Design Concept and Model-based Evaluation of an Exoskeleton for Low Back Support during Lifting Tasks

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***Abstract* – Workers in various context are exposed to physical stress during repetitive movements carrying heavy loads, resulting in musculoskeletal disorders like low back pain and lumbar disc herniation. Furthermore, consequences are direct and indirect costs due to rehabilitation and a high amount of days of absent. To counteract the physical workload, support systems like exoskeletons are developed for different settings like transferring patients in care. Nevertheless, positive and negative biomechanical effects of exoskeletons to the human body are not comprehensively examined and need to be evaluated by using different methods like simulation and modelling of the human-system interaction.**

This paper describes the process of development of a back support exoskeleton for the example of nursing care regarding context-based requirements, design concept and evaluation of the aimed biomechanical effects. In order to systematically evaluate the design, a combined model of human and exoskeleton is developed. Focusing on drive technology, dimensioning quantities are derived from the combined model. Further, an outlook on the analysis of effects of the design on stress to the human body is provided.

Keyword – Exoskeleton, Development Process, User-centered Design, Nursery Care, Modeling, Biomechanics

I. INTRODUCTION

Due to ongoing technological advances and other general needs (e.g., demographic changes), physical support systems such as exoskeletons are becoming increasingly popular [1]. In recent years, numerous exoskeletons have been developed to support different body regions and functions of the human body in various applications [2]. Examples include exoskeletons for supporting overhead work (e.g., [3], [4]) and walking e.g., [5]), and hand exoskeletons (e.g., [6]). In the medical field, such systems are often used for prevention and rehabilitation of upper and lower limb impairments [7]. They are used, for example, to enable a wearer to relearn movements by prescribing movements through an exoskeleton, e.g., as a result of a stroke or accident [8]. Support systems are also used to compensate for the loss of muscle strength for people who can no longer access their full muscle strength due to age or illness. For industrial, nursing care or military purposes, exoskeletons are used to prevent musculoskeletal disorders due to the overuse of muscles and joints in repetitive physically demanding tasks, as well as to extend the wearer's capabilities such as lifting and transporting heavy loads [9], [10].

Exoskeletons are diverse in their function and design. Depending of the intention of use, the user movement can be e.g. stabilized, expanded, supported, strengthened or facilitated. Basically, exoskeletons can be classified differently. This is often done with regard to the support areas (e.g., back and shoulder). Of greater technical relevance is, on the one hand, the morphological structure (type and manner of force transmission, in which biomechanically equivalent structures and end effector structures are to be fundamentally distinguished) inclusively the rigidity of the force transmitting structure (differentiation between soft and hard structures or structures with and without compliance/flexibility) as well as on the other hand the kind of support and functionalities (context-based characteristics of support). Depending on the approach and application purpose, different actuator technology, sensor technology and interfaces (for operation and force transmission) are also used. Today's market-ready products use rotating or linear actuators that transmit forces to humans via appropriate mechanics. Electric, pneumatic or hydraulic actuators are used [9]. Textile systems use stretch or similar structural elements.

In the project “Intelligente Auslegung und Optimierung von KI basierten, physischen (Körper-) Unterstützungssystemen mit moderner Antriebstechnologie” (KIKU), founded by dtec.bw, the purpose is to design a back support system for caregivers based on an electric drive, which is optimized and built using intelligent model-based methods. These methods can be applied to the design of the system, taking into account electromagnetic and mechanical system properties. Hence, intelligent algorithms efficiently optimize extensive multi-physical models. Optimization goals are, for example, the selection of the best possible number and design of drive units acting in parallel and their arrangement in relation to each other in the overall system.

To be able to validate models for later optimization, a demonstrator platform for a lower back exoskeleton is developed. The development process, without extensive use of

models, is described in section II. Further, a modeling approach is presented in section III and applied to the lower body exoskeleton in section IV.

II. Practice-oriented Developtment of Lower Body Exoskeleton

A. Motivation

Caregivers and nurses are confronted by a variety of mental and physical workplace exposures [11]. The day of care workers is characterised by a high demand of social interaction with patients within time pressure and physical strain resulting from long time standing, lifting and transferring patients. Therefore, they are exposed by a high strain of the musculoskeletal system resulting in muscular disorders in shoulder and neck and low back pain (LBP) [12]. Because of risk factor like many working hours, repetitive lifting tasks, trunk flexion and rotation, LBP is common in 37 % of caregivers [13]. Besides physical disorders, negative consequences are also direct costs for rehabilitation and physical therapy and indirect costs during days of absence and lower productivity. Additionally, the demographic change lead to higher demand for nurses and caregivers. Good working conditions are needed to maintain healthiness until retirement within the care setting.

