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Human Sit-to-Stand Transfer Modeling for Optimal Control of Assistive Robots

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Abstract— **Sit-to-stand (STS) transfers are a common human task which involves very complex sensorimotor processes to control the highly nonlinear musculoskeletal system. In this paper, typical unassisted and assisted human STS transfers are formulated as optimal feedback control problem that finds a compromise between task end-point accuracy, human balance, jerk, effort, and torque change and takes further human biomechanical control constraints into account. Differential dynamic programming is employed, which allows taking the full, nonlinear human dynamics into consideration. The biomechanical dynamics of the human is modeled by a six link rigid body including leg, trunk and arm segments. Accuracy of the proposed modelling approach is evaluated for different human healthy subjects by comparing simulations and experimentally collected data. Acceptable model accuracy is achieved with a generic set of constant weights that prioritize the different criteria. The proposed STS model is finally used to determine optimal assistive strategies to be performed by a robotic mobility assistant suitable for either a person with specific body segment weakness or a more general weakness.**

I. INTRODUCTION

The rapidly ageing society and the continuous decrease of nursing specialists call for new assistive devices that fit elderly and patient demands. Human sit-to-stand (STS) transfers are a frequently exercised daily activity, which highly influences the quality of life of people who are not able anymore to accomplish normal STS transfers due to a specific or more general muscle weakness.

Only few assistive robotic devices focused on supporting human STS transfers so far and their control can be grouped into three categories: motion control, force control and switching control.

The group of Hirata and Kosuge presented different types of walking helpers that use a basic admittance controller to control the motion of the platform during STS transfers to take over e.g. a certain amount of the knee torque [1]. In [2], [3] sit-to-stand transfers are guided by the trajectory of a support plate mounted on the developed robot called SMW by two predefined trajectories. Médéric and Pasqui developed a mobility assistant equipped with 2 DoFs actuated handles that support patients in STS transfers. They fitted pre-recorded hand paths with cubic splines and tested a series of pre-parametrized trajectories [4].

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Force control was employed by Médéric and Pasqui who evaluate the Zero Moment Point (ZMP) for a simplified human model and control the interaction force between user and robot to stabilize the configuration. They solved an optimization problem that minimizes the interaction force taking ZMP-based balance constraints into account [5].

Finally, also switching controllers have been investigated. A new STS rehabilitation system consisting of a 3 DoF support pad that the patient must lean on during STS transfers was proposed in [6]. Analyzing the different phases of STS movements by means of multi-body computer simulations, they realized an admittance controller with force reference implementing damping control for the lifting body phase and compliant impedance control for all other phases while using a pre-computed reference trajectory based on real human STS transfers [7]. Pasqui again presented a fuzzy controller to ensure stability of the patient during assisted sit-to-stand transfers [8]. They subdivided the sit-to-stand transfer into several phases and defined fuzzy rules that evaluate the center of pressure and the horizontal component of the handle force to guarantee stability for the patient by switching between controllers implementing different variations of admittance control.

All these STS assistive strategies hardly incorporate any computational model of natural STS transfer motions to drive the robotic mobility assistant. So far STS transfers were mainly studied and analyzed in hypothesis-driven experiments, which led to a considerable amount of findings summarized in reviews like [9] and [10]. Chair, subject and strategy-related determinants of STS-transfers have been investigated by analyzing measured video, motion capture, force plate, force-torque, accelerometer, dynamometer, and EMG data as well as non directly measurable data like joint torques determined with the help of inverse dynamic models.

While this way a huge variety of data has been analyzed by various researchers, only few computational models to study human STS transfers have been presented so far. In [11], [12] authors investigated an optimal LQR formalism in the context of an optimal tracking controller combined with a fuzzy biomechanical model, which interpolates between two linearized models of the nonlinear four segment/bipedal dynamics around the sitting and standing position. They optimized physiological costs when tracking a predefined ankle, knee, hip, and pelvis reference trajectory [11], [13]–[16].

In [17] authors employed dynamic optimization to determine optimal STS trajectories by considering a cost function that minimizes joint torques, torque change and the difference between left and right ground reaction forces based on sequential quadratic programming (SQP). They determined different weights of the single criteria for unassisted STS transfers of healthy subjects as well as amputees, but did not study assisted STS transfers. Moreover, critical balance criteria were not considered in their approach. Further, SQP is considered a method of local optimization and thus, may lead to suboptimal solutions, while global methods based on the Hamilton-Jacobi-Bellmann equations and dynamic programming typically suffer from the curse of dimensionality. Both is problematic when considering biomechanical problems, as they are typically high-dimensional and involve model uncertainties [18].

Differential Dynamic Programming (DDP) and Iterative Linear-Quadratic Gaussian (ILQG) have been proposed in literature to overcome those aforementioned limitations. They solve the optimization problem by dynamic programming, and lead to feedback control laws. Both are methods based on Optimal Feedback Control (OFC) that have shown to be a powerful tool to study biological movements and interpreting human motor behavior [18].

