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Kinesiological Description of Hippotherapy as a Treatment Modality

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ABSTRACT

Hippotherapy as a treatment modality relies on patient-equine dynamic interaction to enhance physical abilities in a range of neuromuscular diseases. The modality takes advantage of external stimulations in the form of kinetic and kinematic inputs to patient's upper body. Current practices and procedures could be greatly enhanced by an objective approach to session planning based on a predictive neuromuscular model. Individualization of the treatment program is both subject-specific and equine-specific.To this effect, kinesiological aspects of the three main upper body flexor-extensor muscles which are directly affected by this treatment modality are presented in a biomechanical model. Events and phases of this dynamic interaction are identified and described using a phase plane analysis. Physical interpretations of coefficients in the movement differential equation illustrates that the proposed approach and mathematical modeling have the potential to be tailored for various musculoskeletal or neuromuscular disorders. Validation results show that the model has the ability to simulate kinematic response and muscle forces of the patient upper body during a hippotherapy session. This predictive ability could provide the therapist with a tool to estimate the effects prior to therapy sessions and choose the most suitable combination of horse and exercises.

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NOMENCLATURE

m	Total mass of UB	K_{RA} & K_{ES}	Linear spring coefficient of Hill muscle model for RA & ES respectively
l	Distance between UB center of mass and COR	K_{TP}	Torsion spring coefficient of Hill muscle model for Psoas
θ_g^*	Initial inclination of UB center of mass (as defined in Figure 1)	C_{RA} & C_{ES}	Linear dashpot coefficient of Hill muscle model for RA & ES respectively
F_{ES}	ES muscle force	C_{TP}	Rotary damper coefficient of Hill muscle model for Psoas
F_{RA}	RA muscle force	x_{RA}^* , x_{ES}^* & θ^*	Active initial length for RA, ES & Psoas muscles respectively
F_p	Psoas muscle force	b_1 , b_2 & r_P	Anthropometric measures as defined in Figure 2

1. INTRODUCTION

Hippotherapy (HT), as a treatment modality (TM) is performed by specifically trained physio/occupational therapists. The observation-based procedures in hippotherapy could be considered as a form of physical manipulation which take advantage of horse movements as well as exercises performed by the patient during horseback riding. The TM targets some of the symptomatic problems faced by patients suffering from cerebral palsy (CP), multiple sclerosis (MS), motor control problems, neuromusculoskeletal (NMS) disorders and a wide range of other dysfunctions [1, 2].

The horse gait at walk exerts repetitive three-dimensional pseudo-sinusoidal mechanical stimuli on the patient, enhancing postural stability and maintenance of balance [3–6]. Spatiotemporal inputs affect the patient upper body (UB) through her/his hip. The UB is

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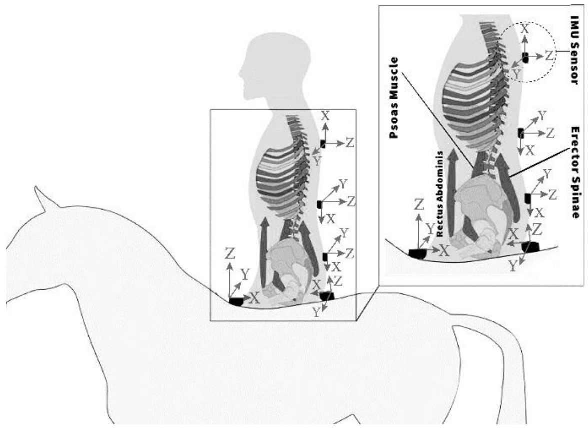

Figure 1. IMU sensors position and UB Flexor/Extensor muscles in sagittal plane

consequently experiencing a complex movement caused by a combination of posterior/anterior pelvic tilts (frontal-transversal plane), rotational pelvic movements (frontal-sagittal plane) and side-flexion (coronal-transversal plane) [7]. The complex motion is very similar in nature to hip movements during normal human walking gait [7, 8]. The modality hence provides a realistic dynamic simulation of hip movements for a patient who is otherwise incapable of autonomous normal gait, while simultaneously stimulating core and UB muscular structures [8, 9]. Here, even small changes in both the horse gait and patient riding position along with a myriad of possible exercises that can be performed during a therapy session, provide the therapist with tremendous opportunities in the manipulation of kinetic/kinematic inputs to patient hip [10].