To avoid high strain in spine and lumbar load on intervertebral disks, professional caregivers learn "spine-friendly" lifting techniques in trainings and educations [14]. Nevertheless, there is poor evidence regarding the efficacy for these interventions in preventing low back pain [15]. Therefore, physical support is crucial to lower heavy strain. There are several approaches including lifting aids like ceiling-based patient lifts, sliding boards or body worn devices called exoskeletons. These assistive devices can help to ease physical demanding and high intensity work. Not least because of the social interaction between nurses and patient, the purpose is not to substitute human work, but to support the user while doing his work.

Exoskeletons can be used early in "working life" as a primary prevention to avoid the incidence of physical complaints due to heavy load. Kermavnar et al. [16] reported a reduction in back muscle activity and spinal compression forces, which could lead to a reduction of LBP. Because exoskeletons are relatively new on the market and LBP is already present in many caregivers, they may also help to prevent a worsen of existing disorders in terms of the secondary prevention.

There are several systems already developed to assist lifting tasks and support the back muscles [17], but not all of them are appropriate to apply them into care setting. Because there are no specific developed exoskeletons for nurses or caregivers regarding there special needs, we evaluate the Scoot Pivot Transfer [18] as one of the crucial lifting techniques for patients using motion capture and electromyography to derive requirements for exoskeleton development.

B. Biomechanical Analyses of Lifting Tasks in Nursery Care

In order to evaluate the requirements, which are given by the lifting tasks in care setting, we conducted a preliminary experiment using motion capture and electromyography. To measure the kinematics during the lifting tasks, a mobile Xsens MTw Awinda System was used (Xsens Technologies BV, Enschede, Netherlands). With a Myon 320 system (myon AG, 275 Schwarzenberg, Swiss), the muscle activity of the m. erector spinae was recorded and scaled relatively to the maximum voluntary contraction (MVC). The ergonomic evaluation tool Industrial Athlete (IA, scalefit) was used to estimate the lumbar disc compression of L5-S1 during lifting.

We present the upper body flexion angle in comparison to the muscle activation of m. erector spinae when performing the Scoot Pivot Transfer.

The Scoot Pivot Transfer is a typical method to move patients from the bed into a wheelchair and is depicted in Figure 1, whereby the first three picture show the movement cycle. The kinematic flexion angle during the transfer is shown in the first diagram. The second diagram presents the muscle activation of the low back muscle (m. erector spinae). The last curve displayed the compression force of the lumbar spine.

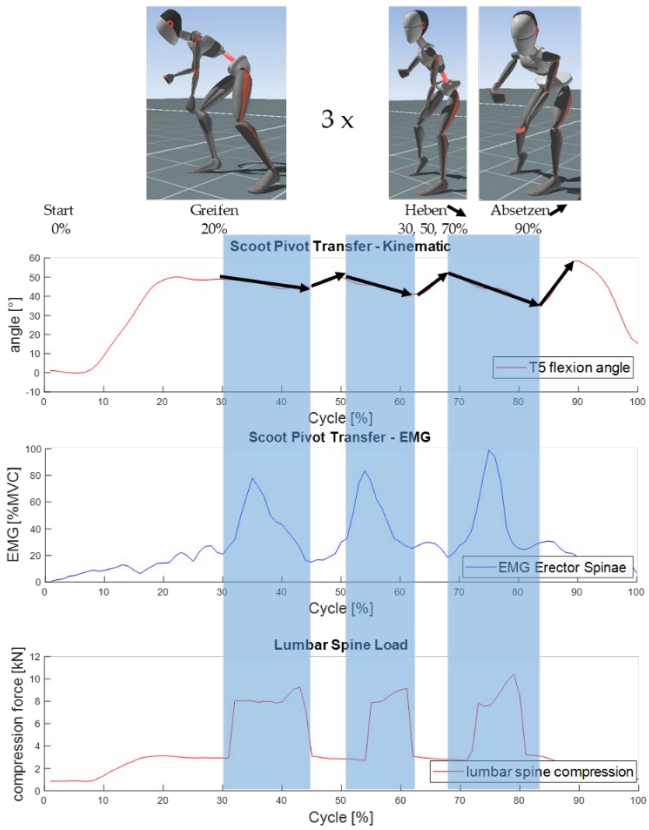

Figure 1: Scoot Pivot Transfer and analysis of joint angle (T5), electromyography and lumbar spine compression.

