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Conception and Design of a Hardware Simulator for Restoring Lost Biomechanical Function

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***Abstract*—The Prosthesis-User-in-the-Loop simulator concept represents an approach to integrate users to prosthetic development by a holistic simulation of gait with a prosthesis. It aims at a more user-centered design of lower limb prosthetic devices by utilizing user experience and assessment. As this requires a complex mechanical robot design and sophisticated control strategies that allow for restoring lost biomechanical function, this paper presents the conception and design of a hardware simulator for proof-of-concept studies of those issues. For those investigations, the ankle joints of healthy praticpants are locked mechanically to induce a temporary disability. The task of the simulator is to provide a simulation of physiological gait by artificially restoring ankle functionality. Therefore, the biodynamic behaviour of the locked ankle joint and the enviroment have to be mimicked mechnically. After introducing Prosthesis-User-in-the-Loop simulator idea, the conception of a proof-of-concept simulator is presented. From this, an analytical model is derived and inverse dynamics simulation are used for design. The resulting mechanism is limited to sagittal plane movements and thus has three degrees of freedom. The actuators are dimensioned to meet the requirements of walking motions in the human subject with maximum body height among the test population.**

***Index Terms*—Prosthetics, User-centered Development, in-the-Loop Evaluation, Biomechanics, Robotics.**

I. INTRODUCTION

In the last decades the development of lower limb prostheses led to mechatronic devices with increased biomechanic performance and decreased physical effort for the users [1], [2]. State-of-the-art solutions support users either semi-actively by adapting the mechanical characteristics of the system - e.g., stiffness and damping - or actively by introducing energy to locomotion by actuators - e.g., an electric motor [2], [3]. Anyhow, a high demand for better functionality and an improved situation of the users remains. An increase of functionality is required in gait flexibility or stair climbing for example [1], [4], [5]. At the same time, users might not be satisfied with their prosthesis, since they experience gait as being directed by the prosthesis [5], need to learn specific gait strategies [5], [6] or have an higher metabolic effort than persons without amputation [7]. To cope with this situation users could be involved in every step of prosthetic development. This is usually attempted during early development by evaluating surveys and interviews and transfer their results into technical requirements as preformed in [1] on the one hand. On the other hand, the evaluation of prototypes in clinical trials with amputees integrates users in the end of this process as carried out in [3]. Yet, there is a void of utilizing experiences and knowledge of users between assessing requirements and their participation in prototype evaluation.

Among present gait and prosthetic simulators, the one from [8] aims at decreasing the development time in prosthetic design trough reproducible experiments. The mechanical design allows to simulate the horizontal, vertical and rotatory movements of hip and stump to the investigated prosthesis while the ground is simulated by a treadmill. Despite of the possibility to perform biodynamic examinations, many other simulators and test rigs aim on the investigation of prosthetic materials and components as the one in [9] that provides addtioanl degrees of freedom compared to [8]. In contrast to testing scenarios, biodynamic examinations are conducted in fundamental biomechanical research aiming at insights in human gait, but not at developing new prosthetic concepts. The simulator in [10] is designed to evaluate the kinematics und kinetics of foot and ankle during stance phase considering cadaveric shanks and comparing those to below-knee prostheses. Another application of gait simulators is the rehabilitation of harmed limbs or functions by externally induced movements or haptic interactions [11], [12]. An approach to investigate locomotion with different prostheses based on computational models is the prosthesis presented in [13] that is limited to simulate passive or semi-active solutions. An important example for haptic locomotion interfaces is the Hapticwalker [14]. This rigid hybrid parallel-serial robot is designed to be applied in gait rehabilitation. Beyond those, the Prosthesis-User-in-the-Loop simulator concept presented in [15], [16] aims at closing the gap in user-centred prosthetic design by a holistic simulation of gait with the investigated prosthesis for the participant.

A detailed description of the fundamental simulator idea combining biodynamical and visual simulation is presented in Section II. Subsequently in Section III, the conception of the proof-of-concept hardware simulator for the investigation of restoring biomechnical function is given. A kinematic and a dynamic model are established in Section IV and used for an inverse dynamics simulation based on human gait data

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in Section V. Subsequently, the mechanical design of the system is derived and the actuators are dimensioned with the simulations results and biomechanical requirements. Finally, a conclusion and an outlook are given in Section VI.

