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Enhanced assessment of limb neuro-mechanics via a haptic display

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Abstract

Current state-of-the-art quantitative assessments of abnormal neuro-mechanics (e.g., spasticity, rigidity, dystonia) require sophisticated measurement systems that, together with the lengthiness of the data acquisition, make these approaches impractical for the clinical setting. Our long-term goal is to create a tool that enhances the capability of therapists to diagnose abnormal neuro-mechanics without the need of lengthy procedures and sophisticated equipment. Towards this direction, we present a haptic display of abnormal limb neuro-mechanics. This haptic display can be used as a training device for clinicians to understand the causality of the haptic sensation associated with neurological impairments and abnormal biomechanics during manipulation of a patients' limb. Other uses include the testing of operational research algorithms to maximize the exchange of haptic information, reproducing conditions of human-human interaction, and a test bed for developing novel assessment techniques and targeted interventions. Experimental results support the hypothesis that the best assessment of abnormalities can be obtained via specific manipulation strategies.

Keywords: Biomechanics; Haptic display; Hypertonia

Background

A common consequence of many neurological disorders like cerebral palsy, Parkinson's disease, or stroke (to name a few) is the appearance of atypical neuro-mechanical conditions such as an increase in muscle tone (hypertonia) due to active contraction of the muscle, hyperexcitability of motoneurons causing excessive co-activation, muscle contractures, etc. It is common in today's clinical practice to manipulate the patient's limbs to diagnose these abnormal neuro-mechanical changes. Through such physical interaction, clinicians extract as much information as possible to infer the patients' condition, such as in the assessment of hypertonic conditions [1-3] (e.g., spasticity, rigidity, dystonia, among others). This approach is subjective [4], which may be a limitation when inferring the nature and gravity of the impairment since the effect generated by pluri-articular muscles to multi-joint mechanics can be misjudged. A misjudgment on the

initial or intermediate assessments can lead to selecting a non-optimal intervention, which can have negative consequences, such as increasing intervention time, rising in treatment cost, or even worsening the condition. Hence, the capability to discriminate between different types and levels of impairments is important for the clinical intervention to be chosen and planned accordingly.

Traditional engineering approaches to assess differences in limb mechanics such as non-parametric [5,6] and parametric [7,8] system identification techniques provide a less subjective, quantitative measure of biomechanical variables. However, the necessity of sophisticated measurement systems, such as stiff robotic devices, makes most of these classical engineering methods impractical for the clinical setting due to a high cost/benefit ratio. In addition, the majority of studies applying such purely robotic-driven approaches have been limited to single joints (e.g., knee, ankle, elbow) as more complex joints (e.g., shoulder) pose significant challenges [9]. Yet, abnormal neuro-mechanical conditions such as hypertonia can encompass alteration of inter-muscular (heteronymous) reflexes [10] or abnormal multi-joint couplings due to increased rigidity [11,12]. Consequently, some of the

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most important clinical problems have been ignored in these engineering approaches.

Our approach

A recent review on spasticity assessment [13] concluded that there is a need for simple instruments that can provide reliable measures in a practical setting. Therefore, our goal is to extend the interactive nature of traditional assessment techniques by providing clinicians with a method that enables them to appreciate and quantify hypertonic phenomena during manipulation of the affected limb in multiple degrees of freedom (DOF). Towards this end, we propose a haptic display, in which a model of an impaired limb is haptically rendered on a robotic device in which clinicians can physically interact with. This haptic display allows us to evaluate the threshold of human perceptions to different pathological neuro-mechanics at different levels of severity, which in turn can translate into different strategies that enhance the human perception of these impairments. This will likely provide therapists with better training and improved classification, measurement, and treatment of these disorders.

