Universität Rostock
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Elaine
electrically active implants
SFB 1270

Towards Efficient Design Optimization of Thermoelectric Generator via Model Order Reduction and Submodeling Technique

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Introduction

Thermoelectric generators (TEG) convert the thermal energy into electrical energy, and are under investigation as power supplies for medical implants. Currently, the design optimization of TEG is based on time-consuming finite element simulations. This work aims to speed up the design optimization process.

The assembling setup of an electrically active implant is shown in Fig.1. It contains a TEG, an energy buffer and an application-specific integrated circuit (ASIC). In this work, the geometry of the TEG model was constructed based on a commercially available TEG (see Fig.2).

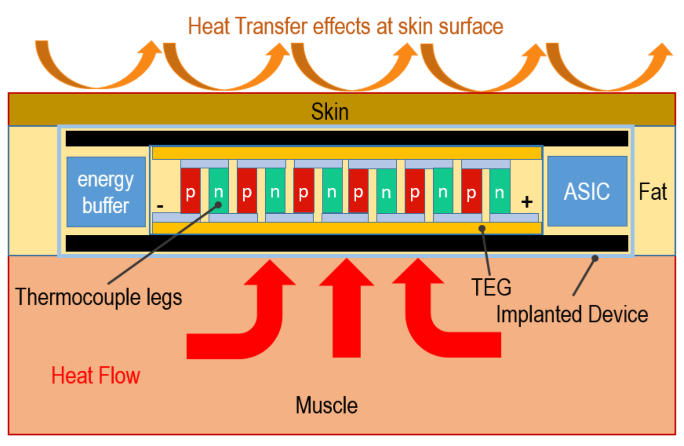

Fig.1: Assembling setup of TEG integrated electrically active implants inside human tissue.

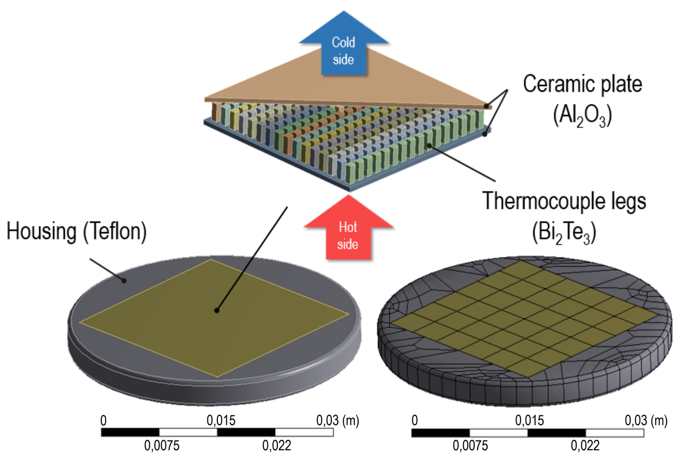

Fig.2: TEG model with 16×16 thermocouple legs and a disc-shaped housing.

Human Torso Thermal Model

A human torso model consisting of realistic geometry of the solid internal organs, skeleton, and main vessels, as well as muscle, fat, and skin layers, was constructed in ANSYS (See Fig.3). Realistic thermal data and physiologically correct material parameters were assigned to the various tissue sections. Subsequently, the TEG model was placed in the fat layer of the human torso model in the chest region (see Fig.4).

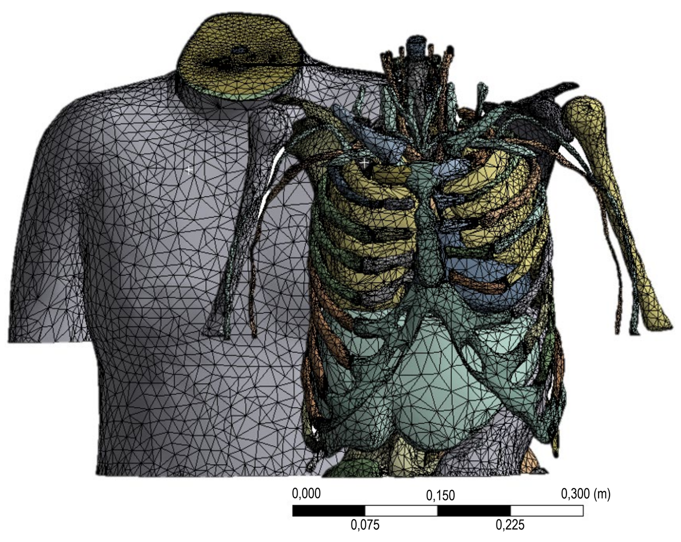

Fig.3: Half human torso model constructed in ANSYS.

