

Individualized Interactive Home-Based Haptic Telerehabilitation

We present a haptic telerehabilitation framework for patients with upper-limb dysfunction that is well-suited for deployment in patients' homes. Specifically, a commercial-off-the-shelf (COTS) haptic force-feedback driving wheel interfaces with a PC to create a haptic Virtual Driving Environment (hVDE). Coupling this framework with parametric exercise/movement protocols—structured as driving exercises along paths of varying complexity—is the key to the creation of an inexpensive, immersive, and yet individualized personal-movement trainer.

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According to the National Stroke Association (<http://www.stroke.org>), in the US alone, there are more than 750,000 people experiencing a new or recurrent stroke each year, leading to motor disabilities. There is considerable evidence which directly links functional recovery from a stroke to the duration, frequency, regularity, and intensity of physical (motion and force) interactions in a rehabilitation regimen.^{1,2}

To this end, we wanted to develop the architecture and algorithms for an inexpensive haptic telerehabilitation framework that extends the individualized interactive nature of traditional rehabilitation therapies to patients' homes. We implemented this in the form of the haptic Virtual Driving Environment (hVDE), with an intended audience of patients with upper limb (UL) dysfunction (secondary to a cerebrovascular event, such as a stroke or physical injury). For a brief background on important considerations as well as the state of work in the field, see the sidebar, "Home-Based Rehabilitation Programs."

Architecture

The hVDE, shown in Figure 1 (on page 34), serves as an illustrative example of an individual-

ized interactive haptic telerehabilitation framework by allowing us to integrate multiple aspects of our research. At the same time, it lets us identify the issues with development, implementation, and deployment of a flexible, reconfigurable, inexpensive, portable telerehabilitation tool, suitable for setup in patients' homes and outpatient clinics. Finally, the development of such a rehabilitation tool in the context of driving, one of the higher activities of daily living, can serve to enhance the motivation and compliance aspects of a therapeutic regimen. However, we note that this hVDE is intended to serve as a network-based tool for assessment and rehabilitation of UL physical motor dysfunction and not as a driving simulator for cognitive assessment. (See for comparison Systems Technology's STISIM Drive simulator at <http://www.systemstech.com/index.php?pid=22>.)

Patient interface

The hVDE consists of a patient interface (ultimately intended to be home based) and a therapist interface (ultimately intended to be at a remote central hospital location) that are connected through the Internet. The patient interface serves both as the data-acquisition framework as well as the exercise-deployment framework. It consists of force-feedback kinesthetic-interface devices coupled with a variety of exercise scenarios implemented in the form of immersive driving activities within a haptics-enabled virtual environment.

Kinesthetic interface devices

A careful selection of the kinesthetic interface is important because it serves to stimulate the sense of touch and movement, while creating quantitatively measurable and customizable patterns of user motions and forces. We focused on selecting and validating the use of low-cost, mass-produced devices (such as the Microsoft Sidewinder force-feedback steering wheel and hobby-rate gyros) with simplified PC interfaces (Universal Serial Bus-based versus explicit data acquisition). Additionally, such commercial, off-the-shelf gaming devices employ standard software interfaces making them easy to control from a PC environment. For example, we took advantage of the extensive DirectX libraries of force-feedback (FFB) effects (available free at <http://www.microsoft.com/windows/directx/default.aspx>). These can be composed, within our specially developed Matlab/DirectX FFB interface

Home-Based Rehabilitation Programs

Home-based rehabilitation programs have gained importance because of their flexibility in individualizing the intensity and duration of the rehabilitation therapy. However, studies have also shown that newer neurorehabilitation techniques such as constraint-induced therapy¹ can bring about a significant acceleration in restoring functional use. Such approaches, however, require intensive and supervised assessment and therapeutic procedures administered by a clinician working with a single patient at a time, which limits the applicability to clinical settings.

