Musculoskeletal Simulation Based Optimization of Rehabilitation Program

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Abstract— **Rehabilitation is a complex multifaceted process with complexity and variability that depends not only on human patients and/or specialized equipment but also on the nature of their functional interaction. Rapid and effective customization of the functional interactions between the patient and the rehabilitation device thus becomes critical for any rehabilitation program. Two principal dimensions govern the effectiveness of such functional interactions: geometric placement of user-device (ergonomics) and exercise selection and performance (regimen). In this paper, we discuss aspects of creation of a Virtual Design Environment, leveraging tools from musculoskeletal analysis, optimization, and simulation-based design, which will permit a therapist to rapidly evaluate and systematically customize various candidate rehabilitation programs. Specifically, this framework: (i) permits study of parametric performance variability due to ergonomic or regimen variability; and (ii) facilitates use of all design tools such as optimization to determine the best program. We illustrate various aspects of this customization using an illustrative case-study of a motor-rehabilitation haptic virtual driving environment.**

I. INTRODUCTION

EACH year, a growing number of people suffer from diminished motor capabilities, due to disease (muscular dystrophy and stroke), aging or simply disuse. There is considerable evidence which directly links functional recovery to the duration, frequency, regularity and intensity of the rehabilitation therapy [1, 2]. Many of these studies have confirmed that significant improvement is possible by following a dedicated therapeutic regimen (even several years after the initial incident). In almost all these cases, physical rehabilitation has typically entailed performance of structured repetitive physical exercises, typically with specialized equipment, in order to rebuild strength/functional capacity.

Rehabilitation is a complex multifaceted process [3] with complexity and variability that depends critically on the nature of the functional interaction between the human users and the specialized equipment. Innate variation of performance and function exist across individuals in a population, based on sex/age/race. However, beyond this, one can also witness considerable variability in muscle conditioning based on stage of disease/rehabilitation therapy. Similarly, there is variability in exercise programs available to help improve performance. Unassisted exercises have traditionally formed the mainstay of rehabilitation therapies, especially in home-based settings. However, issues of compliance with a specified therapeutic regimen can limit their overall utility. Hence, most effective rehabilitation programs employ machine-assisted exercises to improve the functional capacity (and strength). Machine-assistance also allows prescription of a greater variety of exercises ranging from isometric, isotonic or isokinetic contractions of the affected skeletal muscles.

Many of the newer Computerized Exercise Systems (CES) [4] possess numerous novel features that allow them to be adaptable to the users. This adaptability is realized by allowing *changes in geometry of user-device interaction* as well as *changing parameters of exercise performance*. On one hand, the geometry of user-device setup governs the biomechanical-interactions and thus becomes extremely vital in order to realize the benefits from the exercise equipment. On other hand, the newer CES also allow tailoring of specific rehabilitation/exercise regimen in terms of performance parameters (such as amplitude, velocity, frequency, resistance) that can be adjusted. It is beneficial to be able to create, test, evaluate and refine the rehabilitation program to be deployed on such CES by a Virtual Prototyping (VP) methodology [5, 6]. Virtual prototyping refers to functional simulation, quantitative performance analysis and iterative refinement of suitable products and processes in a virtual environment. By permitting realistic, accurate and quantitative testing of multiple intermediate models within a virtual environment, virtual prototyping, also known as Simulation-Based Design (SBD), has rapidly gained popularity and become a crucial part of most engineering design processes.

In this paper, we examine the application of this VP methodology to study and refine the interactions of a patient with the rehabilitation equipment and regimen completely in a virtual environment. This is realized by the integration of engineering support tools (such as visualization, musculoskeletal analysis, and optimization) and structured therapist involvement within a Virtual Design Environment (VDE) to facilitate the evaluation and the subsequent refinement of the of rehabilitation regimen in the presence of variability. Such framework leveraging computational, visualization, and decision support tools has immense value for rapidly performing what-if type analyses, or parameter-sensitivity-studies. Specifically, in this paper, we will focus on: (i) demonstrating the extent-of-variability that can be attributed to these two causes; (ii) develop a design tool to systematically

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study effects of variability in each; and ultimately (iii) enhance this design framework to allow an automated determination of the 'best' geometries and regimen. We will use a case-study of a driving simulator [7, 8] to illustrate these three aspects in more concrete terms.