In this paper we formulate first unassisted and then assisted STS transfers as optimal feedback control problems and solve them using an iterative optimal control approach to derive optimal robot trajectories. We employ DDP that iterativelly quadratically approximates the nonlinear system dynamics and the optimal cost-to-go function around the current trajectory. It takes physical control constraints like torque limitations into account, while stability-related criteria are considered in the cost function. The obtained model is evaluated for different human healthy subjects by comparing simulations and experimentally collected data. Finally optimal assistive strategies for subjects characterized by a specific or more general muscle weakness are studied, and optimal trajectories are derived which can be realized with robotic mobility assistants.

II. STS MODELING AS AN OPTIMIZATION PROBLEM

In the following subsections the STS transfer task is formulated as an optimal feedback control problem with a nonlinear cost function subject to control constraints. An approximative optimal control approach based on DDP [19], [20] is employed to allow for an efficient solving of this optimization problem.

A. Human Biomechanical Model

While a triple inverted pendulum has been widely studied as a simplified biomechanical model of the human in biomechanics and biomedical literature (e.g. [21]), in this paper a model consisting of five joints and six rigid bodies [1] involving foot, lower leg (shank), upper leg (thigh), trunk (torso and head), lower and upper hand is considered, which moves in the sagittal plane as shown in Fig. 1. The ankle, knee, hip, shoulder and elbow joint torques are used to control the motion of the model. The equations of motion are derived using the Euler-Lagrange method. The nonlinear dynamics of the biomechanical model is given by

$$\boldsymbol{M}(\boldsymbol{\theta})\ddot{\boldsymbol{\theta}} + \boldsymbol{C}(\boldsymbol{\theta}, \dot{\boldsymbol{\theta}}) + \boldsymbol{G}(\boldsymbol{\theta}) = \boldsymbol{\tau} + \boldsymbol{\tau}_{ext} = \boldsymbol{\tau}_{tot} \quad (1)$$

where $\boldsymbol{M}(\boldsymbol{\theta}) \in R^{5\times 5}$ is the positive definite symmetric inertia matrix, $\boldsymbol{C}(\boldsymbol{\theta}, \dot{\boldsymbol{\theta}}) \in R^5$ the vector of Coriolis and centripetal forces, and $\boldsymbol{G}(\boldsymbol{\theta}) \in R^5$ the gravitational force vector, while $\boldsymbol{\theta} \in R^5$ refers to the joint angle vector with ankle ($\boldsymbol{\theta}_1$), knee ($\boldsymbol{\theta}_2$), hip ($\boldsymbol{\theta}_3$), shoulder ($\boldsymbol{\theta}_4$) and elbow ($\boldsymbol{\theta}_5$) angles, $\boldsymbol{\tau} \in R^5$ to the joint torques and $\boldsymbol{\tau}_{ext} \in R^5$ to the torque due to external assistive generalized forces applied to the human.

The equations can be written as first order dynamic system with $\boldsymbol{x} = [\boldsymbol{\theta}, \dot{\boldsymbol{\theta}}]^T \in R^{10}$

$$\dot{\boldsymbol{x}} = \boldsymbol{f}(\boldsymbol{x}, \boldsymbol{\tau}) = \begin{pmatrix} \dot{\boldsymbol{\theta}} \\ -\boldsymbol{M}(\boldsymbol{\theta})^{-1}(\boldsymbol{C}(\boldsymbol{\theta}, \dot{\boldsymbol{\theta}}) + \boldsymbol{G}(\boldsymbol{\theta}) - \boldsymbol{\tau}_{tot}) \end{pmatrix}. \quad (2)$$

Considering $\boldsymbol{F} \in R^m$ external generalized forces applied to a specific point on the human model, and $\boldsymbol{J}_k(\boldsymbol{\theta}) \in R^{m\times 5}$ the Jacobian associated to this point, then $\boldsymbol{\tau}_{ext}$ is given by

$$\boldsymbol{\tau}_{ext} = {\boldsymbol{J}_k}^T(\boldsymbol{\theta})\boldsymbol{F}. \quad (3)$$

Please note that in the unassisted case, we adopt a simplified version of this model controlled by three joint torques (hand segments not actuated). Moreover, in case of assisted STS transfers we study two different supporting points based on the level of the patient's demand advised by nursing specialists: i) under the arms close to the shoulder and ii) at the hands.

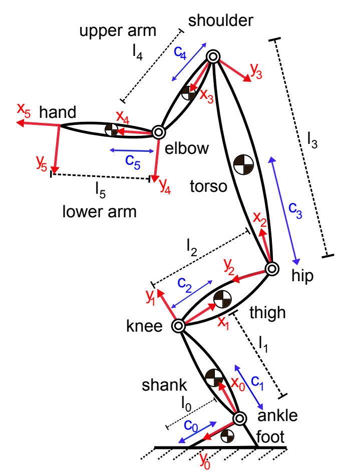

Fig. 1. Rigid body biomechanical model of the human, l_i and c_i represent the length and center of gravity of the segments while x_i,y_i are the reference frames attached to each joint.