Over the past decade, robotics and automation technologies have proven their ability to create customizable movement-therapy rehabilitators and allow seamless capture of quantitative patient-performance information during the neurorehabilitation process. Researchers have developed numerous robot-assisted therapy devices² to physically interact with patients and assist in movement therapy. Such robotic devices now take the role of a therapist in guiding the patient through the intensive, repetitive practice of functional movement with documented successes.³⁻⁶

Additionally, traditional assessment methods involved a clinician-/therapist-based evaluation of physiological characteristics (such as speed, range of motion, and strength) using subjective and semiquantitative tests such as the Rivermead motor assessment score.⁷ In contrast, the sensitive, instrumented user interface provided by a robotic device can aid a transparent and automated quantitative-assessment process.^{3,8} It can also bring forth other desirable functional assessment capabilities, such as

- specificity (to distinguish between different diagnoses),
- sensitivity/resolution (for finer gradation), and
- repeatability/stability (observer-, spatial- and temporal-invariance).

However, despite these obvious benefits, costs and deployment logistics have hindered the transition of these advances over to the home-based rehabilitation arena.^{9,10}

In recent years, numerous truly low-cost, mass-produced commercial-off-the-shelf (COTS) haptic/force-feedback devices have become increasingly available, commonly for gaming

applications. Thus, we believe that a class of problems exists¹¹ where coupling low-cost COTS therapy devices with quantitative rehabilitation assessment/therapy protocols will facilitate widespread deployment as home-based telerehabilitation personal-movement trainers. In this article, we examine the critical research and development issues entailed in successfully implementing this vision.

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(available free at <http://www.mathworks.com/matlabcentral/fileexchange/loadFile.do?objectId=4939>), to create an entire assistive/resistive exercise regimen.

The software deployed on the patients' home-computer reads these quantitative user-inputs (motions and forces) from the force-feedback interface device (and other sensors). An appropriate kinesthetic dynamic-interaction (haptic)

model^{3,4} generates motions and forces to be fed back to the user and updates the virtual environment. Further, the vast amount of data collected needs to be effectively digested and analyzed prior to presentation to the clinician for interpretation. This takes the form of development of performance measures based on the captured motions and force, which we discuss later in the article.

Figure 1. Haptic Virtual Driving Environment (hVDE) for individualized interactive telerehabilitation.

