Conversely, the rate of temperature decrease during cooling should be rapid to avoid excessive grain growth and minimize thermal stress on the components.

Due to the thermal mass of the assembly rig, the temperature of the TEG inside the assembly rig may lag behind the desired setting temperature of the oven during soldering. In addition, incorrect temperature control can potentially result in failures such as overheating and cold soldering, which are common issues in experiments of this nature.



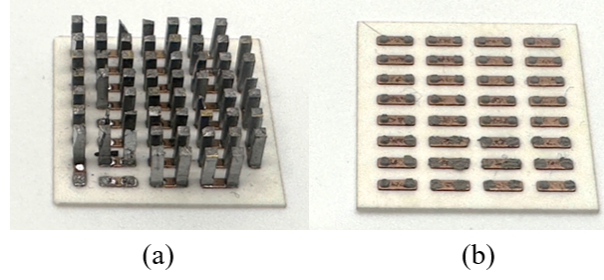


Figure 50: Failures caused by inappropriate temperature setting in the soldering process. (a) Disjointed thermolegs caused by overheating. (b) Unmelted solder caused by cold soldering.

Overheating can accelerate oxidation on metal surfaces and burn away the flux, impeding proper joint formation. Additionally, as previously explained, brittle Sn–Te intermetallic compounds can form between the solder and thermolegs at temperatures around 250 °C, which reduces bonding strength. Conversely, if the peak solder temperature is lower than the melting temperature, or if the reflow zone duration is insufficient, cold soldering may occur, which leads to failed joints or high electrical resistance. Figure 50 (b) depicts a failure resulting from overheating, where the connections break apart during the removal of the assembly rig. Finally, Figure 50 (c) illustrates a failure due to cold soldering, where the solder paste on the interconnects remains unmelted.

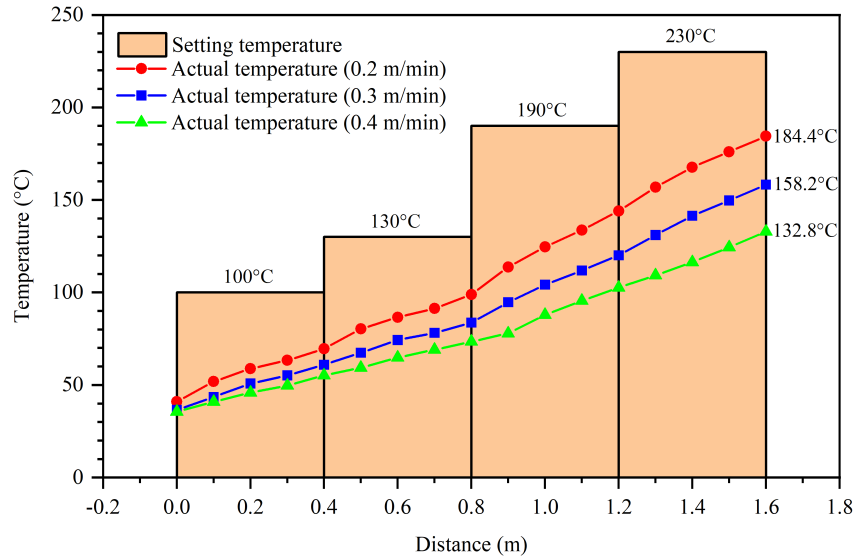


Figure 51: Actual temperature in the assembly rig during soldering in the reflow oven.

To mitigate this issue, a temperature sensor (Heraeus M310) was affixed to the copper interconnects and integrated into the assembly rig. This setup enabled accurate temperature monitoring and regulation of the soldering process. The thermal profile of the reflow soldering process is depicted in Figure 51. Each heating zone was set to temperatures of 100, 130, 190 and 230 °C. A higher transfer velocity reduces the duration of each heating zone, resulting in lower solder temperatures. When the transfer velocity is increased to 0.4 m/min, the peak solder temperature is 132.8 °C, which is lower than the melting

temperature of approximately 138 °C, leading to cold soldering. Transfer velocities of 0.2 and 0.3 m/min result in successful soldering.

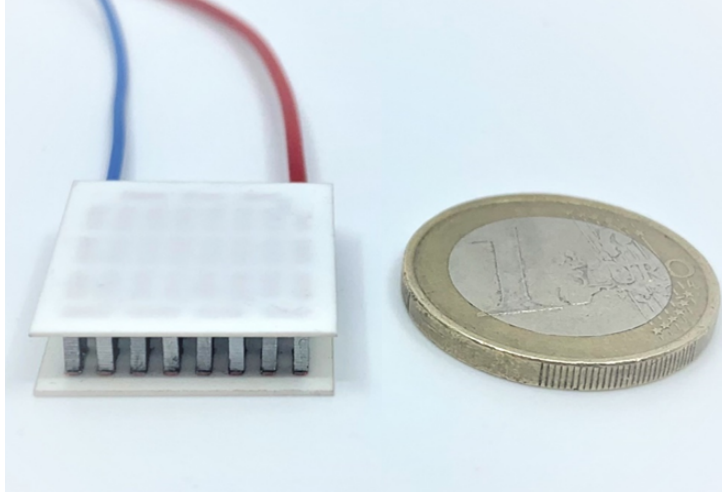


Figure 52: TEG-1 ( $20 \times 20 \text{ mm}^2$ ) with wire connections, featuring 62 thermolegs with dimensions of  $1 \times 1 \times 4 \text{ mm}^3$ .

Consequently, all connections in TEG-1 were established reliably and repeatedly after a single soldering operation. Any failures that might occur during the soldering process due to melted solder flowing, thermal expansion, uneven forces applied on the TEG, and component errors were avoided by using the assembly rig and the applied weight on the top substrate. Figure 52 displays the successfully soldered TEG-1 after the removal of the assembly rig and the completion of the wire connections. Each of the two wires was connected to the interconnects, which were only in contact with a single thermoleg.

### 4.3 TEG with High-Aspect-Ratio Thermolegs

As studied in Sections 2 and 3, increasing the number of thermolegs while maintaining the optimum fill factor enhances the voltage output while maintaining maximum power generation. Therefore, another design, referred to as TEG-2, will further explore the manufacturability of TEGs with higher integration densities. This TEG configuration comprises a  $16 \times 16$  array of thermolegs ( $0.5 \times 0.5 \times 4 \text{ mm}^3$ ) with a pitch of 1.1 mm. Retaining the same fill factor as TEG-1, this configuration features a four-fold increase in the number of TE legs. Consequently, TEG-2 is anticipated to deliver the same power as TEG-1 but with a four-fold increase in voltage output.

#### 4.3.1 Grinding the Thermolegs

$\text{Bi}_2\text{Te}_3$  exhibits a layered rhombohedral crystal structure with strong covalent bonding within the layers and weak van der Waals interactions between them [151]. This structural characteristic facilitates cleavage along specific crystallographic planes, causing the

material's brittleness. Thus, wire-cutting is not applicable to manufacture thermolegs with intended dimensions. As a solution, a grinding procedure is implemented to reduce the cross-section of the  $\text{Bi}_2\text{Te}_3$  legs from  $1 \times 1 \text{ mm}^2$  to  $0.5 \times 0.5 \text{ mm}^2$  (see Figure 53).

A custom grinding jig was designed to facilitate the size reduction of  $\text{Bi}_2\text{Te}_3$  legs, which consist of an aluminum base and two stainless-steel stencils. The stencils have a thickness of  $250 \text{ }\mu\text{m}$  and 40 openings with a cross-section of  $1 \times 5 \text{ mm}^2$ . They were stacked and attached to the aluminum base using a strong adhesive (5 Minute Epoxy, Faserverbundwerkstoffe). Thereby, the grinding jig features  $0.5 \text{ mm}$  deep notches that securely hold the legs during grinding and simultaneously act as a depth gauge. Grinding was performed using a polisher (TegraPol-15, Struers) and a silicon carbide foil (Struers) with a grain size of  $15 \text{ }\mu\text{m}$ . The grinding was performed with a rotation speed of  $60 \text{ rpm}$ , with water added for cooling and lubrication.

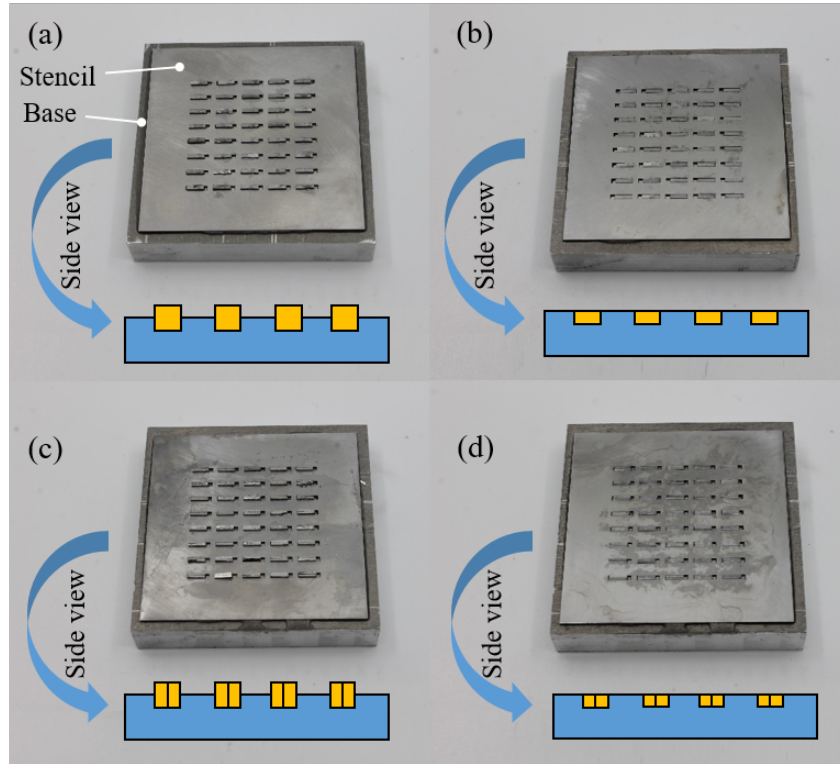


Figure 53: Grinding process of  $\text{Bi}_2\text{Te}_3$  thermolegs. (a) Thermolegs ( $1 \times 1 \times 4 \text{ mm}^3$ ) are attached to the grinding jig. (b) Thermolegs after first grinding ( $0.5 \times 1 \times 4 \text{ mm}^3$ ). (c) Ground thermolegs are combined and attached to the grinding jig. (d) Thermolegs after second grinding ( $0.5 \times 0.5 \times 4 \text{ mm}^3$ ).

Initially,  $\text{Bi}_2\text{Te}_3$  legs with dimensions of  $1 \times 1 \times 4 \text{ mm}^3$  were attached to the grinding jig using a water-soluble adhesive (Crystalbond<sup>TM</sup> 555) (see Figure 53(a)). After the first grinding step, the cross-section of the legs was reduced to  $0.5 \times 1 \text{ mm}^2$  (see Figure 53(b)). Subsequently, the jig was soaked in hot water to dissolve the adhesive, which allowed the legs to be repositioned.

The second grinding step aims to further reduce the leg cross-section to  $0.5 \times 0.5 \text{ mm}^2$ . However, using  $0.5 \text{ mm}$  wide notches designed for this size often leads to an uneven cross-

section (see Figure 54(a)). To address this problem, two semi-grounded legs were combined and reattached to the jig using adhesive (see Figure 53(c) and Figure 54(b)). With this approach, the legs were successfully reduced to a regular cross-section of  $0.5 \times 1 \text{ mm}^2$  (see Figure 53(d)).

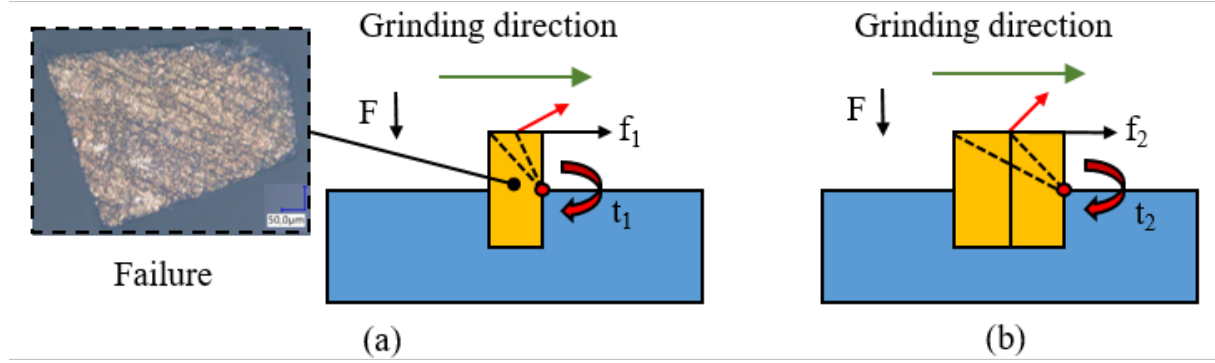


Figure 54: Stress analysis of the Bi<sub>2</sub>Te<sub>3</sub> thermolegs during the second grinding process using different jigs. (a) Grinding jig using notches with a cross-section of  $0.5 \times 0.5 \text{ mm}^2$ . (b) Grinding jig using notches with a cross-section of  $1 \times 0.5 \text{ mm}^2$ . After first grinding, every two ground thermolegs are combined and attached to the jig.

Figure 54 illustrates the rationale behind the improvement observed during the second grinding. Assuming an identical normal force applied to the Bi<sub>2</sub>Te<sub>3</sub> legs for each grinding approach, using the 0.5 mm wide jig results in higher pressure on the leg due to less contact area between the polisher and the leg. This leads to higher friction at each contact point ( $f_1$ ) compared to the friction ( $f_2$ ) using the 1 mm wide jig. Consequently,  $f_1$  generates more torque ( $t_1$ ) using the exposed part of the leg and the contact point of the jig as a pivot point, which can potentially cause failures such as breakage or crushing. Combining the legs reduces the torque, which ensures stability and quality during grinding. Alternatively, separating the second grinding into two rounds, including a grinding step using 0.75 mm deep notches, can further mitigate this issue.

Figure 55 presents the original and the ground cross-sections of Bi<sub>2</sub>Te<sub>3</sub> legs. The ground legs exhibit regular and complete profiles, which underscores the effectiveness of the grinding approach.

### 4.3.2 Thermolegs Assembly

A new substrate and corresponding alignment fixtures were specifically designed to match the array configuration of TEG-2. These fixtures were fabricated from 100  $\mu\text{m}$  thick stainless steel and feature a customized shape that complements the TEG-2 array. However, laser cutting caused deformations due to the high aspect ratio teeth. This tilting stems from localized heating during laser cutting. The heat induces a slight expansion in the teeth, and as they cool down and contract their shape deviates from the ideal form. This

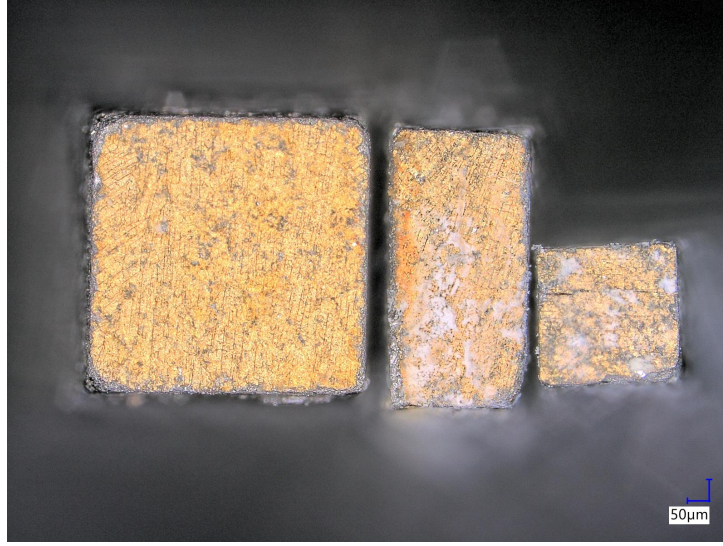


Figure 55: Cross section of  $\text{Bi}_2\text{Te}_3$  thermolegs before grinding ( $1 \times 1 \text{ mm}^2$ , left-hand panel), after first grinding ( $0.5 \times 1 \text{ mm}^2$ , middle panel), and after second grinding ( $0.5 \times 0.5 \text{ mm}^2$ , right-hand panel).

might lead to the disconnection of the thermolegs or the tilting of TEG-2, as demonstrated in Figure 56(a).

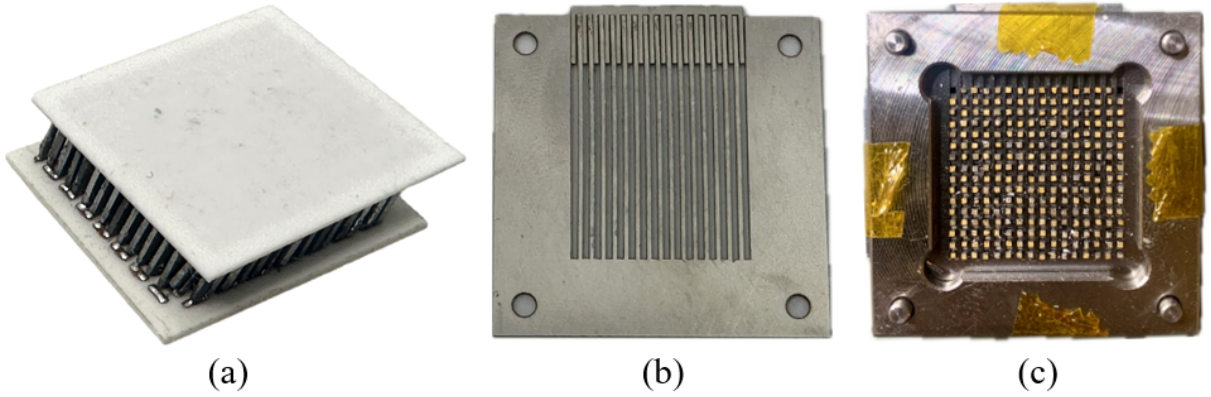


Figure 56: (a) Failure of TEG-2 caused by tilted teeth of the alignment fixtures. (b) A teeth guide is used to interlock the teeth of the alignment fixture. (c) Thermolegs ( $0.5 \times 0.5 \times 4 \text{ mm}^3$ ) are positioned into the improved assembly rig.

A teeth guide is designed to address this issue and guarantees that the thermolegs are properly positioned. This guide interlocks with the teeth of the alignment fixtures to ensure their precise alignment (see Figure 56(b)).