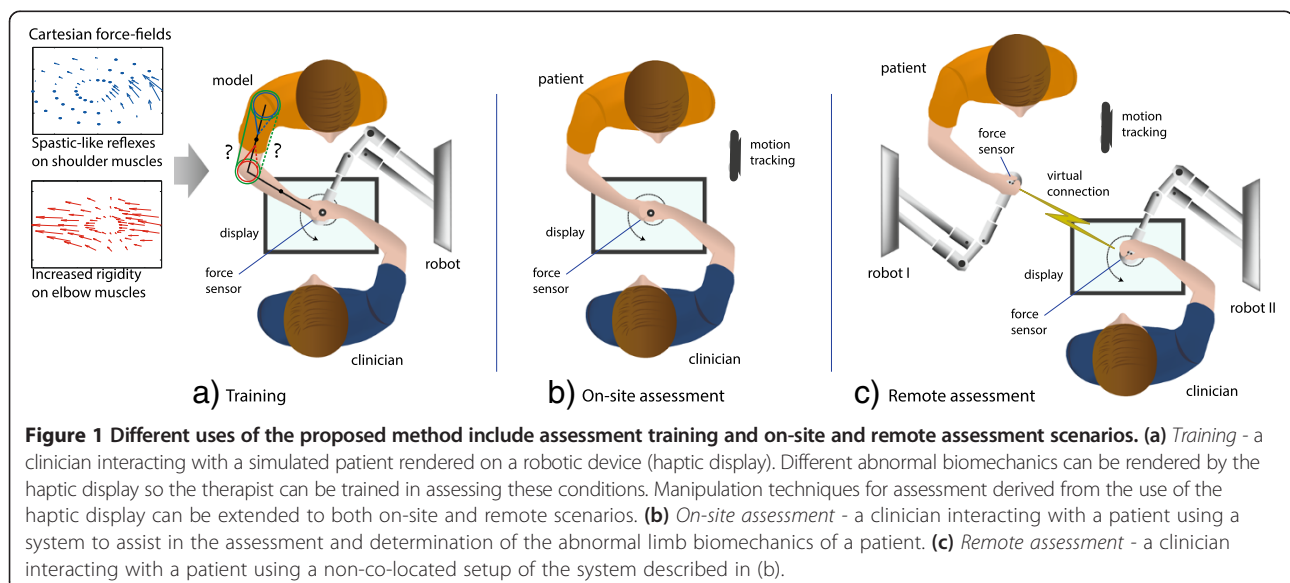
Haptic rendering of physical impairments of stroke survivors have been reported previously (e.g., [14,15]). Such approaches, however, have been limited to only one DOF and have focus on mimicking current clinical assessment techniques (e.g., Ashworth test). It is essential to extend the manual assessment of hypertonia to more than one DOF so as to capture the effect of plur-articular muscles and multi-joint coupling. Differently from the previous work, our approach allows us to characterize how clinicians perceive different types of hypertonia in a multi-DOF environment. The interaction in our haptic display occurs through a single haptic port under the

rationale that assessment techniques derived from the use of this tool should be directly applicable to both on-site and remote scenarios (Figure 1). Extending traditional manipulation methods in tele-interaction scenarios requires the use of an exoskeleton on the patient side and a robotic mock-up of the patient's limb on the clinician side so that the different limb segments can be manipulated independently. We suggest the implementation of simpler configurations where the interaction occurs through a single port from both sides, allowing the use of commercially available robots. Our hypothesis is that the individualization of abnormalities through a single interaction port in patient-clinician physical interaction can be done more objectively via specific manipulation strategies (*probing motions*) derived from understanding the forces at the point of interaction produced by the different types of impairment. The force-motion generated by the patient-clinician interaction along these motions could uniquely characterize the nature and severity of the impairment. We expect that our haptic display will help us better understand how to improve human perception during physical manipulation.

Methods

Haptic display of limb neuro-mechanics

The haptic display comprises a multi-joint robotic manipulator that is programmed to output mechanical forces that match the forces a clinician would feel when moving a limb (e.g., arm, leg, hand) of a patient (*model*). To interact with the model, subjects seat in front of the robot and hold the robot's end effector. The end effector of the robot is virtually connected to the model via a two-way connection that can be rendered, for instance, through a shared virtual environment which includes a simulated mechanical element [16,17]. The limb model



is rendered as if the patient were seated opposite the participants or at any other desired configuration. Both the subjects' hand and the robotic arm can be covered by an opaque horizontal screen, on which an image of the rendered virtual limb is projected. A superimposed image representing the workspace can also be projected. This configuration can also be reproduced during tele-assessment, where instead of a simulated arm, the users interact remotely with another person.

Neuro-mechanical model of a human limb

The force produced by muscles can be expressed in terms of passive (i.e., intrinsic physical properties) and active (i.e., reflexes and voluntary activation) components. The combination of forces generated by different muscle groups gives rise to characteristic force patterns at a limb's endpoint. The set of forces associated with a set of limb states (e.g., position, velocity) in the Cartesian space is a *force field* [18,19]. In our haptic display, the force of each muscle group is modeled as a linear combination of non-linear passive and active components. The passive component is position dependent and it is a function of the muscle length, whereas the active component depends on reflexive activation and is both position and velocity dependent. The resulting force field is the linear superposition of the two force components. This simple, yet representative, approach allows us to simulate different normal and hypertonic-like forces at the point of interaction by modifying few weighting coefficients. The different abnormal force fields generated by a hypertonic limb can be rendered on a robotic device (*haptic display*) and so the haptic perception to the nature and severity of hypertonic phenomena can be investigated systematically (e.g., [20,21]).