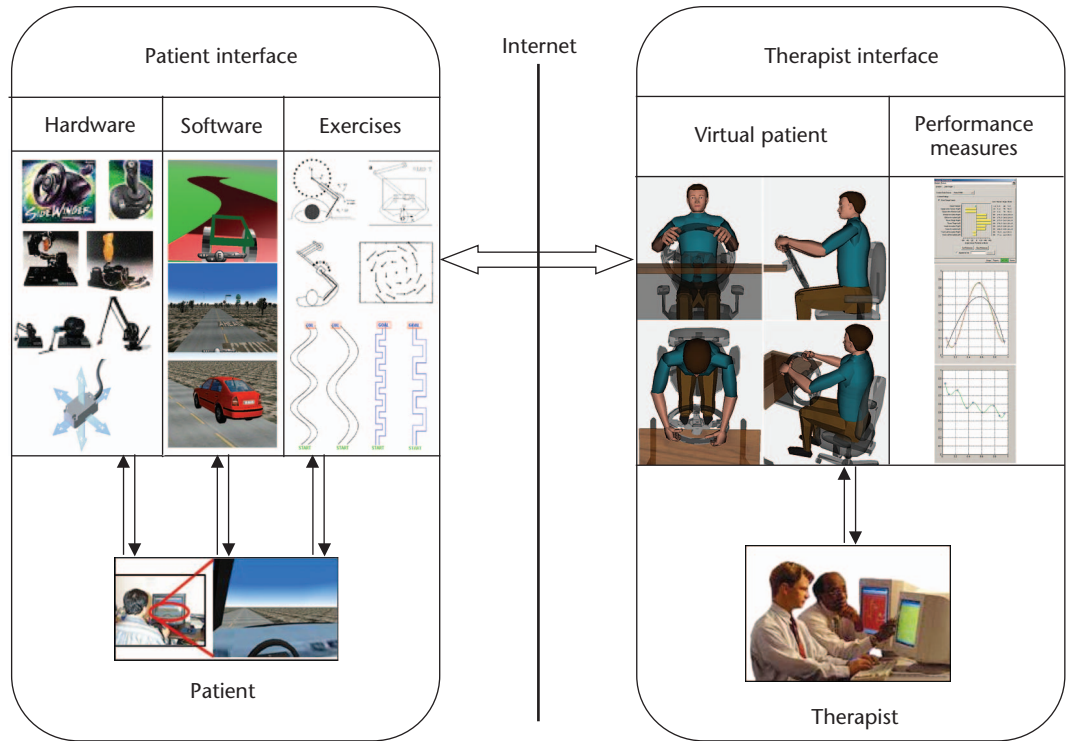
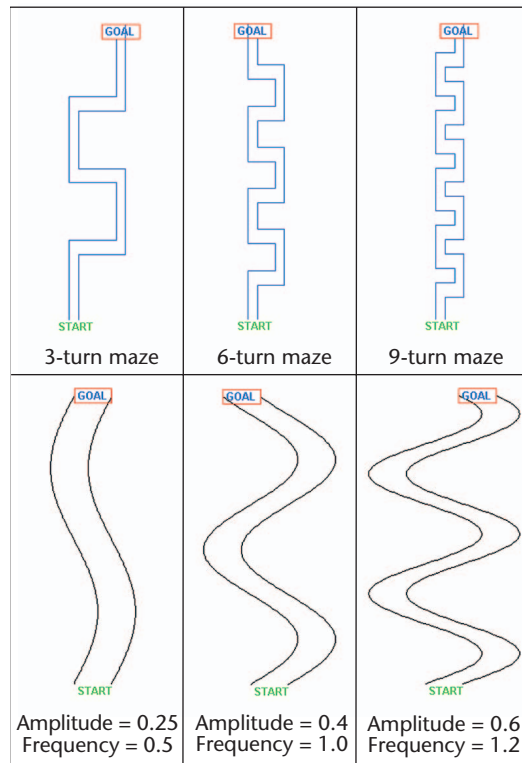


Figure 2. Parametric library of labyrinthine maze-style and sinusoidal paths.



We employed a variable level-of-detail implementation to facilitate mixing and matching the various levels of modeling and simulation fidelities—both for visualization (simple 2D GUIs ver-

sus detailed 3D environments) and haptic dynamic simulation (kinematic versus dynamic vehicle models). Such an interactive environment can function in a standalone manner; in this form it resembles any game that would be available on the market.

Parametric library of exercises

The creation of a parametric library of exercises and subsequent parametric assessment/therapy sets the stage for the individualized interactive rehabilitation. We created this library of exercise routines with various parameterized driving scenarios (such as roads of increasing curvature, sharp turns, and so on) to serve as templates to generate the desired exercise therapeutic regimen. For example, Figure 2 shows a set of such parameterized sinusoidal and labyrinthine maze-style paths.

Therapists can parametrically generate the sinusoidal paths by specifying the amplitude and frequency. Alternatively, they can generate the labyrinthine mazes by specifying the mean value of free straight-line paths and/or the number of turns between a desired start and finish location. More complex paths can be expressed in terms of these parametric basis functions (sinusoids/step-functions) with additional analysis (Fourier/wavelet decompositions), without significantly increasing the computational burden.

Such a parameterized set of paths offers a low-order parameterization of the infinite dimensional set of exercises, permitting a therapist to easily control the complexity of the proposed assessment/therapy regimen. Furthermore, we envision creating a test suite using the design of experiments methodology that progressively and interactively varies the complexity of the tests. This mitigates the need to explicitly present the user with the complete test suite, thereby speeding up the assessment process. Additionally, this can also facilitate creating a finer resolution of tests to enhance the differential assessment process.