II. SIMULATOR IDEA

The fundamental idea of Prosthesis-User-in-the-Loop is to support user-centered prosthetic design by allowing users to directly participate during the development process. For that purpose, different prosthetic concepts are simulated for the user in several gait scenarios mechanically and visually. The functional units used to create a virtual room and to simulate gait and interactions with the specific investigated prosthesis are shown in Figure 1. While the environment is simulated to the intact leg of the participant, the stump of the harmed leg is attached to an actuated and instrumented biodynamic simulation unit. This robotic device simulates the biodynamical behaviour of the prosthesis by applying the mechanical interactions between the body of the user and the prosthesis to the participant. The system is controlled based on simulations of gait and the prosthesis using software models and the acquired sensor data. The models represent the dynamic behaviour of the prosthesis as well as the gait chracteristics of the participant as in [16]. In order to investigate different types of prostheses, the models in the real time control system can be changed. A holistic illusion of walking with the simulated prosthesis is established by adding a visual simulation unit as proposed in [15] and hiding the components for the biodynamical feedback in a blackbox. The combination of biomechanical and visual simulation should provide the experience of physical integrity for the user while testing the real behaviour of the investigated prosthesis in virtual reality. Hence, the functionality of the prosthesis can be isolated from other possibly correlated factors for an assessment by the user [15]. During the experiments the user is secured in a user safeguarding unit. Further potential of the simulator concept lies in the elaboration of prosthetic devices that are in pre-prototype status or cannot be realized with present technologies. This can be enabled by implementing a high-performance combination of kinematics and actuation especially in the biodynamical simulation unit. Beyond research and development purposes, the simulator can also be used for gait training and psychological support of amputees [17]. The requirements arising from this are discussed in [17] and a test rig for the investigation of psychological issues in prosthetic development is presented in [18]. A possible implementation of the Prosthesis-User-in-the-Loop concept using a hexapod platform as the biodynamic simulation unit and a treadmill for the simulation of the environment is presented in [17]. For the visual simulation, a frontal projection showing the way of locomotion and a horizontal screen depicting the virtual representation of the locomotor system are proposed.

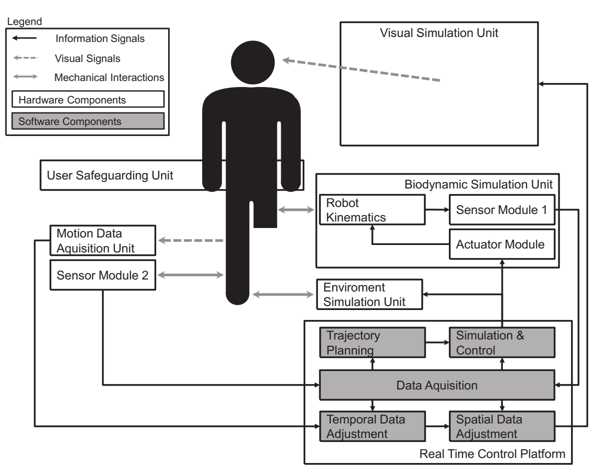

Figure 1. Functional concept and units of the simulator.

This simulator concept requires a complex mechanical design and sophisticated control strategies to restore lost biomechanical function. The reduced simulator hardware presented in this paper aims at proof-of-concept studies in those fields. Therefore, the ankle joints of healthy particpants are locked mechanically to induce an artificial and temporary disability. The simulator ought to simulate physiological gait by restoring ankle functionality in sagittal plane with appropriate motions. Thus, the biodynamic behaviour of the locked ankle joint and the enviroment have to be simulated mechnically. The mechanism presented in this paper allows to investigate control concepts for biodynamic simulation with lower effort in design and computation as well as the selection of participants.

III. CONCETPION

To simulate the required motions of the ankle joint in sagittal plane, the robotic system has to provide three degrees of freedom as presented in Figure 2. Those are the horizontal direction of motion x, the vertical movement direction z and the joint rotation θ. By establishing the motions in these directions and accomodating static and dynamic loads during gait simulation, the biomechanical functionality of the human ankle joint is replaced. To provide an appropriate simulation of gait, the system is further required to have no components disturbing the possible range of motion. Due to this and to sustain the visual impression of physiological walking, placing the whole setup below the participant is assessed to be advantageous.