As a proof of concept, we emulated a simple representation of the human arm with six different muscle groups that are embedded in a planar, two-joint mechanism. The model considers rigid body dynamics and muscle intrinsic and active properties. The model was implemented in Simulink and was rendered on a two revolute joint planar robot running in real-time using xPC Target at a rate of 1 kHz. The dynamics of the virtual arm moving in a horizontal plane while interacting with the subject were modeled as

$$H[q]\ddot{q} + C[q, \dot{q}]\dot{q} = J_q^T[q] \cdot F_{\text{external}} - J_\lambda^T[q] \cdot \left(\Phi[\lambda, u[\dot{\lambda}]] + \Psi[\lambda] \right) \quad (1)$$

where $H[q]$ represents the arm inertia matrix of a double pendulum system; q denotes the vector of shoulder and elbow joint angles [rad]; $C[q, \dot{q}]\dot{q}$ is the term corresponding to Coriolis and centripetal forces; $J_q[q]$ is the

Jacobian matrix transforming an endpoint force, F_{external} , into joint torque; λ and $\dot{\lambda}$ correspond to the muscle stretch and muscle stretch velocity, respectively; Φ is a vector of muscle forces produced by motor signals u ; and Ψ is a vector of muscle forces that are intrinsic to the biomechanical properties of the muscle. Simulation-specific parameters for the limb centers of mass and inertia were estimated using anthropometrical tables from Winter [22]. The Jacobian matrix $J_\lambda[q]$ transforms the muscle force into joint torque and contains the muscle moment arms ρ at any particular position, q :

$$J_\lambda[q] = \begin{pmatrix} -\rho_{sf} & \rho_{se} & 0 & 0 & -\rho_{bf_1} & \rho_{be_1} \\ 0 & 0 & -\rho_{ef} & \rho_{ee} & -\rho_{bf_2} & \rho_{be_2} \end{pmatrix}^T \quad (2)$$

The sub-indexes correspond to sf, shoulder adductors (deltoid anterior, coracobrachialis, pectoralis major clav.); se, shoulder abductors (deltoid posterior); ef, elbow flexors (biceps long, brachialis, brachioradialis); ee, elbow extensors (triceps lateral, anconeus); bf, bi-articular flexors (biceps short); and be, bi-articular extensors (triceps long) muscle groups.

The force of each muscle group is modeled as a linear combination of active (i.e., produced by a motor command u) and passive (i.e., produced by intrinsic rigidity of the muscles and connective tissue) components [23,24] associated with intrinsic rigidity of the muscles and connective tissue. The force produced by each muscle in response to the motor signal u is

$$\begin{aligned} \Phi[\lambda, u[\dot{\lambda}]] &= (\varphi_{sf} \quad \varphi_{se} \quad \varphi_{ef} \quad \varphi_{ee} \quad \varphi_{bf} \quad \varphi_{be})^T \\ \varphi_i &= \max \left(0, \alpha_i u[\dot{\lambda}_i] \cdot \Gamma \cdot e^{-\Gamma^2} \right) \\ \Gamma &= \left(\frac{\lambda_i - \lambda_{\min,i}}{\lambda_{\max,i} - \lambda_{\min,i}} - \frac{\lambda_{\text{rest},i}}{\lambda_{\max,i}} (1 - u[\dot{\lambda}_i]) \right) \end{aligned} \quad (3)$$

The term u corresponds to an active motor signal (e.g., reflexes, voluntary commands); the term α corresponds to the stiffness of the muscle when the system is linearized at its equilibrium position when the activation is maximum (i.e., $u = 1$).

Typical assessment of hypertonia is performed by asking the patient to not voluntarily intervene. Under this assumption, the term $u[\dot{\lambda}]$ corresponds to an active motor signal that depends on the muscle stretch velocity $\dot{\lambda}$ (e.g., tonic reflex), which we defined as

$$u[\dot{\lambda}_i] = \beta_i \min \left(1, \max \left(0, \frac{\dot{\lambda}_i}{\lambda_{\max,i}} \right) \right) \quad | \quad \beta_i \in [0, 1] \quad (4)$$

where β corresponds to a 'stretch gain'. The variables

$\lambda_{rest,i}$, $\lambda_{max,i}$ and $\lambda_{max,i}$ correspond to the length of the i th muscle group at rest, its maximum length and maximum rate of length change, respectively. The maximum rate of length change was calculated as the one obtained by moving the endpoint of the virtual arm along a circle of 35 cm in diameter at a frequency of 2 Hz. The maneuver was assumed to be a good compromise between the amplitude and frequency of a very fast movement in endpoint space.

The force produced by the intrinsic rigidity of the muscles and connective tissue is a function of the muscles' lengths:

$$\Psi[\lambda] = (\psi_{sf} \ \psi_{se} \ \psi_{ef} \ \psi_{ee} \ \psi_{bf} \ \psi_{be})^T$$

$$\psi_i = \max \left(0, K_{m,i} (\lambda_i - \lambda_{rest,i}) e^{-(\lambda_i - \lambda_{rest,i})^2} \right) \quad (5)$$

The term $\Psi[\lambda]$ can be multiplied by a generalized logistic function to avoid sharp discontinuities when adding both passive and active components in Equation 1. The term λ represents the muscle length, and K_m corresponds to the muscle stiffness linearized at the muscle's rest position.