Therapist interface

From the therapist's point of view (see Figure 1), this telerehabilitation system facilitates effective visualization and quantification of the patients' motions and associated pathologies as the patient follows a prescribed exercise regimen. We used JACK, a digital human modeling and simulation software, to develop synthetic human user models—these consisted of articulated rigid-body models that reflect the user's geometry and kinematics.⁵

This JACK model forms the virtual model (the "avatar") of the patient with which the therapist interacts within this virtual environment. The system customizes each digital model to reflect the specific performance characteristics of each individual patient (for example, ranges of motions and strength), as determined by our biomechanical identification efforts. We report elsewhere⁶ some of our early efforts in identifying the kinematic parameters of the upper-arm building with online kinematic calibration. The end goal is a system that is capable of adaptively estimating these parameters based solely on the streaming measurements without requiring expensive and explicit calibration.

Therapists can use the remotely collected data to replay the patient's driving (exercise) session on the digital human model and review it from various viewpoints. Further, the interface can also provide the therapist with additional computed, postprocessed information (such as graphs of computed ranges of motion, comfort indices, and so on) to aid the assessment process.

Some of our efforts in developing invariant and quantitative performance measures (which we discuss further in the next section) are relevant here. The therapist can now appropriately modify the therapeutic regimen and download a new therapeutic regimen back to the patients' machine.



Figure 3 shows the implementation of the overall prototype system where data acquired from the patient's arm movements are replicated by the digital JACK model on the therapist's computer (see also the video "Immerse Virtual Environment for Individualized Progressive Telerehabilitation"; <http://mechatronics.eng.buffalo.edu/research/hapticdrivingsimulator>).

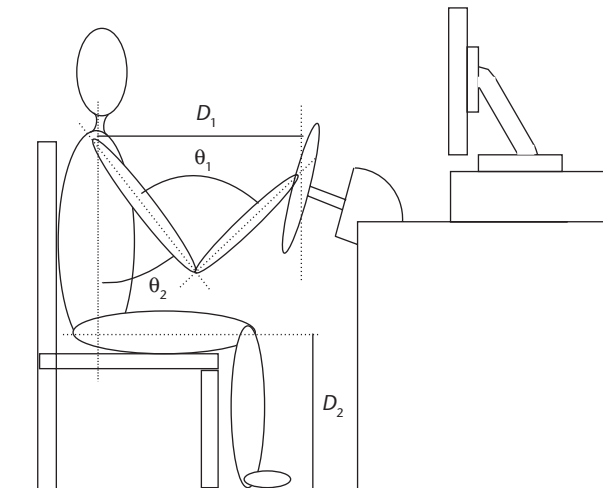
Experimental testing

The hvDE framework collected quantitative data from the kinesthetic interface (FFB wheel) while five able-bodied subjects performed the parametric exercises. In this section we provide brief overviews of the conduct of the testing and the results from offline computation of performance measures, along with anticipated observations and extrapolations to a clinical setting. Further details are provided elsewhere.^{3,4}

Testing protocol

We developed a standardized experimental protocol, in consultation with an exercise physiologist, to mitigate the influence of environmental conditions. We restrained each subject in the chair with a seat belt and a shoulder harness. We measured the shoulder and elbow flexion angles (θ_1 and θ_2 , as Figure 4a on the next page shows) at the nominal rest position with a manual goniometer. Then we adjusted the height of the chair and the distance from the wheel (D_1 and D_2) to attain the desired shoulder and elbow flexion angles of 45 and 120 degrees, respectively. We instructed the subjects to grip the steering wheel in the 9–3 clock position, with their thumbs aligned along the grooves provided on the wheel (see Figure 4b). For future testing, we plan to use

Figure 3. Implementation of network-connected patient and therapist interfaces.



(a)



(b)

Figure 4. Experimental test setup:

(a) schematic with relevant parameters; (b) photograph taken during testing.

an instrumented jacket with relatively inexpensive resistive-bend sensors mounted at elbows.