The high aspect ratio of the thermolegs poses a challenge for the manipulator's picking mechanism. To overcome this problem, precision tweezers were employed to carefully insert the p- and n-type legs into the alignment fixtures. Figure 56(c) illustrates the assembly rig after this successful pick-and-place process. Leveraging the fabrication strategy of TEG-1, here the TEG-2 design has been successfully fabricated (see Figure 57).



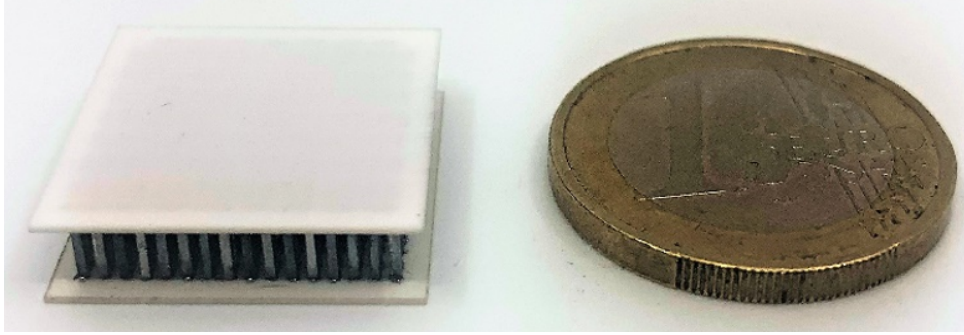


Figure 57: TEG-2 ( $20 \times 20 \text{ mm}^2$ ) with 254 thermolegs with dimensions of  $0.5 \times 0.5 \times 4 \text{ mm}^3$ .

#### 4.4 TEG with Integrated Boost Converter

Although TEG-2 had TE legs with an aspect ratio of 1:8, the achieved voltage output is still insufficient for direct use in IMDs. Increasing the aspect ratio of the thermolegs even further would require a more time-consuming and error-prone assembly process. For instance, to maintain the power output and obtain a 2 Volt voltage output, it would be necessary to incorporate around 6000 thermolegs with dimensions of  $0.1 \times 0.1 \times 4 \text{ mm}^3$ . However, achieving this configuration using the semi-manual fabrication approach is not feasible due to the challenges associated with assembly at that scale and the limited processing time of the solder paste. To address this limitation, a commercially available boost converter is introduced for voltage up-conversion. This approach can meet the voltage requirements of IMDs without the need for excessively high aspect ratios or an impractical number of thermolegs. The boost converter's circuit is routed directly on the alumina substrate of the TEG. This approach minimizes the footprint and satisfies the restricted space requirements of IMDs. Fewer wired connections also improve system reliability and thus reduce the potential for failure.

Integrating a boost converter within a TEG requires a more meticulous thermal management design. The boost converter (ECT 310, EnOcean) that was used in the experiment has a dual-layer circuit. Routing the circuit across both substrates would require vertical interconnects between both substrates. This introduces highly conductive heat paths, which could negatively impact the temperature gradient across the thermopile and reduce TEG performance. To avoid thermal shunting between the substrates, the boost converter's circuit layout is routed into a single layer (see Figure 88 in the Appendices). The single-layer circuit is routed around the center of a round alumina DBC substrate (referred to as the bottom substrate) with a diameter of 30 mm, as shown in Figure 58(a). The copper interconnects surrounding the boost converter circuit are prepared for the assembly of 62 thermolegs with dimensions of  $1 \times 1 \times 4 \text{ mm}^3$ . The top substrate connects the TE legs (see Figure 58(b)). Figures 58(c) and (d) present the patterns of the adapted stencils prepared for screen printing of solder paste onto the bottom and top substrate, respectively.

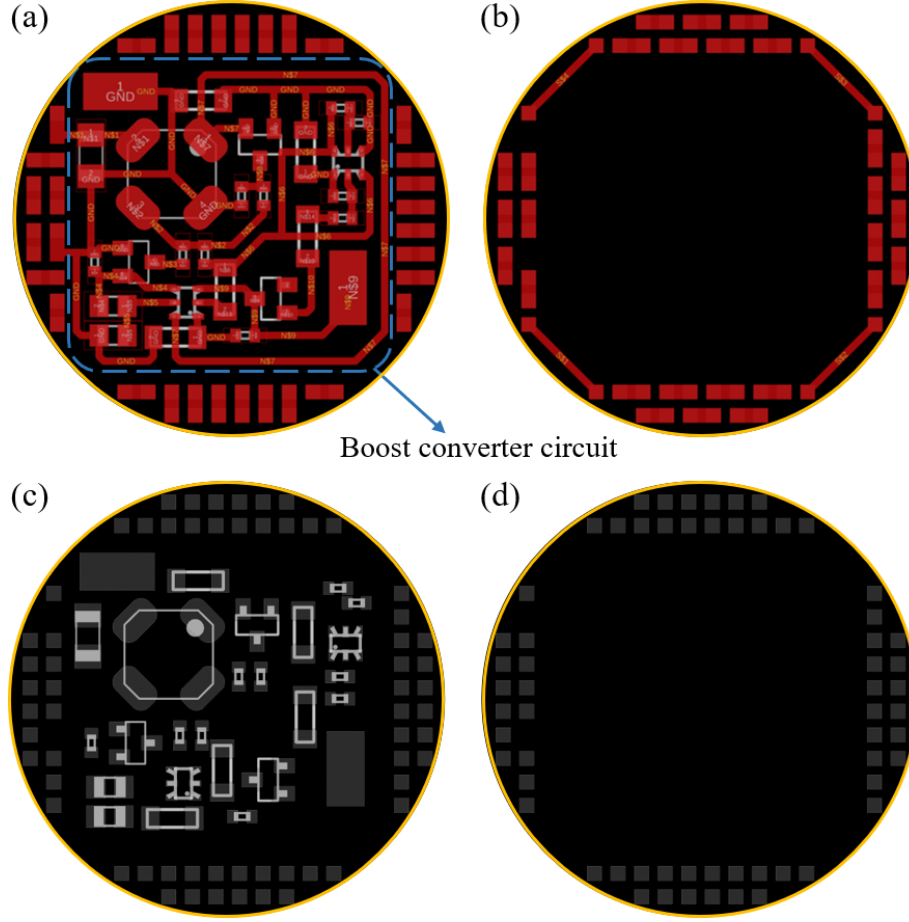


Figure 58: Design of the DBC alumina substrates and stencil that were used to fabricate TEG-3. (a) The bottom substrate with copper interconnects (red) is used to connect the thermolegs and the boost converter electronics. (b) The top substrate with copper interconnects (red) is used to connect the thermolegs. (c) Stencil with openings (grey) for solder pasting on the bottom substrate. (d) Stencil with openings (grey) for solder pasting on the top substrate.

Figure 59(a) illustrates the bottom substrate of TEG-3, where the electronic components of the boost converter were soldered and tested. Notably, the electronics in the boost converter have a maximum height of 3.7 mm, which is less than the height of the TE legs being 4.0 mm. Thus, a small air gap remains between the electronic components and the top substrate, which provides thermal isolation between the two substrates. The assembly process for TEG-3 is similar to the assembly process explained in Figure 47. The assembly rig was adapted to accommodate round substrates, which provided a suitable platform for the placement and alignment of the thermolegs. Appropriate assembly fixtures were utilized to securely hold the thermolegs in place during the soldering process. As shown in Figure 59(b), the middle teeth of the alignment fixture are shortened to leave space for the already present electronic components. Unlike the assemblies of TEG-1 and TEG-2, the TEG-3 assembly requires four alignment fixtures in different orientations for the bottom and top fixtures. The holder and spacers in the adapted assembly fixtures have centered, line-shaped notches on each of their four edges, which ensures precise align-

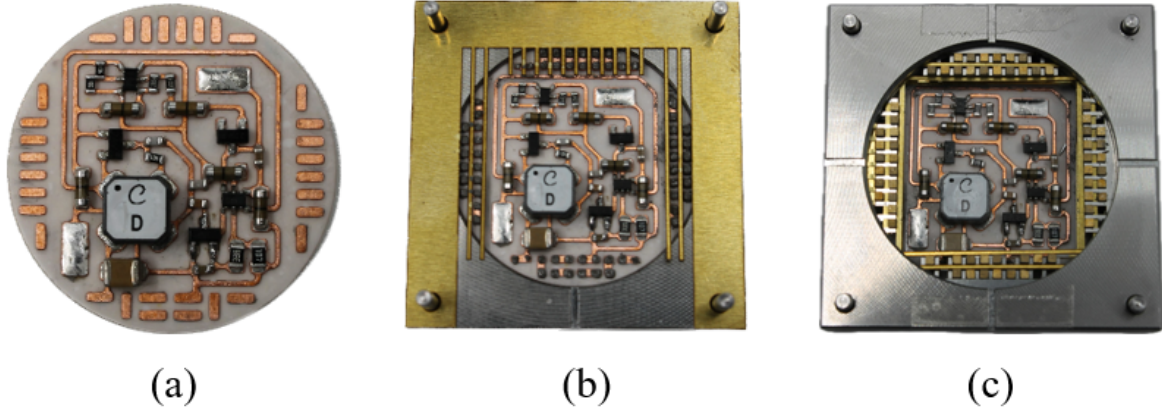


Figure 59: Fabrication process for TEG-3. (a) The boost converter electronics are soldered onto the bottom substrate. (b) One of the adapted alignment fixtures is mounted by the pins. (c) The thermolegs are positioned into the adapted assembly rig. Figure (a) and (b) are reprinted from [118].

ment between the substrates and thermolegs. Figure 59(c) illustrates the assembly after the placement of the thermolegs.

Unlike the assemblies of TEG-1 and TEG-2, the TEG-3 assembly requires four alignment fixtures in different orientations for both the top and bottom. The holder and spacers in the adapted fixtures have centered, line-shaped notches on each of their four edges, ensuring precise alignment between the substrates and thermolegs.



Figure 60: TEG-3 with a diameter of 30 mm, featuring 62 thermolegs with dimensions of  $1 \times 1 \times 4$  mm<sup>3</sup>. Reprinted from [118].

This fabrication approach ensured that the boost converter could be seamlessly integrated into the TEG assembly, which enhanced the overall functionality and efficiency. The TEG-3 with integrated boost converter has a diameter of 30 mm and a height of 5 mm, as depicted in Figure 60.

## 4.5 Summary

Chapters 2 and 3 emphasized the importance of free-standing high-aspect-ratio thermolegs in ITEGs, highlighting the associated manufacturing challenges. In this chapter, a novel fabrication strategy for such TEGs was proposed, introducing an assembly rig for the



soldering process. This rig includes dual-layer alignment fixtures with open teeth, which enable the simultaneous establishment of all joint connections in a single soldering process. Failures that might occur during the soldering process were able to be avoided due to the alignment and constraints of the TEG components using the assembly rig. After soldering, the assembly rig can be disassembled and reused.

Several TEGs were designed and fabricated with optimized thermal resistance to match the implantation environment described in Chapter 2, maximizing power output. Although these TEGs share the same optimized thermal resistance, they are expected to produce different voltages due to variations in array configurations and integrated boost converter.

- TEG-1: A  $20 \times 20 \text{ mm}^2$  device featuring 62 thermolegs, each measuring  $1 \times 1 \times 4 \text{ mm}^3$ .
- TEG-2: Also  $20 \times 20 \text{ mm}^2$ , this design contains 254 thermolegs ( $0.5 \times 0.5 \times 4 \text{ mm}^3$ ), improving voltage output while maintaining maximum power. A custom grinding jig was developed to address the brittleness of  $\text{Bi}_2\text{Te}_3$  during the production of smaller thermolegs.
- TEG-3: A round-shaped device (30 mm diameter) with 62 thermolegs ( $1 \times 1 \times 4 \text{ mm}^3$ ). A boost converter, integrated by routing circuitry on the ceramic substrate, further enhances voltage output to meet IMD requirements.

## 5 ITEG Characterization

In the previous chapters, the simulation and fabrication of ITEGs were thoroughly presented. This chapter will describe their characterization, which is a crucial step in assessing fabrication reliability and validating simulation accuracy. The specific setups that were designed and fabricated for measurements and experiments will be introduced in detail. Additionally, the measured material properties of  $\text{Bi}_2\text{Te}_3$  thermolegs will be presented, which is essential for validating the TEG characterization results. Various methods for electrical and thermal characterization of the TEGs will be employed and meticulously analyzed. To evaluate whether the defined objectives have been successfully achieved, the optimized TEGs that were described in Chapter 4 are subjected to comprehensive characterization and comparison.

### 5.1 Methods Review

As studied in the previous simulations, thermal design is crucial for optimum performance of ITEGs. To maximize their power output, the fill factor of the TEG should be optimized to achieve thermal matching between the TEG and its implantation environment. Therefore, the measurement of the thermal resistance of the TEGs is most important in the characterization and should be treated with care. The common methods to measure thermal resistance can be classified into two types: transient and steady-state methods. Laser flash analysis (LFA) is a typical transient method and is used to measure the thermal diffusivity<sup>1</sup> of the sample. Although this method is very fast and suitable for high thermal conductivity materials, it requires a homogeneous and well-prepared sample. Steady-state methods measure the thermal resistance by acquiring the temperature difference and heat flow in steady-state. Thus, these methods require a longer acquisition time but are compatible with samples with complex structures or multiple components. This measurement is achieved by controlling either the heat flow or the temperature difference.

#### **Laser Flash Analysis**

The working principle of LFA is illustrated in Figure 61. Measurement samples shall feature uniform thickness. A homogeneous laser pulse heats the front side of the sample and leads to a gradual temperature rise from the front side to the back side of the sample. The higher the thermal diffusivity of the sample, the faster the energy reaches the backside. A detector receives the infrared radiation emitted from the backside of the sample. With higher temperatures at the backside, more infrared energy is radiated [152]. Therefore, a time-dependent temperature at the backside can be obtained by processing the electrical signal from the detector.

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<sup>1</sup>Thermal diffusivity measures the speed at which heat propagates through a material.

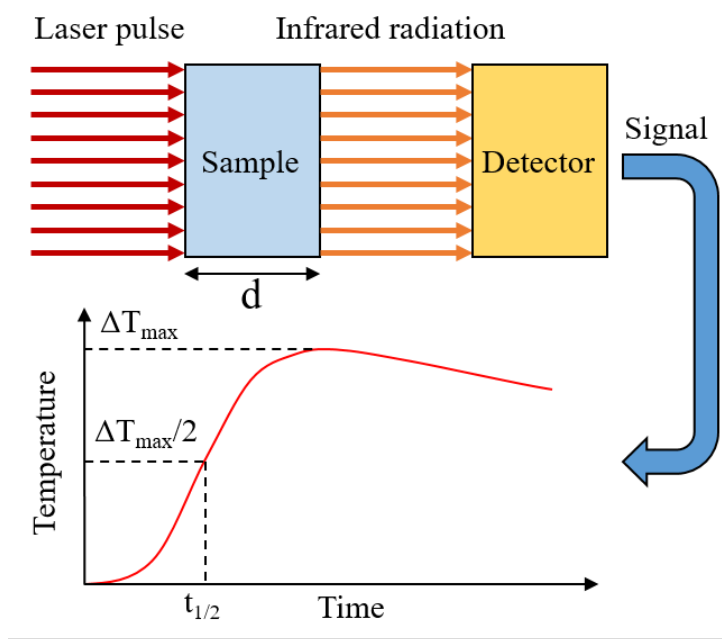


Figure 61: Working principle of LFA for thermal characterization. A laser pulse hits the front side of the sample and causes a temperature rise. The backside temperature of the sample is measured by an infrared detector.

If the density  $\rho$ , specific heat capacity  $c$ , and thermal diffusivity  $\alpha_d$  of the samples are known, the thermal conductivity of the material  $\kappa$  can be calculated as:

$$\kappa = \alpha_d \cdot \rho \cdot c \quad (5.1)$$

$\alpha_d$  can be calculated as:

$$\alpha_d = 0.13879 \cdot \frac{d^2}{t_{1/2}} \quad (5.2)$$

where  $d$  is the thickness of the sample.  $t_{1/2}$  represents the time it takes for the temperature of the sample back side to reach half of its maximum temperature as shown in Figure 61.

### Measurement with Controlled Heat Flow

Figure 62(a) illustrates the working principle of a setup using a reference meter bar to measure the thermal conductance of a specimen. Both meter bars, fabricated from a material with known thermal conductivity, carry temperature sensors at well-defined positions. The specimen is positioned between the two meter bars. A heater and a cooling unit are placed on the upper and lower side, respectively. The hot and cold side temperature of the specimen can be extrapolated from the temperature gradient along the meter bar:

$$T_{hot} = T_1 - \frac{d_2}{d_1}(T_1 - T_2) \quad (5.3)$$

$$T_{cold} = T_3 + \frac{d_4}{d_3}(T_3 - T_4) \quad (5.4)$$

If a resistive heater is employed and good heat insulation is guaranteed, then all generated heat passes through the specimen and the heat flow passing through the specimen can be calculated as:

$$Q = V \cdot I \quad (5.5)$$

where  $V$  and  $I$  are the voltage and current applied to the resistive heater, respectively. Alternatively, the heat flow can be calculated as:

$$Q = (T_1 - T_2) \frac{\kappa A}{d_1} \quad (5.6)$$

or

$$Q = (T_3 - T_4) \frac{\kappa A}{d_4} \quad (5.7)$$

where  $\kappa$  is the thermal conductivity of the reference meter bar.  $A$  is the cross-sectional area of the reference meter bar. The thermal conductance of the specimen can be calculated using the obtained temperature difference and heat flow:

$$K = \frac{Q}{T_{hot} - T_{cold}} \quad (5.8)$$

For accurate measurement, the heat transfer from the setup to the environment should be minimized. This can be achieved by conducting the measurement in vacuum. A heat insulator or a guard heater surrounding the setup also support measurement accuracy.

### Measurement with Controlled Temperature Difference

As illustrated in Figure 62(b), the heat flow through the specimen can also be obtained by using a heat flux sensor. This sensor is typically a thermopile that is embedded in a thin, thermally conductive material. When heat flows through the sensor, it creates a temperature difference across the thermopile. This temperature difference generates a voltage that is proportional to the heat flow rate. The sensor is calibrated against known heat flow standards, which allows the measured voltage to be directly converted to heat flux or flow. Typical standards and techniques used for calibrating the heat flux sensors include NIST SP-250-65 [153], ISO 14934-2:2013 [154], and ASTM C177 [155].

## 5.2 Design and Fabrication of the Measurement Setups

This section describes the design of the measurement setups that will be used for TEG characterization. One experimental setup was developed to precisely control the temperature at both sides of the specimen, enabling reproducible measurements. As a variation of this setup, another similar setup was built to measure the heat flux through the TEGs.