Haptic rendering of abnormal limb neuro-mechanics

Haptic rendering of the correct ranges of forces that impaired individuals can exhibit is essential so that valid psychometric studies on haptic perception can be done in a haptic display. These ranges of forces depend on both the intrinsic physical properties (rigidity) of the limb and the influence of muscle reflexes on the limb mechanics. However, the separation of these two terms is an open, difficult, and controversial issue [6,25-28]. However, assuming that both intrinsic and reflexive stiffness are linearly combined, we could get a good approximation of the order of magnitude of these two terms by means of measuring *postural* and *stiffness* maps. A *postural map* represents the stationary force field (i.e., with zero velocity) produced by the intrinsic properties of the limb measured at specific locations [11,29,30]. It is representative of the intrinsic properties of the arm. The integral of this force field with respect to the position workspace is the potential energy associated with the passive mechanics of the limb. A *stiffness map* represents the Hessian of the potential of active muscle force at a specified velocity obtained by perturbing the limb at different positions and frequencies. Given the rapid change in velocity during the acquisition of these maps, the reflexive stiffness is expected to rise and be somehow distinguishable from the intrinsic rigidity, as demonstrated in [11]. By integrating each map twice with respect to the position, a potential describing the active component of the muscles at the specified velocity is obtained.

Given a database of different force fields generated by different impaired individuals in the form of *postural*

and *stiffness* maps, one can use optimization algorithms (e.g., simulated annealing) to tune the sensitive parameters of the model, such as the stiffness of the connective tissue and the reflex gain for each muscle, to generate a force field that matches the measured ones (see Figure 2). Therefore, one could create and reproduce any arbitrary mixture of representative reflex and intrinsic stiffness by means of our haptic display.

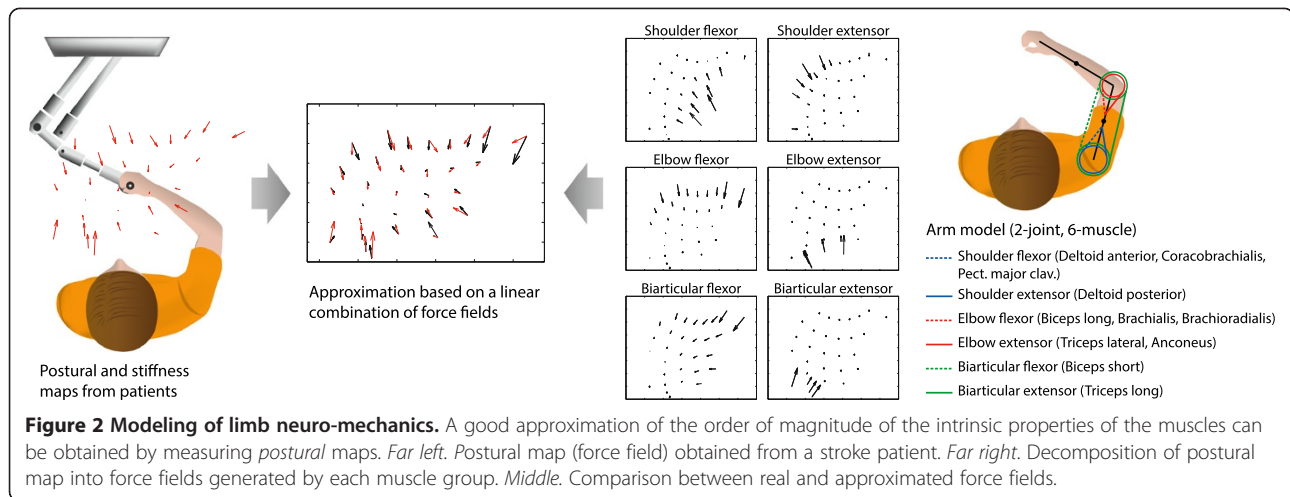
Haptic rendering of hypertonic-like forces

Among the numerous factors that characterize hypertonia, we are interested in training the ability to discriminate forces produced by increased muscle rigidity and the non-linear phenomena associated with velocity-dependent stiffness. Another way of simulating hypertonic-like forces is to assume that these could be achieved by increasing both α and K_m in Equations 3 and 5. Measurements of joint rigidity during passive movements are available in the literature for both stroke survivors and unimpaired individuals [8,31]. Based on such data, the average joint passive linear stiffness of unimpaired individuals can be taken as the *lower boundary* of joint rigidity. The *upper boundary* of joint rigidity corresponds to the passive linear stiffness recorded on stroke survivors with the Modified Ashworth Score equal to 4 [11,32]. To render intermediate levels of severity, one could consider the just noticeable difference (JND) or the Weber fraction, which is an important index representing the sensitivity of the subject to stiffness stimuli. Different stiffness JND have been obtained empirically, depending on the experimental protocol used. For palpation with a fixed displacement, the value of stiffness JND is around 8% [33]. For free exploration, the JND is much higher and can be up to 67% [34]. Since the rigidity discrimination in the clinical setting is performed with a free movement, a stiffness JND = 60% could be chosen to segment the whole range of rigidity between the two aforementioned boundary conditions.