The rest of this paper is organized as follows: In Section II, we discuss some of the critical challenges involved in creating the proposed design framework. Section III presents an overview of the various individual modules that come together to form the Virtual Design Environment. Many of these aspects are explained further in the context of an illustrative case-study of a motor-rehabilitation haptic virtual driving environment in Section IV. Finally, we conclude the paper with a summary and directions for future work in Section V.

II. CRITICAL CHALLENGES

The adoption of a computational-analysis paradigm is beneficial from the viewpoint of allowing us to think of a rehabilitation regimen in the form of a 'design problem' [9]. In such a setting, there is a clear emphasis on systematic generation, evaluation and elimination of design choices. SBD tools in engineering have capitalized on setting-up and solving such design problems by coupling parametric models with functional simulation tools and optimization methods.

The ability to perform functional evaluations depends on the availability of functional simulation tools. However, the effectiveness of the functional simulation tools is limited by the extent of capture of the underlying physics, the modeling and analysis fidelities and ultimately computational power. In developing rehabilitation regimen, the effective simulation of interactions between the human user, the device and the environment becomes critical. Varying levels of such simulations are possible – e.g. several digital-human modeling tools such as Jack allowed *kinematic modeling* of human-device-environment interactions. The ability to monitor internal human variables, such as joint angles and torques, from the virtual avatar formed the basis of a user-customized development of human-worn products [5, 6]. In recent times, a number of computational musculoskeletal-analysis tools such as SIMM [10] or AnyBody [11] have become available. Musculoskeletal systems consist of numerous bones connected together at joints, activated by muscles and thus can be treated within framework of Articulated Multi-Body Systems (AMBS) [12]. Unlike traditional engineering systems, musculoskeletal systems inherently possess considerable redundancies, which can be resolved typically leveraging an optimization approach [12]. Several packages such as SIMM focus on the forward dynamic simulation problem while others such as AnyBody implement an inverse dynamic approach [11-13]. Such tools can now allow monitoring the *dynamic of internal human variables* (such as muscle-forces and muscle activities).

Hence, in this paper, we will use this 'insight' into the human to help study the effects of variability. For example, a rehabilitation routine might aim to target specific muscle groups of a patient. A wide variety of musculoskeletal analysis results (ranging from lengths, forces, reactions of muscles, tendons, joints, etc), can be accessed. However, this raw data needs to be further processed to create performance measures that can be of utility to the therapist. Hence, in this paper, (i) peak muscle forces; (ii) average muscle forces; and (iii) muscle force fluctuation (difference between maximum/minimum muscle forces) will serve as our performance measure. Such performance measures allows a therapist to directly monitor the effects of a regimen on target muscle groups and allow changes to rehabilitation program based on patient condition. Alternatively, other higher-level abstracted measures may be developed from the physical measurements of kinematic and dynamic quantities available from the model.

Parametric designs simplify the process of systematic generation of choices especially in a computer-based implementation. However, the appropriate selection of design variables which are both biomechanically relevant and are of significant to the therapist poses challenges. Design variables can encompass geometry as well as regimen parameters for the user-device interaction. Geometric parameters could include individual user geometries such as limb length, etc [14, 15], device properties for customizable devices [5, 6], as well as properties determined by the ergonomics of placement of the user with respect to the device. Regimen parameters such as dynamically adjustable springiness, damping, as well as frequency, amplitude of a desired motion can also serve as design variables. These design variables can be used to systematically explore the space of feasible alternatives. Coupled with selection of a suitable performance measure, an optimization problem can be setup to determine the 'best' set of values for a rehabilitation regimen, form a potentially large set of choices thereby serving as a decision-support for a therapist tool.

III. OUR FRAMEWORK

Our paradigm for virtual rehabilitation program refinement brings together several tools within a VDE as shown in Fig. 1. The VDE consists of a Central Therapist Interface with which additional human musculoskeletal modeling and optimization modules interact.

A. User-Device-Environment Modeling:

The rehabilitation environment model (user and device) is setup using the AnyBody Modeling System. Parametric models of the human patient can be extracted from libraries and potentially customized to reflect the specific patient characteristics (the bones maybe re-dimensioned or the peak muscle forces can be customized). Similarly, the model of the rehabilitation device can be created within the same framework. The model of the user and the model of the device can thus be co-simulated and the obtained results can be used for refinement.