B. Balance and Task End-Point Accuracy Criteria

To determine human balance and postural stability during STS transfers, the virtual zero moment point (for abbreviation ZMP) is evaluated [22]. ZMP is a point on ground level where the pressure between the foot and ground is replaced by a force which can balance active forces acting on the human dynamics during the motion. ZMP can be computed from the vertical component of contact moment $\boldsymbol{T}$ and the horizontal component of contact force $\boldsymbol{F}$ as follows:

$$\boldsymbol{p}_{zmp} = \frac{\boldsymbol{T}}{\boldsymbol{F}}. \quad (4)$$

[1] Stiffness of the human segments, specially arms, is neglected in the model assuming that the human willingly accomplishes the STS task and thus, reacts very stiff to external forces.

Task end-point accuracy is determined using the center of mass (COM):

$$\boldsymbol{p}_{com} = \frac{\sum_{i=1}^{6} \boldsymbol{m}_i \hat{\boldsymbol{p}}_i}{\sum_{i=1}^{6} \boldsymbol{m}_i}, \tag{5}$$

where $\boldsymbol{m}_i$ is the mass of the i^{th} segment and $\hat{\boldsymbol{p}}_i$ the position of its center of gravity.

C. Formulation of Optimization Problem

The STS optimal control problem is formulated as follows: The human sitting position with zero joint velocities is considered the initial state at time $t = 0$ and the position of the COM in the steady-state standing position is considered the desired final state of the system at time $t = T$. The main goal is to find a control law $\boldsymbol{\tau}^* = \pi(x, t)$ that stays within joint torque limits and that drives the system states smoothly from the initial to the final configuration while minimizing a given cost function.

We propose a cost function for the STS transfer task, which aims for a compromise between task end-point accuracy, human balance, effort, jerk, and torque change. The minimum effort term tries to achieve a minimum time response as joint torques are much lower in the standing than in the sitting configuration (when neglecting the interaction forces with the chair). The minimum torque change term assures a smooth control, while the minimum jerk term improves smoothness of the resulting motion. As humans automatically try to stabilize their movement patterns, the human balance criteria are included as well. The following combination of criteria is used to model the STS transfer task:

$$\boldsymbol{J}_{total} = \boldsymbol{J}_{final}(\boldsymbol{x}) + \int_0^T \Big(\sum_{i=1}^{6} \boldsymbol{C}_i\Big) dt \tag{6}$$

with

$$\begin{aligned}
\boldsymbol{J}_{final} &= \boldsymbol{J}_{f1} + \boldsymbol{J}_{f2} \\
\boldsymbol{J}_{f1}(\boldsymbol{x}(T)) &= |\boldsymbol{p}_{com}(\boldsymbol{x}(T)) - \boldsymbol{p}_{com}^{tar}|^2_{\boldsymbol{W}_{f1}} \\
\boldsymbol{J}_{f2}(\boldsymbol{x}(T)) &= |\dot{\boldsymbol{\theta}}(T)|^2_{\boldsymbol{W}_{f2}} \\
\boldsymbol{C}_1(\boldsymbol{x}(t), \boldsymbol{\tau}(t)) &= |\boldsymbol{p}_{zmp}(\boldsymbol{x}(t), \boldsymbol{\tau}(t)) - \boldsymbol{p}_{zmp}^{max}|^2_{\boldsymbol{W}_1} \\
&\quad + |\boldsymbol{p}_{zmp}^{min} - \boldsymbol{p}_{zmp}(\boldsymbol{x}(t), \boldsymbol{\tau}(t))|^2_{\boldsymbol{W}_1} \\
\boldsymbol{C}_2(\boldsymbol{x}(t)) &= |\dddot{\boldsymbol{\theta}}(t)|^2_{\boldsymbol{W}_2} \\
\boldsymbol{C}_3(\boldsymbol{\tau}(t)) &= |\boldsymbol{\tau}(t)|^2_{\boldsymbol{W}_3} \\
\boldsymbol{C}_4(\boldsymbol{\tau}(t)) &= |\dot{\boldsymbol{\tau}}(t)|^2_{\boldsymbol{W}_4} \\
\boldsymbol{C}_5(\boldsymbol{x}(t)) &= |\max(0, \boldsymbol{x}(t) - \boldsymbol{x}_{max})|^2_{\boldsymbol{W}_5} \\
&\quad + |\max(0, \boldsymbol{x}_{min} - \boldsymbol{x}(t))|^2_{\boldsymbol{W}_5} \\
\boldsymbol{C}_6(\boldsymbol{F}(t)) &= |\boldsymbol{F}(t)|^2_{\boldsymbol{W}_6}
\end{aligned}$$