Psychophysical experiments

We used our haptic display to investigate whether specific exploration patterns or probing motions can be used to discriminate the salient features of a given set of force fields associated with different hypertonic-like conditions. In particular, we present a study that evaluates the capabilities of (non-therapists) individuals in discriminating between different hypertonic virtual arms with diverse muscle groups affected. Three different muscle groups were tested - *i*) hypertonic shoulder muscles, *ii*) hypertonic elbow muscles, and *iii*) hypertonic bi-articular muscles - at four levels of severity - *i*) very mild, *ii*) mild, *iii*) moderate, and *iv*) severe, following a Weber Fraction of 0.6.

Twelve subjects (non-therapists), naïve to the experimental conditions, participated in the study. Subjects



gave informed consent prior to participation. Experiments were approved by the Northwestern University's Institutional Review Board. Participants were divided in two groups: i) non-instructed and ii) instructed. *Non-instructed* subjects were required to discriminate between the different conditions using their own-developed manipulation technique during a familiarization phase. *Instructed* subjects were required to manipulate the haptic display with specific probing motions that highlight the salient features of each abnormal condition (pulling and semi-elliptical motions - see 'Selection of specific probing motions').

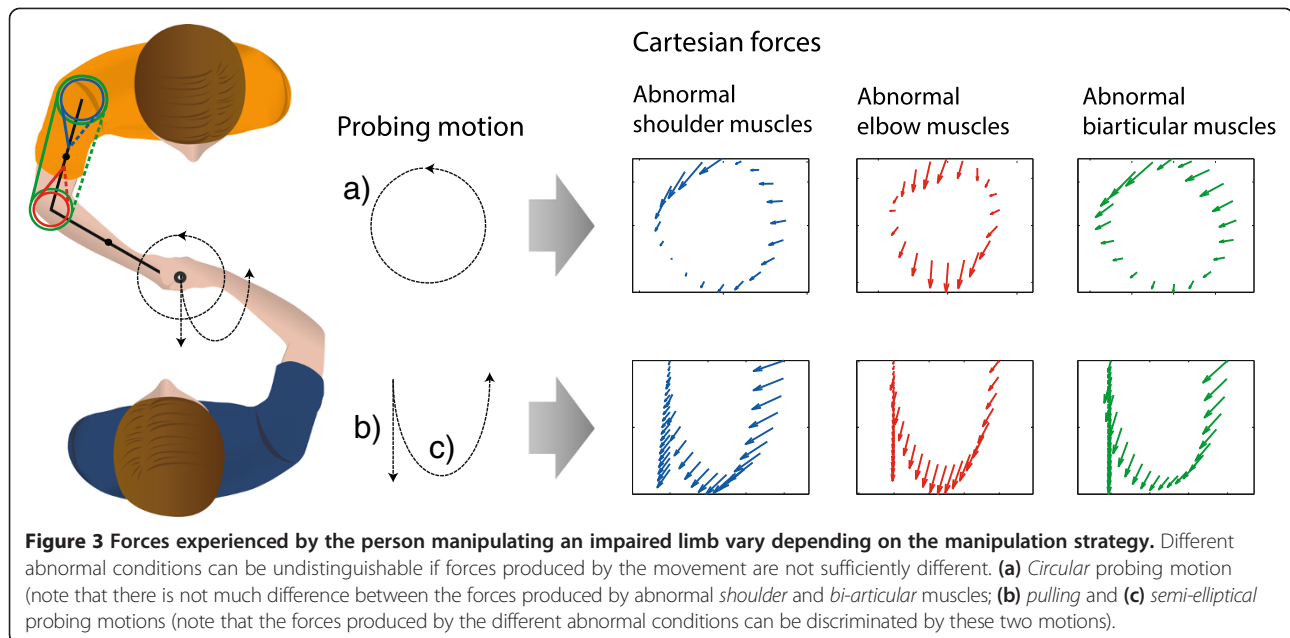
Experimental protocol

The experiment was divided in two consecutive phases: *a)* familiarization and *b)* assessment. During *familiarization*, participants interacted with four different virtual arms: *i)* unimpaired arm, *ii)* maximally impaired shoulder, *iii)* maximally impaired elbow, and *iv)* maximally impaired bi-articular muscles. Subjects interacted sequentially with each condition in blocks of 30 s for eight trials (8 trials \times 4 conditions = 32 trials). During this phase, a legend appeared on the top right corner of the screen indicating the condition that the user was experiencing. In order to familiarize with the different haptic sensations produced by the hypertonic muscle groups, *non-instructed* subjects were free to interact with those four conditions as they wished, while *instructed* subjects were specifically asked to move the arm with a pulling and semi-elliptical probing motions. During *assessment*, all subjects were randomly presented with different abnormal virtual arms of varied hypertonic severity. Their task was to identify which group of muscles was impaired. All subjects were presented each condition nine times in trials of 15 s (3 hypertonic conditions \times 4 levels of severity \times 9 trials = 108 trials). The only information available