B. Central Interface:

However, the interface for interaction with the AnyBody system is not particularly therapist-friendly, i.e., non-computational scientists would face difficulty in using it. Hence, we created a Graphical User Interface (GUI) which allows for visualization and provides a restricted means for therapist interaction with AnyBody settings (e.g. using sliders). Further, using the GUI, the therapist can select the rehabilitation regimen, the objective function, as well as customize patient geometry information. The therapist can then perform parametric studies by varying the appropriate design variables.

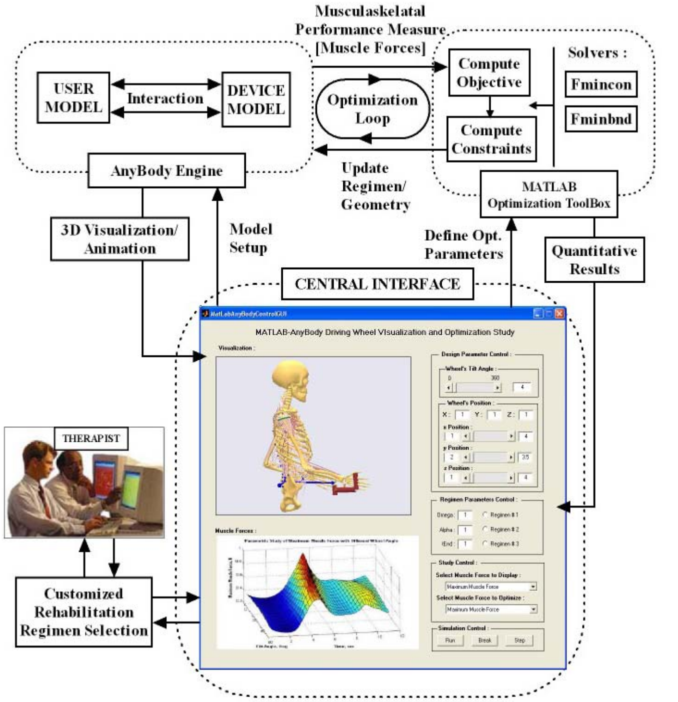


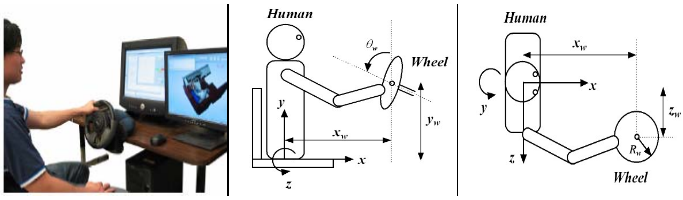
Fig. 1: Paradigm of our framework for VP of rehabilitation device: A MATLAB GUI that serves as the Center Interface that allows the therapist to examine the effects of different regimen and determines the 'best' regimen based on user's geometric information. The AnyBody engine is responsible for the computation of the muscle forces, while the optimization routine is handle using MATLAB's optimization toolbox.

C. *Optimization:*

However, as the numbers of parameters grows, it can be very difficult for therapist to interactively vary all the parameters. Thus, decision support tools leveraging optimization were created to aid the process of finding the 'best' regimen. The optimization process and data manipulation are handled using MATLAB, wherein a variety of standardized optimization algorithms are available. The MATLAB optimization routine interacts with the AnyBody system through the console environment [16] and the results of the optimization are displayed back within the GUI.

IV. CASE STUDY

We use a motor-rehabilitative Haptic Virtual Driving Environment (hVDE) [7, 8] to illustrate various aspects of the study of variability in a rehabilitation regimen. This hVDE was developed by integrating a commercial-of-the-shelf (COTS) force-feedback steering wheel with parameterized rehabilitation therapies to serve as a home-based inexpensive personal-movement-trainer (as shown in Fig. 2(a)).