The required degrees of freedom x, z and θ can be implemented by mechanisms with three or more joints. Considering only three-joint solutions, there are six setup permutations, as prismatic and rotational joints could be exchanged in order. Further, there are different technical solutions to realize those joints. Since the horizontal direction of motion represents the main direction of motion and thus requires the highest joint powers resulting in a heavyweight actuator, it is located first in the kinematic chain to unload the other degrees of freedom. Further, the power required for motions in z direction are higher than the ones in θ. Thus, the order of the kinematic chain is chosen to be x, z, θ. The previously selected arrangement of the robots degrees of freedom is implemented as depicted in Figure 2. In this, the horizontal movement x is realized by a toothed belt axis. For the vertical direction z, a scissor lift table is used while the rotational degree of freedom θ is implemented by a revolute joint. An orthosis fixing the ankle joint is attached to the third link, induces the artificial and temporary disability and represents the human-machine interface.

IV. MODELING

The joint motions of the three-joint mechanism are represented by the vector $q = [q_1 \quad q_2 \quad q_3]^T$. For the transformation of measured human motions into the desired joint trajectories of the robot, an inverse kinematics model of the mechanism is presented in this section. Additionally, a dynamic model is derived for the design of the system and the dimensioning of the actuation by simulation.

A. Inverse Kinematics

The forward kinematic function $\varphi(q)$ of the system is determined from the geometry given in Figure 2 and Table I as shown in [19].

The Jacobian J of the robot is obtained by the derivation of this function with respect to the joint coordinates q and inverted analytically. With this, the relation between the kartesian human and the robot joint velocities is

$$\begin{bmatrix} \dot{q}_1 \\ \dot{q}_2 \\ \dot{q}_3 \end{bmatrix} = J^{-1} \begin{bmatrix} \dot{x} \\ \dot{z} \\ \dot{\theta} \end{bmatrix} = \begin{bmatrix} 1 & 0 & J_{i,13} \\ 1 & J_{i,22} & J_{i,23} \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} \dot{x} \\ \dot{z} \\ \dot{\theta} \end{bmatrix}, \qquad (1)$$

where the elements of the inverted Jacobian are

$$J_{i,13} = -l_5\, sin(q_3) + l_4\, cos(q_3)\,,$$
$$J_{i,23} = \sqrt{l_8^2 - (l_7 - q_2)^2}\,(l_7 - q_2)^{-1}\,,$$
$$J_{i,23} = (l_5\, cos(q_3) + l_4\, sin(q_3))\sqrt{l_8^2 - (l_7 - q_2)^2}\,(l_7 - q_2)^{-1}\,.$$

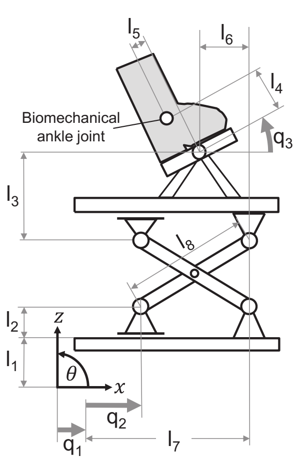

Figure 2. Sketch of the simulator mechanics.

B. Dynamics

The dynamics model of the system is given by the joint space dynamics equations of the system

$$\tau = M(q)\ddot{q} + C(\dot{q}, q) + G(q), \qquad (2)$$

derived by Lagrange equations of the second kind [20]. In this, the centers of gravity of the links are assumed to be located in the geometric center. Therefore, the kinetic energies of the joints are determined with the masses and inertias of the system given in Table I. For the potential energies, displacements in z and θ are considered. Due to the three-joint mechanism, the vector $\tau = [\tau_1 \quad \tau_2 \quad \tau_3]^T$ gives the joint torques induced by actuation and ground reaction forces (GRF). The matrices $M(q)$, $C(\dot{q}, q)$ and $G(q)$ describe inertial, coriolis and gravitational effects. In an iterative process, the masses and inertias are estimated by simulations assuming rigid geometric bodies and using the data sheets of the actuators that are subsequently selected to cover the operational space and the powers required by the trajectories taken from [16], [21]. As the model in [16] uses a coordinate system fixed to the hip of the participant, hip motions are not considered and thus results might slightly differ from the real motion. The mass m_2 equals the sum of m_{21} and m_{22} representing the masses of the components of third joint and scissor lift table. Beyond those assumptions, impacts of elasticities, friction and levels of efficiency are not considered in the dynamic equations. The static and dynamic influences of the human subject to the mechanism are modeled as resulting ground reaction torques using the ground reaction forces and their center of pressure (CoP) measured in [21].