The transfer technique can be divided into three phases. During Phase 1, the caregiver holds the patient around the back with the right hand and at the thigh with the other hand. When holding the patient safety, the caregiver lifts the person in phase 2, moves him a small distance to the side by squatting and rotating and sets him down. This phase is repeated three times to increase the moved distance. At phase 3, the patient is set down into his wheelchair.

As can be seen, the physical strain onto the lower back rises sharply when lifting the patient. The muscle activation peaks around 80 % of the maximal voluntary contraction and the compression force of the lumbar disc amounts up to 10 kN. The exoskeleton to be developed is intended to reduce the maximum load to the body. The blue box plots show the periods, where the device must actively support the user during the task.

This experiment allows insights about the requirements, the exoskeleton has to fulfil regarding the mechanical structure, the actuator and control algorithm. Therefore, an actuator with high acceleration and torque is needed, due to the fast slope of muscular and tissue stress when lifting the patient. It needs to be activated three times during 15 % of the whole lifting cycle, which lasts about 10 seconds. The control unit must ensure the real-time requirements to make sure a timely actuation.

A more detailed analyses of requirements to be considered for the design of exoskeletons are given below.

C. Requirements

To the best of our knowledge, there are only two specific developed exoskeletons for nurses [10, 19]. Most devices were developed for back support during unspecific lifting tasks and where also tested in laboratory with healthy subjects [16], [2]. Nevertheless, there are different preferences and characteristics, that are beneficial for a medical setting when working with patients in comparison to an industrial setting.

The exoskeleton needs to be designed regarding to general requirements for human-system interactions, as well as to specific requirements, that are given in order to the context of nursery care and for the work with patients. Besides the mechanical design, properties like acceptance, wearing comfort and safety are essential for design optimization. Some important characteristics are shown afterwards in detail.

The **purpose/function** of the exoskeleton is to support care workers during lifting and transferring patients. Regarding the stress to the lumbar spine, it has to reduce load by actively assist the degree of freedom of hip and upper body extension during static posture and movements.

The system is supposed to ensure the users **kinematic** conditions by mimicking the user movements in ways, that he is able to move freely without any restrictions when wearing the exoskeleton. It has to be verified, that no compensatory movements are made when using the exoskeleton, in order to accomplish tasks that are not feasible in an “natural” manner.

Safety is elementary to consider and has to be observed from two perspectives. One for the user itself, and the other regarding to the patients, who are also in direct or indirect contact with the device. Therefore, hardware security elements like an emergency stop button and software control algorithms that continuously record data in order to stop the device at adverse events are required to ensure safety. This includes mechanical restrictions, to not exceed any intended ranges of motion. As the patients are in contact with the device, it must be ensured, that he is not able to grasp into the system and get bruises or any else injuries.

The **control** unit, usually a microcontroller or –computer, uses input sensor information like force from Force Sensing Resistors (FSR) and joint angles, speed or acceleration from Inertial Measurement Units (IMU). Based on this information, the control algorithm operates the actuation system and activate or deactivate the supporting function dependent to the user motion and intention. For this, the user should always retain sovereignty over the system, without becoming externally determined. This is done by a feedback mechanism with ongoing target-performance comparison. Nevertheless, the major challenge and current research in the field of human-system interaction is to predict the intentions of the users prior to the movement, to assure a fast system acting in real time.

The **actuation** module generates the external force in order to support the users lifting task. Generally, there are several possible technologies like electro motors, pneumatic, hybrid or hydraulic actuators [20]. Special drive technologies like artificial polymer muscle fibers [21] or tubular linear motors [22] may be used in exoskeletons. Every drive system has its own advantages and disadvantages, expressing e.g. in power to weight ratio, dynamics, heat generation, efficiency or power consumption.

Acceptance is one of the most crucial requirements when developing system for human users, because a perfectly working device is nothing worth, if it is not used. Therefore, subjective determinants should be considered for the design process by integrating users into the development as soon as possible. To enhance the acceptance of an exoskeleton for the non-industrial context, the design should be ideally invisible and wearable under working clothes. Moreover, the system should be put on quick and easy, because of the time pressure during work. Major demands are made regarding the wearing comfort, as it has to be worn during the working day without disturbing the user while doing his working routine.