during this phase was the configuration of the arm and the rendered force at the end effector of the haptic display. After physical manipulation of the virtual arm, subjects were required to select one of the three options presented on the screen - *shoulder*, *elbow*, or *both*. Participants could take as much time as they wanted to give their assessments and start a new trial. The whole experiment lasted for about 45 min. Simulation-specific parameters were the same as the ones described in detail in [20].

Selection of specific probing motions

As a first approximation, we planned the probing motion for this experiment considering only the effect of the position-dependent muscle force Ψ . Given the three force fields associated with *severe* hypertonic *shoulder*, *elbow*, and *bi-articular* muscle groups, our first goal was to find the minimum number of probing motions that, executed within each force field, would produce a different force profile. Such force profiles along the trajectories ought to be easily identifiable from one another. One possible strategy is to include the locations of the workspace in which the minimum and maximum force magnitude occurs for each field in each probing trajectory. We called this strategy the *min-max paradigm*. The set of probing trajectories should be designed so that, along the same probing motion, the minimum and maximum force magnitude occurs at different locations for each force field. Figure 3(a) represents the passive force Ψ for severe hypertonic muscle groups, generated by a circular probing motion. It can be seen that the minimum and maximum force occurs in different locations along the circles for hypertonic *shoulder* and *elbow* force fields. Yet, there is not much difference in the location between hypertonic *shoulder* and *bi-articular*. Hence, an extra probing motion would allow disambiguation between conditions. It can be demonstrated that given n types of force fields, when using the *min-max paradigm*, it is



always possible to identify all impairments by exploring $n - 1$ trajectories.

While circles could completely characterize the location and intensity of the maximum force associated with a specific impairment, the path to be covered is quite long. This is a limitation since the effect of the reflexes is attenuated after prolonged repetitive movements. To avoid this, we tried to reduce the length of the movements by incorporating simple strokes. According to the *min-max paradigm*, the arm's equilibrium position (i.e., the center of each circle) ought to be included in the probing motion as it is the point of absolute minimum force for all conditions. On the other hand, it should be noted that the magnitude of the force is not the only variable that can be discriminated. The orientation of the force is also important as it allows for an unequivocal haptic cue. Pulling the endpoint of impaired arm from its equilibrium position along the mid-sagittal plane in a fast and straight motion (Figure 3(b)), we can find that the orientation of the force profile generated by the hypertonic *shoulder* force field is completely different from those produced in the other two fields. Hence, we chose this movement for discriminating the shoulder force field from all the other. It is important to notice that the orientation and magnitude of the force would change depending on the impairment. We refer to these probing strategies as the *orientation paradigm*. By pulling the endpoint of impaired arm along a semi-elliptical trajectory as shown in Figure 3(c), the orientation of the corresponding force profile is quite different if the movement is executed when interacting with hypertonic *elbow* or *bi-articular* conditions. This is the second trajectory that we chose to complete the set of probing

motion. Since the number of impairments was restricted to three in the current experiment, we hypothesize that subjects would be able to discriminate between the conditions by using only two probing motions.