Potentially, both the parametrically generated desired path and the user-advancement speed can be continuously adjusted during an exercise regimen. However, for the testing procedure that we outlined, we chose three sinusoidal paths of increasing amplitude and frequency, and three labyrinthine paths of increasing complexity from a parametric library. We applied a damping torque, proportional to the rate of change of the steering angle, to provide resistance to user motions. Subjects were now required to guide the “vehicle” along these paths, remaining as close as possible to the center line, with three preset forward speeds.

We used the GUI (see Figure 5a) to present the patients with a test suite of paths of varying complexity drawn from the library of paths. However, we would like to note that using this parametric library did not preclude the use of detailed 3D visual interfaces for the immersive environments (see Figures 5b and 5c).

We conducted tests on five male subjects between the ages of 25–30. The general assumptions made regarding the subjects included

- similar prior driving background;
- similar mental and physical state; and
- no significant variation in terms of height or weight.

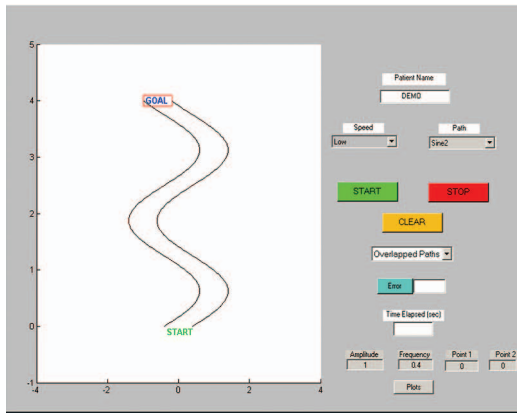
We conducted the tests on each subject at a stretch without any rest breaks, since all the

tests were of a short duration (10–20 seconds). We allocated no prior training time and we presented the sinusoidal- and maze-style paths alternately to eliminate the possibility of conditioning (even within the short testing period). We recorded 18 sets of data (six paths at three speeds) for each subject for a grand total of 90 sets (five subjects) in all. We repeated these tests on four of the initial test subjects after an interval of more than six months to also assess the repeatability of the test setup.

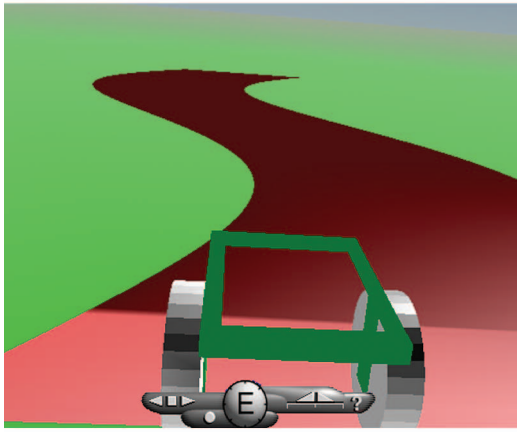
Performance measures

The time history of the error-value parameter (EVP), calculated as the difference between the subject-generated path and the nominal path, and other measures derived from this time history serve as our principal assessment measures. We examined the staged extraction of measures of increasing complexity from collected data. This included using simple statistical measures (such as mean, standard-deviation, and kurtosis) as well as performing additional analysis (by way of Fourier decomposition and principal component analysis).

Figure 6a (on page 38) is a plot of the EVP versus time for all subjects traversing the sine1 path at the low speed, while Figure 6b depicts the EVP versus time for one of the subjects in two trials six months apart. In Figure 6a, we observe that even within a small group of healthy subjects a significant variation exists in the EVP and can be captured by our setup. We anticipate that these differences will be further exaggerated in populations with UL dysfunction, permitting us to distinguish between various individuals as well as classes of functional impairment. Early and dis-



(a)



(b)



(c)

Figure 5. (a) Example of a 2D GUI that allows conduct of the experiment and provides immediate relevant statistical feedback. Enhanced 3D visual interfaces for the hVDE with (b) simple, parametrically generated 3D paths and/or (c) realistic texture-mapped environments.

tinctive detection of the type and extent of functional impairment can significantly help a therapist.