That is why hippotherapy is used by specifically trained physiotherapists to improve posture, enhance body motion and maintain balance in both children and adult populations [11]. Other physiological effects of HT, such as cardiorespiratory responses and pelvic kinematics have also been addressed in studies on youth with and without cerebral palsy [12]. Although the study results indicated that HT did not effectively affect cardiorespiratory fitness, it is argued that it could facilitate improvement in functional outcomes such as gait, balance and posture. Other studies have investigated the sensorimotor, as well as psychomotor effects, where it is suggested that HT could save and normalize muscle tone for a longer period (up to three months), compared with traditional methods of physiotherapy [13]. Hippotherapy has also been shown to exhibit more beneficial effects, in certain circumstances, than traditional physiotherapy exercises in healthy older adults [14]. The modality has also been used to tackle issues faced by Autistic, Down's Syndrome and other NMS disorders [15–19].

Robotic HT and robotic physiotherapy has recently become a widespread clinical application to facilitate postural core stabilization [20–22]. The study results provide evidence of safety and efficacy of this treatment for postural instability control and sitting balance dysfunction that mitigates the risk of falls in CP. Long-term effects of a more intense robotic HT on a CP patient were also assessed in other studies [23]. Here, higher speeds of horse movement such as trot, canter and gallop as well as walk were simulated and the results showed significant improvement in postural muscle size in addition to improvements in static and dynamic stability.

While most studies provide clinical evidence of HT efficacy in addressing physical disabilities, the literature lacks suggestion of why and how these benefits occur; a point confirmed by the other studies [1]. It has also been shown that direct assessment of patient movements by physiotherapists (using a combined accelerometer-gyroscope device called an actimeter) is important for estimation of therapeutic efficacy [24]. Furthermore, like many other modalities, personalization of treatment is an issue increasingly addressed on the field. Individualization in hippotherapy requires recognition and identification of causal interactions between horse and the patient. The dynamic interactions take place through a series of phases and events, the summation of which, could be shown using state space trajectories for an intended landmark [25]. It, therefore, is essential to understand and be able to explain the events representing kinesiological and biomechanical characteristics of this external stimulus on patient's neuromuscular structures. Dynamic interaction is also significantly affected by characteristics such as equine gait which is in turn determined by the equine physical parameters. The effect of horse pace on spatiotemporal parameters of gait in children with CP and the effect of different types of saddles has also been studied [26, 27]. An extensive review on the subject has also suggested that more studies are required to identify all involved muscles and the degree of activation [1].

The literature on the subject has therefore not addressed a number of fundamental questions. The first issue is that an infrastructure in the shape of a combined musculoskeletal model for the interaction between the two systems is required so that the following questions could be addressed quantitatively and not qualitatively:

1. How does equine walking velocity, gait frequency, changing accelerations, physical parameters such as height, width and length, affect the dynamics of this interaction and how should such parameters be used in a patient specific and personalized approach?
2. How does alternative physical activities, such as sitting position, hand and/or body movements, reaching or throwing exercises, affect dynamic stability or core muscle activity of the patient and

what is the expected effect of each exercise on the patient muscular system?

3. How does the patient disability affect the dynamic interaction? (i.e. what happens when two patients with similar anthropometric parameters, but suffering from different disabilities, ride the same horse?)
4. Is it possible to quantitatively assess the positive or negative effects of this treatment modality during an HT session?

A detailed biomechanical description of the events taking place during this modality could contribute towards wider acceptance amongst mainstream medical professionals [1]. This study should hence attend to these concerns by suggesting a preliminary and expandable mathematical model capable of describing the fundamental biomechanical and dynamic interaction between the horse and patient . The current study intends to portray an image of both the structural and functional characteristics of kinetic/kinematic interactions between the horse and the patient starting from a suitable foundation for mathematical description of movement phases and events.