and $\boldsymbol{W}_{f1}$, $\boldsymbol{W}_{f2}$ weighting matrices for the terminal costs evaluated at the desired human COM position $\boldsymbol{p}_{com}^{tar}$ in a standing position at time T with zero joint velocities, $\boldsymbol{W}_1$ the weighting matrix for the human balance term that aims to satisfy $\boldsymbol{p}_{zmp}^{min} \le \boldsymbol{p}_{zmp}(\boldsymbol{x}, \boldsymbol{\tau}) \le \boldsymbol{p}_{zmp}^{max}$, and $\boldsymbol{W}_2 = diag(w_{2a}, w_{2k}, w_{2h})$, $\boldsymbol{W}_3 = diag(w_{3a}, w_{3k}, w_{3h})$, $\boldsymbol{W}_4 = diag(w_{4a}, w_{4k}, w_{4h})$ the weighting matrices for the human jerk, effort and minimum torque change terms respectively, where the term $diag(.)$ represents a diagonal matrix[2] and $|\mathrm{v}|^2_{\boldsymbol{W}} = \mathrm{v}^T \boldsymbol{W} \mathrm{v}$. The weighting matrix $\boldsymbol{W}_5$ is responsible for the human joint position and velocity boundaries $\boldsymbol{\theta}_{min} \le \boldsymbol{\theta}(\boldsymbol{t}) \le \boldsymbol{\theta}_{max}$ and $\dot{\boldsymbol{\theta}}_{min} \le \dot{\boldsymbol{\theta}}(\boldsymbol{t}) \le \dot{\boldsymbol{\theta}}_{max}$. The weighting matrix $\boldsymbol{W}_6$ is considered to minimize the interaction forces exchanged between assistive robot and human and therefore is considered equal to zero for the case of unassisted human STS modeling. This cost function is subject to constraints like the system dynamics formulated in (2) and control constraints, i.e. $\boldsymbol{\tau}_{min} \le \boldsymbol{\tau}(\boldsymbol{x}, \boldsymbol{t}) \le \boldsymbol{\tau}_{max}$.

D. Optimal Feedback Control

We solve this optimal control problem using Differential dynamic programming (DDP) first proposed in [20]. This approach iteratively, quadratically approximates the costs and the nonlinear system dynamics around the current trajectory. Then, an approximately optimal control law is found by designing an affine controller for the approximated system that enforces formulated control constraints.

For our specific STS transfer problem we consider pure gravity compensating forces as an initial guess of the control sequence, which is then iteratively improved by the algorithm with respect to the formulated cost function.

The algorithm shows quadratic convergence in the vicinity of a local minimum, similar to Newton's method [23] and returns the optimal control and the corresponding state sequences.

III. VALIDATION OF UNASSISTED STS MODEL

A. Data Capturing

In order to test the quality of the STS model against real measurements, we performed STS transfer experiments with three healthy male subjects. Their body measurements are listed in Table I.

TABLE I
ANTHROPOMETRIC DATA OF SUBJECTS

subject	age	weight $[kg]$	height $[m]$
S1	26	74	1.72
S2	25	80	1.84
S3	29	70	1.83

Participants were instructed to perform a few practice trials in order to find a comfortable feet placement. They were asked to keep their feet fixed to the ground, their arms crossed over the chest, and their upper body straight during the whole experiment (see Fig. 2). Each subject was asked to repeat five STS transfers at a natural speed while the seat heights were adjusted to fit the lower leg length, and the initial upper body inclination was decided to be about 30 degrees, where zero represents the upright trunk position. An armless office chair was adjusted vertically for each subject.

An Xsens MVN inertial motion capture system [24] was used for full-body human motion capture. Kinematic data

[2]Please note that a same value on diagonal elements is considered for each weighting matrix which is not represented with $diag(.)$.

including segment position and orientation, velocity and acceleration were captured with a sampling rate of 120 Hz.

Every STS transfer was assumed to start from the static configuration in the sitting position and to finish when the user arrives at the fully standing position with zero joint velocities. The average STS transfer movement time for all subjects was in the range of 1 to 2.5 seconds.

B. Evaluation of STS Model

Next, the proposed optimal control approach for simulating natural STS transfers was evaluated by comparing simulations with measurements. Captured data was preprocessed to remove noise using a low-pass filter with 5 Hz cut-off frequency. Parameters of the biomechanical model were estimated for healthy subjects using regression formulas provided in literature [25], see Table II for results of subject one. For each STS transfer, the joint angles, torques and the COM trajectories were estimated based on the human inverse dynamics. Under the aforementioned experimental conditions of initial upper body inclination and chair height, simulated STS trajectories were derived using the optimization approach described in Sec. II.