V. DESIGN

The system is simulated for the selection of the kinematic configuration given in Section IV and for the dimensioning of

Table I
PARAMETERS OF THE SIMULATOR MECHANISM

	Parameter	Value	Unit	Description
mechanics	l_1	0.10	m	
	l_2	0.05	m	
	l_3	0.05	m	
	l_4	0.07	m	
	l_5	0.07	m	
	l_6	0.15	m	
	l_7	0.30	m	
	l_8	0.30	m	
	m_1	40.0	kg	mass link 1
	m_2	25.0	kg	mass link 2
	m_{21}	21.0	kg	mass joint 3
	m_{22}	4.0	kg	scissor mass
	m_3	5.0	kg	mass link 3
	I_3	0.400	kgm^2	inertia link 3
	$I_{m,1}$	0.023	kgm^2	inertia actuator 1
	$I_{m,2}$	0.011	kgm^2	inertia actuator 2
	$I_{m,3}$	0.003	kgm^2	inertia actuator 3
	i_1	21.8	m^{-1}	gear ratio 1
	i_2	28.0	m^{-1}	gear ratio 2
	i_3	2.2		gear ratio 3
	g	9.81	$\frac{m}{s^2}$	gravity

the actuators. With the inverse Jacobian (1) human movements are transformed into desired trajectories of the robot. For the determination of the required actuator torques and powers, an inverse dynamics simulation of (2) is performed.

A. Human Data

The human data obtained from [21] contains measurements for joint motions and torques as well as ground reaction forces inlcuding center of pressure. One gait cycle of walking on level ground at a velocity of $1.6\,\mathrm{ms}^{-1}$ resulting in a cycle time of $0.978\,\mathrm{s}$ is captured. Figure 3 shows the ankle trajectories resulting from hip, knee and ankle motion transformed to x, z and θ. The corresponding velocities and accelleration are determined numerically. The gait cycle starts with heel contact and stance phase, transfers to swing phase at about $0.65\,\mathrm{s}$ and ends with toe off. The corresponding horizontal and vertical ground reaction forces between human foot and the environment as well as their center of pressure are given in Figure 4. Due to gravitation, the vertical load is higher than the horizontal one. During swing phase starting at about $0.65\,\mathrm{s}$ the foot is not in contact with ground and thus no ground reaction forces appear.

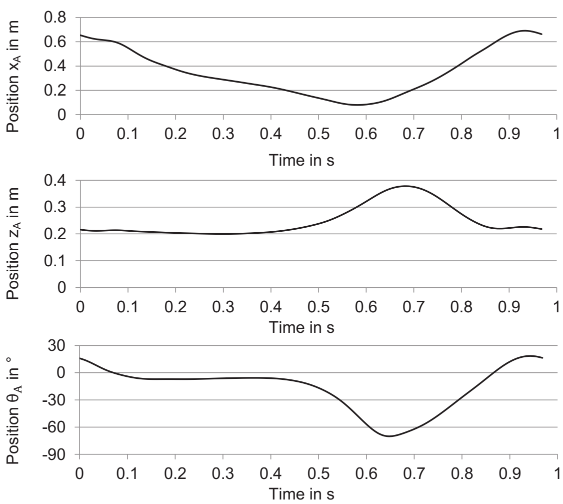

Figure 3. Trajectories from subject with maximum body height.