2.METHODS

2. 1. Musculoskeletal Model The current study adopts a combination of inverse and forward dynamic approach towards musculoskeletal modeling to determine kinematic parameters as well as muscular forces. A quasi-quantitative study of the interaction between the main stabilizing muscles and the UB kinematic behavior during hippotherapy could provide a biomechanical description of this treatment modality. Due to unique situation of the problem, where neither the kinematics of UB nor any of the muscle forces are known, existing available software such as OpenSim cannot be used. The foundational model will include head and neck, arms and hands, the spine, rib cage and scapula as well as pelvic and femoral bone. However, in many hippotherapy sessions, side walkers could restrain the patient's legs to enhance stability. This diminishes the dynamic effect of legs on UB and thus allows for the exclusion of legs in the corresponding preliminary model. It is also assumed that the patient is discouraged from moving his/her upper extremity or head relative to the trunk. The UB could hence be represented by a single link connected through a joint to the pelvic link in this preliminary model.

Three force-bearing members representing the three primary muscle groups connect the upper body and pelvic links as shown in Figure 2. Two of these members represent Rectus Abdominis (RA) and Erector Spinae (ES) substituted by linear muscle models (linear spring-dashpot) and the third, represents Psoas muscle

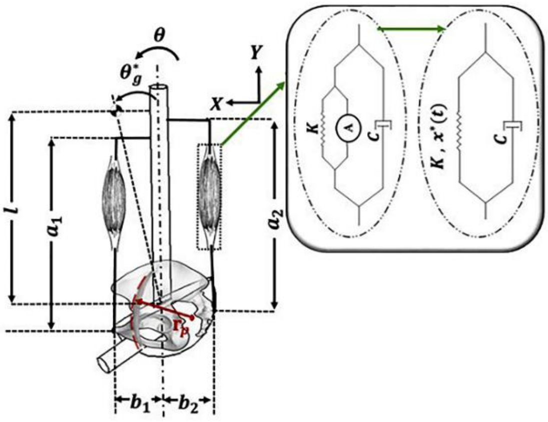

Figure 2. Two-dimensional dynamic model of the patient on sagittal plane

substituted by a rotational muscle model (rotational spring-dashpot).

The motion of the upper body during hippotherapy takes place in sagittal, transverse and frontal planes. The dynamic movements on the transverse plane are limited during hippotherapy and could be neglected at this stage. Linear and angular velocities are small and hence Coriolis acceleration effects could also be neglected. In the absence of Coriolis acceleration effects, it is possible to assume UB dynamics in two sagittal and frontal planes to be independent. To establish and understand the governing mechanisms, the preliminary model would focus on system dynamics in the sagittal plane.

Hill muscle model is used to substitute the three force-bearing members as shown in Figure 2.

The function of the active element $F_A(t)$ is not known at this stage. The effect of this active element is hence combined with spring passive force resulting in Equation (1):

$$F = K(x - x_0) + F_A(t) + C\dot{x} \tag{1}$$

$$Assume\colon\ F_A(t) = Kd^*(t) \tag{2}$$

$$\left.\begin{matrix}(1)\\(2)\end{matrix}\right\} \rightarrow F = Kx - Kx_0 + Kd^*(t) + C\dot{x} \tag{3}$$

$$Assume\colon\ \left(x_0 - d^*(t)\right) = x^*(t) \tag{4}$$

$$\left.\begin{matrix}(3)\\(4)\end{matrix}\right\} \rightarrow F = K(x - x^*(t)) + C\dot{x} \tag{5}$$

where $x^*(t)$ in Equations (4) and (5) is a substitute variable to cater to F_A (t) which was removed from Equation (1). It should be noted that x^* could also be defined as "active initial length" of the muscle. Parameter $d^*(t)$ in Equations (2) to (4) (defined as $F_A(t)/k$) is a purely mathematical step in development of the model.

Solving forward dynamic equations for movement of UB center of mass by utilizing internal muscle forces based on Equation (5), leads to a second order nonlinear differential equation with variable coefficients for θ:

$$A(\theta)\ddot{\theta} + B(\theta)\dot{\theta} + C(\theta)\theta + D(\theta) = 0 \quad (6)$$

$$A(\theta) = \frac{4ml^2}{3} \quad (7)$$

$$B(\theta) = C_{TP} + C_{RA}b_1^2 cos^2\theta + C_{ES}b_2^2 \quad (8)$$

$$C(\theta) = K_{TP} + K_{ES}b_2^2 \quad (9)$$

$$D(\theta) = -K_{TP}\theta^* - ml\big(a_{oy}\sin\big(\theta - \theta_g^*\big)\big) - mla_{ox}\cos\big(\theta - \theta_g^*\big) - mlg\sin\big(\theta - \theta_g^*\big) - K_{ES}b_2x_{ES}^* + K_{RA}b_1(x_{RA}^* + b_1\sin\theta)\cos\theta \quad (10)$$