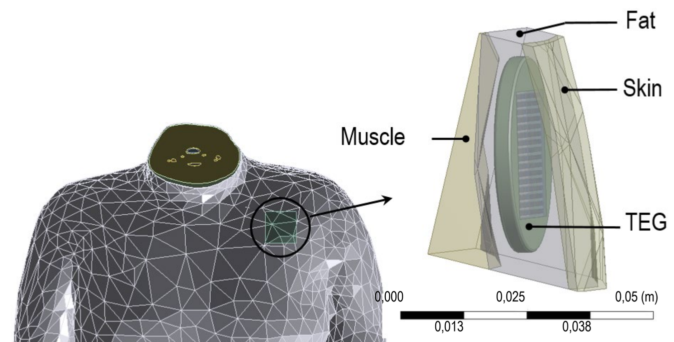

Fig.4: TEG positioned in the fat layer of human torso in the chest region.

Bioheat Modeling

To characterize the internal heat transfer in human tissue, the Pennes bioheat equation is used [1]:

$$\nabla(\kappa \nabla T) + \underbrace{\rho_b c_b \omega (T_a - T)}_{Q_b} + Q_m = \rho c \frac{\partial T}{\partial t} \qquad (1)$$

where ρ,c and κ are the density, specific heat capacity and thermal conductivity properties of different tissue. T is the unknown nodal temperature distribution of the human tissue and T_a is the arterial blood temperature. Q_b and Q_m are the blood perfusion and metabolic heat generation rates.

The heat transfer between the skin surface and the environment can be described as follows [2]:

$$q_{skin} = \underbrace{h_c(T_{skin} - T_{amb})}_{q_{conv}} + \underbrace{\sigma\epsilon\left(T_{skin}^4 - T_{amb}^4\right)}_{q_{rad}} + \underbrace{h_e(P_{skin} - \phi P_{sa})}_{q_{eva}} \qquad (2)$$

where q_{conv}, q_{rad} and q_{eva} are the convection, radiation, and evaporation heat fluxes normal to the boundary skin surface. T_{skin} is the unknown temperature at the skin surface and T_{amb} is the ambient temperature. h_c, h_e are the convection and evaporation heat transfer coefficients. σ and ϵ are the Stefan-Boltzmann constant and emissivity. P_{skin} and P_{sa} are the saturated vapour pressure at the skin surface and saturated vapour pressure, respectively. ϕ is the relative humidity.

Linearization

In Eq.(1) and Eq.(2), the nonlinearities exist in the blood perfusion, radiation and evaporation effects. These nonlinear effects can be linearized as follows:

- „Convection-type“ blood perfusion heat generation rate: $q_b = -\rho_b c_b \omega (T_a - T)$
- Linearized radiation effect: $q_{rad} = 4\sigma\epsilon T_{amb}^3 (T_{skin} - T_{amb})$
- Average evaporation flux trough each node at skin surface: $\bar{q}_{eva} = \sum_{j=1}^{r}(\sum_{i=1}^{s} w_i q_{i,j})/r$