Wearing comfort referring to the human-system interfaces. They have two inherent functions: (1) transfer forces to the body and (2) ensure the greatest possible comfort. Whereas the material for the first function is supposed to be stiff in the direction of force, it needs to be soft and flexible vertical to the body to avoid pressure peaks at the contact area and therefore facilitate a comfortable wear throughout the use of the exoskeleton. Moreover, regulation of heat and humidity needs to be considered, as well as the weight and the geometry of the system, as it should be designed lightweight and near the body without overhanging structures. That includes the positioning of the major weight near the centre of the body.

An easy **handling and usability** include a short time in taking the system on and off without any help. Regarding the adaptability to different users, the exoskeleton has to be adjusted to fit the anthropometries of the user and needs to be adjusted to guarantee the alignment of body and system joints. A user control panel is mandatory to switch on/off and to regulate the required supporting force. The entirety adjustment functions should be characterized by their simplicity and intuition in use.

The **compatibility** of the supporting device is necessary due to the setting of nursery care. Therefore, it has to be compatible to the working clothes, environment and equipment. A protection class of minimum IP44 is obligatory in order to protect users and surrounded persons from the drive system or other moving parts.

Maintenance is preferred as low as possible. Nevertheless, with daily use, wear and defective parts are not completely unavoidable. An easy repair can be provided by a modular system with interchangeable parts.

The **adaptability** to individual and context specific requirements must be taken into account. Therefore, a software based differentiation of various contexts like dynamic tasks when lifting patients or static demands resulting from holding and walking with the patient must be distinguished. Moreover, adjustment to the magnitude of supporting force must be enabled e.g. with a rotary knob to fulfil individual demands for physical relief.

D. Design Concept

Exoskeletons are usually distinguished regarding the materials used for the structure in rigid exoskeleton and soft exosuits. Whereas rigid devices are primary made out of material with high stiffness like metal and carbon, soft exosuits comprises mostly fabrics and flexible plastics. Due to the utilized material, the advantages of exosuits are a comfortable and lightweight wear, while rigid systems beneficial in transferring higher loads by simultaneous redirecting compression forces from body into the system. In contrast, textiles are suitable only to transfer tensile forces but no compression forces, which could lead to additional strain to joints and passive body tissue like discs and cartilage. Regarding the mechanical and kinematic structure of rigid systems, the movements are primary possible at the predefined joints of the system, whereas soft exosuits allows more wide-ranging degrees of freedom without restricting user movements. Nevertheless, the goal is to combine the advantages of both types into a hybrid exoskeleton with high wearing comfort due to soft fabrics at the body-system interfaces as well as lightweight and stiff material to transfer forces without harmful effects to the body.

A second distinctive feature between exoskeletal systems are the used elements for actuation. An active device (powered) uses actuation modules like electric motors, pneumatic cylinders or linear actuators with additional energy source and controlling unit. Contradicting, passive devices (unpowered) uses elastic materials like rubber band or springs, which are storing energy that is introduces by the user during preceding movements. Due to the fixed torque profile, it is not possible to vary the given support regarding different needs like various lifting techniques or strain magnitudes resulting from different load weight. Moreover, there is no automatic deactivation when supporting torque is not intended like bending forward without extern stress or for ancillary tasks.

The fundamental design concept of our exoskeleton is an actively controlled system that should comprises the advantages of soft and rigid structures. It is characterized by four main modules of the basic structure and the involved drive system and sensors (see FIGURE 2).

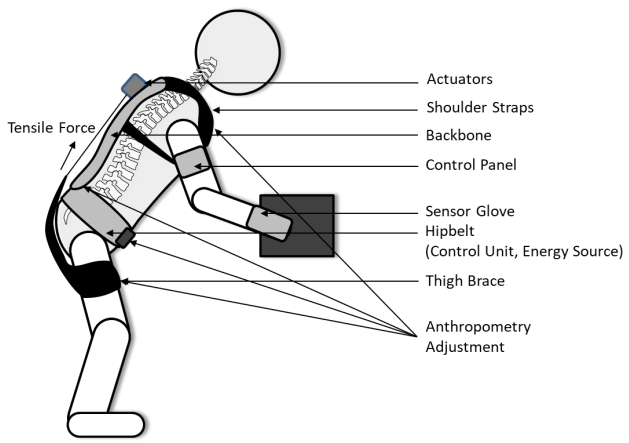

FIGURE 2. CONCEPT OF EXOSKELETON STRUCTURE.