$$\theta^* = \frac{e_1K_{RA}b_1x_{RA}^*}{K_{TP}} \quad (11)$$

$$x_{ES}^* = \frac{K_{RA}x_1^*b_{RA}(1+e_1)+mgl\sin\theta_g^*}{K_{ES}b_2} \quad (12)$$

$$F_{Rectus\ Abdominis} = F_{RA} = K_{RA}(x_{RA}^* + b_1\sin\theta) + C_{RA}b_1\dot{\theta}\cos\theta \quad (13)$$

$$F_{Erector\ Spinae} = F_{ES} = K_{ES}(x_{ES}^* - b_2\theta) - C_{ES}b_2\dot{\theta} \quad (14)$$

$$F_{Psoas} = F_p = \left({}^{1}/{}_{r_P}\right) * \left(K_{TP}(\theta - \theta^*) + C_{TP}(\dot{\theta})\right) \quad (15)$$

Equation (6) is a nonlinear second order differential equation solved using numerical methods in MATLAB (R2017b).

2. 2.Validation Model validation takes place through a single subject study to obtain upper body kinematic data as well as the horse gait characteristics. This is performed using a combination of inertial measurement units (IMU) and EMG sensors (Figure 3). Horse gait data along with rider personalized measurements were used as inputs to model simulation. Validation takes place through comparisons made between simulation and experimental results.

2. 3. Test Procedure Tests were conducted on a healthy adult (25 years old, 174 cm height, 64kg weight, and BMI=21.14). The procedure was conducted in accordance with basic hippotherapy protocol. Informed consent was obtained and test procedure was approved by university ethics committee on human and animal tests. The experiment took advantage of a walking horse gait for a distance of 20 m in a shuttle test. Linear accelerations and angular velocities in three dimensions were recorded using five IMU sensors (Xsens-Netherland). Three sensors were placed on bony landmarks of S1, T2, and T11 and the other two were

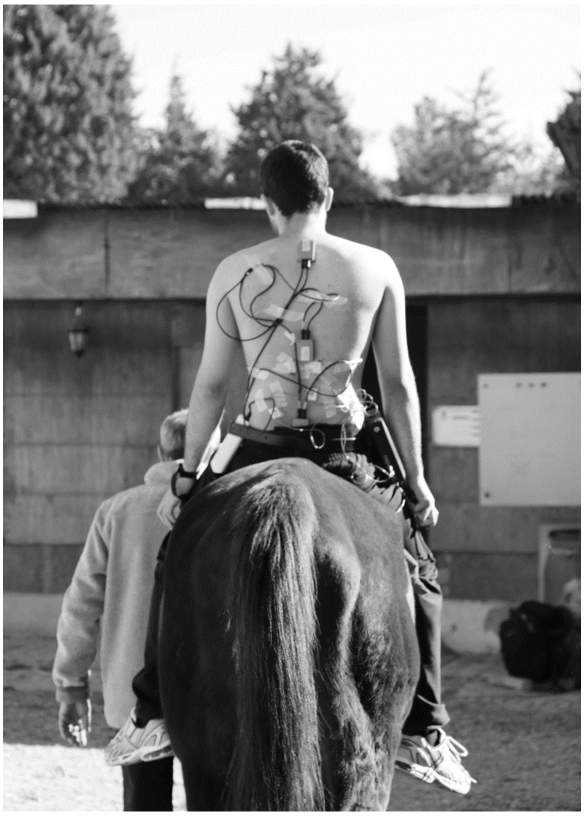

Figure 3. Test subject with IMU and EMG sensors attached

placed on the equine back region just before withers at points immediately in front and behind the rider hip area (Figure 1). Data acquisition was performed at 50 Hz for IMU sensors. The acquired data were transferred to the global coordinate frame using rotation matrices. Artifact elimination was achieved through the application of 4th order low pass filter with a cutoff frequency of 10 Hz [28, 29]

Activation signals associated with RA and ES were recorded at 1000 Hz using an 8-channel electromyography system (Biometrics Ltd). Electrode placement was based on a relevant reference [30]. Forth order low-pass filter was adopted at 20 Hz to eliminate the noise. An audio signal was used for manual synchronization and test initiation.