Figure 6b is representative of the relative stability of the EVP performance measure on data

Table 1. Mean path deviation of the five subjects for traversing at three speeds along three sinusoidal paths.

Speed	Path			Subject	Ranking
	Sine1	Sine2	Sine3		
Low	12.06	33.362	46.32	1	
	17.13	31.68	51.80	2	
	14.52	23.32	42.56	3	
	19.92	29.84	50.88	4	Poorest
	10.41	18.50	44.39	5	Best
Medium	13.66	22.33	50.29	1	Best
	19.72	38.49	46.90	2	Poorest
	19.64	31.35	49.50	3	
	15.76	29.08	51.74	4	
	17.37	28.18	47.26	5	
High	14.42	25.78	47.20	1	
	15.43	33.28	49.20	2	
	15.10	39.68	45.06	3	
	19.05	28.79	52.48	4	Poorest
	14.17	25.80	47.46	5	Best

collected from an able-bodied subject six months later (in the absence of an active therapeutic regimen). Preliminary analysis^{3,4} with other subjects (not shown in this article) indicates that this natural temporal variability is well bounded. Significantly greater changes are anticipated during an active rehabilitation regimen. Thus, such measurement can help monitor the progress and efficacy of the regimen (and patient compliance); this aspect is being examined in some of our current work.

We carried out statistical analyses for various combinations; for example, we tabulated the mean path deviation of the various subjects in Table 1 and show the time bar-plot for sine1 at all three speeds in Figure 6c. By comparing the data obtained from the time bar-plot and the corresponding mean deviation data from Table 1, we can see the correlation between the maximum deviations from the nominal path and time to task completion. Thus, even a relatively simple measure such as mean time to completion can serve as a good measure of performance. Some of the other developed error measures—such as the ratio of the subject velocity to nominal velocity for traversing a particular path—are discussed elsewhere.³

Finally, Figure 6d depicts the results of the principal component analysis of the same data. We note that the results from the principal component data analysis are potentially attractive from the viewpoint of their selectivity and reso-