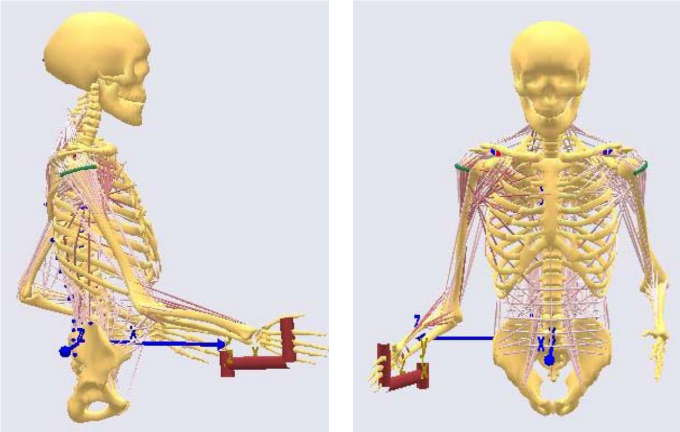
(a) (b) (c)

Fig. 2: (a) the actual setup that is being modeled and studied [8]; (b) Front view of the user and driving wheel (rehabilitation device) arrangement in the case study; and (c) the corresponding top view of the same arrangement.

Specifically, users are instructed to perform structured-rehabilitation-exercises in the form of driving-tasks along prescribed parametric paths while holding the driving wheel with one- or both-hands. However, a very important question arises in such a setting – what is the effect of variability in terms of (i) how the equipment is setup (ergonomic); and (ii) how the exercise is performed on the actual functional rehabilitation (regimen). These questions are critical to evaluate the effectiveness of a rehabilitation device.

Our framework allows the therapist to answer these questions quantitatively using a range of (i) parametric sweep; and (ii) optimization studies to be discussed next. Ultimately, such quantitative evaluation is intended to help the therapist to determine the 'best' combination of the two for a specific user.

The musculoskeletal model, shown in Fig. 3, can help a therapist model, analyze and visualize various aspects the patient's interaction with the surrounding environment. If desired, such as model can also be customized with specific patient geometry and muscular characteristics. However, for computational efficiency, we restricted the human model to the upper-body bone-tendon-muscle from the pelvis up much like others [17].



(a) (b)

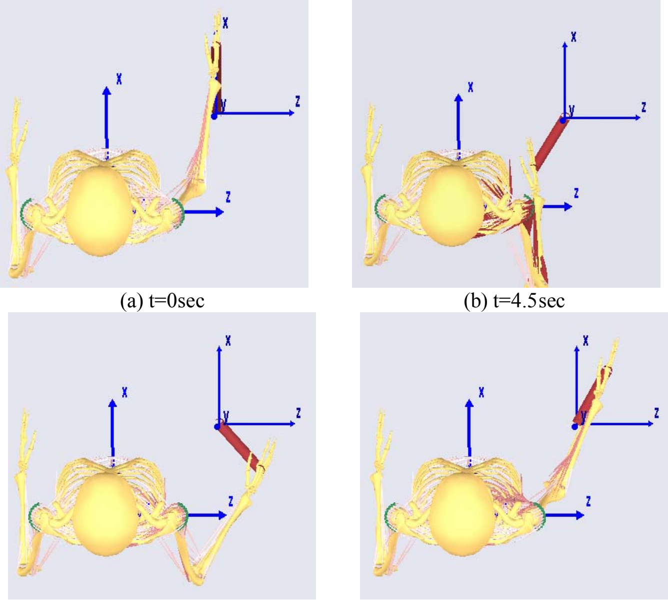
Fig. 3: The human musculoskeletal model, the rehabilitation device, and their interaction used in the case study were modeled using AnyBody Modeling System. Shown here is the (a) Side view and (b) Front view of the model.

A model developed using the bottom-up approach (from scratch) is a very time consuming process. Alternately, one can take advantage of the human model repository [18]. The designer can download the human model at various fidelities and add necessary design components to complete the model This top-down modeling approach simplify the overall modeling process tremendously and is the recommended approach [16]. Further, musculoskeletal software tends to be very complex in general and hence we intentionally limit the therapists' interactions through the central design interface.

A. *Parametric Studies*

The five geometry parameters involved in this study, $x_w, y_w, z_w, R_w, \theta_w$, pertain to the ergonomics of placement of the wheel with respect to the patient. These are spatial location of the center of the steering wheel (x_w, y_w, z_w) ; the radius of the steering wheel (R_w) ; and the tilt angle of the wheel with respect to the global y-axis (θ_w), as shown in Fig. 2(b) and (c). We can then perform parametric study for each of selected parameters within the reachable region of the user. This process allows one to analyze the effect of one parameter while others were held fixed. While any of these five parameters can be used as design variables, we choose to study only two of them (x_w, θ_w) in this paper.