The exoskeleton is built likewise a backpack. The **shoulder straps** act as human-system and system-system interface. They transmit the force to the upper body due to high stiffness in direction of force by simultaneously cushion the forces vertical to the body to maintain high wearing comfort. Additionally, the actuators and a backbone force transmitting structure are connected with the straps.

An artificial **backbone** made of aluminum is connected between the shoulder straps and a hip belt to transmit the supporting force parallel to the human spine. The backbone is flexible in direction of flexion and extension but does not allow compression upright to the spine. It is important to redirect forces that are not acting in the direction of the intended assisted degrees of freedom to avoid additional compressive forces to joints and lumbar discs.

In equivalence of the shoulder straps, the **hipbelt** is connected with the end of the backbone and transmit forces like a backpack onto the hip bones. Furthermore, an interface guides a steel cable that transmit the force from the actors at the upper back to the thighs. Small integrated pockets contain the energy source and the control unit. To make anthropometric adjustments, the shoulder straps, backbone and hipbelt comprise corresponding elements.

To generate a hip joint moment, the force is applied to a **thigh brace** around the left and right leg. A moment arm of approximate 0.1 m is built by the distance from the hip center of rotation to the point where the brace makes contact to the posterior waist [23].

In addition to the basic structure, an **actuation** module actively actuate the exoskeleton by two electrical EC-Motors, where a steel cable is connected to a pulley. An IMU **Sensor** Net determine the body postures to deliver input signals for the control unit to activate the drive system in order to support the upper body extension during lifting tasks. To recognize the load when grapping a patient, a **glove** with integrated pressure sensors is planned as an auxiliary to improve the control algorithm through the input signals.

III. MODELLING CONCEPT

Modelling is conducted with the goal to develop insights in influences on the function of the lower back exoskeleton. Eventually, a mathematical approach can lead to an integrated design methodology where relevant influences and dependencies of, e.g., the control law and the drive system can be optimized in an early phase of product development.

The interaction between human and exoskeleton is modeled as differential algebraic equation. A combined model, comprising of a human multibody model, an exoskeleton multibody model and an electromagnetic drive system model is suggested. The combined model is depicted in FIGURE 3. As can be seen, the human multibody model consists of rigid bodies, called body segments. They are connected by ball joints and are arranged in a tree structure. The root joint is at the hip body segment, which is highlighted in FIGURE 3.

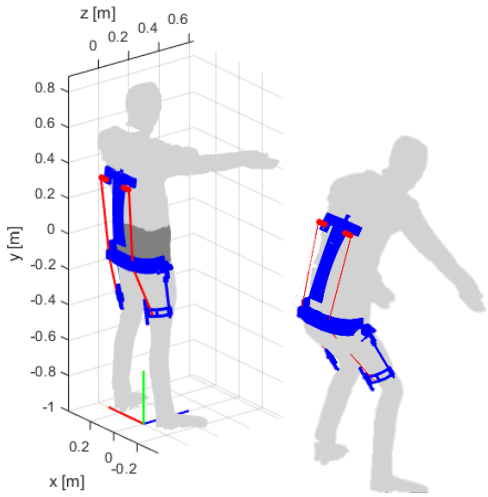

FIGURE 3: COMBINED MODEL IN FINAL (LEFT) AND INITIAL (RIGHT) POSE.

In the following, the equations for this model are mostly presented without extensive explanation. Relevant topics in multibody modeling are reviewed in [24]. The derivation of the human multibody model from 3D body scans is presented in [25]. For modelling of electromagnetic actuators, it is referred to [26] and [28]. The combined model comprises of the equations

$$\begin{bmatrix} M_{hum}(q_{hum}) & 0 \\ 0 & M_{exo}(q_{exo}) \end{bmatrix} \begin{bmatrix} \ddot{q}_{hum} \\ \ddot{q}_{exo} \end{bmatrix} + \begin{bmatrix} k_{hum}(q_{hum}, \dot{q}_{hum}) \\ k_{exo}(q_{exo}, \dot{q}_{exo}) \end{bmatrix} - C^T \lambda = \begin{bmatrix} g_{hum}(q_{hum}) \\ g_{exo}(q_{exo}) \end{bmatrix} + \begin{bmatrix} \tau_{hum} \\ \tau_{exo} \end{bmatrix} \tag{1}$$

$$c(q_{hum}, q_{exo}) = 0 \tag{2}$$

$$C = \frac{\partial c}{\partial q} \tag{3}$$

$$\tau_{exo}^T = \begin{bmatrix} 0 & .. & \frac{\partial \Psi_{m,j}^T}{\partial q_{drive,j}} i_j & .. \end{bmatrix} \tag{4}$$

where (1) represents the multibody dynamics, (2) and (3) the coupling constraints that are necessary to join both multibody models and (4) the electromagnetic torque production.