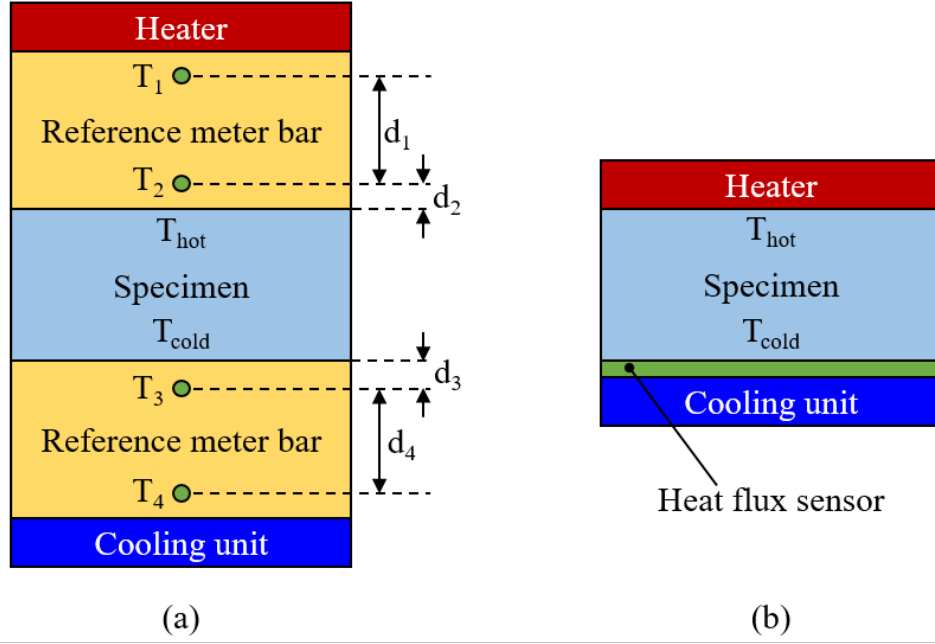


Figure 62: Steady-state methods for thermal characterization. (a) The temperature difference across the specimen is measured by reference meter bars with a known heat flow through the specimen. (b) The heat flow through the specimen is measured by a heat flux sensor whereas the temperature difference across the specimen is maintained constant.

### 5.2.1 Experimental Setup for Temperature Control

The experimental setup incorporates several key components to ensure precise temperature control on both sides of the TEG. A CAD model of the setup is illustrated in Figure 63. The fabricated setup is demonstrated in Figure 64. A breakdown of the components and their functions follows:

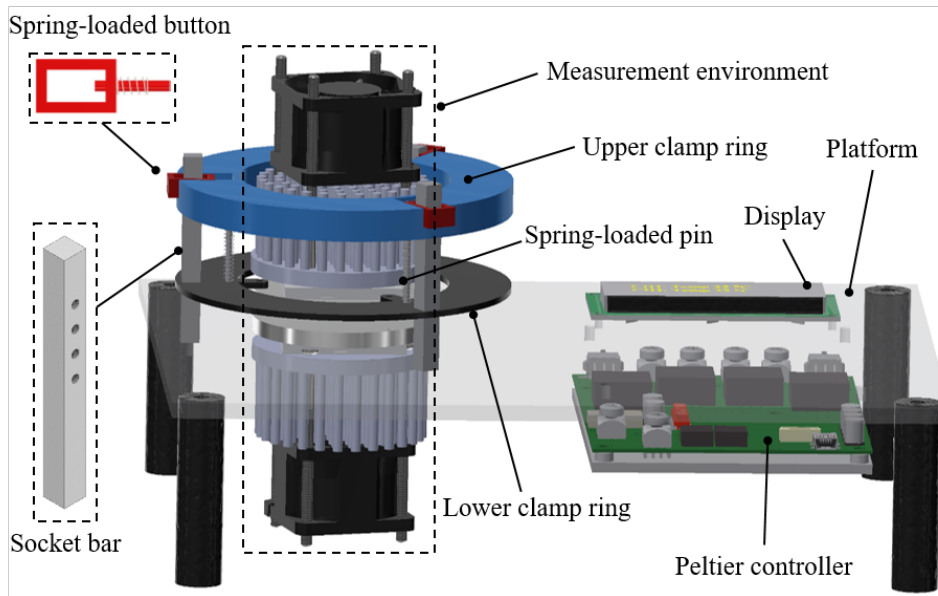


Figure 63: CAD model of the experimental setup, which consists of a platform, a clamping system, a measurement environment incorporating temperature sensing, and a Peltier controller.

- **Platform:** The setup platform accommodates the measurement environment, a Peltier controller, and a display. The display shows temperature readings from integrated sensors, allowing for real-time monitoring.
- **Clamping System:** A set of clamps is designed to apply pre-set and uniform clamping force to the specimen. Three circularly symmetric distributed socket bars secure these clamps to the platform. Spring-loaded buttons allow for adjustment of the upper clamp ring's position along the socket bars, providing different clamping force applied to the top aluminum block and the specimen.
- **Measurement Environment:** As illustrated in Figure 65, the measurement environment houses two Peltier modules (TEC1-12706) attached to aluminum blocks. The Peltier modules utilize the Peltier effect to transfer heat from one side to the other. Heating or cooling rates are primarily determined by the applied current and voltage. Heat sinks (Dualled-5830, Wakefield Thermal) and fans dissipate the heat generated by the Peltier modules to maintain their proper operation.



Figure 64: Experimental setup for temperature control during TEG characterization.

- **Temperature Sensing:** Various sensors are strategically placed to monitor different components and facilitate temperature sensing within the experimental setup. Two platinum temperature sensors (M 310, Heraeus) are glued to the top and bottom aluminum blocks to sense the temperature of the specimen. These sensors provide essential feedback for temperature control and regulation. Additionally, two thermistors (NTC, 10 Kohm, 0.2%) are attached to the top and bottom fans

(Delta FFB0424VHN-TZT4). These serve as temperature sensors for the hot side of the Peltier modules. If passive cooling provided by the heat sinks is insufficient to reduce the temperature of the hot side, the fans enable active cooling.

- **Peltier Controller:** The Peltier controller (TEC-1122-SV, Meerstetter Engineering) enables independent feedback control of the upper and lower side temperatures. It employs a proportional-integral-derivative (PID) control system for temperature control. For this purpose the input voltage and current of the Peltier modules are adjusted accordingly.
- **Thermal Attachment:** All attachments utilize thermally conductive adhesive (Keratherm-Bond 100 RT, KERAFOIL) to minimize heat loss, thereby enhancing measurement accuracy.

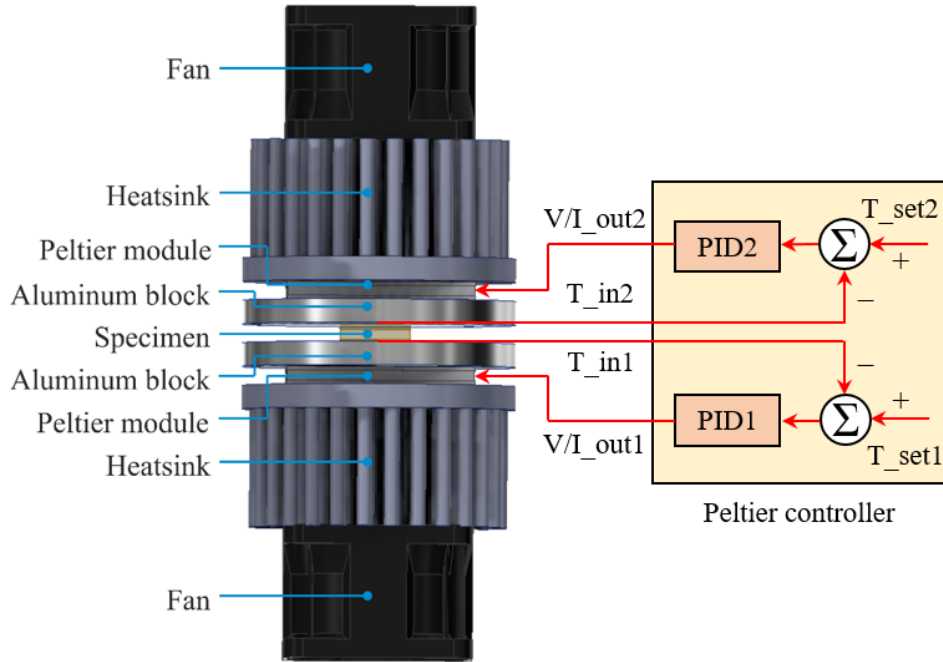


Figure 65: Measurement environment in the experimental setup. The Peltier modules are controlled by a PID Peltier controller.

### 5.2.2 Measurement Setup for Thermal Characterization

An additional measurement setup was adapted from the experimental setup described above for the thermal characterization of TEGs. Figure 66(a) illustrates the modification to the measurement setup, which incorporates a thin-film heat flux sensor attached to the bottom aluminum block. An aluminum sample stage, which matches the cross-section of TEG-1 ( $20 \times 20 \text{ mm}^2$ ), was attached to the top surface of the heat flux sensor using thermally conductive adhesive. This ensured the precise positioning of the TEG samples

within the sensing area of the heat flux sensor. Compared to the top surface of the heat flux sensor, the surface of the stage is flat, which improves thermal contact between the specimen and the measurement setup. To accommodate the measurement of the round-shaped TEG-3, a stage adapter with a diameter of 30 mm (depicted in Figure 66(b)) was utilized.

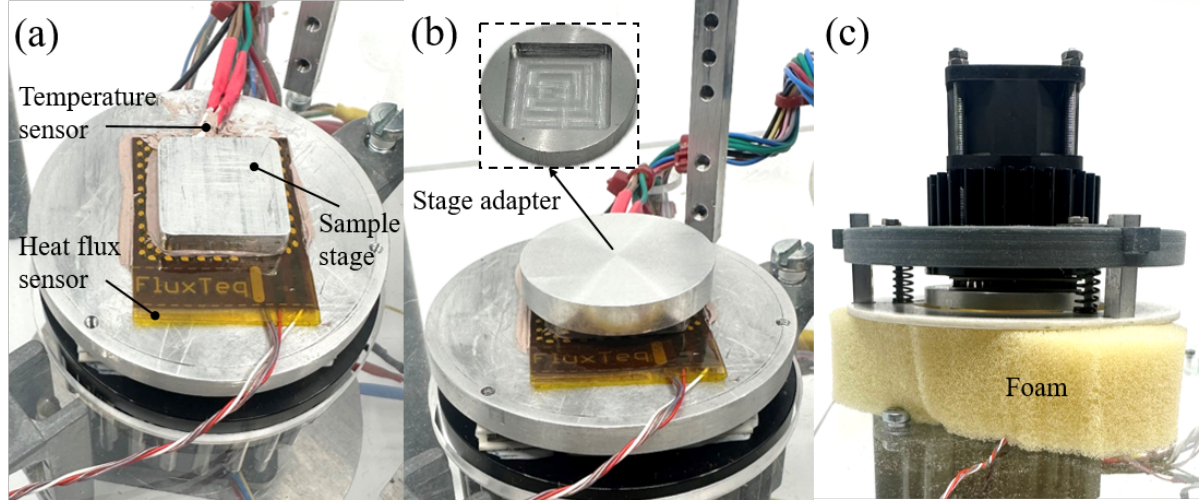


Figure 66: Measurement setup for thermal characterization. (a) Integration of a heat flux sensor and a sample stage. (b) A stage adapter to accommodate the measurement of TEG-3. (c) Foam to enclose the measurement environment.

A temperature sensor is attached to the sample stage to record the temperature on the bottom side of the sample. To minimize environmental disturbances during measurements, the measurement environment was enclosed by plastic foam, which served as a thermal insulator, as depicted in Figure 66(c).

### 5.3 Material Properties of $\text{Bi}_2\text{Te}_3$ Thermolegs

This section presents the characterization of the p- and n-type  $\text{Bi}_2\text{Te}_3$  material of the TEG thermolegs. Table 4 presents the material properties of the thermolegs at a temperature typically found in implant locations (35 to 36 °C). The Seebeck coefficient and electrical resistivity were experimentally measured, while the thermal conductivity was sourced from [156]. This data was used to calculate the expected values of the TEG characteristics in Section 5.6.1 for verification purposes.

Table 4: Material properties of p-type thermolegs ( $\text{BiSbTe}$ ) and n-type thermolegs ( $\text{BiTeSeS}$ ) at human body temperature (35 to 36 °C).

| Material                  | $\rho$ ( $\mu\Omega\cdot\text{m}$ ) | $\alpha$ ( $\mu\text{V}/\text{K}$ ) | $\kappa$ ( $\text{W}/\text{m}/\text{K}$ ) |
|---------------------------|-------------------------------------|-------------------------------------|-------------------------------------------|
| $\text{BiSbTe}$ (p-type)  | 5.9                                 | 183.6                               | 1.5                                       |
| $\text{BiTeSeS}$ (n-type) | 14.9                                | -232.0                              | 1.5                                       |



### Seebeck Coefficient Measurement

The Seebeck coefficients of p- and n-type  $\text{Bi}_2\text{Te}_3$  material were measured using a Linseis LSR-3 setup (see Figure 67). The  $\text{Bi}_2\text{Te}_3$  sample was cut into a regular shape and positioned between two electrodes. The entire measuring setup was arranged in a furnace, which can control the measuring temperature to a specific value. Two heaters above and below the electrodes created a defined temperature gradient. Two thermocouples contacted the hot and cold side of the sample and measured their temperature,  $T_1$  and  $T_2$ , respectively. One wire of each thermocouple was connected to a voltage meter to measure the open-circuit voltage ( $U_{oc}$ ) that was generated from the sample section between two thermocouples.

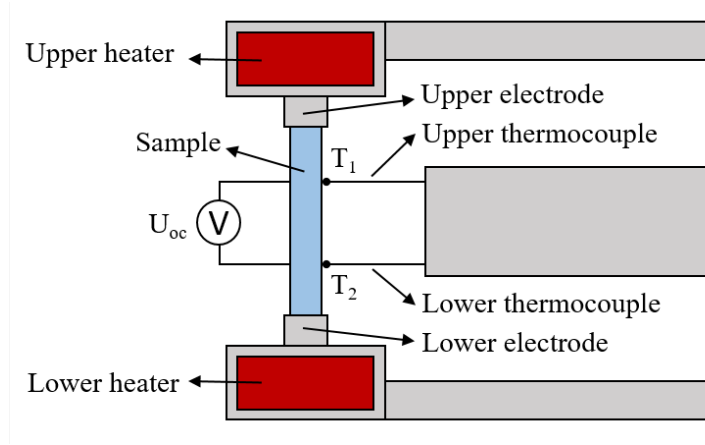


Figure 67: Principle of the Seebeck coefficient measurement using Linseis LSR-3.

The Seebeck coefficient of the sample can be calculated by:

$$\alpha = \frac{U_{oc}}{T_1 - T_2} \quad (5.9)$$

### Electrical Resistivity Measurement

The electrical resistivity of the p- and n-type  $\text{Bi}_2\text{Te}_3$  material was measured using a different configuration by Linseis LSR-3, as depicted in Figure 67. This configuration utilizes the four-terminal method to eliminate the influence of contact and wire resistances, which allows the contact and wire resistances to be neglected. A constant current ( $I_{dc}$ ) is fed through the sample using the upper and lower electrodes. The corresponding voltage drop ( $U_{\Omega}$ ) along the sample is measured by a voltage meter that was connected to the wires of the thermocouples. During the measurement process, precise temperature control was maintained within the furnace. This prevented any temperature gradient along the sample, thereby avoiding the generation of a Seebeck voltage from the BiTe sample, which could potentially disturb the voltage meter readings.

Knowing the probe's distance ( $L$ ) and the sample's cross-sectional area ( $A$ ), the elec-

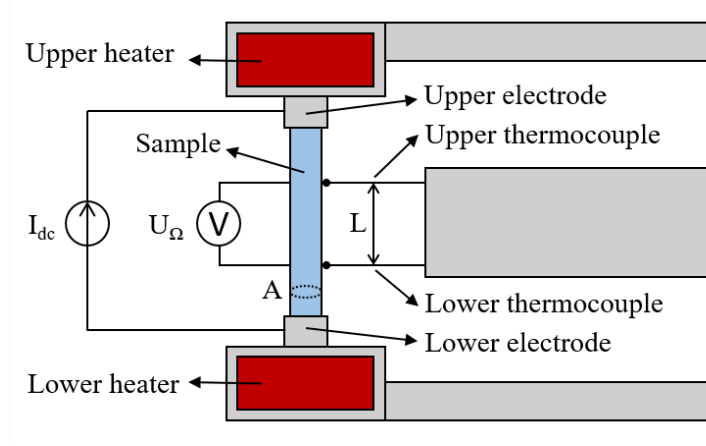


Figure 68: Principle of the electrical resistivity measurement using Linseis LSR-3.

trical resistivity of the sample can be calculated by:

$$\rho = \frac{U_{\Omega}}{I_{dc}} \cdot \frac{A}{L} \quad (5.10)$$

The electrical properties of p- and n-type BiTe were measured under different temperature conditions, as listed in Table 5. The power factor ( $PF$ ) represents the ability of TE material to generate electrical power from a temperature difference, which is expressed as:

$$PF = \frac{\alpha^2}{\rho} \quad (5.11)$$

Table 5: Material properties of p-type thermolegs (BiSbTe) and n-type thermolegs (BiTeSeS) measured by LINSEIS LSR-3.

| T (°C)      | $\rho$ ( $\mu\Omega \cdot m$ ) | $\alpha$ ( $\mu V/K$ ) | PF (mW/m/K <sup>2</sup> ) |
|-------------|--------------------------------|------------------------|---------------------------|
| p-type BiTe |                                |                        |                           |
| 35.0        | 5.9                            | 183.6                  | 5.7                       |
| 42.1        | 6.2                            | 198.1                  | 6.3                       |
| 52.6        | 6.6                            | 193.7                  | 5.7                       |
| 62.7        | 7.0                            | 189.6                  | 5.1                       |
| n-type BiTe |                                |                        |                           |
| 36.4        | 14.9                           | -232.0                 | 3.6                       |
| 41.2        | 15.1                           | -242.9                 | 3.9                       |
| 52.7        | 15.8                           | -238.1                 | 3.6                       |
| 62.9        | 16.3                           | -235.7                 | 3.4                       |

## 5.4 Electrical Characterization of TEG

The ability of TEGs to generate electrical power from temperature differences is reflected by their electrical properties. This section presents the characterization methods of the electrical properties of the TEG. The open-circuit voltage per temperature difference was measured using a temperature control method. The measurement of the electrical resistance was achieved both by the temperature control method and an alternating current (AC) measurement method.