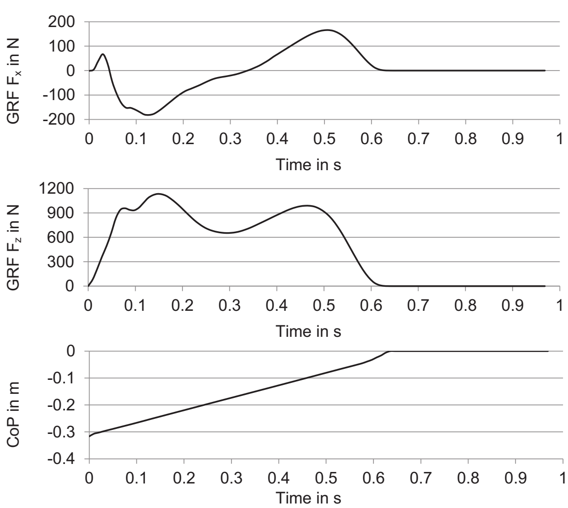

Figure 4. Ground reactions from subject with maximum body height.

B. Acutator Dimensioning

For the dimensioning of the actuators the motion data of the participant with the maximum body height of $1.91\,\mathrm{m}$ and a body weight $97.3\,\mathrm{kg}$ is considered to examine the loading of the system in a worst case scenario. To keep the powers required from the actuators on a low level, those are dimensioned for walking on even ground with $1.6\,\mathrm{ms}^{-1}$ only and are thus limited regarding gait velocity. The required actuator powers are determined by an inverse dynamics simulation of (2). From this, the drive sided motions $q_{m,j} = i_j\, q_j$, torques $\tau_{m,j} = i_j^{-1}\, \tau_j$ and powers $P_j = \tau_j\, \dot{q}_j$ are evaluated for every joint j to select appropriate actuators. The power plots for the three joints of the mechanism are shown in Figure 5. The maximum power requirement for x direction is $6823\,\mathrm{W}$ and thus at a higher level than the ones required for vertical movement ($943\,\mathrm{W}$) and rotation ($667\,\mathrm{W}$). As the maximum required torques show to be $154\,\mathrm{Nm}$, $108\,\mathrm{Nm}$ and $56\,\mathrm{Nm}$ for x, z and θ, torque motors are assessed to be feasible actuators. To match the required joint velocities, the gear ratios of the toothed belt axes are chosen to be $i_1 = 21.8\,\mathrm{m}^{-1}$ and $28\,\mathrm{m}^{-1}$ for x, z while an addtional transmission with a gear ratio of 2.2 is introduced at the third joint. For the simulations, the data of the drives $MST210C-0050$, $MST160C-0050$ and $MST130E-0020$ from Bosch Rexroth AG, Lohr am Main, Germany and the toothed belt axes $QSZ125$ and $QSZ100$ from GETOtec, Munich, Germany are used. The latter ones are selected to bear the belt forces of $4311\,\mathrm{N}$ and $3130\,\mathrm{N}$ for the x- and z-axis as well as vertical loads of $1844\,\mathrm{N}$ $1467\,\mathrm{Nm}$ resulting from design and simulation. Compliance that might be introduced by the toothed belts is not considered in this paper. Beyond possible negative impact on the positioning accuracy, this compliance could also be beneficial for user security.

C. Mechanical Design

Figure 6 shows a conceptional $3D$-model of a possible implementation of the robotic mechanism. To cover a high bandwith of subjects in the geometric design of the mechanism, the data of the participants with minimum and maximum body height of $1.63\,\mathrm{m}$ and $1.91\,\mathrm{m}$ are considered. In the lower part of Figure 6, the toothed belt axes moving the translational degrees of freedom x and z can be seen. The upper one of those moves the scissor lift table realizing the z-movement. On top of this the revolute joint moving the orthosis indicated by a schematic boot is located. By realizing the vertical axis with the scissor lift table, the mechanism can be placed completely below subject without disturbing the simulation. Beyond this, the required range of motion in the second joint is reduced due to the chosen lever ratio. This is designed to perform the adaptation to the specific body heights of the subjects and to cover the required working range of $0.18\,\mathrm{m}$ for vertical motion from the human trajectories. With the kinematic setup and toothed belt axis, a working range of $0.28\,\mathrm{m}$ is constituted. In x, a range of $0.61\,\mathrm{m}$ is required while the axis provides up to $0.80\,\mathrm{m}$. As the revolution in θ is not limited considerably, the solution shows to meet the workspace requirements. For structural integrity, the scissor levers and the plate reinforcing the orthosis are dimensioned regarding bendingm, while the scissor shafts are dimensioned regarding torsion. The loads are assumed from human and robot weight as well as the resulting forces and torques resulting from dynamics simulation. To obtain realistic results, the parameters in Table I are the final ones from iteratively adjusting simulation results and structural design. The adaptation to the foots of subjects, commerically available orthoses allow for a certain adjustment - e.g., by pumping mechanisms. To enable investigations of psychological factors like body scheme integration, the requirements given in [18] are considered.