The speed and amplitude at which the patient turns the wheel serve to parameterize the regimen of the rehabilitation. Increasing the speed of the turning allows a quicker completion of the regimen but requires greater muscle forces for completion of the task. In this paper, the rehabilitation regimen is such that the patient turns the wheel at an angular velocity of $30^\circ/\sec$ for amplitude of 330° turn, thus it takes 11.0 seconds to perform such task (shown in Fig. 4(a)-(d)).

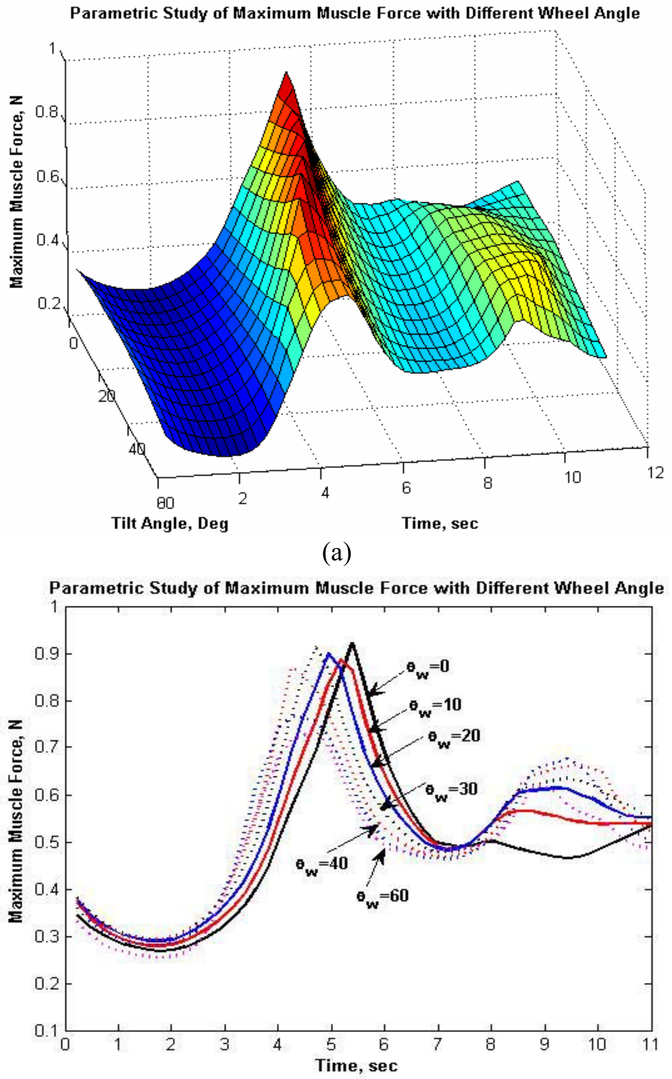


(a) t=0sec (b) t=4.5sec

(c) t=8.0sec (d) t=11.0sec

Fig. 4: (a)-(d) The hand placement on the steering wheel at different time instant in the parametric study, view from the top of the virtual model. (a) The initial position; (b)-(c) the intermediate positions; and (d) the final position.

In particular, we perform parametric sweep of x_w (corresponding to the distance of the steering wheel in front of the user) from 0.15 m to 0.45 m, at a constant angular velocity of $30^\circ/\sec$ for a total of 330° turn. The results are shown in Fig. 5(a) and (b). On the other hand, to study how the muscle force changes with respect to the steering tilt angle, we fixed the steering wheel at $(0.25, -0.1, 0.25)$ and vary the tilt angle from 0° to 60°. The results are shown in Fig. 6(a) and (b).