### Temperature Control Method

The electrical characterization of the TEG was conducted using the experimental setup described in Section 5.2.1. In this measurement, the TEG under examination was subjected to a cold side temperature of 25 °C and a hot side temperature of 35 °C. The TEG was connected to a variable load resistor and a digital voltmeter, as depicted in Figure 69(a).

To determine the open-circuit voltage  $U_{oc}$ , the load resistor was disconnected and the digital voltmeter recorded the voltage across the TEG terminals. The internal resistance  $R_{in}$  was then determined through load matching. This involved adjusting the load resistor  $R_L$  until the voltage  $U_L$  was half of  $U_{oc}$ . At this point,  $R_{in}$  equaled  $R_L$ . It is crucial to note that voltage readings should only be taken after the temperature at both sides of the TEG stabilizes to ensure accurate measurements.

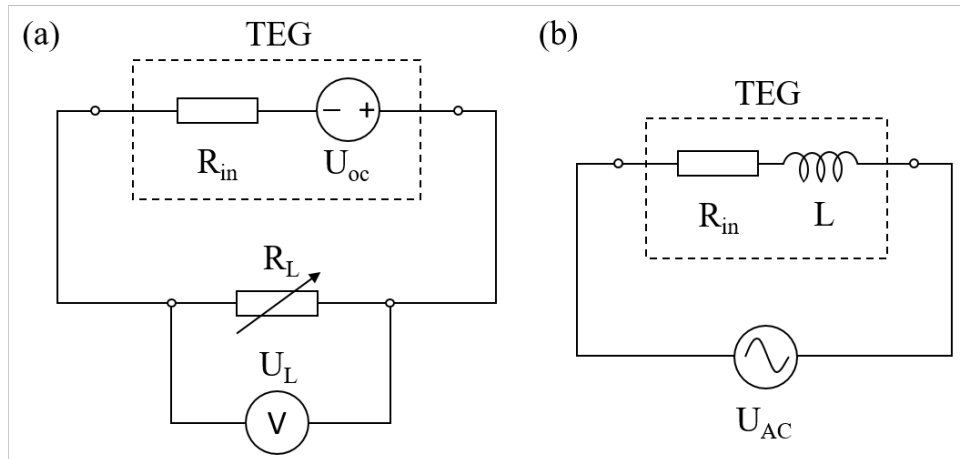


Figure 69: Equivalent circuit of the methods that were used to measure TEG electrical resistance. (a) Temperature control method. (b) AC measurement method.

### AC Measurement Method

It is important to recognize that an ohmmeter may not provide accurate measurements of the internal resistance of a TEG due to the generation of a Seebeck voltage when subjected to direct current. This voltage can interfere with the ohmmeter's test voltage, leading to inaccuracies in the readings. To address this issue, an AC measurement method

was employed to measure the internal resistance of the TEG for verification purposes. This approach minimizes the temperature difference across the TEG without temperature control, thereby reducing the generation of the Seebeck voltage. In this method, the TEG was connected to a measurement system (Gamry Reference 600), which provided an RMS AC voltage ( $U_{AC}$ ) of 100 mV with a variable frequency ranging from 100 Hz to 100 kHz. An equivalent circuit of the TEG was modeled in the system to analyze its impedance, as depicted in Figure 69(b). In this circuit representation, the TEG's impedance was modeled as an internal resistor ( $R_{in}$ ) connected with an inductor. The inductor component accounted for the inductive effects arising from the internal wiring of the TEG. The parasitic capacitor was neglected in the equivalent circuit.

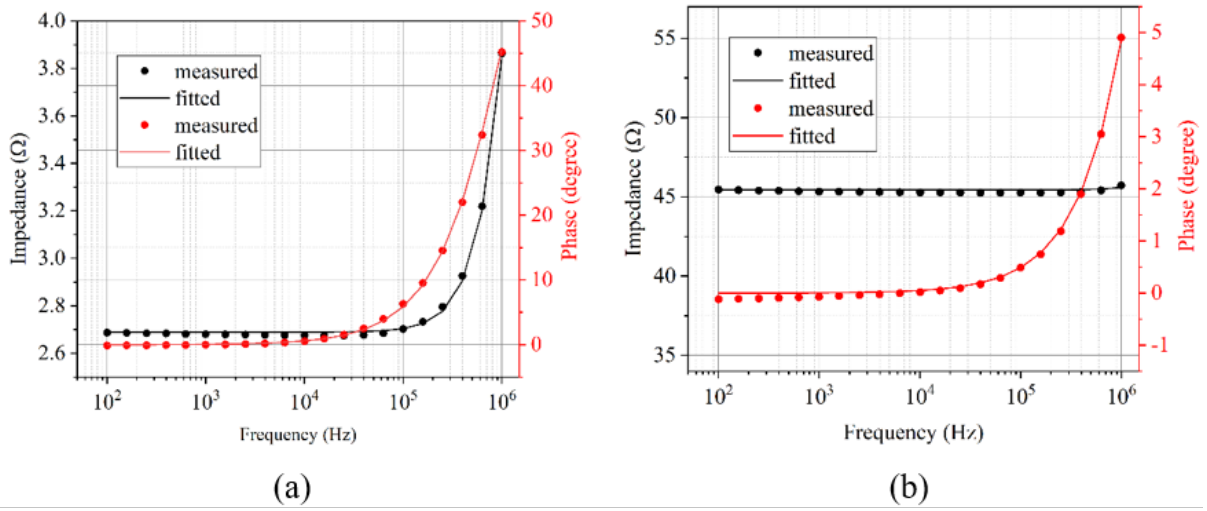


Figure 70: TEG impedance spectrum obtained from the AC measurement method. (a) TEG-1 with 62 thermolegs in dimensions of  $1 \times 1 \times 4 \text{ mm}^3$ . (b) TEG-2 with 254 thermolegs in dimensions of  $0.5 \times 0.5 \times 4 \text{ mm}^3$ .

The inductive reactance ( $X_L$ ) can be calculated as :

$$X_L = 2\pi f L \quad (5.12)$$

where  $L$  is the inductance and  $f$  is the frequency of  $U_{AC}$ . The phase angle ( $\phi$ ), representing the amount by which the current lags the voltage, is calculated by:

$$\phi = \arctan \frac{X_L}{R_{in}} \quad (5.13)$$

The impedance ( $Z$ ) of the RL circuit can be expressed using the following formula:

$$Z = \sqrt{X_L^2 + R_{in}^2} \quad (5.14)$$

Fitting the experimental data spectrum to the equivalent model in Figure 69(b) was accomplished using the Levenberg—Marquardt algorithm [157]. Figure 70 illustrates the measured and fitted values of TEG-1 and TEG-2 using the AC measurement method.

Table 6 lists the resistance and inductance of TEG-1 and TEG-2, along with their corresponding errors. Notably, TEG-2 exhibits a significantly higher  $R_{in}$  compared to TEG-1, which results in a less pronounced phase increase with frequency for TEG-2.

Table 6: Measured resistance and inductance of TEGs and their standard deviations.

| Parameter                 | TEG-1 | TEG-2 |
|---------------------------|-------|-------|
| $R_{in} (\Omega)$         | 2.736 | 45.43 |
| $\sigma(R_{in}) (\Omega)$ | 0.107 | 0.175 |
| $L (\mu\text{H})$         | 0.293 | 0.615 |
| $\sigma(L) (\mu\text{H})$ | 0.008 | 0.121 |

## 5.5 Thermal Characterization of TEGs

The ability of TEGs to obtain temperature differences is reflected by their thermal properties. The thermal characterization of the TEGs was conducted using steady-state methods. This was achieved through two methods: using a heat flux sensor and employing a fluid-cooled heat sink (see Figure 71). Both methods were calibrated by measuring Teflon and PEEK reference samples with a known thermal conductivity and a cross-section of  $20 \times 20 \text{ mm}^2$ .

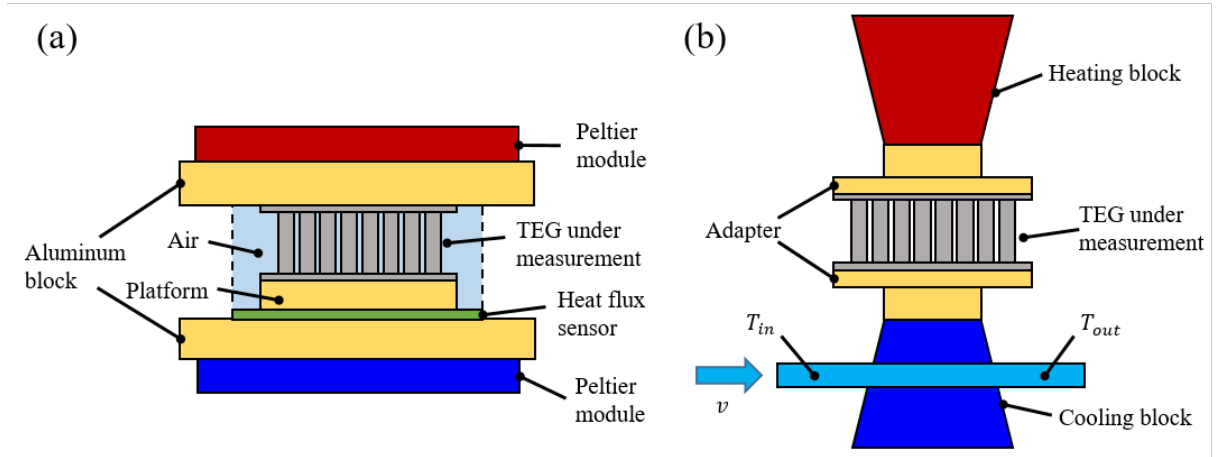


Figure 71: TEG thermal characterization. (a) Measurement setup using heat flux sensor. (b) Mini-PEM using a fluid-cooled heat sink. Reprinted from [118].

### Measurement Using a Heat Flux Sensor

This method utilized the measurement setup described in Section 5.2.2. The heat flux sensor measures the heat flux passing through its sensing area for a constant temperature difference across the TEG, as demonstrated in Figure 71(a). The aluminum platform and TEG substrates ( $\text{Al}_2\text{O}_3$  plates and copper interconnects) exhibit high thermal conductivity, so their thermal resistance is negligible. The temperature sensors were attached to

the top aluminum block and the platform. The thermal resistance between the heat flux sensor and the upper aluminum block can be approximated as:

$$R_{th} = \frac{R_{TEG}R_{air}}{R_{TEG} + R_{air}} + R_{contact} \quad (5.15)$$

where  $R_{TEG}$  is the thermal resistance of the TEG,  $R_{contact}$  denotes the contact resistance between the TEG and the measurement setup, and  $R_{air}$  denotes the thermal resistance of the air surrounding the TEG, which can be calculated by:

$$R_{air} = \frac{h_{leg}}{\kappa_{air}(A_{sense} - A_{leg})} \quad (5.16)$$

where  $h_{leg}$  is the height of the thermolegs,  $\kappa_{air}$  is the thermal conductivity of air,  $A_{sense}$  is the sensing area of the heat flux sensor, and  $A_{leg}$  is the total cross-sectional area of the thermolegs.

The measured thermal resistance within the sensing space is expressed as:

$$R_{th} = \frac{\Delta T}{qA_{sense}} \quad (5.17)$$

where  $\Delta T$  is the temperature difference across TEG measured by the temperature sensors.  $q$  is the heat flux measured by the heat flux sensor.

Neglecting  $R_{contact}$ , the conductance of measured TEG can be calculated by:

$$K_{TEG} = \frac{qA_{sense}}{\Delta T} - \frac{\kappa_{air}(A_{sense} - A_{leg})}{h_{leg}} \quad (5.18)$$

The TEG under examination was subjected to a cold side temperature of 25 °C and a hot side temperature of 35 °C. Therefore,  $\kappa_{air}$  was evaluated at 30 °C, which is 0.02588 W/m/K [158]. The sensing area  $A_{sense}$  of the heat flux sensor is 645 mm<sup>2</sup>. A stage adapter was used to adapt the cross-section of TEG-3, as illustrated in Figure 66(b). In this configuration,  $A_{sense}$  is extended to the cross-sectional area of the stage adapter (706.86 mm<sup>2</sup>).

### Measurement Using Fluid-cooled Heatsink

This method used a commercial evaluation system (Mini-PEM, ADVANCE RIKO) to measure the temperature difference for a given heat flow. A heat source and a fluid-cooled heat sink on either side established the temperature difference across the TEG specimen. The Mini-PEM system is designed to characterize devices with cross-sectional areas less than 10×10 mm<sup>2</sup>. To adapt the TEGs to the measurement system, two mounts were affixed to the heating and cooling blocks, as shown in Figure 71(b). The measurement was performed in a vacuum environment. Once the thermal system reaches a steady state,

the heat flow passing through the TEG specimen can be calculated by:

$$Q = (T_{out} - T_{in})cv \quad (5.19)$$

where  $T_{out}$  is the temperature of the outlet fluid,  $T_{in}$  is the temperature of the inlet fluid,  $c$  is the specific heat capacity of the cooling fluid, and  $v$  is the mass flow rate of the fluid. Measuring the temperature difference ( $\Delta T$ ) across the TEG through the temperature sensors in the Mini-PEM system, the thermal conductance can be obtained:

$$K_{TEG} = \frac{Q}{\Delta T} \quad (5.20)$$

## 5.6 Characterization Results

This section presents the characterization results of the TEGs using the methods described above. TEGs with different array configurations were measured to assess the robustness of the fabrication strategy outlined in Chapter 4. Additionally, a thermal cycling experiment was conducted to further demonstrate the robustness of the fabrication process. TEG-1, TEG-2, and TEG-3, with the optimum fill factor as determined by simulations in Chapter 2, were characterized and compared.

### 5.6.1 Experimental Verification

A set of TEGs featuring array configurations of  $6 \times 6$ ,  $8 \times 8$  (TEG-1), and  $10 \times 10$  was fabricated for experimental verification, as illustrated in Figure 72. These TEGs incorporate thermolegs of identical dimensions, measuring  $1 \times 1 \times 4 \text{ mm}^3$ .

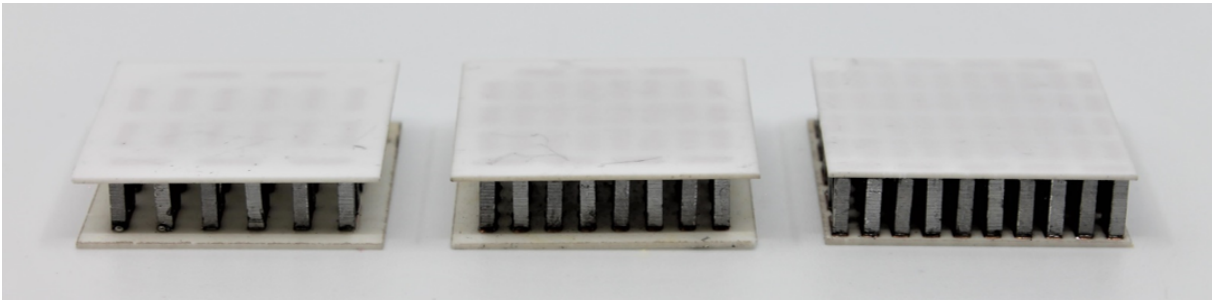


Figure 72: TEGs with different array configurations:  $6 \times 6$  (left),  $8 \times 8$  (middle), and  $10 \times 10$  (right). The dimension of thermolegs is  $1 \times 1 \times 4 \text{ mm}^3$ .

The TEGs were evaluated with a cold side temperature of  $25^\circ\text{C}$  and a hot side temperature of  $35^\circ\text{C}$ . The open-circuit voltage per temperature difference, electrical resistance, and thermal conductance of the fabricated TEGs were acquired, following the procedures outlined in Sections 5.4 and 5.5. The results were compared with data calculated from the geometry of the TE legs and the material properties of  $\text{Bi}_2\text{Te}_3$ , as presented in Table 4. These calculations considered only the effects of the TE legs and were derived

using Equations 2.6, 2.8, and 2.9. Figure 73 illustrates the measurement results for the open-circuit voltage, which shows excellent agreement with the calculated values. The open-circuit voltage exhibits a linear and positive correlation with the number of TE legs, which is consistent with the relationship described by Equation 2.6.

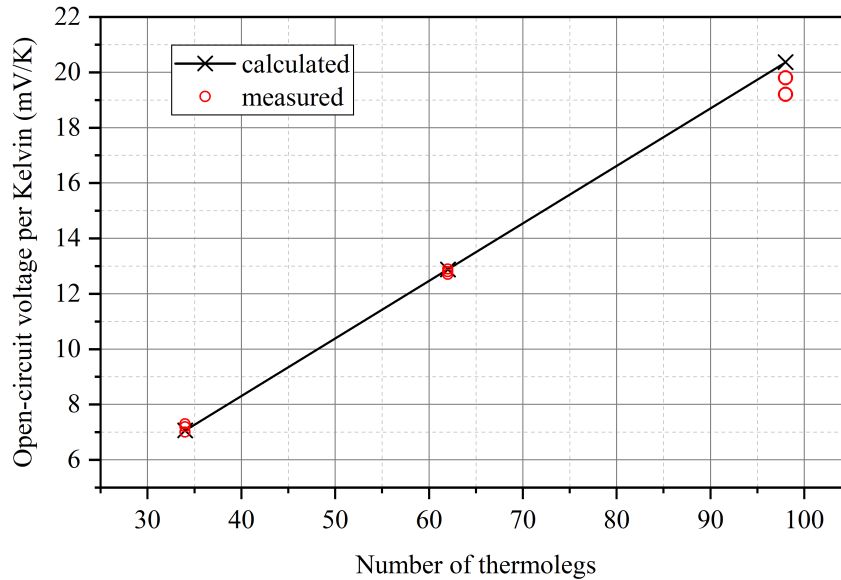


Figure 73: Open-circuit voltage of TEGs dependent on the number of thermolegs. Reprinted from [118].

Figure 74 shows the measurement results of the electrical resistance of the TEGs. Two measured results were obtained using the methods described in Section 5.4, where Data<sup>1</sup> was obtained using the temperature control method and Data<sup>2</sup> was obtained using the AC measurement method. Both measured results are linearly and positively correlated with the number of thermolegs, which is consistent with Equation 2.9. Data<sup>2</sup> matches the calculated results. However, apparent discrepancies can be observed in Data<sup>1</sup> when measuring TEGs with  $8 \times 8$ , and  $10 \times 10$  array configurations.