Model Order Reduction

Through the finite element method, the thermal human torso model can be presented by a large scale ordinary differential equations (ODEs) system. To speed up the simulations, parametric model order reduction is applied to generate a boundary condition independent compact thermal model [3]:

$$\Sigma_r \begin{cases} \underbrace{V^T E V}_{E_r} \cdot \dot{x}(t) = \underbrace{V^T(A_0 + h \cdot A_1)V}_{A_r(h)} \cdot x(t) + \underbrace{V^T B}_{B_r} \cdot \underbrace{\begin{bmatrix} Q_m \\ h \cdot T_{amb} \\ \bar{q}_{eva} \end{bmatrix}}_{u} \\ y(t) = \underbrace{CV}_{C_r} \cdot x(t) \end{cases} \qquad (3)$$

where $E \in \mathbb{R}^{N \times N}$ is the global heat capacity matrix and A_0, $A_1 \in \mathbb{R}^{N \times N}$ are the parameter-independent heat conductivity matrices. u is the input vector and $B \in \mathbb{R}^{N \times m}$,$C \in \mathbb{R}^{p \times N}$ are the input and output matrices. m and p are number of inputs and user-defined outputs. The full-scale human torso finite element model is projected onto a lower dimensional subspace $V \in \mathbb{R}^{N \times n}$, with n = 31 and N = 921.336. The full-scale temperature state vector is approximated as: $T(t) \approx V \cdot x(t)$.

Submodeling Technique

The reduced human torso model is applied within a thermal submodeling approach. Its temperature distribution results are used as cut-boundaries for the detailed TEG submodel, which is further used for efficient design optimization (see Fig. 5).

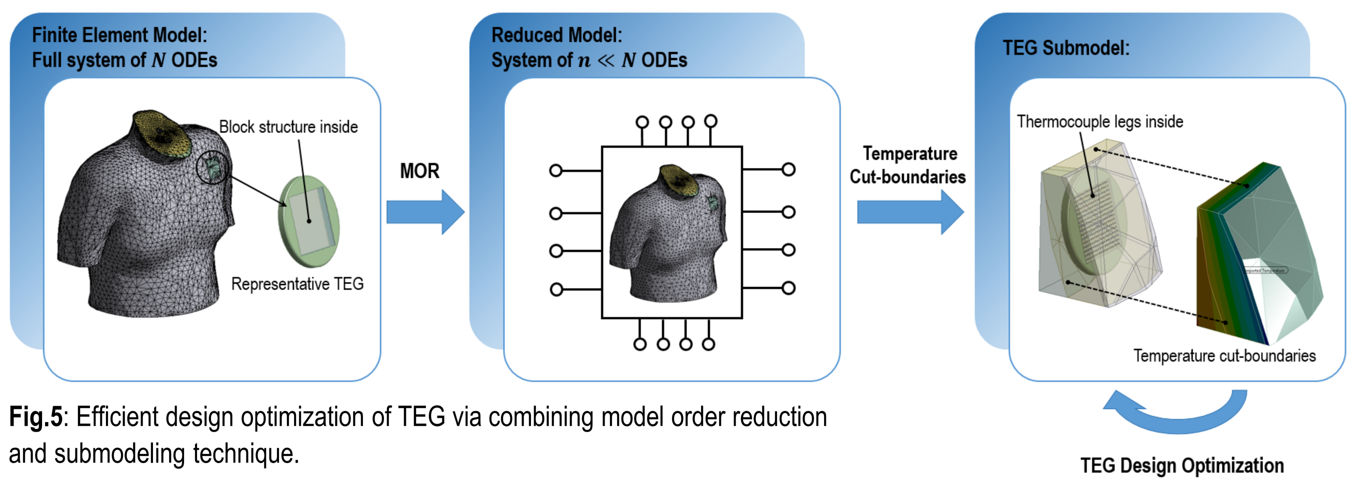

Fig.5: Efficient design optimization of TEG via combining model order reduction and submodeling technique.

Simulation Results

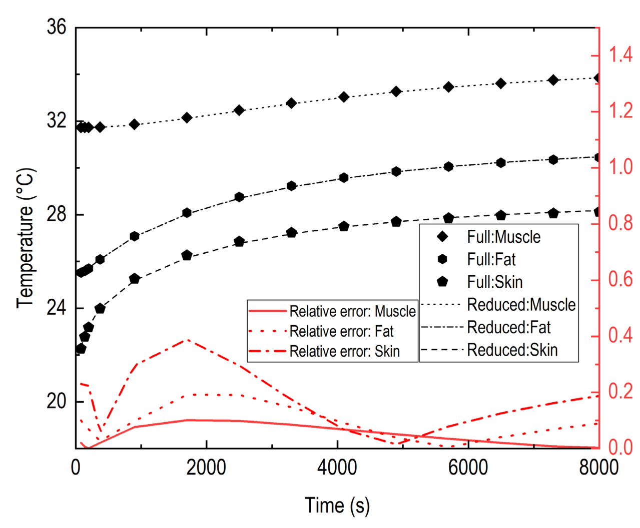

Fig.6: Comparison of the temperature results between full and reduced human torso models (921,336 DoF vs. 31 DoF).