3. RESULTS

3. 1. Validation and Sensitivity Analysis Results of the validation process for upper body kinematic simulation of the model is presented in Figure 4. Flexion/extension changes of UB center of mass predicted by the simulation as well as the corresponding data measured at T11 landmark (T11 is the closest bony landmark to the UB center of mass) are shown and compared in Figure 4.

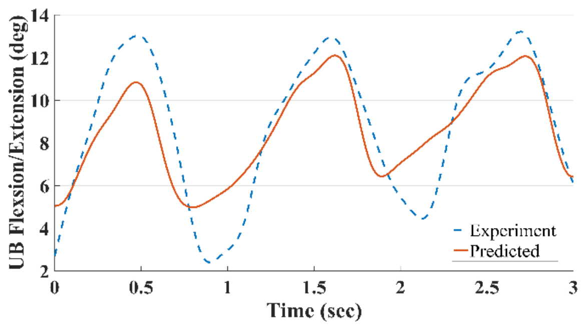

Figure 4. Predicted UB flexion/extension (dashed line) and measured flexion/extension (solid line) of UB center of mass

The sensitivity of the simulation results to uncertainties on the magnitude of muscle parameters is shown in Figure 5. The solid line in the middle shows the predicted normalized UB Flexion/Extension profile in two consecutive gait cycles (initially presumed muscle coefficients k and c). The dashed line shows the same parameter when muscle coefficients were decreased by %50 and the dotted line shows simulation results when muscle coefficients were increased by %50.

The sensitivity of simulation results to uncertainty in geometry (anthropometric measures) is also shown in Figure 6. A quasi-quantitative approach adopted in this study, investigates the profile of changes in kinematic and dynamic parameters, rather than their exact values. Therefore, normalized forms of the parameters are delineated. The normalized form is a dimensionless parameter calculated as: $\frac{Flexion/Extension}{Max_{(Flexion/Extension)}}$. In both figures, normalized predicted UB Flexion/Extension changes in two consecutive gait cycles are illustrated. The solid line in the middle shows simulation results with rider's initial anthropometric data. Dashed line in Figure 6 shows simulation results when there was a 20% increase or decrease in b1 value, while the dashed line shows the simulation results when there was a ±20% uncertainty in b2 value.

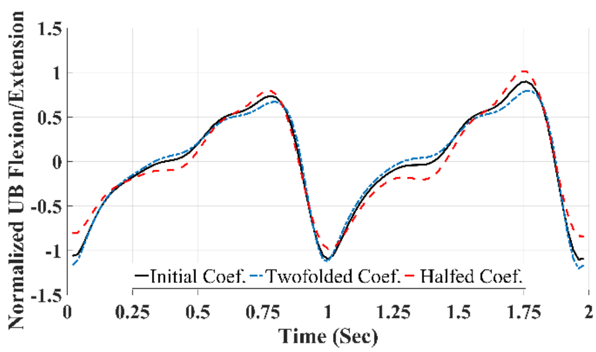

Figure 5. Sensitivity of simulation output to ±%50 changes in muscle properties (k & c). Changes on dimensionless normalized UB Flexion/Extension against time.

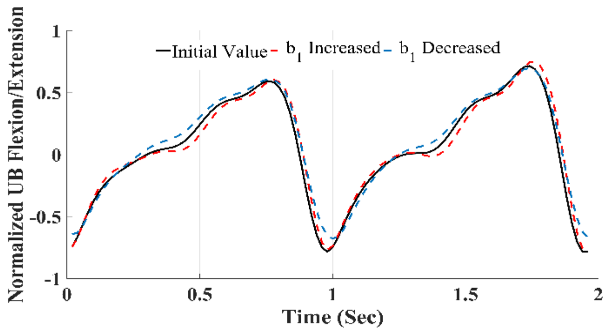

Figure 6. Simulation of the effect of uncertainty in anthropometric measures (b1) upon output results. Changes on dimensionless normalized UB Flexion/Extension against time.