To analyze this error, the equivalent thermal and electrical circuits of a TEG are illustrated in Figure 75. Due to the thermal contact resistance ( $R_{contact}$ ) between the TEG and the experimental setup, the actual temperature difference across TEG ( $\Delta T$ ) is smaller than the difference between the set point temperatures ( $T_h$  and  $T_c$ ). The measurement of  $U_{oc}$  is accurate because no current passes through the TEG. However, the connection of the resistive load ( $R_L$ ) introduces current and leads to Peltier heat, and thus lower effective thermal resistance of the TEG ( $R_{th}$ ), as explained in Chapter 2.3.2. Therefore, when  $R_L$  equals the inner resistance of TEG ( $R_{in}$ ), the Peltier effect causes lower  $\Delta T$ , and the voltage reading is less than half of the open-circuit voltage. The  $R_L$  needs to be further increased until the voltage  $U_L$  equals the open-circuit voltage. This explains why the measured results in Data<sup>1</sup> are higher than the calculated results. Additionally, a TEG with more thermolegs has lower  $R_{th}$ , and thus the discrepancies caused by  $R_{contact}$  are more significant.



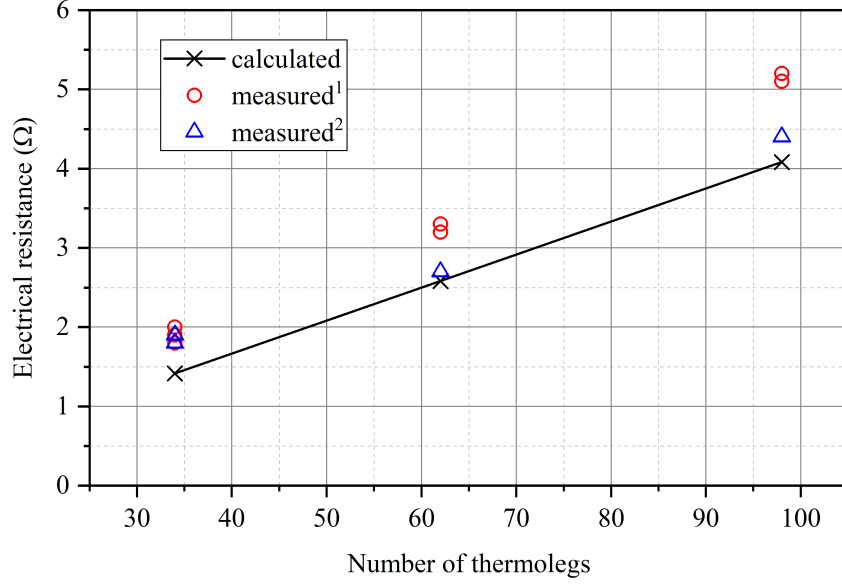


Figure 74: Electrical resistance of TEGs dependent on the number of thermolegs. Data<sup>1</sup>: temperature control method. Data<sup>2</sup>: AC measurement method. Reprinted from [118].

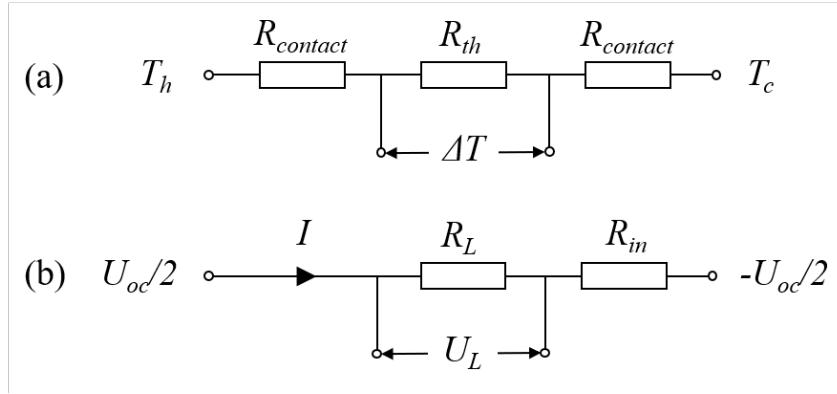


Figure 75: Equivalent circuits of a TEG under measurement using the temperature control method. (a) Thermal circuit. (b) Electrical circuit.

To assess the impact of solder joints and copper interconnects on the electrical resistance of the TEGs, a set of devices was fabricated for evaluation purposes. These devices utilize copper instead of BiTe thermolegs with dimensions of  $1 \times 1 \times 4 \text{ mm}^3$ , arranged in an  $8 \times 8$  array as shown in Figure 76. The TEG-shaped devices exhibited an average electrical resistance of  $0.07 \text{ } \Omega$ . Subtracting the electrical resistance of the copper legs ( $4.16 \text{ m}\Omega$ ), the solder joints and interconnects contributed approximately 2% of the total resistance measured in TEG-1. This suggests that the impact of the solder joints and copper interconnects on the overall electrical resistance is negligible.

However, these TEG-shaped devices are unable to reflect the electrical contact resistance between the  $\text{Bi}_2\text{Te}_3$  legs and the copper interconnects. The contact resistance describes the deviation of an actual contact from its ideal condition [159]. The reasons for this deviation are complex and can be largely attributed to the Schottky barrier formation [160,161] and surface imperfections [162]. The contact resistivity between the  $\text{Bi}_2\text{Te}_3$

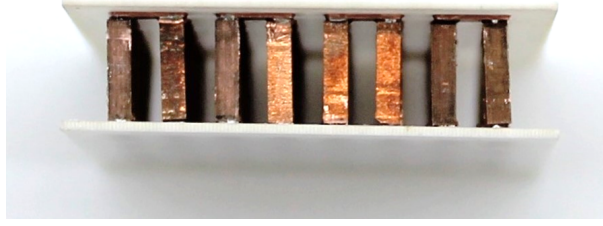


Figure 76: TEG-shaped device in TEG-1 configuration with copper legs with dimensions of  $1 \times 1 \times 4 \text{ mm}^3$ .

material and metal is about  $3 \times 10^{-10} \Omega \cdot \text{m}^2$  according to the measurement results [163]. The inner resistance of the copper-based configuration was determined to be  $1.2 \mu\Omega$ , which can be neglected compared to the inner resistance of TEG-1.

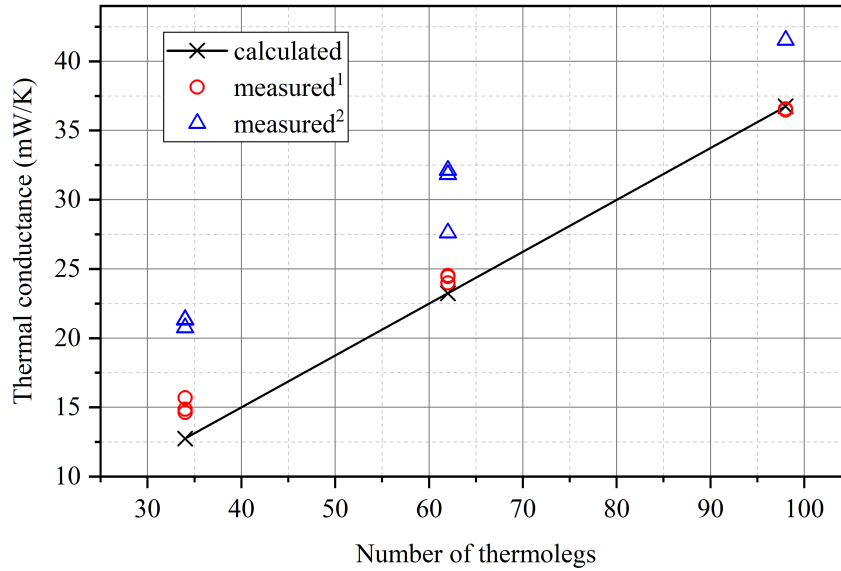


Figure 77: Thermal conductance of TEGs dependent on the number of thermolegs. Data<sup>1</sup>: measured by custom measurement setup. Data<sup>2</sup>: measured by Mini-PEM. Reprinted from [118].

Figure 77 shows the measurement results of the thermal conductance. Two result sets were obtained using the approaches described in 5.5. Data<sup>1</sup> was obtained through the heat flux sensor method and Data<sup>2</sup> was obtained through the fluid-cooled heat sink method. Both data sets are linearly and positively correlated with the number of thermolegs, as expected by Equation 2.8. Data<sup>1</sup> follows the calculated values. The apparent discrepancies observed in Data<sup>2</sup> arise from the introduction of adapters in the Mini-PEM system, which resulted in measurement errors. The adapters in the Mini-PEM system caused nonhomogeneous heat flow, with not all heat passing through the thermolegs. As a result, the measured thermal conductance of TEGs is higher than the calculated values.

### 5.6.2 Thermal Cycling

Due to the minimal temperature difference across the TEG in the implantation scenario, the thermomechanical stress within the TEG's components has a limited impact on mechanical strength. Nonetheless, temperature fluctuations can aid in detecting weaknesses in junctions, which are imperceptible to the naked eye.

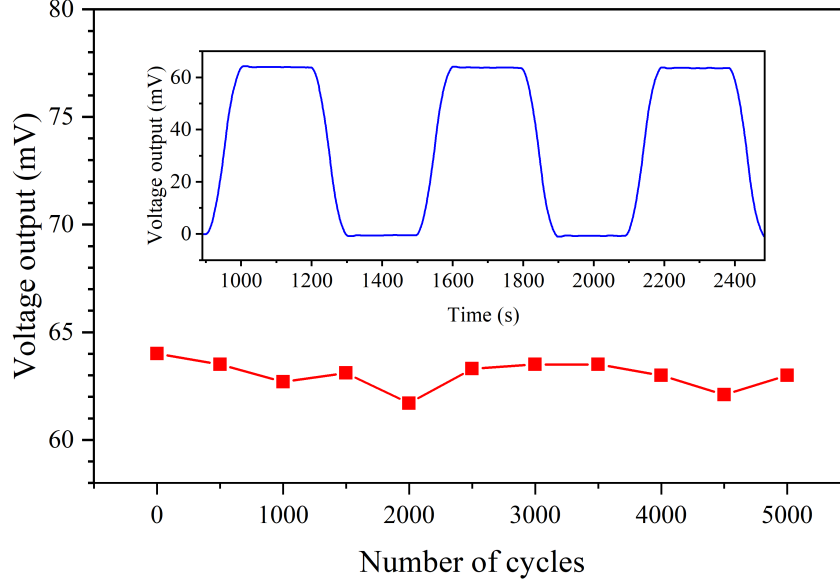


Figure 78: Thermal cycling experiment conducted on TEG-1. The temperature difference across TEG-1 fluctuates from 0 to 10 K with a cycle time of 10 minutes. Reprinted from [118].

To assess the robustness and reliability of TEG-1, a thermal cycling experiment was conducted using the experimental setup detailed in Section 5.2.1. The Peltier controller was used to apply a periodic temperature difference across TEG-1. The temperature difference ranged from 0 to 10 K, corresponding to a fluctuating temperature range of 25 to 35 °C at the hot side. The period time was chosen to be 10 minutes, with a dwell period of 200 seconds to stabilize the temperature. The ramp rate for temperature change was set at 0.1 K/s. Throughout the thermal cycling experiment, a matched resistive load was connected to TEG-1. The load voltage, as illustrated in Figure 78, remained relatively constant at 64 mV. This stable load voltage was maintained even after 5000 cycles, which is equivalent to approximately 35 days.

### 5.6.3 Comparison of Optimized TEGs

TEG-1, TEG-2, and TEG-3 were designed to exhibit identical thermal conductance, matching the surrounding tissue environment. Consequently, the simulation results in Section 2.4.3 revealed a consistent temperature difference of around 3 K across these optimized TEGs. Hence, the properties of these TEGs were measured under simulated thermal conditions, maintaining the cold side at 33 °C and the hot side at 36 °C, as detailed in Table 7.

Table 7: Characterization of TEGs with a cold side temperature of 33 °C and a hot side temperature of 36 °C. The thermal conductance is measured in open circuit condition, where Data<sup>1</sup> is obtained using the custom measurement setup and Data<sup>2</sup> is obtained using the Mini-PEM. The matched load resistance is measured using the temperature control method. The inner resistance is measured using the AC measurement method. The voltage and power output are measured with a matched resistive load.

|                                  | TEG-1                 | TEG-2                     | TEG-3                 |
|----------------------------------|-----------------------|---------------------------|-----------------------|
| Number of thermolegs             | 62                    | 254                       | 62                    |
| Dimension of thermolegs          | 1×1×4 mm <sup>3</sup> | 0.5×0.5×4 mm <sup>3</sup> | 1×1×4 mm <sup>3</sup> |
| Boost converter                  | without               | without                   | integrated            |
| Thermal conductance <sup>1</sup> | 24.3 mW/K             | 24.8 mW/K                 | 26.2 mW/K             |
| Thermal conductance <sup>2</sup> | 31.8 mW/K             | 32.7 mW/K                 | 33.0 mW/K             |
| Matched load resistance          | 3.2 Ω                 | 55 Ω                      | 113 kΩ                |
| Inner resistance                 | 2.7 Ω                 | 45.4 Ω                    | not applicable        |
| Voltage output                   | 19.9 mV               | 79 mV                     | 2.5 V                 |
| Power output                     | 117.0 μW              | 113.2 μW                  | 53 μW                 |

Due to the increased number of thermolegs, TEG-2 exhibited a voltage output that was four times higher than TEG-1. However, since the fill factor remained unchanged, the power delivered to a matched resistive load remained almost constant at 113 μW. TEG-2's inner resistance was measured to be 16 times that of TEG-1, which is consistent with Equation 2.9. The matched load resistance, measured by the temperature control method for both TEG-1 and TEG-2, exceeded the internal resistance obtained via the AC measurement method. This discrepancy is primarily attributed to inaccuracies in  $\Delta T$  across the TEG, introduced by thermal contact resistance in the temperature control method.

The integrated boost converter of TEG-3 provided a voltage of 2.5 V to the matched load. However, the conversion efficiency of the boost converter reduced the available power output to 53 μW. While the boost converter's incorporation led to a reduction in available power, it represented a beneficial trade-off given that the up-converted voltage output met the requirements of typical IMDs, such as pacemakers and neurostimulators.

TEG-3 was also measured at a temperature difference of 31 to 35 °C, as illustrated in Figure 79. The boost converter required a TEG output voltage of no less than 20 mV, which is present at a temperature difference of 1.5 K. Consequently, the boost converter delivered an open-circuit voltage ranging from 3.3 to 4.2 V, at temperature differences between 2 and 4 K. For higher temperature differences maximum power delivery is observed for smaller matched loads resistances. The conversion efficiency remained nearly constant at 45% over the temperature difference range of 2 to 4 K.

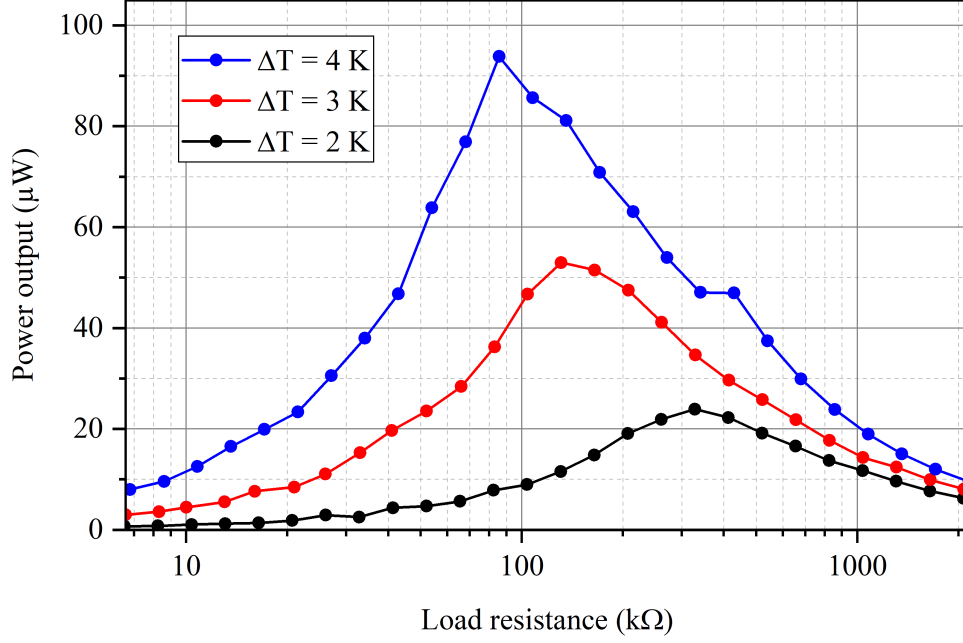


Figure 79: Power output of TEG-3 dependent on the load resistance. Reprinted from [118].

## 5.7 Summary

This chapter introduced various approaches for TEG characterization, with custom setups designed and fabricated for this purpose as well as commercially available instrumentation. TEGs with different numbers of thermolegs were fabricated and measured, revealing positive correlations between Seebeck voltage, electrical resistance, and thermal conductance with the number of thermolegs. The measured values are in good agreement with the expected values calculated from the material properties of the  $\text{Bi}_2\text{Te}_3$  thermolegs. These characterization results demonstrate the reliability and robustness of the fabrication approach outlined in Chapter 4. The applicability of the fabrication approach is exemplified by its ability to accommodate variations in array configurations without compromising TEG quality or yield. However, there are opportunities to improve the setups and measurement approaches for greater accuracy:

- The thermal contact resistances between TEG specimen and the setups impact both the thermal and electrical resistance measurements. This effect can be mitigated by measuring the thermal contact resistance using a precise pressure control system.
- Thermal characterization is sensitive to ambient temperature fluctuations. To mitigate this problem, measurements should be conducted in a vacuum environment or with the use of a guard heater to maintain consistent temperature, particularly when significant differences exist between the measurement temperature and ambient conditions.

TEG-1, TEG-2 and TEG-3 designs were optimized in Section 2.4.3 and further characterized and compared. TEG-1 and TEG-2 exhibited similar power outputs due to identical

thermal conductance, while TEG-2 provided a higher voltage output owing to its increased number of thermocouples. TEG-3 integrated a voltage boost converter and provided a significantly higher load voltage that was suitable for typical IMDs. This comes at a reduced power output due to conversion losses. Overall, the performance of these TEGs met the design's objectives and expectations.

## 6 ITEG Packaging

As described in Chapter 3, the integration of metal thermal interfaces with the PEEK housing enhances the power delivery of ITEGs. However, this packaging approach degrades hermeticity, which is crucial for implants.

This chapter examines and analyzes the PEEK-based package, especially its thermal properties, biocompatibility, and hermeticity. An equivalent moisture penetration model was developed to predict the package's lifespan. To prevent moisture ingress through PEEK, Parylene C was coated on the PEEK surface. Helium leak tests and aging tests were conducted to evaluate the package's lifetime (LT).

### 6.1 Methods Review

PEEK is increasingly used for long-term implantation in various medical applications due to its excellent properties, such as biocompatibility, pronounced mechanical strength, and high chemical resistance. It is commonly used in spine [164], orthopedics [165], trauma [166], and dental [167] applications, serving as fusion cages, bone replacements, and fixation devices. Due to its superior heat insulation compared to typical IMD packaging materials such as titanium and ceramic, PEEK significantly enhances the performance of ITEGs.