TABLE II

ESTIMATED ANTHROPOMETRIC LIMB DATA FOR SUBJECT ONE

	length $[m]$	COG $[m]$	mass (half body) $[kg]$	inertia (half body) $[kg.m^2]$
foot	0.11	(0.115,0.01)	1.434	–
shank	0.414	0.257	3.346	0.0476
thigh	0.459	0.224	9.56	0.2431
trunk	0.736	0.431	22.66	2.6967

Simulation data was compared with captured data from the instance where subjects left the chair as we did not consider the effect of the chair support in our simulations. The joint configuration at this instance with zero velocity was used as initial condition for the optimization algorithm. We considered a sampling time of $\Delta t = 0.12$ s and a total simulation period of $T = 1.8$ s based on the mean value of the experimental STS transfer time observed in experiments.

TABLE III

WEIGHTING FACTORS OF COST FUNCTION

weights	assigned value
$\boldsymbol{W}_{f1}$	$w \times 10^5$
$\boldsymbol{W}_{f2}$	$w \times 10^2$
$\boldsymbol{W}_1$	6×10^5
$\boldsymbol{W}_2$	$10^{-6} \times diag(1,1,1)$
$\boldsymbol{W}_3$	$diag(8,0.5,0.4)$
$\boldsymbol{W}_4$	$10^{-3} \times diag(1,1,1)$
$\boldsymbol{W}_5$	10^6

The comparison of simulation results with the obtained measurement data indicates that the proposed weighted sum of minimum effort, minimum torque change, minimum jerk and balance control criteria can replicate recorded STS transfers quite well. The largest weighting factors were specified for the joint position and velocity limits ($\boldsymbol{W}_5$) to effectively remove unfeasible motions. To guarantee human balance (specifically for the case of assisted STS, see Sec. IV), the corresponding weighting factors ($\boldsymbol{W}_1$) were given a high priority too. Lower values were specified for the minimum jerk, minimum effort, and minimum torque change terms. Choosing the proper weight for the minimum jerk term ($\boldsymbol{W}_2$) was found to be a challenging task as small changes resulted in a rather high variation of simulated STS transfers; reducing the value resulted in a relative high velocity impulse close to the end of the STS transfer and increasing the value resulted in smoother trajectories with a comparable deceleration at the end of the motion. The weighting factors of the minimum effort term ($\boldsymbol{W}_3$) for the knee were selected smaller than for the hip and both were selected smaller than for the ankle as the highest and lowest contribution for a STS transfer were observed to come from the knee and ankle, respectively. Concerning the minimum torque change term ($\boldsymbol{W}_4$), the weighting factor was decided to be rather low since such low values resulted in smoother motions and control profiles while larger values produced non-human like behavior.

Final term conditions ($\boldsymbol{W}_{f1}$, $\boldsymbol{W}_{f2}$) in the cost function were also found to be a very important factor in the optimization. Selecting low values, no control in the sagittal plane was possible. On the other hand, very large values overruled all other factors in the cost function and thus, led to an immediate termination of the optimization as no improvement over iterations could be achieved.

Table III shows the found weighting matrices of the cost function (eq. 6) that led to a good correspondence of simulation and experimental data with w the subject's total weight[3].

A series of snapshots showing a STS transfer of one subject and corresponding simulation results are reported in Fig. 2. As can be observed the user leaves the chair while having almost 45 degree upper body inclination.

Representative simulations and measurements for the COM position, joint angles and joint torques are shown in Fig. 3. Simulations of the COM position were compared with captured experimental trajectories when the subject was asked for five STS transfers. For the sake of presentation, trajectories of the joint positions and torques for one repetition is depicted.[4] As can be seen all 3 joints as well as COM smoothly converge to their stable final configurations, which is well captured by the model. The initial errors for the user joint torques between simulations and corresponding experiments resulted mainly from neglecting the supportive chair effect, more specifically from neglecting initial subject velocities and accelerations at the instance of leaving the chair.

In order to validate the generalizability of the modeling approach, we investigated invariance of weighting factors

[3]Please note that no correlation analysis has been performed on other weighting factors of the cost function

[4]Although the subjects were asked to minimize variation, still non-negligible differences were observed, especially for initial upper body inclination and feet positions.