3. 2. Simulation Results Equations (6) to (14) were solved using 4th order Runge-Kutta in MATLAB (R2017b). Input parameters (a_{ox} and a_{oy}) were obtained from the horse gait during the experimental test. Upper body kinematic behavior ($\theta(t)$) and three muscle group forces (F_{RA}, F_{ES}, F_p) in response to horse gait inputs were calculated/predicted. Figure 7 shows the normalized phase plane for rider UB kinematics ($\|\dot{\theta}\|$ Vs. $\|\theta\|$) for one gait cycle of the horse. The dimensionless normalized form is calculated by dividing the parameter's value by it's maximum value in the cycle. Figure 8 illustrates the normalized force of the extensor muscle group (Erector Spinae) against one of the input signals, which is horse linear acceleration in forward direction (a_{ox}). A number of kinematic or kinetic significant points are marked in both Figures 7 and 8 simultaneously to describe and track various events in different movement phases.

4. DISCUSSION

4. 1. Modeling Concerns and Assumptions Currently available software such as OpenSim are designed to solve either forward or inverse dynamic problems. The nature of input stimulation exerted to a single link, limits the adoption of such commercially/open source available software.

Instantaneous center of rotations of individual vertebrae and the associated three dimensional rotations of intervertebral fibrocartilage could be the subject of series of studies. Considering the narrow range of motion in this particular activity (during hippotherapy), where the range of Flexion/Extension of thorax in sagittal plane were less than ±6 degrees (measured at T2, Figure 4), the assumption of fixed center of rotation can be acceptable.

4. 2. Validation and Sensitivity Analysis Validation of simulated kinematic responses is shown in Figure 4. Comparison between predicted changes in UB

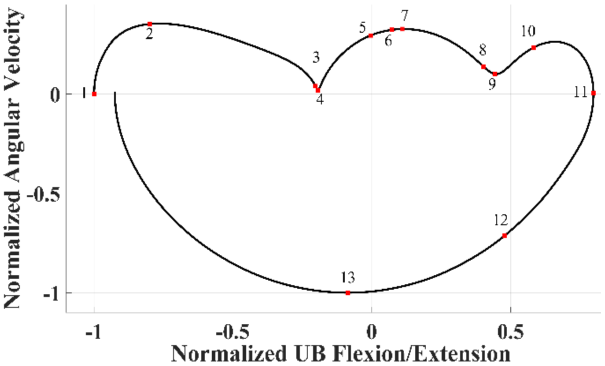

Figure 7. Normalized UB angular velocity Vs. normalized UB angle of the rider in sagittal plane. Both vertical and horizontal axes are dimensionless parameters.

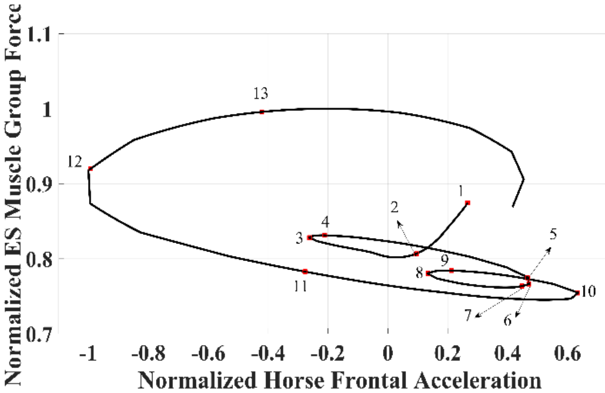

Figure 8. Normalized Erector Spinae force Vs. normalized acceleration of the horse in walking direction. Both vertical and horizontal axes are dimensionless parameters.

angle in the sagittal plane and the actual measured changes of this angle during the test indicates consistencies between predicted and experimental results. Accuracy of the predicted kinematics associated with the preliminary model is not of immediate concern due to deliberate model simplifications. However, similarities in dynamic profiling between the simulation result and empirical measurement shows that the adopted modeling approach provides a sound basis for demonstration of dynamic behavior of the patient's body during HT session. The horse kinematic parameters measured at locations mentioned in Figure 1 shows similar profiles to thoese presented in literature [7].

The sensitivity of the simulation results to changes in muscle model parameters is shown in Figure 5. Although the value of UB angle changes with the change of muscle model parameters, the trend of changes in the profile remains stable. The same phenomenon can be observed in Figure 6, where the sensitivity of the simulation results to uncertainties in anthropometric measures is illustrated. Figures 5 and 6 indicate that although the value of the simulation result is sensitive to model parameters, normalized profile of the results does not alter significantly.