While PEEK is widely employed for implants, its use as a packaging material for IMDs is less common and requires careful consideration. Hermetic sealing is crucial for IMDs to prevent moisture, bodily fluids, and ions from penetrating the device's housing, thereby protecting the electronics from corrosion, electrical shorts, and malfunctions.

However, achieving hermeticity is challenging for polymer-based packages. Moisture ingress can occur through the polymer material itself and along the seams. Polymers are organic materials with amorphous regions, which create diffusion pathways for small gas and liquid molecules. Coating technologies such as plasma-enhanced chemical vapor deposition (PECVD) and ALD can significantly improve the hermeticity of IMDs by coating PEEK with Parylene C and alumina [168].

Research indicates that a 10 nm thick alumina film deposited by ALD can reduce the water vapor transmission rate by over three orders of magnitude [169]. Dahan et al. found that the LT of a PEEK capsule in water at 37 °C was improved by 136% after ALD coating [170].

There are three joining methods for polymers: mechanical fastening, adhesive bonding, and direct bonding [171]. Mechanical fastening methods such as clipping, clamping, screwing, or riveting do not guarantee hermetic sealing. Adhesive bonding is unsuitable for long-term implantation due to degradation and potential safety issues [172–174]. Direct bonding techniques such as laser welding, friction welding, and electromagnetic welding achieve permanent fusion by using localized heat and pressure, which results in a seam-

less bond with similar hermeticity and mechanical strength as the bulk polymer [171]. Reliable bonding between metal and polymer can also be achieved through laser welding [175]. Unlike polymer-to-polymer laser welding, polymer-to-metal bonding melts only the polymer. The melted polymer surface forms tiny bubbles or spikes that anchor into the metal surface, thereby improving the strength of the joint [176, 177].

## 6.2 Housing Fabrication

The packaged TEG described in Section 2.4.3 was fabricated. The dimensions of the housing design are detailed in Figure 80. The lid and sidewalls were made of PEEK. A feedthrough was incorporated into the sidewall to allow wires to exit. The heat conductors were made of brass, an alloy primarily composed of copper and zinc. Over time, zinc can leach out into the body, which can potentially cause adverse reactions or toxicity. To ensure biocompatibility, the surfaces of the heat conductors were coated with a thin layer of gold ( $30\text{ }\mu\text{m}$ ).

Each part of the housing includes a connection structure to increase the bonding area between the components. Epoxy was used as the adhesive for these connections. TEG-1 was integrated inside the housing, with its surfaces attached to the heat conductors using a silicone-based adhesive (Keratherm-Bond 100 RT, KERAFOIL), which has a thermal conductivity of  $1.5\text{ W/m/K}$ . Fabrication tolerances were maintained within  $0.5\text{ mm}$ .

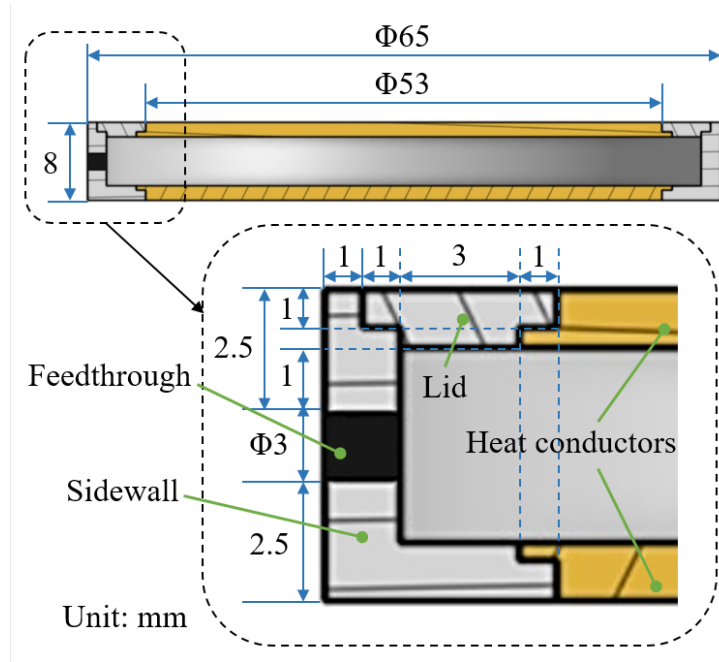


Figure 80: Dimension drawing of the housing in midplane section view. The housing incorporates a lid and a sidewall made of PEEK, and two heat conductors made of gold-coated brass. A feedthrough on the sidewall is prepared for the wire connection.



### 6.3 Thermal Properties of the Packaged TEG

The thermal properties of the packaged TEG are critical for the ITEG's performance. This section measures the overall thermal resistance and evaluates heat transfer within an agar section. Due to the small temperature difference across the TEG, mechanical stress from thermal expansion is negligible. Furthermore, as shown in Chapter 3, Joule heating from TEG operation minimally impacts the thermal distribution within the human body.

#### 6.3.1 Thermal Resistance Measurement

The measurement setup (Section 5.2.2) was adapted to measure the thermal resistance of the packaged TEG. Due to the larger cross-sectional area of the packaged TEG ( $3318 \text{ mm}^2$ ) compared to the previous heat flux sensor's sensing area ( $645 \text{ mm}^2$ ), 80 mm diameter copper adapters with thermal pads were added to the alumina blocks for accurate measurement. The heat flux sensor was replaced by a version with a larger sensing area ( $88 \times 95 \text{ mm}^2$ , PHFS-09e, FluxTeq). The adapted measurement setup was calibrated using a PEEK disk with a diameter of 80 mm and a height of 8 mm.

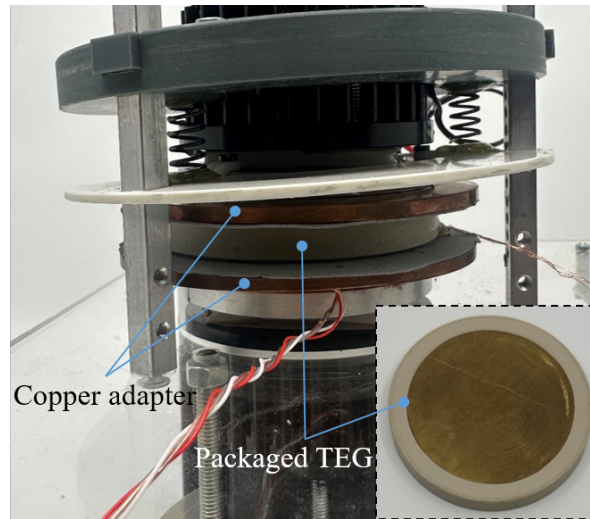


Figure 81: Adapted measurement setup for thermal resistance measurement of the packaged TEG. The specimen is clamped with two copper adapters measuring 80 mm.

The simulated thermal conductance of the packaged TEG was  $54.6 \text{ mW/K}$ , while the experimental measurement was slightly lower ( $48.5 \text{ mW/K}$ ). This discrepancy is attributed to imperfect contact between the packaged TEG and the adapters, which is likely caused by slight unevenness in the surface of the device and the use of epoxy glue. Despite polishing and the use of thermal pads, contact resistance was not fully eliminated.

### 6.3.2 ITEG Embedded within Agar Tissue Model

To evaluate the heat transfer of the packaged TEG embedded in the tissue, the simplified tissue geometry described in Section 2.3.1 was replicated using a 3% water-agar gel (see Figure 82), with dimensions of  $100 \times 100 \times 45 \text{ mm}^3$ . Agar, a seaweed extract, is a common medical substitute for tissue due to its similar properties [178]. The measured thermal conductivity ( $0.54 \text{ W/m/K}$ ) of the used agar is comparable to that of muscle, fat, and skin [98].

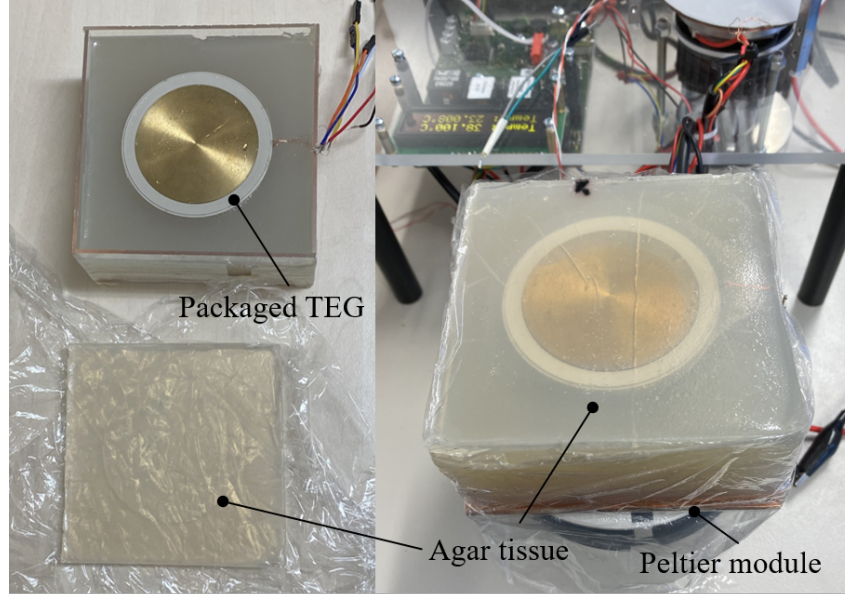


Figure 82: Thermal performance evaluation of the packaged TEG embedded within the agar tissue model. The packaged TEG was embedded into the agar tissue (left-hand panel). During the measurement, a 2 mm thick agar block was placed on the top of ITEG and the sides of the agar block were wrapped in cling film (right-hand panel).

Two temperature sensors were attached to the inner surfaces of the heat conductors within the packaged TEG, which was then embedded in the agar 2 mm below its top surface. Four acrylic plates were used to shape the agar tissue. A Peltier module (TEC1-12706) heated the copper plate upon which the agar was mounted. A Peltier controller (TEC-1122-SV, Meerstetter Engineering) maintained the temperature of the copper plate at  $37.7^\circ\text{C}$ . To minimize convection on the sides, the model's sides faces were wrapped in cling film after removing the acrylic plates. Initial measurements were taken with an air-filled package at an ambient temperature of  $23^\circ\text{C}$ .

To simulate this agar tissue setup, the simplified tissue model was adapted by replacing human tissue regions with agar. The bottom surface temperature of the muscle section was set to  $37.7^\circ\text{C}$ . By matching the simulated and measured temperatures on the packaged TEG surfaces, the convection coefficient applied to the top surface of the agar was determined to be  $17.66 \text{ W/m}^2/\text{K}$ .

The packaged TEG was filled with krypton (4.0 MINICAN, Linde) and re-embedded within the agar. The temperature was measured under the same environmental conditions,

Table 8: Simulated and measured temperatures of the ITEGs embedded in the agar tissue. The filler materials in the ITEGs are air and krypton, respectively. The ambient temperature is 23 °C. The temperature applied to the bottom surface of the agar is 37.7 °C. The convection coefficient applied to the top surface of the agar is 17.66 W/m<sup>2</sup>/K.

| Filler material | Experiment                     |                             |                              | Simulation                     |                             |                              |
|-----------------|--------------------------------|-----------------------------|------------------------------|--------------------------------|-----------------------------|------------------------------|
|                 | $T_{bottom}(^{\circ}\text{C})$ | $T_{top}(^{\circ}\text{C})$ | $\Delta T(^{\circ}\text{C})$ | $T_{bottom}(^{\circ}\text{C})$ | $T_{top}(^{\circ}\text{C})$ | $\Delta T(^{\circ}\text{C})$ |
| Air             | 31.217                         | 26.590                      | 4.627                        | 31.178                         | 26.594                      | 4.584                        |
| Krypton         | 31.419                         | 26.280                      | 5.139                        | 31.371                         | 26.335                      | 5.036                        |

as presented in Table 8. The measured temperatures aligned well with the simulations when filling the packaged TEG with krypton. The temperature difference across the TEG was approximately 0.5 K higher when compared to the TEG filled with air due to krypton’s lower thermal conductivity (0.00943 W/m/K).

## 6.4 Hermeticity of Packaged TEG

The hermeticity of the packaged TEG significantly impacts its safety, reliability, and longevity. To assess hermeticity, an equivalent moisture permeation model was developed to predict moisture ingress. The PEEK housing was coated with Parylene C to enhance hermeticity, and both helium leak and aging tests were performed on the packaged TEG.

Importantly, the thermal conductivity of air demonstrates negligible variation around human body temperature, even with changes in humidity [179]. Thus, any potential moisture ingress will not affect the TEG’s thermoelectric power generation significantly.

### 6.4.1 Biocompatibility

Biocompatibility of implants refers to their ability to function within the human body without causing harm, essentially making the implant seem like an original part of the body. To achieve this, implants should be designed to be nontoxic, nonreactive, non-immunogenic, non-thrombogenic, noncarcinogenic, and nonirritant [180].

The biocompatibility of PEEK has been proven through in vivo experiments [181]. The thermal conductors used in the ITEG package were gold-coated brass. Although gold is biocompatible, the integrity of the gold coating could potentially be compromised over the long term due to wear and tear within the body, which would increase the risk of zinc leaching from the brass. To mitigate this risk, biocompatible metals such as gold, silver, platinum, and tantalum can be used as heat conductors without impacting the TEG performance due to their high thermal conductivity.

Biocompatibility concerns extend beyond the package to the entire implant, which is also related to the hermeticity of the implant. The thermoelements in the TEG are made of Bi<sub>2</sub>Te<sub>3</sub>, which is generally considered to have low toxicity [182]. The solder (Sn<sub>42</sub>Bi<sub>57</sub>Ag<sub>1</sub>)

used for TEG connections is lead-free and nontoxic. Additionally, the materials used in the electronics (e.g., metals, plastics) may not be biocompatible, and the interaction between the electronics and bodily fluids could cause chemical reactions or toxicity in the tissues.

General requirements for IMDs, including biocompatibility, are delineated in directives and standards such as 90/385/EEC [183], EN 45502-1 [184], and ISO-14708 [185].

### 6.4.2 Equivalent Moisture Permeation Model

The moisture permeation through polymeric package is governed by Fick's second law [186]:

$$\frac{dC}{dt} = \nabla(D\nabla C) \quad (6.1)$$

where  $C$  is the concentration of the moisture,  $t$  is time, and  $D$  is the diffusion coefficient.

Fick's first law describes the relationship between the flux of diffusing substance  $J$  and the concentration gradient  $\nabla C$ :

$$J = -D\nabla C \quad (6.2)$$

A practically equivalent quasi-steady-state (QSS) model for a cavity package with identical material and thickness was developed by Tencer [187], giving a solution for Fick's law equations:

$$RH_t = RH_i + (RH_a - RH_i)(1 - e^{-\frac{t}{\tau}}) \quad (6.3)$$

where  $RH_t$  is the relative humidity (RH) at time  $t$  in the cavity,  $RH_i$  is the initial RH in the cavity,  $RH_a$  is the ambient RH (when the package is immersed in water the  $RH_a$  is 100%). RH describes the ratio of the absolute moisture concentration  $C$  over the saturation moisture concentration at the specific temperature.  $V$  is the cavity volume of the package,  $d$  is the thickness of the package wall, and  $A$  is the surface area of the package.  $\tau$  is the time constant, which is defined by:

$$\tau = \frac{Vd}{SDA} + \frac{d^2}{2D} \quad (6.4)$$

$S$  is the solubility and represents the amount of water per unit of volume  $V_{water}$  that can be absorbed and stored by the polymetric package per unit of volume  $V_{polymer}$ :

$$S = \frac{\rho \Delta m V_{mol}}{m_0 m_{mol}} \quad (6.5)$$

where  $\rho$  is the mass density of the polymeric package,  $\Delta m$  is the mass gain of the package,  $V_{mol}$  is the molar volume of the water,  $m_0$  is the initial mass of the package, and  $m_{mol}$  is the molar mass of the water.

Dahan et al. proposed an approach to make Tencer's QSS model applicable for a cavity package made of different thicknesses and materials [131]. This approach analogized the

diffusion model as a circuit consisting of resistors and capacitors, as shown in Figure 83. In this circuit, the voltage  $H_e$  is equivalent to the external RH and the current  $i$  is equivalent to the diffusion rate of water. The resistance  $R_p$  represents the resistance of the package to the moisture ingress. The capacitors  $C_v$  and  $C_p$  are the ability of the internal space and the package to store moisture in volume.  $H$  is the RH in the internal space.

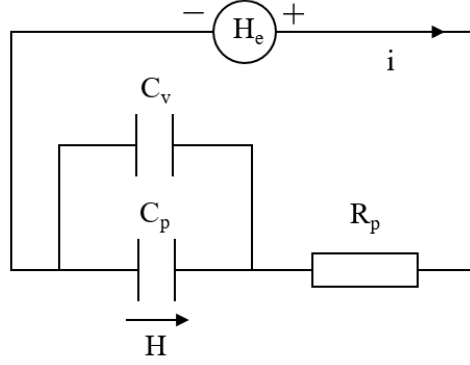


Figure 83: Equivalent moisture permeation model.  $H_e$  is the external RH,  $i$  is the diffusion rate of water,  $R_p$  represents the resistance of the package to the moisture ingress,  $C_v$  and  $C_p$  are the ability of the internal space and the package to store moisture in volume.

The balance of RH can be described by:

$$H_e = iR_p + H \quad (6.6)$$

In quasi-steady-state, Equation 6.6 can be written as:

$$\frac{dH_e}{dt} = R_p \frac{di}{dt} + \frac{dH}{dt} = 0 \quad (6.7)$$

Since

$$i = (C_v + C_p) \frac{dH}{dt} \quad (6.8)$$

$$\frac{di}{dt} + \frac{1}{R_p(C_v + C_p)} i = 0 \quad (6.9)$$

When  $t=0$ ,

$$i = \frac{H_e - H_0}{R_p} \quad (6.10)$$

where  $H_0$  is the existing amount of moisture when the package is sealed (initial RH), the solution of Equation 6.9 is given as:

$$H = H_e - (H_e - H_0) R_p e^{-\frac{t}{\tau}} \quad (6.11)$$

$$\tau = R_p(C_v + C_p) \quad (6.12)$$

Associating Equation 6.4 and 6.12,  $R_p$  and  $C_p$  are expressed as:

$$R_p = \frac{d}{SDA} \quad (6.13)$$

$$C_p = \frac{ASd}{2} \quad (6.14)$$

Since the PEEK package described in Section 6.2 used different materials with different thicknesses, the calculation of  $R_p$  and  $C_p$  can be divided into three parts:

1. PEEK on the top and bottom.
2. PEEK on the sides.
3. Epoxy between the joints.

The moisture will not penetrate the thermal conductors. The measured and calculated parameters in the equivalent moisture permeation model at 32 °C are listed in Table 9.