In the equations, indices hum and exo stand for human resp. exoskeleton, M represents a mass matrix, k a vector for centrifugal and Coriolis forces, g a vector for gravitational forces, τ a vector for control forces and torques and q a vector for generalized coordinates. Further, c represents coupling constraints and λ a vector of Lagrange multipliers that enforce the coupling in equation (1) and represent interaction forces. The column vector $\Psi_{m,j}$ represents the flux linkage due to permanent magnet excitation, column vector i_j the current in each strand and $q_{drive,j}$ is the generalized coordinate of the rotor of electric drive j. As the electric drives are part of the exoskeleton, the generalized coordinate $q_{drive,j}$ is part of the vector for exoskeleton generalized coordinates q_{exo}.

For model-based analysis of the human system interaction, it is assumed that human generalized coordinates q_{hum} differ only negligibly with and without exoskeleton. This reflects the goal to minimally restrict human motion. With this assumption, inverse dynamics of the combined model, i.e. determination of human control torques τ_{hum} can be performed on tracking data gathered with unapplied exoskeleton. Hence, the interaction of human and exoskeleton is primarily evaluated regarding human control torques τ_{hum}.

A. Exoskeleton Drive System Model

As the drive system is regarded in particular, modelling of it is introduced more thoroughly. Each electric drive is able to generate the generalized force

$$\tau_{drive,j} = \left(\frac{\partial \Psi_{m,j}^T}{\partial q_{drive,j}} \right) i_j. \tag{5}$$

As shown in equation (5), the current-force relation of the drive system is modeled. The vector of flux linkages due to permanent magnet excitation is

$$\Psi_{m,j}^T = \hat{\Psi}_{m,j} \begin{bmatrix} cos(p_j a_j) & cos(p_j b_j) & cos(p_j c_j) \end{bmatrix} \tag{6}$$

$$a_j = q_{drive,j}, b_j = q_{drive,j} - \frac{2\pi}{3}, c_j = q_{drive,j} + \frac{2\pi}{3} \tag{7}$$

where p_j is the number of pole pairs of the electric drive j and $\hat{\Psi}_{m,j}$ is the fundamental amplitude flux linkage due to permanent magnet excitation. The dynamic system from voltages to current is not regarded at this point. Further note, that with this model only permanent magnet synchronous machines with three phases and without reluctance, ripple torque and iron saturation can be modeled.

For a given drive torque $\tau_{drive,j}$, the choice of strand current vector i_j is ambiguously. To reproduce reality adequately, it is leaned on the commonly used procedure of field oriented control [26], [27]. A transformation is applied on equation (5), such that

$$\tau_{drive,j} = \left(\frac{\partial \tilde{\Psi}_{m,j}^T}{\partial q_{drive,j}} \right) i_{dq0,j} \tag{8}$$

$$\frac{\tilde{\Psi}_{m,j}^T}{\partial q_{drive,j}} = \left(\frac{\partial \Psi_{m,j}^T}{\partial q_{drive,j}} \right) K^{-1} = \begin{bmatrix} 0 & \hat{\Psi}_{m,j} p_j & 0 \end{bmatrix} \tag{9}$$

$$K^{-1} = \begin{bmatrix} cos(p_j a_j) & -sin(p_j a_j) & 1 \\ cos(p_j b_j) & -sin(p_j b_j) & 1 \\ cos(p_j c_j) & -sin(p_j c_j) & 1 \end{bmatrix} \tag{10}$$

with $i_{dq0,j}$ as the transformed current vector. It can be seen easily, that the torque $\tau_{drive,j}$ only depends on the second component $i_{q,j}$ of the transformed current vector. Hence, in practice, strand currents i_j are chosen such that only $i_{q,j}$ is created. A mapping from drive torque $\tau_{drive,j}$ to strand currents i_j can now be established as,

$$i_j^T(\tau_{drive,j}) = -\begin{bmatrix} sin(p_j a_j) & sin(p_j b_j) & sin(p_j c_j) \end{bmatrix} \sqrt{2} \hat{\imath}_{drive,j} \tag{11}$$

$$\hat{\imath}_{drive,j} = \frac{\tau_{drive,j}}{\sqrt{2} \hat{\Psi}_{m,j} p_j} \tag{12}$$

where $\hat{\imath}_{drive,j}$ is the effective current amplitude of electric drive j.