4. 3. Phase Identification Phase planes shown in Figures 7 and 8 could be used to identify dynamic characteristics of individual phases or events. The following provides examples of phase and event descriptions;

From point 1 to 2, UB angle and angular velocity are increasing in the positive direction (Extension in the sagittal plane). The former diminishes the extensor muscle forces (Erector Spinae) due to muscular elastic properties and, the latter diminishes muscle forces due to viscous properties. Therefore, the overall erector spinae muscle force should show depletion in this period as confirmed by the graph in Figure 8.

From point 2 to 3, UB angle is increasing while the angular velocity is falling. The elastic component of muscle force is decreasing while its viscous component is increasing. This causes a functional conflict, resulting in an initial reduction followed by a rise in muscle force as shown in Figure 8.

When the horse is accelerating forward, patient's body inertia causes upper body center of mass to stay behind initially, resulting in an increase in angular velocity in the opposite direction. Forward acceleration of the horse starts to increase from point 3 (Figure 8) and the angular velocity rises with a slight delay from point 4 (Figure 7). The horse's forward acceleration begins to decrease from point 5 (Figure 8) which leads to a downturn of UB angular velocity from point 8 (Figure 7). Similar relations are also observed between horse acceleration and riders UB angular velocity in other phases of movement such as points 8, 10 and 12.

The characteristics outlined so far could portray repeatable and stable rules governing the relationship between UB kinematics and muscle forces. Extraction and identification of these rules could contribute towards estimation of changes in generated muscular forces through observation of upper body kinematics during the equine therapeutic session.

It could, therefore, be concluded that the relationship between patient's upper body kinematics and horse input acceleration is governed by relatively stable rules. It would hence be possible to predict patient's UB kinematics based on horse gait. This could then lead to a determination of muscular function from the predicted UB kinematics.

4. 4. Physical Interpretation of Coefficients Coefficients of the differential equation describing θ, shown in Equation (6) could also provide information on system dynamics. The function A(θ) is associated with the characteristics of UB rotational inertia. This coefficient would change as the UB mass alters in circumstances where a separate weight is carried by the patient or when there are morphological changes such as opening arms during prescribed neuromechanical exercises. Any alteration of this coefficient, assuming others remain unchanged, is an indication of increase or decrease of the amplitude of angular variations or changes in the amplitude of muscular forces.

Function $B(\theta)$ is associated with the viscous properties of muscles and it is dependent upon C_{RA}, C_{ES} and C_{TP} parameters. Variations on viscous characteristics of muscle tissue caused by disease, injury or disability, results in deviations in system dynamics through changes in this coefficient. Therefore, the pathological states could be simulated for an individual patient by adjusting parameters in $B(\theta)$.

The function $C(\theta)$ is related to elastic properties of the rotational spring. In other words, this coefficient is dependent upon the K_{TP} of the Psoas muscle and elastic coefficient and the moment arm of ES muscles. Any changes in this coefficient affects the relationship between muscle force and UB angle, thus influencing the range of forces and stability of the system. Rotational spring characteristics representing ES muscles ($K_{ES}\ b_2^2$) in $C(\theta)$, are influenced by the length of the moment arm, the radius of curvature of Sacrum as well as elastic characteristics of muscles. In a similar way, the characteristics of rotational spring representing Psoas muscles are affected by the modulus of elasticity as well as anatomic dimensions such as the angle or radius of curvature of the muscle (r_P).

$D(\theta)$ is a function of: a) linear acceleration inputs from the horse, b) elastic properties of RA and ES muscles, and c) active initial length of the muscles (x^*) . Therefore any changes in any of these three parameters could lead to changes in system behavior by affecting $D(\theta)$. One interesting implication of this function is that linear acceleration inputs could be altered to assist personalized prescriptions leading to improved efficiency of Hippotherapy sessions. The inputs could, for example, change by changing the horse gait. The new gait could be the result of choosing a particular horse for a certain patient. Some diseases or impairments may have the ability to cause variation in the muscular elasticity and thus change $D(\theta)$. Changes in the initial muscle length could be voluntary by stiffening the muscle or could be caused by a disease or disorder like spasticity, muscular damage, or fatigue. All such causes, change function $D(\theta)$ in Equation (6) which in turn affects system behavior. Finally, any change in $\theta(t)$ results in variation of the system behavior and muscle forces as indicated by Equations (13), (14) and (15).