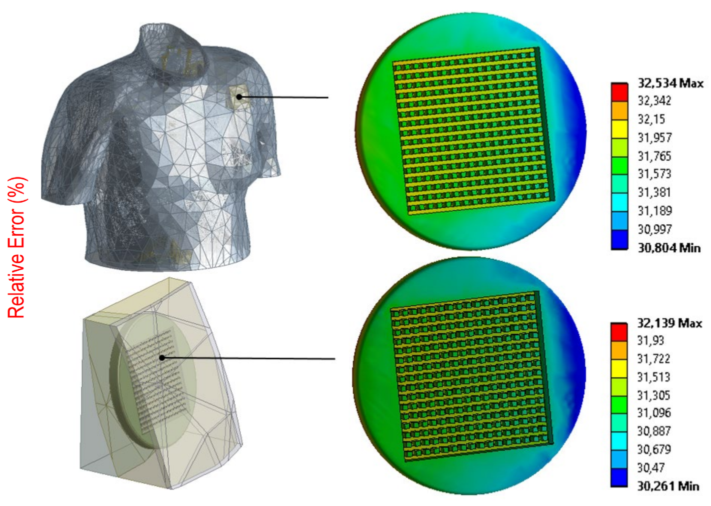

Fig.7: Detailed TEG simulated separately in global human torso model and submodel.

Fig.6 shows the excellent matching between the full and the reduced model. Fig.7 demonstrates the accuracy of the submodel.

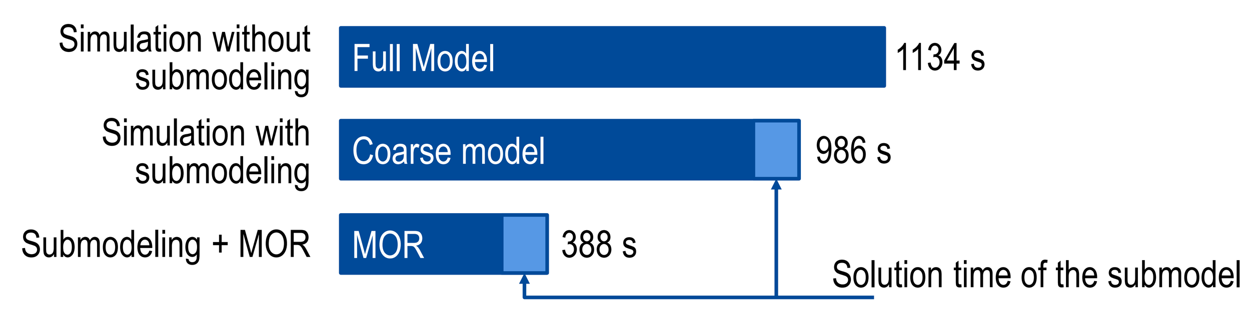

Fig.8: Computational time comparison between the thermal simulations of detailed TEG in submodel and global model. (16× Intel® Xeon® CPU E5-2687W v4 @ 3.00 GHz, RAM 324 GB, VGA NVIDIA Tesla M10)

Fig.8 shows the supreme computational efficiency of the combination of MOR and submodeling compared to “standard approaches”.

References

[1] H.H. Pennes. Analysis of tissue and arterial blood temperature in the resting human forearm. *Journal of Applied Physiology*, 85(1), pp. 5-34, 1998.
[2] K.C. Parsons. Human thermal environments. Taylor & Francis Inc. 2003.
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DFG Deutsche Forschungsgemeinschaft
This research was supported by the German Research Foundation (Deutsche Forschungsgemeinschaft, DFG) within the Collaborative Research Centre 1270 ELAINE.