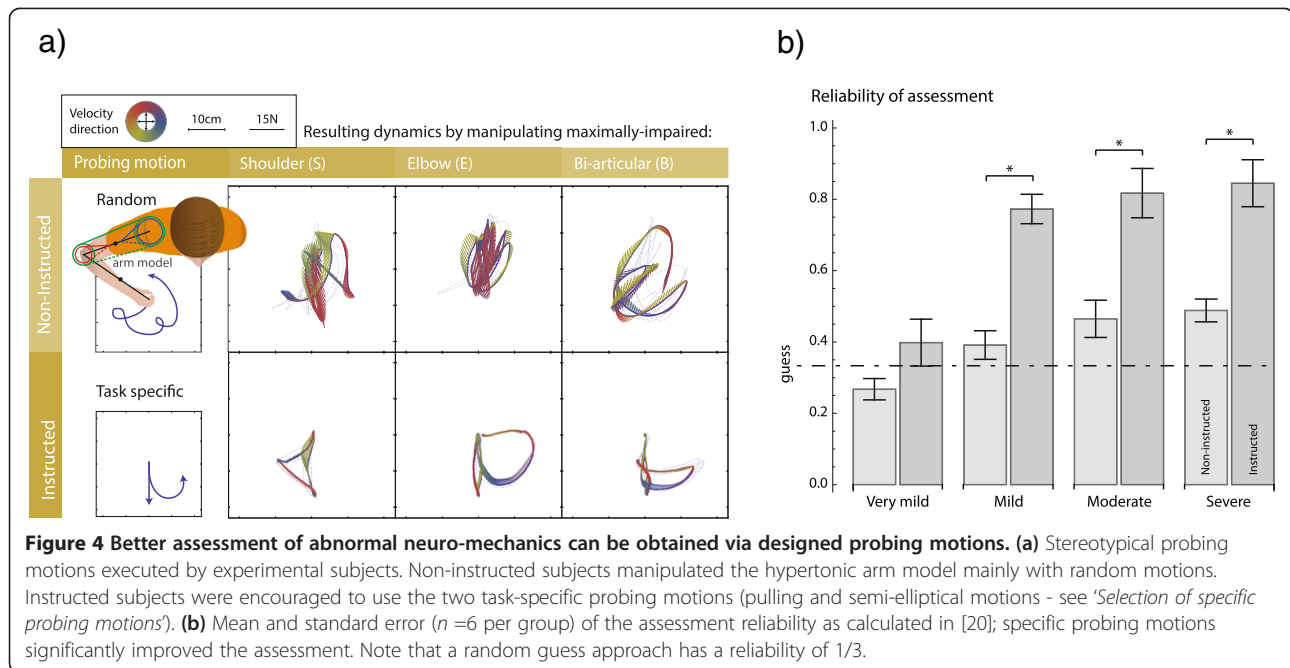
Results and discussion

Psychophysical experiments

Figure 4a shows stereotypical probing motions executed by both groups, and Figure 4b shows a comparison of the reliability between the two manipulation techniques. It can be seen that the reliability of the users' discrimination was better when they followed a specific probing motion as compared to random exploration. These experimental results support the hypothesis that better assessment of abnormalities can be obtained via specific probing motions that highlight haptic features of the different abnormalities, thus enhancing human perception.

Other experiments

In addition to this experiment, we are currently using this haptic display to *i)* assess the capabilities of naïve subjects and expert therapists to correctly identify the nature and severity (see [20,21]) of hypertonic impairments and to *ii)* assess how the perception of hypertonia is affected by the impedance introduced by the virtual connection between a patient and a clinician interacting remotely [35]. The haptic display allows to tackle several practical issues. For example, it can be used to investigate the effects of transmission delays on haptic perception of abnormal biomechanics or to derive specific probing motions that would enhance recognition of such disorders.



Discussion

This paper presents a haptic display for assessing neuro-mechanical pathologies such as hypertonia. An innovative step in our haptic display is to give clinicians the capacity to translate clinical parameters into a haptic sensation. This could also fulfill the need for a training system that can help clinicians to better understand the causality of the haptic sensation experienced during physical manipulation of a patient's limb afflicted by a particular impairment, for instance, to find specific probing motion(s) that are better at discriminating between different pathologies. While in current clinical practice the assessment of different aspects of hypertonia are expressed by the subjective judgment of the therapist [4], to our knowledge, there is no scientific literature that has systematically investigated a therapist's perceptual identification of the nature and severity of multi-joint hypertonic phenomena. Some studies have tried to categorize the sensitivity of therapists to non-biological stiffness [36,37]; however, these studies have been based on identifying the sensitivity of tactile sensations produced by surfaces. Our haptic display allows us to systematically study these conditions.

A guided approach to manipulation

The capability to discriminate between two haptic stimuli is not just a function of touch sensitivity. While direct touch conveys cutaneous information about local surface features at the points of contact, probing movements (such as in hypertonia assessments) convey information mostly from proprioceptors [38]. The manner in which a user chooses to actively probe can influence the

information gained and quality of the assessment. Visually impaired individuals use different exploration strategies which allow them to gain richer information content, and when sighted subjects are required to use similar strategies, their performance is radically improved [39]. Our haptic display is therefore a valuable tool to explore the haptic sensation of different neuro-mechanics and to determine the most informative probing motion(s) to be used in order to discriminate them.

It should be noted, however, that the number of probing motions to distinguish different abnormalities could be quite large and the search could be long and tedious. This poses a problem, as humans can handle only limited amounts of information at once and their memory is often not reliable. Our current research focuses on the development of an operational research algorithm that searches for the 'most informative probing motion', i.e., the one that produces the most characteristic haptic sensation when interacting with different stereotypical force fields generated by patients.

Autonomous guided vehicles operating in dynamic and unknown environments are often faced with the problem of deciding where to go to get the most relevant information [40]. The dynamic environment is assumed to be unknown, but the automaton has a sensor to measure the environment. The environment can be described using a potential function such as an elastic energy, where maxima of the potential are associated with repulsive forces which represent obstacles to avoid. Minima in the potential correspond to attractive forces representing waypoints.