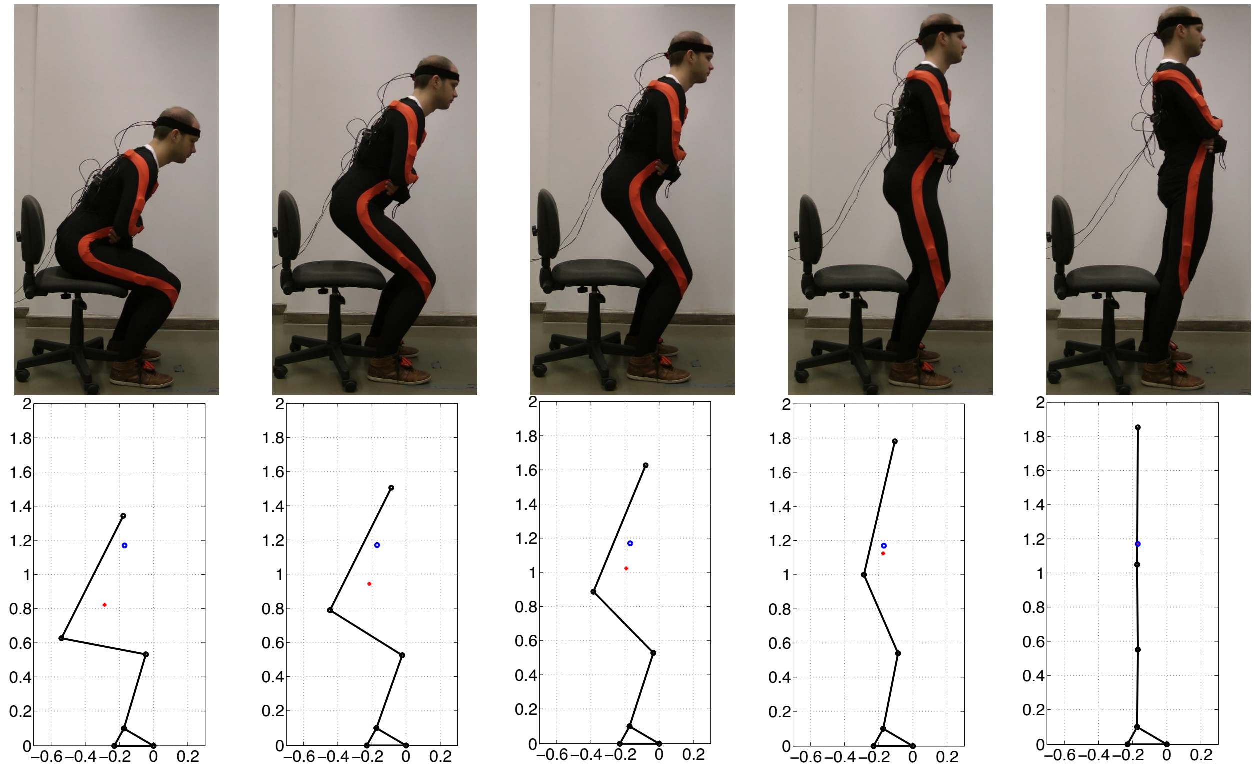

Fig. 2. Snapshots taken during a human STS transfer (first raw) and corresponding simulation results (second raw).

with respect to different subjects with a normal chair height. The user's COM position and joint torques observed during STS transfers and averaged over the 5 captured trials were chosen for comparison of simulation and experiments. A first comparison showed that similar STS strategies were selected by different subjects, but that the body weight strongly influences the final STS model performance. Thus, we defined the weighting matrices $\boldsymbol{W}_{f1}$ and $\boldsymbol{W}_{f1}$ of the final terms in proportion to the user weight w.

To provide a measure for the overall model accuracy, for all subjects the normalized integral of the error between experiments and simulation was computed as

$$e_v = \frac{\int |v_{exp}(t) - v_{sim}(t)|dt}{\int |v_{exp,max} - v_{exp,min}|dt}$$

where v_{exp} and v_{sim} refer to data in experiments and simulation respectively, $v_{exp,max}$, $v_{exp,min}$ the maximum and minimum value of experiments. This error was evaluated over the x and y components of the COM position (e_{comx}, e_{comy}) and the ankle, knee and hip torques ($e_{\tau a}, e_{\tau k}, e_{\tau h}$), see Table IV. Considering the complexity of the problem and the simplified assumptions for the human model, the errors are considerably low and illustrate an overall high agreement of model and measurements.

TABLE IV

NORMALIZED INTEGRATED ERROR BETWEEN SIMULATION AND EXPERIMENTS OF DIFFERENT USER'S STS TRANSFERS.

subject	e_{comx}	e_{comy}	$e_{\tau a}$	$e_{\tau k}$	$e_{\tau h}$
S_1	0.0383	0.1325	0.3434	0.2049	0.1334
S_2	0.1662	0.0962	0.3175	0.1391	0.1261
S_3	0.0275	0.0235	0.1663	0.1007	0.0507

IV. OPTIMAL STS ASSISTANCE

In the following we use the already introduced biomechanical model and optimization approach to calculate proper assistive forces for a robotic assistant that is supposed to support subjects in STS transfers. We implement assistive strategies that are tailored to the specific class and weakness of a certain subject.