5. CONCLUSION

American Hippotherapy Association (AHA) defines hippotherapy as a purposeful manipulation of equine movements to achieve functional outcomes. This study provides the basis for a musculoskeletal modeling approach, which has the capacity for both patient and equine specificity. Validation results show that the resulting simulation has the ability to predict kinematic response and muscle forces of the patient's upper body during a hippotherapy session. This predictive ability could provide the therapist with a tool to estimate the outcome, prior to therapy sessions and thus provide a choice of the most suitable combination of horse and exercises. Physical interpretations of coefficients in the movement differential equation (Equation (6)) illustrates that the proposed approach to mathematical modeling has the potential to adapt to various musculoskeletal or neuromuscular disorders.

The current study has a number of limitations requiring attention in the future efforts:

- The current simulation is based on a 2D model on sagittal plane, whereis the the actual movement is in 3D plane and more muscles are involved.
- Changes in muscle activation due to stability control mechanisems such as muscle reflexes (streatch reflex, golgi tendon reflex and vestibulospinal reflex) are not simulated in this model.
- Upper extrimities (arms and hands) and head and neck are assumed to not move relative to the trunk in this model. This assumption can only simulate the most basic form of hippotherapy practice.
- In the current model, the whole abdomen and thorax together is modelled as one rigid link with a fixed joint against pelvis. In wider ranges of trunk Flexion/Extension, this assumption can lead to unacceptable simulation errors.

This article has presented a kinesiologically based mathematical model which describes the fundamental dynamic interaction between the horse and the patient during hippotherapy. The model could provide a sound basis for future studies of this highly complex treatment modality.

6. CONFLICT OF INTEREST STATEMENT

The authors certify that they have NO affiliations with or involvement in any organization or entity with any financial interest, or non-financial interest in the subject matter or materials discussed in this manuscript.

7. ACKNOWLEDGEMENT

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Persian Abstract

چکیده

هیپوتراپی (اسب درمانی) به عنوان یک روش درمانی، بر تعامل دینامیکی اسب-بیمار تکیه می‌کند، تا توانایی‌های جسمانی را در گستره‌ای از بیماری‌های عصبی- عضلانی ارتقاء بخشد. این روش درمانی از تحریک خارجی که به صورت ورودی سینتیک و سینماتیک بر بالاتنه‌ی فرد بیمار وارد می‌شوند بهره می‌برد. درمان‌ها و پروسه‌های درمانی که در حال حاضر انجام می‌شوند را می‌توان با یک رویکرد عینی برای برنامه‌ریزی جلسه‌درمانی بر مبنای مدل پیش‌بینی کننده‌ی عصبی-عضلانی، به مقدار زیادی ارتقاء بخشید. شخصی‌سازی برنامه‌ی درمانی برای هر بیمار خاص و بر اساس هر اسب خاص انجام می‌گردد. بدین منظور، ویژگی‌های کینزیولوژیکی سه عضله/گروه عضلانی اصلی بالاتنه (فلکسور-اکتنسور) که بصورت مستقیم در این روش درمانی دخیل بوده‌اند در مدل بیومکانیکی درنظر گرفته شده است. رویدادها و فازهای مختلف این تعامل دینامیکی مشخص شده و به کمک تحلیل نمودار صفحه فازی توصیف شده‌اند. تفسیر ضرایب معادلات دیفرانسیلی حرکت نشان می‌دهد که رویکرد پیشنهادی و مدل‌سازی ریاضی، پتانسیل آن را دارد که متناسب با طیف وسیعی از بیماری‌های اسکلتی-عصبی-عضلانی تنظیم گردد. نتایج اعتبارسنجی نشان می‌دهد که مدل ارائه شده توانایی شبیه‌سازی پاسخ سینماتیکی و نیروی عضلات بالاتنه بیمار را در طی یک جلسه اسب درمانی دارد. این توانایی پیش‌بینی، می‌تواند ابزاری را در اختیار درمانگر قرار دهد که به کمک آن اثرات درمان را قبل از اجرای جلسه‌ی درمانی تخمین بزند و به این ترتیب مناسب‌ترین ترکیب از اسب و تمرینات فیزیکی را انتخاب نماید.