B. Static model

Derivation of equations (1) to (4) in symbolic form appears to be burdensome. To be able to gain insight in the system

behavior in a simple way, a static model is derived preliminary. In the model equations (1) to (4), equation (1) is substituted with

$$-C^T\lambda = \begin{bmatrix}\tau_{hum}\\ \tau_{exo}\end{bmatrix} \tag{13}$$

while the other equations remain the same.

This model covers the relation from drive currents in the exoskeleton i_j to torques in the human body τ_{hum} under neglection of dynamic influences, e.g. inertia of the drive system. As no derivatives appear in these equations, it is purely algebraic.

To solve the static model equations (13) and (2) - (4) for the human control torques τ_{hum} with a given trajectory of human generalized coordinates q_{hum} and drive currents i_j, it is required to solve equation (2) for exoskeleton generalized coordinates q_{exo}. Subsequently, equation (13) is solved for Lagrange multipliers λ and human control torques τ_{hum}.

IV. Model-Based Evaluation

In the following, the static model is applied to analyze the effects of a given control law for the generalized control forces on the exoskeleton, i.e.

$$\tau_{drive,1} = \tau_{drive,2} = -0.035\, sin(q_{hum,7}). \tag{14}$$

The trajectory for human generalized coordinates q_{hum} is generated artificially. Both start (lower) and end pose (raised) are depicted in FIGURE 3. In future applications, it will be gathered from motion tracking. The component $q_{\text{hum},7}$ refers to the rotation of the hips body segment around the x-axis.

The fundamental amplitude flux linkage due to permanent magnets $\hat{\Psi}_{\text{m},j}$ is taken from the data sheet of the preliminary applied drive system (Maxon EC-4 Pole 22 24V with Maxon GP22 HP 62:1, Maxon Motor AG, Sachseln, Switzerland). Further, kinematic relations for the exoskeleton are known from manufacturing data.

It is shown exemplarily, how the combined model allows the derivation of dimensioning quantities for the electric drive system. Further, a first approach for the analysis of the human system interaction is presented. Hence, first steps towards an integrated design methodology are demonstrated.

Firstly, trajectories for actuator rotor generalized coordinates $q_{\text{drive},j}$ can be derived. Numerical differentiation with respect to time yields the rotational speeds $\dot{q}_{\text{drive},j}$, which is a relevant dimensioning quantity. Further, drive system modelling allows the calculation of drive currents i_j resp. their effective amplitude $\hat{\imath}_{\text{drive},j}$. Speed and current quantities are plotted over time for electric drive 1 in FIGURE 4. These results indicate that the drive system was chosen correctly, such that it can operate within its specified limits.

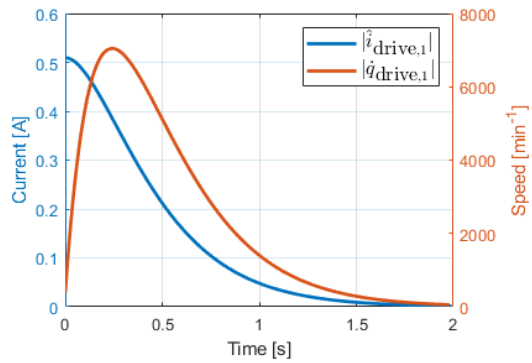

FIGURE 4: ELECTRIC DRIVE CURRENT AND SPEED FROM STATIC MODEL.

Another goal is the estimation of influences on the human. In a first approach, human control torques $\boldsymbol{\tau}_{\text{hum}}$ are considered for this estimation. In context of the static model, they ensure balance of momentum and hence represent the influence of the exoskeleton. In FIGURE 5, trajectories for human control torques are presented.

Human control torques around local x-axes for the right and left hip-upper leg joint $\tau_{\text{hum},43}$ resp. $\tau_{\text{hum},52}$ are regarded. Further, human control torque around global x-axis for the root-hip joint $\tau_{\text{hum},4}$ is regarded.