In [26] a classification scheme for transfer assistance was proposed that considers the request for supervision, type of assistance and participation of targeted persons. Here we focus on the two classes of *maximal assist*, "the patient performs less than 25 % of the activity[5]", and *moderate assist*, "the patient performs at least 50 % of the activity". As proposed by nursing specialists, the most common techniques for assisting persons in STS transfers belonging to the maximal assist class foresee that the caregiver stands in front or slightly beside the person to be assisted, locks the knees or feet of the patient, grips the patient by his/her arms or armpits and lifts the person. Stronger patients belonging to the moderate assist class require less physical assistance, but more balance support. In this case, the caregiver stands in front of the patient, grasps the hands and applies forces to assist in the STS transfer, while simultaneously assisting in keeping the patient's balance.

Finally, the weakness may be either limited to specific segments of the body because of a certain disease or surgery,

[5] Activity refers to completion of one STS transfer.

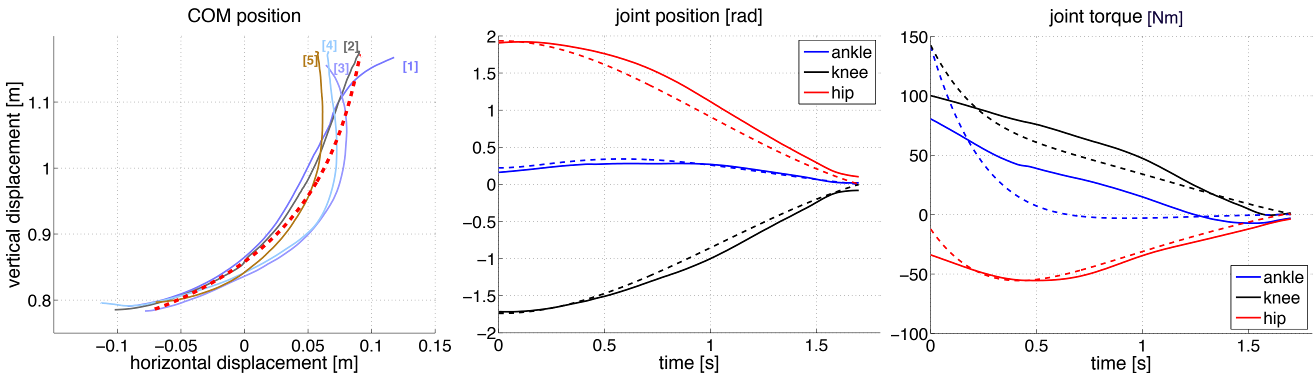

Fig. 3. Simulation and experimental results of the human COM positions (left), joint angles (mid), and joint torques (right) during STS transfers. Left: trajectories of the user's COM from five times repetition of STS under similar conditions represented (solid lines) and corresponding simulation result (dashed red). Middle, right: experimental results of the joint positions and torques (solid line) and simulation results (dashed).

(*case a*), or spread over multiple segments, (*case b*).

The weighting factors for minimization of assistive forces were considered equal to $diag\,(8, 8, 8)$ for all assisted STS transfer simulations.

A. STS Assistance for the Maximal Assist Class

For the maximal assist class the required assistance is typically applied to segments close to the patient's shoulder. Thus, effects of the shoulder and elbow joints were neglected in the biomechanical model. Considering the minimum torques needed to rise successfully without help [27], proper joint torque constraints ($\boldsymbol{\tau}_1 < 75$, $\boldsymbol{\tau}_2 < 75$, $\boldsymbol{\tau}_3 < 25$) and ($\boldsymbol{\tau}_1 < 40$, $\boldsymbol{\tau}_2 < 40$, $\boldsymbol{\tau}_3 < 40$) were introduced to simulate both aforementioned patients of (*case a*) and (*case b*) respectively. Vertical and horizontal external force components (F_x, F_y) as well as an angular momentum (M_z) were considered in the model. Figure 4 shows obtained trajectories of the user's COM, joint position, joint torques, and assistive force/moment supporting patients. As can be observed human weakness is compensated through proper external assistance.

B. STS Assistance for the Moderate Assist Class

For the moderate assist class we assumed that the patient is able to rigidly grasp a robotic device that assists in STS transfers. To simulate this case also the human arm has been considered in the biomechanical model. Higher joint torque limits compared to the maximum assist class and the two sorts of weaknesses were set ($\boldsymbol{\tau}_1 < 100$, $\boldsymbol{\tau}_2 < 100$, $\boldsymbol{\tau}_3 < 50$) for (*case a*) and ($\boldsymbol{\tau}_1 < 55$, $\boldsymbol{\tau}_2 < 55$, $\boldsymbol{\tau}_3 < 55$) for (*case b*). The user's COM, joint positions and torques, and required supportive force/momentum trajectories are shown in Fig. 4. As expected, the required external supportive force/moments are reduced in comparison to the maximal assist class.

V. Conclusion

We have presented an optimal feedback control formulation for the modeling of assisted and unassisted human STS transfers. Compared to previous work based on SQP approaches, we based our optimization on Differential Dynamic Programming that has been shown a powerful tool to study biological movements. It allows to obtain an optimal solution with respect to a defined cost function and considers the nonlinearity of the human biomechanics as well as physical constraints, which are naturally incorporated into the optimization framework. It further shows potential for future online implementation.