Table 9: Parameters of the equivalent moisture permeation model of the packaged TEG at 32 °C.

| Parameter                                          | Top & bottom <sup>1</sup>  | Sides <sup>2</sup>         | Joints <sup>3</sup>      |
|----------------------------------------------------|----------------------------|----------------------------|--------------------------|
| Material                                           | PEEK                       | PEEK                       | Epoxy                    |
| Thickness $d$ (cm)                                 | 0.16                       | 0.205                      | 0.26                     |
| Surface area $A$ (cm <sup>2</sup> )                | 10.93                      | 10.54                      | 0.01                     |
| Solubility $S$ (cm <sup>3</sup> /cm <sup>3</sup> ) | 9.17                       | 9.17                       | 19.2                     |
| Diffusion $D$ (cm <sup>2</sup> /s)                 | $8.2 \times 10^{-9}$ [188] | $8.2 \times 10^{-9}$ [188] | $7 \times 10^{-7}$ [189] |
| $R_p$ (s/cm <sup>3</sup> )                         | $1.95 \times 10^5$         | $2.59 \times 10^5$         | $1.93 \times 10^6$       |
| $C_p$ (cm <sup>3</sup> )                           | 8.018                      | 9.907                      | 0.025                    |

$R_{p1}$ ,  $R_{p2}$ , and  $R_{p3}$  are connected in parallel and their equivalent resistance  $R_p$  is calculated as:

$$R_p = \frac{1}{\frac{1}{R_{p1}} + \frac{1}{R_{p2}} + \frac{1}{R_{p3}}} \quad (6.15)$$

$C_{p1}$ ,  $C_{p2}$ , and  $C_{p3}$  are connected in parallel and their equivalent capacitance  $R_p$  is calculated as:

$$C_p = C_{p1} + C_{p2} + C_{p3} \quad (6.16)$$

$C_v$  equals the cavity volume, which is 12 cm<sup>3</sup>. According to Equation 6.12,  $\tau$  is calculated as 877.5 h at 32 °C.

### 6.4.3 Surface Coating for Extended Lifetime

Parylene C coatings are widely used in biomedical applications due to their improved anticorrosive, biocompatible, anti-infection, and therapeutic functions [190]. Many studies

have shown that Parylene C is effective in preventing water infiltration, thereby extending the life of the equipment [191,192].

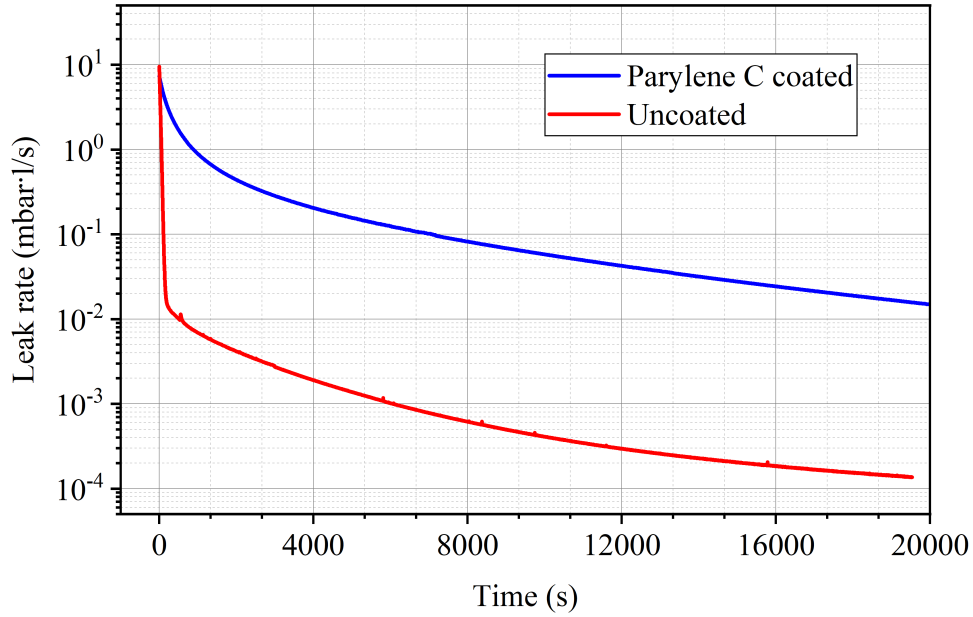


Figure 84: Helium leak test on an uncoated PEEK housing and Parylene C coated PEEK housing. Reprinted from [118].

To enhance the hermeticity of the housing described in Figure 80, the PEEK lid and sidewall were coated with Parylene C with a thickness of 20  $\mu\text{m}$ . The coating service was provided by Plasma Parylene Systems GmbH. A helium leak test was conducted on the housing using a helium leak detector (HLT-560, Pfeiffer). For comparison, an uncoated PEEK housing was used as a control. The housing was filled with helium in a glove bag and then sealed using epoxy in the glove bag. The housing was then placed into a vacuum chamber connected to the helium leak detector. The measurement result is presented in Figure 84. In the beginning, the helium that remained on the housing's outer surface was detected. This explains why the helium leak rate of both housings was 10 mbar·l/s. After the remaining helium was absorbed by the detector, the leak rate of helium from the inner housing was detected. The helium leak rate of Parylene C coated housing is roughly 120 times lower compared to the uncoated PEEK housing.

Although the Parylene C coating significantly improved the hermeticity of the PEEK housing, it is still far from being considered as hermetic. According to the standard for hermetic package qualification, MIL-STD-883H [193], the helium leak rate should be five orders of magnitude lower to be regarded as hermetic.

#### 6.4.4 Moisture Penetration Experiment

The previous section evaluated the hermeticity of PEEK housing using the helium leak test. However, helium atoms are much smaller than water molecules, which makes it easier for them to pass through the molecular gaps and voids in the PEEK polymer structure.

Therefore, a moisture penetration experiment was conducted to evaluate the long-term hermeticity of the PEEK housing against moisture ingress.

This test has three samples: an uncoated PEEK housing, a 20  $\mu\text{m}$  thick Parylene C coated PEEK housing, and a brass housing (see Figure 85). The brass housing replaced the PEEK sidewall and lid with brass components, focusing on the hermeticity evaluation of the joints. The joints of both housings were attached by epoxy. The wires were stretched out through the feedthrough on the sidewall and were wrapped by a silicone rubber tube. To prevent moisture ingress from the feedthrough, the tube was fixed to the sidewall using epoxy and aluminium butyl tape.

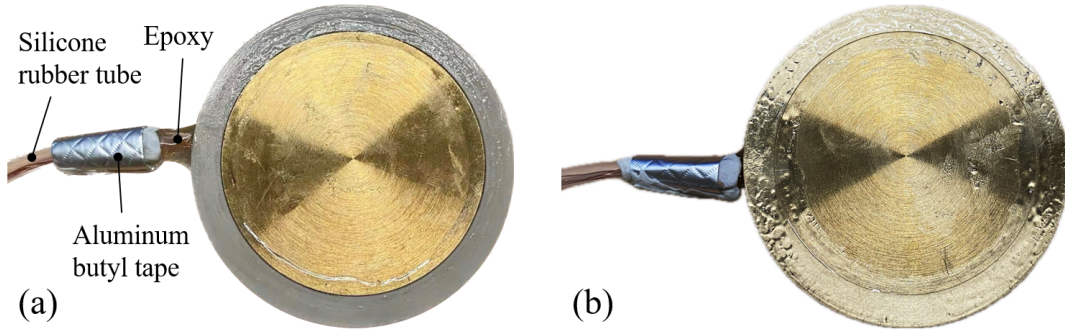


Figure 85: (a) PEEK housing with wires stretched out through the feedthrough on the sidewall. (b) Brass housing as a control sample. Figure (a) is reprinted from [118].

Humidity sensors (SHT31-DIS, Sensirion) were integrated into each housing to measure the RH inside the housing. A single-board computer (Raspberry Pi 4 model B) was used to communicate with the sensor to collect the data.

The experiment was conducted in a water tank, as shown in Figure 86. A temperature-controlled heater was submerged in the water and attached to the sidewall of the water tank. The housing samples were immersed in the water tank. To accelerate the process of moisture ingress and save time for the test, the water temperature was controlled at 53  $^{\circ}\text{C}$ .

The test results are depicted in Figure 87. The dashed lines are the fitted results using Equation 6.3. The equivalent moisture permeation model described in Figure 83 was solved assuming that the moisture passes completely through the package at the beginning. Therefore, the fitting ignored the initial phase of the experimental data, where the moisture was absorbed by the PEEK or epoxy, and thus did not enter the cavity.

Although the fitted values of the uncoated PEEK and brass housings match well with their experimental data, the fitted values do not align well with the experimental data of the Parylene C-coated housing. This discrepancy might be attributed to potential moisture leakages through the silicone rubber tube from the cavity to the outside environment. The fitted  $\tau$  of each housing at 53  $^{\circ}\text{C}$  is listed in Table 10.

The ASTM F1980 standard [194] describes how to calculate the accelerated aging of medical device packages. This standard uses the Q10 rule [195], which is an empirical



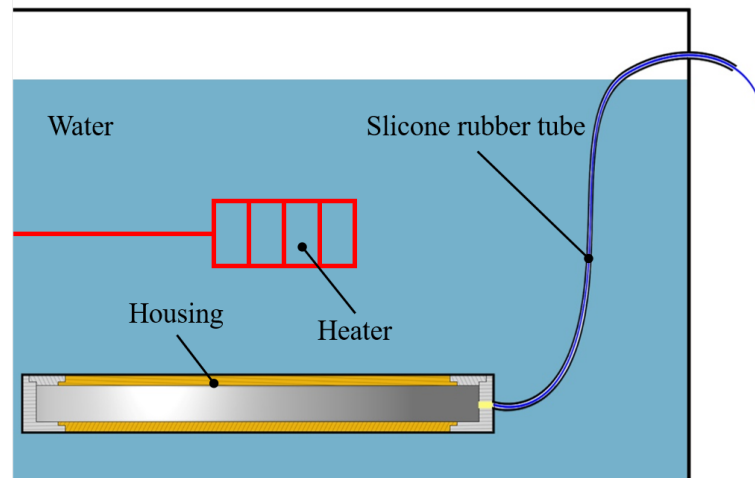


Figure 86: Moisture penetration experiment conducted in a water tank. A heater is used to control the water temperature. Wires are wrapped by a silicone rubber tube and connected to a single-board computer to collect measurement data.

observation that describes the relationship between temperature and the rate of chemical reactions in polymers. It states that for every 10 K increase in temperature, the rate of a polymer degradation reaction roughly doubles. An accelerated aging factor (AAF) between the LTs at different temperatures is calculated as follows:

$$AAF = \frac{LT_2}{LT_1} = Q_{10}^{\frac{T_1 - T_2}{10}} \quad (6.17)$$

Here, the LT of the housing is defined as the time required for the RH inside the housing to reach 85%, with an initial RH of 10%. As described in Section 6.4.2, the  $\tau$  is calculated as 877.5 h at 32 °C, which was verified experimentally in the water tank, with an accuracy of 1.5%. Then, the LT of the PEEK housing at 32 °C is calculated as 1572.3 h using Equation 6.3. The LT of the PEEK housing at 53 °C is calculated as 89.8 h.

Then,  $Q_{10}$  is calculated as 3 using Equation 6.17. By substituting  $Q_{10} = 3$  in Equation 6.17, the LT of the housings at 53 °C can be transferred to the LT at 35 °C (average temperature of ITEG in the human body), as presented in Table 10.

Table 10: The fitted time constant ( $\tau$ ) and calculated LTs of uncoated PEEK housing, Parylene C coated PEEK housing and brass housing.

| Sample     | $\tau$ , 53 °C (hours) | LT, 53 °C (hours) | LT, 35 °C (hours) |
|------------|------------------------|-------------------|-------------------|
| PEEK       | 50.1                   | 89.8              | 648.8             |
| Parylene C | 86.1                   | 154.3             | 1114.8            |
| Brass      | 175.6                  | 314.6             | 2272.9            |

The results indicate that the Parylene C coating prolonged the LT of PEEK housing by 72%. Nevertheless, this improvement falls short of the desired standard. The LT of the Parylene C-coated PEEK housing is only 22 days at 35 °C. From the calculated LT of the

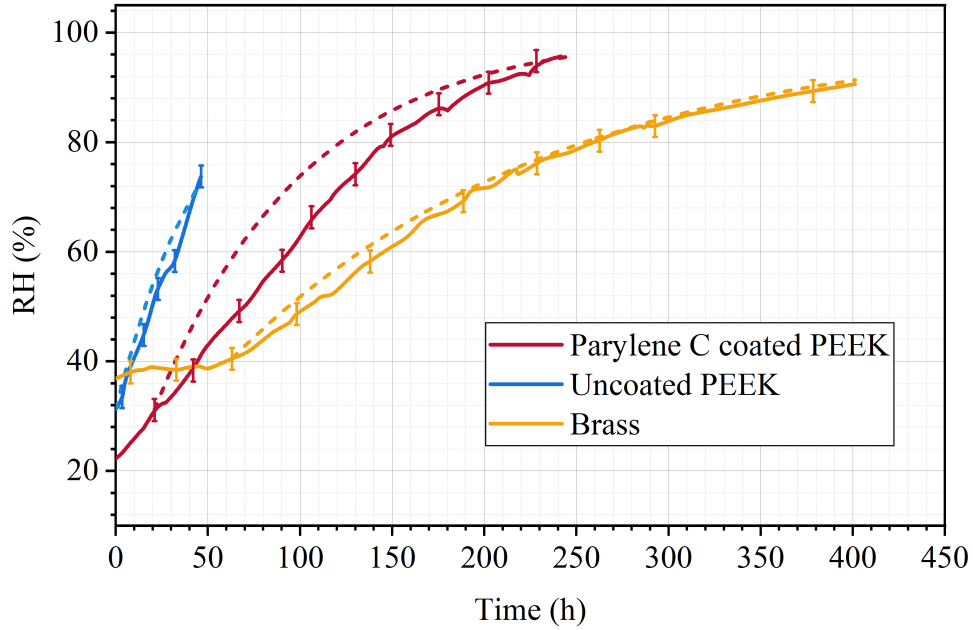


Figure 87: RH of the inner space of the Parylene C coated PEEK, uncoated PEEK, and brass housings immersed in water at 53 °C. Solid line: measured data. Dotted line: calculated values from the fitted time constant ( $\tau$ ). Reprinted from [118].

control (i.e., brass housing as 46 days), it is obvious that moisture ingress predominantly occurs through the epoxy connections because brass is hermetic.

Although the incorporation of a PEEK housing and metal plates leads to a substantial improvement in the TEG's performance, it also introduces a significant challenge concerning moisture ingress. Notably, the thermal conductivity of air with varying humidity levels exhibits only negligible deviations around human body temperature [179]. This implies that moisture ingress is unlikely to exert a significant impact on the power generated by the TEG. To mitigate potential electronic failures stemming from moisture infiltration, we propose the application of Parylene C coating on electronic components. Experimental findings have substantiated the efficacy of Parylene in safeguarding silicon chips, with a demonstrated longevity of Parylene-coated metal at 37 °C exceeding 60 years, as reported in [196]. However, it is crucial to remain vigilant regarding safety concerns arising from interactions between the BiTe material and the implantation environment. This concern can be addressed in two directions. First, efforts should focus on developing nontoxic TE materials that offer comparable performance to BiTe at human body temperature. Second, future research endeavors should prioritize enhancing the PEEK-to-metal seal and polymer encapsulation mechanisms to further fortify device safety. Promisingly, PEEK-to-metal joints can be significantly enhanced by laser welding [175]. In addition, more effective hermetic coating methods, such as ALD, show significant potential in polymer encapsulation [170].

## 6.5 Summary

This chapter discussed and studied the thermal properties, biocompatibility, and hermeticity of the PEEK-metal housing. Compared to conventional packaging materials such as titanium or  $\text{ZrO}_2$  ceramic, PEEK has much lower thermal conductivity. This causes a higher temperature difference across the TEG, and thus higher power output. The packaged TEG was fabricated and characterized thermally via the thermal resistance measurement and an agar tissue experiment that evaluated the heat transfer characteristics of a TEG embedded within a human body.

As a porous polymer, PEEK is not hermetic. This leads to concerns about the reliability and safety of the implant. Therefore, to predict the LT of the PEEK housing in the implantation environment, an equivalent moisture permeation model was developed by analogizing an electric circuit. Parylene C was coated on the PEEK surface to prolong the LT of the implant. A helium leak test was conducted on both uncoated and Parylene C coated PEEK housings, which showed that the helium leak rate of Parylene C coated housing is roughly 120 times lower than that of the uncoated PEEK housing. However, the helium leak test is limited in the long-term evaluation of hermeticity against moisture. Therefore, an aging test was conducted in a water tank to evaluate the long-term performance of the housing against moisture ingress.

The experimental results showed that Parylene C coating increases the time constant by 72%, which shows the potential of Parylene C coating to prevent moisture ingress. However, most of the moisture penetrated the housing through the joints, which makes the moisture ingress much faster than the hermetic package and raises the safety and reliability issues.

Future work should focus on improving the hermeticity of the joints using welding technologies and on the hermeticity of the PEEK using surface coating technologies. Otherwise, a glass-metal housing could be used as an alternative because it sacrifices an amount of power but guarantees hermeticity.

## 7 Summary and Outlook

### 7.1 Summary

This thesis has presented the design and fabrication of ITEGs for powering IMDs. An initial design of an ITEG was proposed, which incorporates a TEG and a housing for encapsulation. The housing consists of PEEK sidewalls and metal heat conductors, which establish thermal contact between the TEG and the tissue. This design focuses on functionality, performance, biocompatibility, and hermeticity.

- **Functionality:** The primary purpose of ITEGs is to power IMDs. Simply considering the TEG within the human body is not sufficient due to the complexities, space constraints, and safety concerns associated with connecting the TEG to IMDs. The housing can encapsulate both the TEG and the electronic components of the IMDs, which allows these components to function with reduced battery size or even achieve battery-free operation with the aid of the TEG.
- **Performance:** An unpackaged TEG provides limited power when embedded in the human body. The housing enhances performance in two ways. First, the PEEK side wall has a lower thermal conductivity than traditional materials such as titanium or ceramic, which reduces thermal shunting and enhances TEG performance. Second, the metal heat conductors improve thermal contact with the tissue, which directs more heat through the TEG.
- **Biocompatibility:** The housing prevents direct contact between the tissue and the TEG or electronics, which ensures biocompatibility. Additionally, both the PEEK and gold-coated metal used in the housing are biocompatible.
- **Hermeticity:** The hermetic housing prevents body fluid ingress, which could cause electronic failure and degrade TEG performance due to corrosion, which ensures long-term operation.