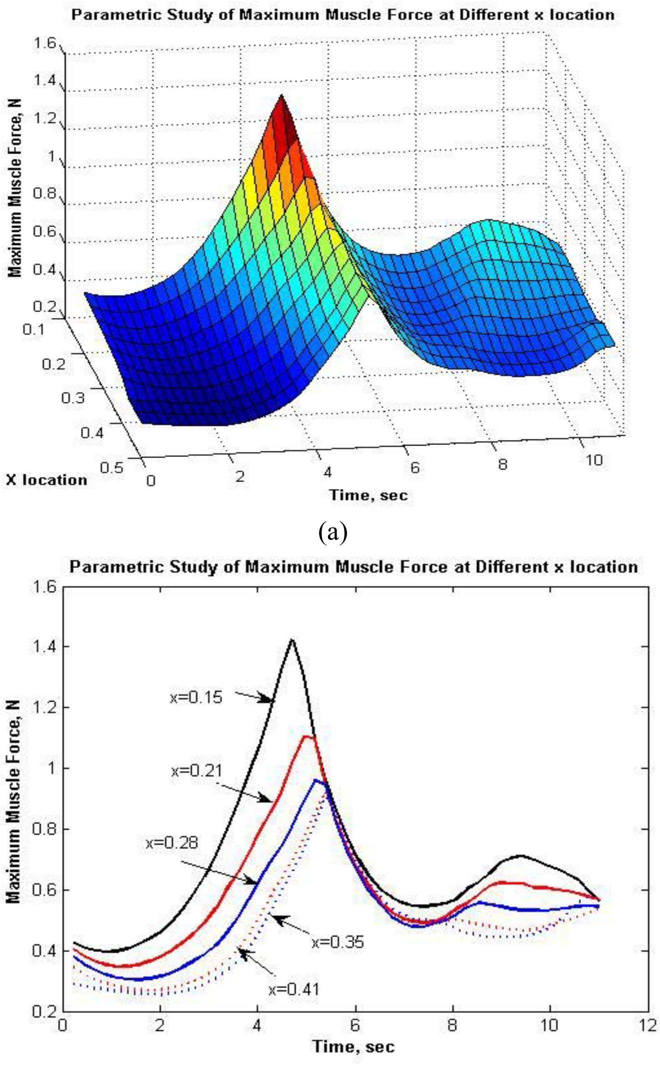
(a)

(c)

Fig. 5: (a) Surface and (b) 2D plot of the maximum combine muscles force for a patient to turn a wheel at a constant angular velocity of 30deg/sec (with 330 deg total movement), for a 0 to 60 degree tilted wheel angle, with wheel center located at (0.25, -0.1, 0.25), measured in meters.

From Fig. 5(a) and (b), we see that the peak of the combine muscle force decreases from about 0.92N to 0.72N as the steering tilt angle increases from 0° to 60° (in region where time = 3 sec to time t = 6 sec). This corresponds to the motion where the hand is at the lower region of the wheel, as shown in Fig. 4(b)-(c). On the other hand, the combined muscle force profile corresponding to different x_w positions is given in **Fig. 6**(a) and (b). In this study, we varied the x_w location from 0.15 m to 0.45m while the y_w and z_w are located at -0.1m and 0.25m respectively. The wheel tilt angle is maintained at 0 deg (no tilt) for this study. From Fig. 5(d), we can see that as the wheel being placed farther away from the user, the maximum muscle force drops from 1.43N (at x = 0.15m) to 0.96N (at x =

0.41m).



(b)

Fig. 6: and (b) Surface and (d) 2D plot of the maximum combine doing the same motion, this time with 0 deg tilted angle (no tilt), for wheel x position varied from 0.1m to 0.45m (far or close in front of the user).

B. Optimization

An optimization study was created to help the therapist to determine the best possible placement of the device (within the feasible region) and the regimen for specific objective. Two different objective functions were studied: (a) *Maximum combined muscle force fluctuation*; and (b) *Average combined muscle force*. For simplicity, we choose only two design variables in this problem, the tilt angle θ_w and wheel's x-position x_w. Thus, both objective functions are function of the x-position and tilt angle, $f(x_w, \theta_w)$. Hence, the general form of the optimization problem may be written as:

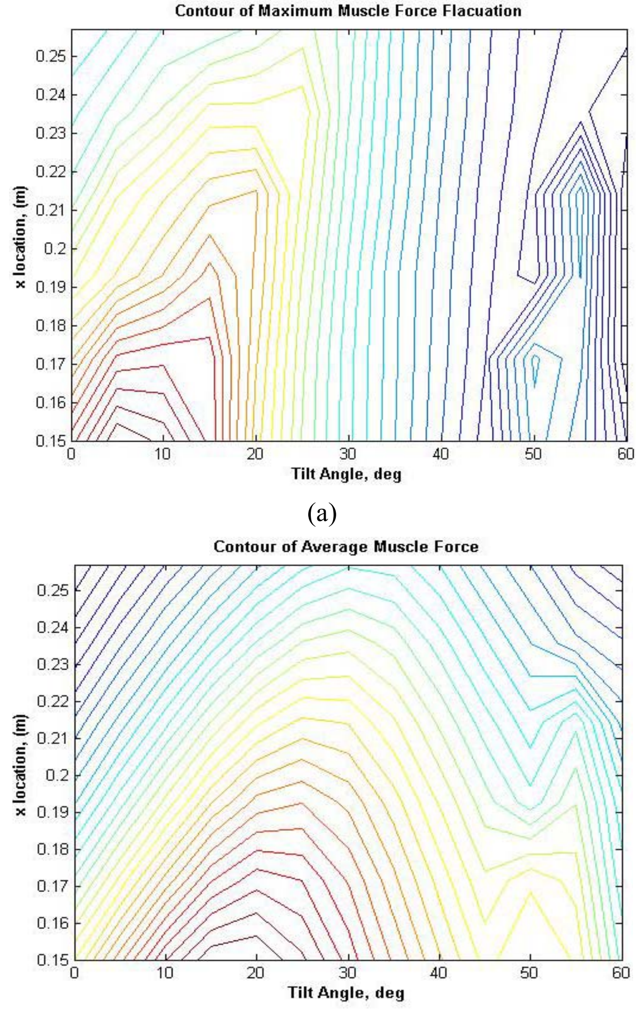
$$\underset{x_w,\ \theta_w}{Min} \quad f_i\left(x_w, \theta_w\right) \tag{1}$$

$$s.t:$$

$$g_1 : 0.15 \le x_w \le 0.25 \tag{2}$$

$$g_2 : 0^\circ \le \theta_w \le 60^\circ$$

Since only two design variables are being studied, the function space of these two objective functions can be explicitly plotted. Fig. 7 (a) and (b) are contour plots of the function space with relatively few data points used (13 data points in the tilt angle direction and 6 data points at the x direction).



(b)

Fig. 7: (a) Function space of maximum combined muscle forces fluctuation as the objective function and steering wheel's x location and tilt angle as the design variables; and (b) Function space of average muscle force as the objective function and steering wheel's x location and wheel's tilt angle as the design variables.

Although the resulting contour of the function space does not look very smooth, these contour plots give us pretty good estimate of the function space and a reasonable good estimate for the location of the design variables.From these two function plots, we can obtain the general rules: (i) to minimize the muscle force fluctuation, one would like to place the wheel closer to the user with a higher tilt angle; (ii) to minimize the combined average force used in performing the rehabilitation routine, one would place the wheel farther away from the user and with small or no tilt angle.

Thus, the setup of the optimization allows creation of a decision-support tool for the therapist. For larger problems, with more design variables, the visualization based solution process finding the 'best' solution may not be viable. However, one can capitalize on the availability of powerful numerical optimization techniques to achieve similar outcomes.

V. SUMMARY

In this paper, we presented an integrated musculoskeletal simulation based design framework that facilitates the rapid design and refinement of the ergonomic and regimen of a rehabilitation device. The integrated Virtual Design Environment serves as a decision support tool for a therapist to determine the 'best' rehabilitation program for a particular user leveraging design and computational tools from musculoskeletal analysis, optimization and visualization. Our preliminary result, demonstrated with a case study of a motor-rehabilitative Haptic Virtual Driving Environment (hVDE). In particular, we present our design framework using a case study which demonstrates the practical implementation of our framework with preliminary result. Much work needs to be done in terms of (i) creating a more seamless VDE interface for the therapist; (ii) developing quantitative performance measures that are well balanced computationally and are of significance/ relevance to the therapist; and (iii) applying the framework to a wider range of settings. We are actively pursuing all these aspects in our current work.

Although the case study used to demonstrate our framework is a driving rehabilitation device, the basic ideas and the design framework in this paper are applicable to the design of variety of other rehabilitation devices where musculoskeletal measure of the user is critical. Such a framework could potentially be used to (i) quantitatively study the effectiveness of existing rehabilitation devices, and subsequently refinement of such devices; and (ii) quantitatively study the effect of variability in the user and/or device to the effectiveness of the rehabilitation regimen, i.e. the robustness of a particular rehabilitation regimen.

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