This framework closely resembles the problem a clinician faces when testing a patient. The total potential, which gradient is composed of the characteristic forces

of the patient's hypertonic arm, is unknown to the clinician (i.e. they need to be assessed). Conversely, the clinician has the capability to 'perceive' the force generated by hypertonic arms, as the autonomous guided vehicles can sense the environment. We wish for the clinician to learn the structure of the patients' total potential by identifying the areas within the subjects' reachable space that are dynamic and the rate of change for these areas. Given the structure of the total potential, we conceive a path for the clinicians to allow them sensing such dynamic areas.

Future directions

Non-robotic assessment techniques

A derivation from the knowledge gained from the studies using our haptic display is an assessment approach in which clinicians and computational algorithms participate in the diagnosis process. The idea is to have clinicians physically interacting with the patient while kinematic and dynamic information is fed into a computational algorithm. The algorithm would suggest optimal manipulation strategies to the clinician for the identification of the impairment. These optimal strategies would consider the non-linear effects of human perception and stereotypical forces produced by the different types of impairment.

An expert clinician's experience should provide an educated guess of what the impairment is. This can be immediately translated into a total potential that can be searched by the informative path planning algorithm, which will suggest to the clinician the best trajectory to follow for the assessment.

Our goal is to close the loop between human experts and a database of stereotypical force fields generated by different types of impairment via a mathematical algorithm, essentially augmenting human perception and interpretation capabilities with computer algorithms. The therapist would be able to select the initial choices of most probable impairments, and the algorithm will suggest a minimum amount of probing for identification. This way, the experience of the therapists will be embedded in a supervised algorithm, where both human and classification algorithms interact. This approach would allow us to identify exploration strategies (easily recognizable in intensity, location, and time course) that maximize exchange of information and can uniquely discriminate among different types of hypertonia. In addition, this would provide clinicians with the capability of testing a patient easily in a routine clinical setting with the ability to support the perceptual findings by quantitative methods. This approach is innovative as it shifts the analysis of impaired behavior from the mere perception of force to the characterization of contact forces, motions, and haptic perception produced in the patient-therapist interaction.

Robotic-aided targeted intervention

Given a set of possible interventions to treat specific impairments, clinicians could analyze their impact directly on the haptic display. For example, two common interventions for the treatment of severe hypertonia in children cerebral palsy are surgery or injections of botulin toxin. The invasiveness of these procedures makes them a one-time intervention. Therefore, careful planning is required. With our haptic display, clinicians could test the biomechanical effects of the different kinds of intervention without even touching the patient. This can be of great help on these cases, helping clinicians to take a more reliable decision and with less margin of error.

Conclusions

We propose a haptic display that can serve as an aiding tool to *i)* synthesize the impairment based on direct measurements of limb mechanics; *ii)* train clinicians in diagnosing different types of abnormal biomechanics, affecting different muscle groups; *iii)* analyze the impact of targeted intervention on specific aspects of limb biomechanics; *iv)* establish and customize a series of most informative probing motions that would maximize the information regarding the impairment so as to assess abnormal biomechanics in a useful, valid, and reliable manner without the need of sophisticated equipment on-site; or *v)* allow the assessment of abnormal biomechanics via remote physical interaction. We expect that this will transform the diagnostic practice by the combination of manual assessment techniques with the computational and mechanical simulation of specific abnormal biomechanics.

Abbreviations

be: bi-articular extensors (triceps long) muscle groups; bf: bi-articular flexors (biceps short); DOF: degrees of freedom; ee: elbow extensors (triceps lateral, anconeus); ef: elbow flexors (biceps long, brachialis, brachioradialis); JND: just noticeable difference; se: shoulder abductors (deltoid posterior); sf: shoulder adductors (deltoid anterior, coracobrachialis, pectoralis major clav).

Competing interests

Part of the work in this manuscript has been used to redact the following provisional patent application: U.S. Patent App. No. 61/991,370, HAPTIC DEVICE AND METHODS FOR ABNORMAL LIMB BIOMECHANICS (2014) to AM-C, DP and FAM-I. JLP has no competing interests. AM-C is a current employee at Hocoma AG, a medical technology company. However, the work presented in this paper has been performed before any relationship between AM-C and Hocoma AG.

Authors' contributions

AM-C and DP contributed equally to this work. AM-C, DP, and FAM-I conceived the idea. AM-C and DP participated in the rendering of the haptic display. DP and AM-C performed the psychophysical experiments. AM-C and DP analyzed the data. AM-C, DP, JLP, and FAM-I contributed the methods and tools. AM-C and DP wrote the paper. All authors read and approved the final manuscript.

Authors' information

AM-C received a B.Sc. degree in Mechatronics and a Pg.Cert in Control Engineering from Tecnológico de Estudios Superiores de Monterrey-Campus Estado de México (ITESM-CEM) in 2006; an M.Sc. degree in Intelligent Autonomous Systems from Aalborg University in 2007, working on the control of powered transfemoral prostheses; and a PhD degree in Bioengineering

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