VI. CONCLUSION

The Prosthesis-User-in-the-Loop concept gives participants the possibility to test prosthetic devices that are not implemented in hardware and to perform fundamental research in their psychological experience. Additionally, reproducibility of the simulated gait scenarios allows for an evaluation with several users in equal situations under laboratory conditions. Further possible applications of the simulator could be psychological support and gait training.

As the implementation of this concept is rather complex, the robotic device introduced in this paper can be used for proof-of-concept studies in design and control of the biodyamic simulation unit. It is designed to work with healthy persons to allow for a better subject acquisition and limited to sagittal plane motions and level ground walking to reduce hardware effort. The device is thus limited to simulate walking as actuation power should be not sufficient for running at higher speeds although the overloading capabilities of the drives. These might be required to tackle additional loads as impacts of elasticities, friction and actuator efficiencies that are not considered in simulation. Due to the limitation to motions in sagittal plane the main aspects of straight walking can be considered, but other types of gait as turning cannot be simulated. Additionally, frontal and transversal plane degrees of freedom should not be locked completely but provide elastic fixations to prevent the participants from being harmed by the simulator. Regarding the actuated degrees of freedom, user safety might be established based on the compliance of the toothed belts and could be increased by introducing series elastic actuation. In sagittal plane, the workspace required to simulate the human data from measurements is covered completely by the investigated mechanism. The deviations from real human movement that might be induced by the fixation of the desired motion trajectories coordinate system to the participants hip, are assessed to be negligible for design issues. This is due to the fact that the hip shows verly small motions in x and z compared knee and ankle joint in level gait.

In their future works the authors will focus on control strategies for the proof-of-concept simulator. Important candidates for this application are methods of force control that might be adapted in parameters to simulate the biomechanical system in motion. Finally, the authors aim at realizing the full Prosthetic-User-in-the-Loop concept.

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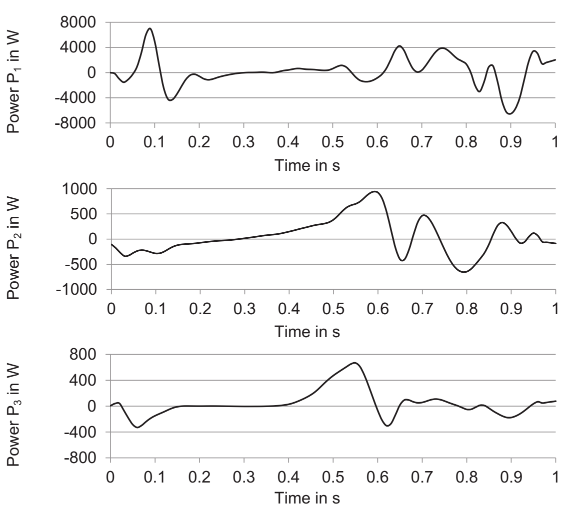

Figure 5. Required actuator powers for the robotic mechanism.

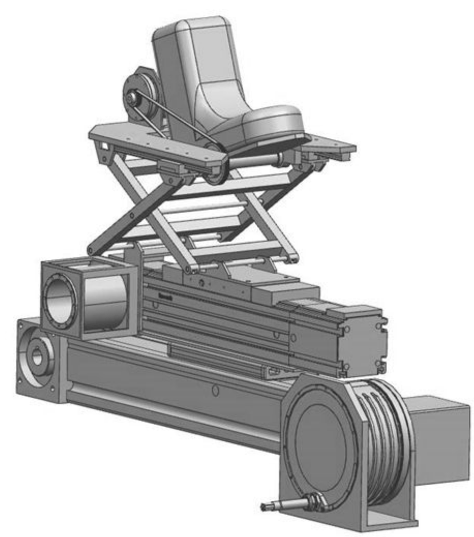

Figure 6. Conceptional $3D$-model of the robotic mechanism.

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