We showed that natural STS transfers could be achieved with the help of a cost function that linearly combines a series of factors including minimum effort, minimum jerk, minimum torque change taking further balance and task end-point accuracy into account. Validation of the proposed approach was performed for different subjects.

The model was extended with external forces and torques and optimal assistive STS strategies were determined considering two types of assistance classes and weaknesses. The resulting trajectories can easily be implemented on a robotic mobility assistant.

Future work will focus on testing these trajectories on our robotic mobility assistant, considering more complex human dynamic models with direct muscle control and three-dimensional models and an inverse optimal control approach for tuning the weighting factors systematically. Moreover, we aim for including chair support forces, which will require switching the model during optimization and the extension to elderly people which might lead to a different weighting of the optimization criteria. Ultimately, we aim for online implementation of optimally assisted human sit-to-stand transfers in cooperation with a robotic mobility assistant.

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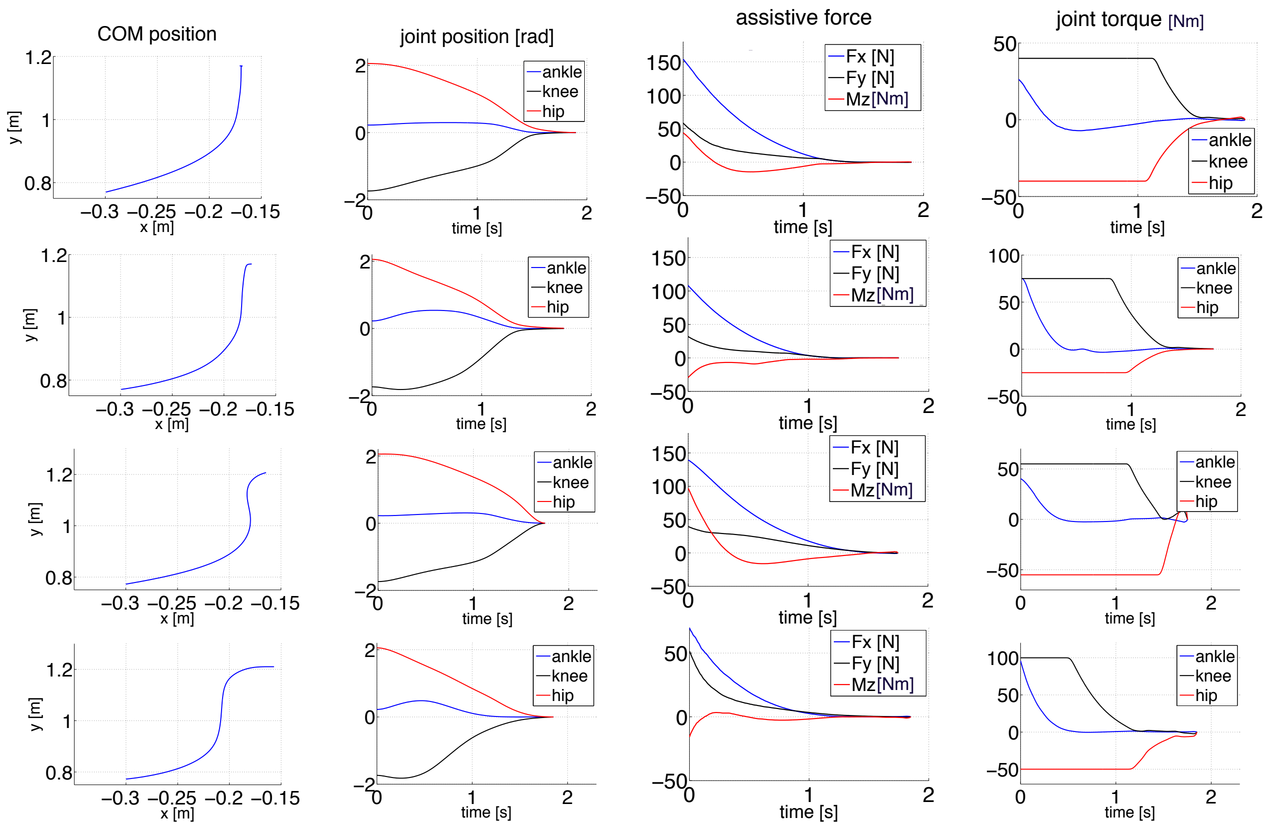

Fig. 4. Columns from left to right: simulation results of the human COM positions, joint positions, external assistive forces and moment, and joint torques during STS transfers for patients belong to the class of maximal assistance and having equal weakness in all joints (first row) and weakness in specific joints (second row), as well as for patients belong to the class of moderate assistance and having equal weakness in all joints (third row) and weakness in specific joints (fourth row).

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