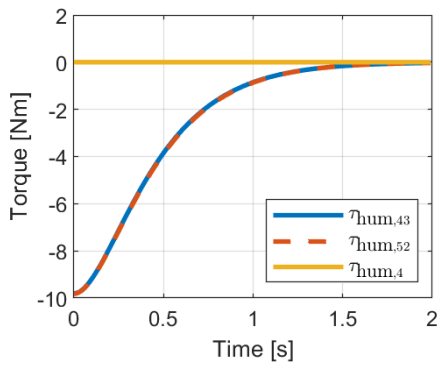

FIGURE 5: HUMAN CONTROL TORQUES DUE TO EXOSKELETON.

Regarding the control torques $\tau_{\text{hum},43}$ resp. $\tau_{\text{hum},52}$, the results indicate a support torque of approx. -10 Nm that decreases with time. Surprisingly, the control torque $\tau_{\text{hum},4}$ is zero. The reason for this is the model structure. As described in Section III, the human multibody model consists of a tree structure, beginning at the hips segment. This renders the forces and torques from the lower part of the exoskeleton to cancel the forces and torques from the upper parts of the exoskeleton in the hips segment. To circumvent this, additional bearing loads b are introduced at the feet. Thus, equation (*13*) is modified again to

$$-C^T\lambda = \begin{bmatrix}\tilde{\tau}_{\text{hum}}\\ \tilde{\tau}_{\text{exo}}\end{bmatrix} + b. \tag{15}$$

Bearing loads b are chosen such that they cancel the control torques $\tau_{\text{hum},43}$ and $\tau_{\text{hum},52}$. Hence, forces and torques from the lower part of the exoskeleton are now transmitted to the feet and canceled by corresponding bearing loads. In FIGURE 6, a trajectory for control torque $\tilde{\tau}_{\text{hum},4}$ is presented. In case of introduced bearing loads, a support torque for the hips body segment of approx. 17 Nm, decreasing with time is observed.

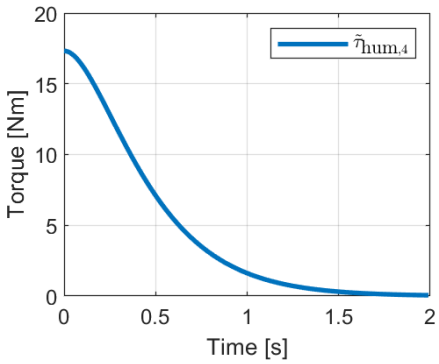

FIGURE 6: QUANTIFICATION OF EXOSKELETON EFFECTS.

The static model does not cover any influences of human masses and inertias. Thus, with this model, it is not possible to find a relation between the load inside the human body during a movement and its reduction due to the exoskeleton. As already mentioned, it is at this point aimed at demonstrating an approach for an integrated design rather than conducting a complete dimensioning.

V. CONCLUSION

This paper provides an overview of the current course of the dtec.bw project KIKU. The approach to develop a hybrid exoskeleton for back support in the setting of nursing care was given. To address the special requirements of care workers, including prevalent tasks like different patient lifting techniques, a preliminary experiment demonstrate the feasibility of biomechanical assessment using measurement devices for three-dimensional kinematic and electromyography analyses. As an example, the Scoot Pivot Transfer Technique shows a high demand of muscle activation and spine compression force up to 80 % MVC and 10 kN, respectively. During working routine and repetitive lifting tasks, this may lead to high stress to the musculoskeletal system, resulting in low back pain and disorders like lumbar disc herniation. To lower the physical strain during lifting tasks, an active driven exoskeleton will be developed by using rigid materials to transmit forces and soft textiles to maintain a high wearing comfort. As first step, a concept design was made, that depicts the main structure of the system including crucial elements of human-system and system-system interfaces, as well as the actuation module and the envisioned path of force. Further, an approach for an integrated design methodology, focusing on drive technology was presented. With a combined model of human, exoskeleton and exoskeleton drive system it is possible to evaluate the requirements of individual motion on dimensioning quantities (e.g. rotational speed) of the drive system. For an exemplary motion, it can be seen, that the requirements are fulfilled by the preliminary selected drive system. Lastly, an outlook on the model-based analysis of human system interaction was explained.

The next step is the construction of a first prototype of the concept design, taking into account requirements such as kinematics, usability, safety or wearing comfort to achieve a high level of acceptance. The physical effects of the exoskeleton on the body will be examined using similar biomechanical methods that has been presented.

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