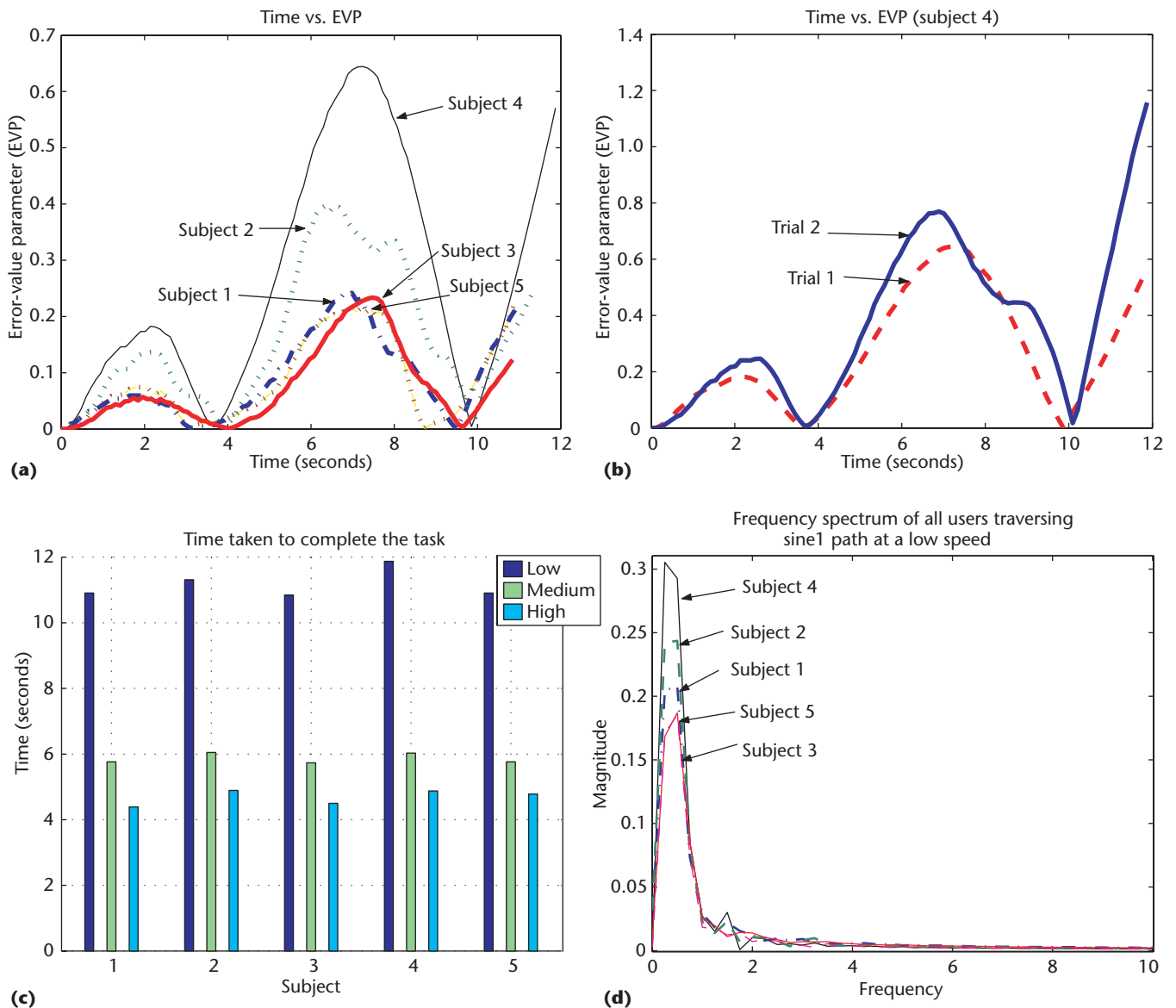


Figure 6. Results collected from the subjects for sine1 at low speed. (a) Error-value parameter (EVP) vs. time; (b) EVP vs. time for the same able-bodied subject after a gap of six months; (c) time taken to traverse paths; and (d) principal component analysis of the collected data.

lution. In addition, the ability to characterize with minimal variability using the amplitude and frequency of sinusoids allows a low-order parameterization as well as a potentially direct link to remedial exercises from the parameterized library. Further, this data suggests that the underlying variation might be a function solely of the ability to modulate the amplitude of the desired exercise effort.

Conclusion

We presented a brief overview of our efforts at developing a framework for low-cost individualized telerehabilitation suitable for home-based deployment. Our testing of the hVDE indicates that such a patient interface possesses adequate fidelity for use as an assessment tool and for pro-

viding controlled motions and forces with minimal modifications or developmental efforts. However, in-depth testing with sample patient populations is critical prior to deployment of this framework and is being pursued.

In future work, we plan to use an instrumented jacket with additional sensing, such as bend sensors and gyros, for online measurements during the performance of the exercises. In addition to eliminating the various manual measurement and calibration stages, this would also offer a means to monitor patient compliance with the regimen. However, there is a critical need for determining appropriate low-dimensional, therapist-relevant performance measures from the high-dimensional collected patient data. We are

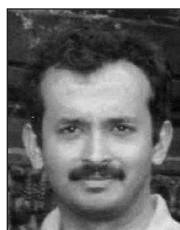
pursuing the development of additional experimental protocols and improved performance measures based on this work. In particular, examining the collected data in the frequency domain, by way of the principal component analysis, is particularly revealing. The analysis (which we are further testing) suggests that the performance variability across subjects depends on their inability to modulate the amplitude of their exercise effort (and not on the inability to match the desired frequency). **MM**

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