To estimate the performance of the ITEG design in an implantation scenario, a simplified tissue simulation model featuring skin, fat, and muscle layers was developed. The model incorporates the effects of metabolism and blood perfusion in each tissue layer and considers convection, radiation, and evaporation (sweating) on the skin surface. To enhance computational efficiency and facilitate parametric design and analysis, the TEG model was simplified by using a representative thermopile instead of detailed thermocouple legs. Changes in TEG design parameters, such as fill factor and the number of thermolegs, were reflected in the material properties of the representative thermopile. The interaction between heat and electricity in the TEG model was decoupled using an

approximation, where the Peltier heat was treated as part of heat conduction. This decoupling enables accurate simulation using solely thermal analysis and paves the way to apply MOR to further reduce the model's complexity.

While the straightforward structure and geometry of the simplified tissue model expedited computation time, it did so at the expense of accuracy. To predict the ITEG's performance in a more realistic biological environment, detailed human torso and head models were developed. These detailed human body models incorporate a variety of tissue and organ types with realistic anatomical geometries and physiological properties. For the first time, such detailed models were used to predict the performance and for parametric analysis of ITEGs, yielding significant and inspiring simulation results.

According to the simulation results, the performance of the ITEG is largely dependent on the thermal management of both the TEG and housing design. For the housing design, several factors enhance the power output of ITEGs, including reduced implantation depth, larger housing dimensions, thinner side walls, and filler materials with lower thermal conductivity. However, larger housing dimensions contradict the restricted space that is available for implantation use. A PEEK housing offers better thermal insulation but is less hermetic when compared to conventional packaging materials such as titanium or ceramic. Gas fillers such as air and krypton provide superior thermal insulation compared to solid fillers such as polymers but they increase the fabrication difficulty for free-standing thermoleg assembly. Therefore, trade-offs between power, miniaturization, hermeticity, and manufacturability must be carefully considered.

For a TEG design within a specific housing, the fill factor ( $A_{thermopile}/A_{ITEG}$ ) should be optimized for maximum power output. Increasing the height of the thermolegs improves power output. For a given fill factor, the number of thermolegs does not affect power output but it does alter voltage output. Increasing the number of thermolegs boosts voltage output but it also increases the aspect ratio of the thermolegs, which makes manufacturing more challenging. Notably, changes in the housing parameters necessitate re-optimization of the TEG's fill factor due to the mismatch between the thermal resistance of the TEG and its surrounding environment (i.e., housing and human tissue).

Understanding the influence of the design parameters on the ITEG's power output allows for the identification of the optimal implant locations in the torso and head. Ideally, implant locations near the skin surface are preferred to enhance TEG performance. In the human torso, ITEGs are suggested for implantation in fat sections with greater thickness (e.g., chest, abdomen, and lumbar), which can accommodate ITEGs with increased height, thereby enhancing power output. With the ambient temperature set to 21°C and the convective film coefficient set to 8.58 W/m<sup>2</sup>/K, the optimized TEG, housed in a PEEK enclosure with a 5 cm diameter (similar in size to a typical pacemaker), can generate 47 µW and 118 µW when embedded in fat layers of 6.2 mm and 14 mm thickness, respectively.

In the human head, where the fat layer is thinner than in the torso, the cranial bone is identified as a potential implant location for effective power generation. The proximity

to both the skin surface and the brain region allows the ITEG to conduct more heat. An optimized ITEG with a 30 mm diameter yields a power output of 66  $\mu\text{W}$ . Further enhancement is achieved by employing a 50 mm diameter heat spreader to cover the upper skull surface, which would boost power output to 91  $\mu\text{W}$ .

According to the simulation results, ITEGs should feature numerous high-aspect-ratio thermolegs within the optimized fill factor for maximum power and sufficient voltage. To fabricate such TEGs, a novel approach using a removable assembly rig was proposed. This rig enabled the fabrication of a TEG with up to 254 thermolegs, each with dimensions of down to  $0.5 \times 0.5 \times 4 \text{ mm}^3$ , in a single soldering process. However, achieving the required voltage for IMDs (typically 1 to 4 V) necessitates a much larger number of thermolegs with a higher aspect ratio, which would pose significant manufacturing challenges. To address this issue, a compact TEG was fabricated with an integrated boost converter for voltage up-conversion. The robustness and reliability of this fabrication approach were validated through the characterization of TEGs with different array configurations, using custom experimental and measurement setups.

The PEEK-metal housing was then fabricated and characterized. The hermeticity tests revealed that moisture ingress occurred at the PEEK and PEEK-metal joints, with the majority occurring at the PEEK-metal joints. Applying a Parylene C coating to the PEEK effectively increased the time constant by 72%, which indicates its potential to enhance hermeticity.

## 7.2 Outlook

In this work, the ITEG simulations were conducted by assuming a bare human body in a stationary thermal state with fixed ambient conditions. Future simulation work should consider more factors in real-world scenarios, such as the effects of body hair and clothing, which may decrease the predicted performance. Additionally, the influence of varied ambient conditions (e.g. temperature, humidity, and air velocity) on the ITEG's performance needs to be further studied. Designing a power management system within an ITEG requires simulations alongside transient thermal analysis. This analysis would account for the fluctuations in human body temperature that occur throughout a year, season, day, or even with a single breath. The body's thermoregulatory responses, such as shivering, vasoconstriction, vasodilation, respiration, and circadian rhythm, should be considered in transient analysis. Understanding how compression and stretching of tissues during movement might impact the ITEG's performance is also important. While these additions will significantly increase the computational cost of the simulations, employing MOR techniques has the potential to accelerate simulation times while maintaining accuracy. Finally, relying solely on core and skin surface temperature measurements is insufficient for a robust validation of the human body model. Integrating precise temperature sensors within the implants themselves, particularly targeting deep tissue temperatures, would

provide valuable data to validate the model's accuracy.

Although a boost converter within the compact TEG can provide sufficient voltage for IMDs, unavoidable power loss occurs in the boost converter's components. To achieve the desired voltage without using the boost converter, an ITEG requires around 10,000 free-standing thermolegs with dimensions of  $0.1 \times 0.1 \times 4 \text{ mm}^3$ . This is challenging to fabricate due to the numerous and intricate pick-and-place operations that would be required for the thermolegs. Compared to mini-bulk TEGs, thin film TEGs offer the advantage of high packing density for thermoelectric elements using precise deposition techniques. However, the height of thin film TEGs is at the micrometer level, which contradicts the need to maximize their height. Stacking micro-TEGs increase the height but also thermal loss at the interfaces. Therefore, novel fabrication techniques should be explored to produce such TEGs.

Furthermore, the hermeticity of the ITEG is crucial for ensuring biocompatibility and reliability. To improve hermeticity, direct bonding techniques such as laser welding should be employed at the joints between the PEEK and metal. Coating techniques such as PECVD and ALD have the potential to enhance the hermeticity of the PEEK housing even further. Hermeticity improvement can be achieved in two directions:

1. Improvements of the Inner Components:

- Coating  $\text{Bi}_2\text{Te}_3$  thermolegs with ALD can prevent interaction with body fluids, although this coating may alter the material properties of  $\text{Bi}_2\text{Te}_3$ .
- Developing biocompatible thermoelectric materials with improved efficiency, specifically optimized for human body temperatures.
- Electronic components can be coated to prevent possible failures caused by moisture ingress.

2. Using a Glass-Metal Housing:

- Glass is a hermetic material and laser welding to metals has proven to yield reliable connections.
- Glass has an inferior thermal conductivity when compared to PEEK. Yet, its conductivity is still lower than ceramic or titanium. Thereby, it causes an acceptable power decrease while ensuring hermeticity.

ITEGs have received considerably less attention compared to other implantable energy harvesters. This is likely due to the misconception that the low-temperature difference within the human body limits their practical use. However, this work demonstrates that ITEGs hold significant promise as a primary or supplementary power source for IMDs.

Simulations on a human body model suggest that an optimized ITEG, comparable in size to a standard pacemaker, could incorporate a thermopile occupying less than 15%

of the volume, significantly smaller than the battery (60-70%) it replaces in a traditional pacemaker. In practical applications, strategically adjusting the volume ratio between the ITEG and the battery inside the implant can further extend the longevity of the device and ensure proper functioning when power generated from ITEG is not sufficient. Hybrid energy harvesting systems that combine ITEGs with other harvesting methods (e.g. kinetic, biochemical, or solar energy) offer the potential for even higher power outputs.

While further research is necessary, particularly regarding biocompatibility testing and long-term in vivo evaluation, this work paves the way for the practical design and implementation of ITEGs in IMDs. I hope that ITEGs will ultimately advance their clinical potential and introduce novel possibilities for healthcare.



## Appendices

A Material Properties in Simulation

B Boost Converter Circuit Diagram of Compact TEG

Table A1: Physical and biochemical properties of materials used in the finite element models.

| Device parameter                | $\kappa$ (W/m/K) | $\alpha$ ( $\mu$ V/K)     | $\rho$ ( $\mu\Omega\cdot\text{m}$ ) |
|---------------------------------|------------------|---------------------------|-------------------------------------|
| Bi <sub>2</sub> Te <sub>3</sub> | 1.5              | 207                       | 12                                  |
| Air                             | 0.0242           | –                         | –                                   |
| PEEK                            | 0.25             | –                         | –                                   |
| Copper                          | 401              | –                         | –                                   |
| Glass                           | 0.9              | –                         | –                                   |
| Ceramic (ZrO <sub>2</sub> )     | 2.5              | –                         | –                                   |
| Titanium                        | 22               | –                         | –                                   |
| Epoxy                           | 0.2              | –                         | –                                   |
| PDMS                            | 0.2              | –                         | –                                   |
| Xenon                           | 0.00565          | –                         | –                                   |
| Krypton                         | 0.00943          | –                         | –                                   |
| Tissue parameter                | $\kappa$ (W/m/K) | $q_m$ (W/m <sup>3</sup> ) | $\omega$ (1/s)                      |
| Skin                            | 0.372            | 1827.1                    | 0.00197                             |
| Fat                             | 0.211            | 461.5                     | 0.00050                             |
| Muscle                          | 0.495            | 988.0                     | 0.00067                             |
| Bladder                         | 0.522            | 1311.9                    | 0.00141                             |
| Bone                            | 0.320            | 295.5                     | 0.00032                             |
| Cartilage                       | 0.487            | 596.0                     | 0.00064                             |
| Heart                           | 0.558            | 42640.2                   | 0.01848                             |
| Intestine                       | 0.542            | 12894.0                   | 0.01388                             |
| Kidney                          | 0.535            | 19247.7                   | 0.06744                             |
| Liver                           | 0.519            | 10712.9                   | 0.01547                             |
| Lung                            | 0.387            | 2446.4                    | 0.00263                             |
| Stomach                         | 0.525            | 7756.3                    | 0.00835                             |
| Trachea                         | 0.487            | 585.4                     | 0.00063                             |
| Grey matter                     | 0.547            | 16230.6                   | 0.01329                             |
| White matter                    | 0.481            | 4497.7                    | 0.00368                             |
| Cerebellum                      | 0.506            | 16373.1                   | 0.01341                             |
| Trachea                         | 0.487            | 585.4                     | 0.00063                             |
| Spinal cord                     | 0.513            | 2669.3                    | 0.00287                             |
| Cerebrospinal fluid             | 0.573            | –                         | –                                   |
| Eyeball                         | 0.595            | –                         | –                                   |
| Tongue                          | 0.495            | –                         | –                                   |
| Blood                           | 0.517            | –                         | 0.17496                             |

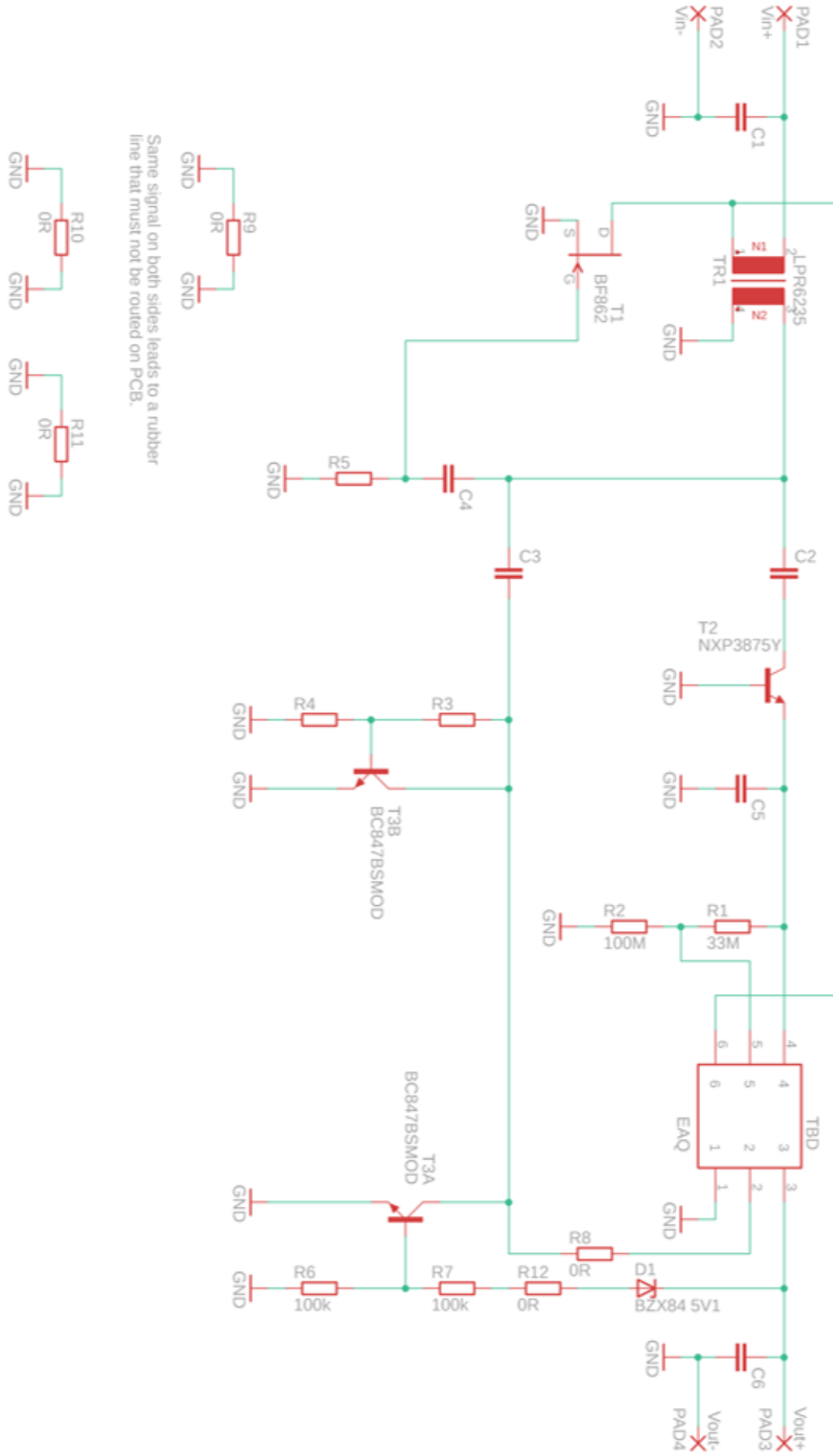


Figure 88: Circuit diagram of the boost converter integrated within the TEG-3.



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## Publication List

The scientific results of this thesis have been partly published in the following journal papers and conference proceedings.

### Journal Publications

**Y. Rao**, C. Xu, M. Voss, P. Ying; H. Reith, K. Nielsch, T. Bechtold, D. Hohlfeld, "Fabrication and Characterization of a Thermoelectric Generator with High Aspect Ratio Thermolegs for Electrically Active Implants", *Advanced Materials Technologies*, vol. 9, no. 1, 2024. <https://doi.org/10.1002/admt.202301157>

**Y. Rao**, T. Bechtold and D. Hohlfeld, "Design Optimization of a Packaged Thermoelectric Generator for Electrically Active Implants," *Microelectronics Reliability*, vol. 139, pp. 114843, 2022. <https://doi.org/10.1016/j.microrel.2022.114843>

**Y. Rao**, C. Yuan, G. Sadashivaiah, D. Hohlfeld, T. Bechtold, "Efficient Design Optimization of a Miniaturized Thermoelectric Generator for Electrically Active Implants Based on Parametric Model Order Reduction," *International Journal for Numerical Methods in Biomedical Engineering*, vol. 37, no. 10, e3517, 2021. <https://doi.org/10.1002/cnm.3517>

**Y. Rao**, T. Bechtold, and D. Hohlfeld, "Design of Packaged Thermoelectric Generators for Implantable Medical Devices: A Comprehensive Parameter Study," *Energy*, 320, p.134932, 2025. <https://doi.org/10.1016/j.energy.2025.134932>

### Conference Publications

**Y. Rao**, M. Voss, T. Bechtold, D. Hohlfeld, "Implantable Thermoelectric Generator with High Aspect Ratio Thermolegs and Integrated Voltage Converter," *22nd International Conference on Micro and Nanotechnology for Power Generation and Energy Conversion Applications (PowerMEMS)*, (Abu Dhabi, United Arab Emirates), 2023. [10.1109/PowerMEMS59329.2023.10417479](https://doi.org/10.1109/PowerMEMS59329.2023.10417479)

**Y. Rao**, T. Bechtold and D. Hohlfeld, "Fabrication Strategy for a Thermoelectric Generator with High-aspect-ratio Thermolegs for Electrically Active Implants", *EASS 2022, 11th GMM-Symposium*, July 2022, Erfurt, Germany.

**Y. Rao**, C. Xu, T. Bechtold and D. Hohlfeld, "Packaged Thermoelectric Generator with High-aspect-ratio Thermocouple Legs for Electrically Active Implants," *2022 23rd International Conference on Thermal, Mechanical and Multi-Physics Simulation and Exper-*

*iments in Microelectronics and Microsystems (EuroSimE)*, 2022, pp. 1-4. [10.1109/EuroSimE54907.2022.9758879](https://doi.org/10.1109/EuroSimE54907.2022.9758879)

**Y. Rao**, S. Bouhedma, T. Bechtold and D. Hohlfeld, "Parametric Simulation of a Packaged Thermoelectric Generator for Electrically Active Implants," *2021 22nd International Conference on Thermal, Mechanical and Multi-Physics Simulation and Experiments in Microelectronics and Microsystems (EuroSimE)*, 2021, pp. 1-4. [10.1109/EuroSimE52062.2021.9410874](https://doi.org/10.1109/EuroSimE52062.2021.9410874)